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Evaluation of Auditory Evoked Potentials as a Hearing aid Outcome Measure

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by

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Abstract

This thesis aimed to explore the applicability of Cortical Auditory Evoked Potentials (CAEPs) and Envelope Following Responses (EFRs) as objective aided outcome measures for use in infants wearing hearing aids. The goals for CAEP-related projects were to evaluate the effect of speech stimulus source on CAEPs, non-linear hearing aid processing on tone-evoked CAEPs, and the effect of inter-stimulus intervals on non-linear hearing aid processing of phonemes. Results illustrated larger amplitude CAEPs with shorter latencies for speech stimuli from word-medial positions than word-initial positions, and no significant effect of the tone burst onset overshoot due to non-linear hearing aid processing. Inter-stimulus intervals in CAEP protocols resulted in significantly lower aided phoneme levels compared to when they occurred in running speech, illustrating potential inaccuracies in representation of relevant hearing aid function during testing.

The major contribution of this thesis includes the proposal and validation of a test paradigm based on speech-evoked EFRs for use as an objective aided outcome measure. The stimulus is a naturally spoken token /susaʃi/ modified to enable recording of eight EFRs from low, mid and high frequency regions. The projects aimed to evaluate previously recommended response analysis methods of averaging responses to opposite polarities for vowel-evoked EFRs as well as sensitivity of the proposed paradigm to changes in audibility due to level and bandwidth in adults with normal hearing and additionally, due to amplification in adults with hearing loss. Results demonstrated a vowel-specific effect of averaging opposite polarity responses when the first harmonic was present, however the averaging did not affect detection in the majority of participants. The EFR test paradigm illustrated carrier-specific changes in audibility due to level, bandwidth and amplification suggesting that the paradigm may be a useful tool in evaluating unaided and aided audibility, and therefore appropriateness of hearing aid fittings. Further validation is necessary in infants and children wearing hearing aids.

In conclusion, CAEPs and EFRs vary in strengths and limitations, and therefore it is likely that a combination of measures may be necessary to address the variety of hearing disorders seen in a typical audiological caseload.

Keywords: Objective outcome measures, aided auditory evoked potentials, Envelope Following Responses (EFR), Cortical Auditory Evoked Potentials (CAEP), natural speech, vowel, fricative, polarity, rise-time, non-linear hearing aid

List of Abbreviations

AAA	:	American Academy of Audiology
ABR	:	Auditory Brainstem Response
AEP	:	Auditory Evoked Potential
AM	:	Amplitude-Modulated
ANSD	:	Auditory Neuropathy Spectrum Disorder
ANSI	:	American National Standards Institute
ASSR	:	Auditory Steady-State Response
BTE	:	Behind-The-Ear
cABR	:	Complex Auditory Brainstem Response
CAEP	:	Cortical Auditory Evoked Potential
CI	:	Confidence Interval
dB	:	Decibel
DAI	:	Direct Audio Input
DFT	:	Discrete Fourier Transform
DSL	:	Desired Sensation Level
EEG	:	Electroencephalogram
EFR	:	Envelope Following Responses
FA	:	Fourier Analyzer
FBW	:	Full Bandwidth
FDR	:	False Discovery Rate
FFR	:	Frequency Following Response
HL	:	Hearing Level
ICC	:	Intraclass Correlation Co-efficient
IEC	:	International Electrotechnical Commission
ISTS	:	International Speech Test Signal
ITU	:	International Telecommunication Union
JCIH	:	Joint Committee on Infant Hearing
LPF	:	Low-pass filter
MUSHRA	:	MUltiple Stimuli with Hidden Reference and Anchor
RAU	:	Rationalized Arcsine Units
RM–ANOVA	:	Repeated Measures Analysis of Variance
RMS	:	Root-Mean-Square
SD	:	Standard Deviation
SE	:	Standard Error
SL	:	Sensation Level
SNR	:	Signal-to-Noise Ratio
SPL	:	Sound Pressure Level
UWO DFD	:	University of Western Ontario Distinctive Feature Differences test
UWO PedAMP	:	University of Western Ontario Pediatric Audiological Monitoring Protocol
WDRC	:	Wide Dynamic Range Compression

The Co-Authorship Statement

This thesis comprises of an introductory chapter (Chapter 1), six integrated manuscripts and a concluding chapter (Chapter 8). Chapters 2 to 4 have been published (Easwar, Glista, Purcell and Scollie, 2012; Easwar, Glista, Purcell and Scollie, 2012; Easwar, Purcell and Scollie, 2012) and Chapter 5 has been submitted for peer review (Easwar, Beamish, Aiken, Choi, Scollie and Purcell, submitted). I, Vijayalakshmi Easwar, am responsible for conception and design of this work, data collection and organization (except Experiment I of Chapter 5), data analyses and interpretation of results. I, Vijayalakshmi Easwar, am the primary author of all chapters. I am the sole author of the introductory and concluding chapters. These chapters were reviewed by Dr. David Purcell and Dr. Susan Scollie prior to inclusion in the thesis. Chapters 2 and 3 were co-authored by Dr. Danielle Glista, Dr. David Purcell and Dr. Susan Scollie. Dr. Susan Scollie and Dr. David Purcell provided guidance on project design and Dr. Susan Scollie provided guidance on statistical analysis. Dr. Danielle Glista and Dr. Susan Scollie provided guidance on obtaining ethics approval. All co-authors reviewed both manuscripts prior to submission. Chapter 4 was co-authored by Dr. David Purcell and Dr. Susan Scollie, who provided guidance on project design and statistical analysis, and reviewed the manuscript prior to submission. Chapter 5 was co-authored by Laura Beamish, Dr. Steven Aiken, Dr. Jong Min Choi, Dr. David Purcell. Laura Beamish collected data and completed response analysis for Experiment I. Dr. Steven Aiken and Dr. David Purcell provided guidance on stimulus design. Dr. Steven Aiken, Dr. David Purcell and Dr. Jong Min Choi provided tools for response analysis. Dr. David Purcell provided guidance on statistical analysis. All co-authors reviewed the final manuscript prior to submission. Chapter 6 and 7 were co-authored by Dr. David Purcell, Dr. Steven Aiken, Dr. Vijay Parsa and Dr. Susan Scollie. Dr. David Purcell, Dr. Steven Aiken and Dr. Susan Scollie provided valuable inputs on stimulus and experimental design. Dr. Vijay Parsa provided tools and guidance for the sound quality rating measure. Dr. David Purcell, Dr. Steven Aiken and Dr. Vijay Parsa provided tools for response analysis. Dr. Susan Scollie and Dr. David Purcell provided support on statistical analysis.

Dedication

To my dearest family

Acknowledgments

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Chapter 1

Outcome measures in infants wearing hearing aids

1.1 Early identification, current trends and impact

The goal of early intervention programs is to diagnose permanent hearing losses by three months of age and provide intervention services such as hearing aid fitting by six months of age (Joint Commitee on Infant Hearing [JCIH], 2007). The implementation of newborn hearing screening programs has reduced the age of identification of hearing loss and thereby age of intervention significantly (Sininger et al., 2009; Uus & Bamford, 2006), the median age of fitting now ranging between four (California, USA: Sininger et al., 2009, England, UK: Uus & Bamford, 2006) and six months (Ontario, Canada: Bagatto, Scollie, Hyde, & Seewald, 2010, NewYork, USA: Spivak, Sokol, Auerbach, & Gershkovich, 2009). With widespread implementation of early intervention programs, emphasis on measures to evaluate the impact of hearing aid fittings in clinical practice has increased. This thesis focused on the evaluation and development of a subset of such measures.

1.2 Outcome measures

1.2.1 Need for outcome measures

Hearing evaluation and intervention such as hearing aid fitting is an individualized process. The process of hearing aid fitting involves selection and verification of hearing aid output to match prescribed targets for a given hearing loss (American Academy of Audiology [AAA], 2013; JCIH, 2007). Hearing evaluation using behavioral test methods is a challenge in young infants and therefore hearing aids are usually fitted based on estimated hearing thresholds (Bagatto et al., 2005). Estimated hearing thresholds are derived from tone burst Auditory Brainstem Responses (ABR) or Auditory Steady-State Responses (ASSR) thresholds using frequency-specific correction factors (Bagatto, 2008; King, 2010). These correction factors, based on the mean difference between ABR/ASSR and behavioral thresholds (Rance et al., 2005; Stapells, 2000; Stapells, Picton, Durieux-Smith, Edwards, & Moran, 1990), improve the accuracy of the fitting process substantially. However, inaccuracies may persist due to individual variations associated with these differences (Picton, 2011), and the difficulty differentiating severe and profound degrees of hearing loss (Stapells, 2000). Outcome measurement could therefore play an important role in asserting the accuracy of hearing evaluation and hearing aid fitting. Outcome evaluation is one of the main components recommended as a part of the hearing aid fitting process (AAA, 2013; Bagatto et al., 2010; JCIH, 2007). Timely evaluation of hearing aid benefit, or lack thereof, could lead to early and improved intervention for an infant with hearing loss, thereby improving effectiveness of early intervention programs.

1.2.2 Types of outcome measures

Several outcome measures have been proposed to evaluate the benefit of amplification in infants wearing hearing aids. These may be placed on a continuum ranging from subjective to objective measures. Subjective outcome measures involve behavior observation by parents or guardians of infants wearing hearing aids. An example of a subjective outcome measurement protocol is the University of Western Ontario Pediatric Amplification Protocol (PedAMP; Bagatto et al., 2011). The PedAMP is an evidence-based clinical protocol that assesses auditory behavior using parental questionnaires. Objective outcome measures, on the other end of the continuum, refer to tests that estimate hearing acuity while requiring minimal participation from the infant and the parent. An example of an objective aided outcome measure is the Cortical Auditory Evoked Potential (CAEP; Golding et al., 2007; Purdy et al., 2005). In general, most Auditory Evoked Potentials (AEPs) are considered objective methods because they do not require active participation of the individual undertaking the test. Other outcome measures such as aided speech detection or recognition measures fall between the two extremes of the continuum. Although they are measured objectively, they require the patient's behavioral responses. In this thesis, the term objective outcome measures refers to AEP tests that are used as an outcome measure.

1.2.3 Subjective, objective, or both?

Subjective and objective outcome measures attempt to evaluate aided auditory performance. A comparison of the advantages and limitations of each type of measure could help highlight the unique information each measure could offer (adapted from Andresen, 2000):

- **Nature of information provided:** Objective outcome measures indicate transmission of specific stimuli to specific generators in the auditory system whereas subjective outcome measures reflect behavioral performance of an infant, in response to auditory stimuli in a real world context. Therefore, information provided by objective measures is relatively fine-grained whereas information provided by subjective measures is relatively holistic, when the two measurement types are compared.

- **Bias in information obtained:** A primary advantage of objective measures is that they are not subject to bias, and observation and reporting skills of an interviewee, as would be the case for subjective measures. In objective measures, tester bias may also be reduced by the use of automated or statistical response detection methods.

- **Timing:** Since objective measures do not rely on a child's behavioral response repertoire, they could provide earlier indications of hearing aid benefit compared to subjective outcome measures.

- **Ecological validity:** Protocols using objective measures use a representative sample of speech stimuli, dictated by the type of AEP and at times, available test time. While use of specific stimuli reduces ambiguity in interpretation, speech stimuli may not necessarily be presented in the form encountered by the infant in the real world. Since non-linear hearing aids are sensitive to the nature of input stimuli (e.g., Scollie & Seewald, 2002), ecological validity of such measures may also be dependent on the hearing aid. Subjective outcome measures evaluate the infants performance in their natural listening environment and therefore this may not be of concern.

- **External validity:** The specific nature of stimuli in objective measures may limit generalization to the range of speech sounds that vary in level and frequency. External validity may not be a concern for subjective measures, because they reflect an infant's performance based on audibility of a range of speech sounds present in their environment. This type of distinction could be compared with capacity versus performance-level measurements in the World Health Organization's International Classification of Functioning, Disability and Health (World Health Organization, 2002). Capacity measures are made under controlled conditions, whereas performance measures are assessed in real world conditions. Both types of measures have inherent value, but differ in external validity.

- **Practical considerations:** Subjective outcome measures present multiple advantages such as ease of administration, lower administration costs and time, and minimal co-operation from the infant during clinic visits. In contrast, objective outcome measures include the need for specific recording equipment, knowledge of equipment use, and co-operation of the infant to be awake/asleep during testing. Language of administration could be a barrier for questionnaires, but may be overcome with an interpreter.

- **Applicability across hearing disorders:** An advantage of using subjective measures is its application in different hearing disorders such as those with Auditory Neuropathy Spectrum Disorder (ANSD[1], e.g., Bagatto et al., 2011). Utility of objective outcome measures may be limited by the nature of the presenting hearing disorder. For example, in the case of ANSD, CAEPs may be useful in demonstrating hearing aid benefit (Rance, Cone-Wesson, Wunderlich, & Dowell, 2002), while AEPs such as ABR and ASSR have limited utility (Rance, 2005; Rance et al., 2005).

Based on the above factors, it is evident that a limitation in one type of measure may be compensated by the other type of measure. The primary motive of outcome

[1]ANSD refers to a type of hearing disorder with normal outer hair cell function but disordered inner hair cells or afferent neural pathway (see Rance, 2005 for a review)

evaluation is to identify areas of sub-optimal performance to optimize intervention. This may not be satisfied with either type of measure exclusively. In conclusion, valuable information from both objective and subjective measures is likely to complement and supplement each other to help maximize the utility of outcome measures.

1.3 Objective aided outcome measures

Objective measures suggested for aided evaluation (in isolation or in conjunction with other measures) include: (1) CAEPs (Carter, Dillon, Seymour, Seeto, & Van Dun, 2013; Golding et al., 2007; Purdy et al., 2005; Van Dun, Carter, & Dillon, 2012); (2) complex ABRs (cABRs; Anderson & Kraus, 2013); (3) Envelope Following Responses (EFRs) using modulated tones (ASSRs, Picton et al., 1998) and natural speech (Choi, Purcell, Coyne, & Aiken, 2013; Dajani, Heffernan, & Giguere, 2013); and, (4) Frequency Following Responses (FFR; Aiken & Picton, 2006; Dajani et al., 2013). The distinction between FFRs and EFRs is the stimulus *characteristic* that the auditory system follows; EFR follows the stimulus envelope whereas FFR follows the stimulus fine structure.

AEPs can be broadly classified into transient and steady-state responses (Picton, 2011). Transient responses are onset (or offset) responses, representing summed synchronous firing of neurons to the stimulus onset (or offset). Steady-state responses occur in response to an ongoing, periodically changing stimulus. Of the four AEPs above, CAEPs could be categorized as transient potentials whereas EFRs and FFRs could be categorized as steady-state potentials. cABRs elicited by speech sounds such as /da/, are a combination of transient and steady-state potentials (Skoe & Kraus, 2010).

1.3.1 Comparative evaluation of objective aided outcome measures

Interest in using AEPs to assess hearing aid function has prevailed since the 1960s (e.g., Rapin & Graziani, 1967). Previous studies have evaluated numerous aspects to illustrate the suitability of AEPs for aided evaluation. The framework shown in Figure 1-1 integrates multiple factors deemed important for aided applications. Although these factors are categorized into three sections in an apparent unidirectional way, there is considerable inter-dependency of factors across categories.

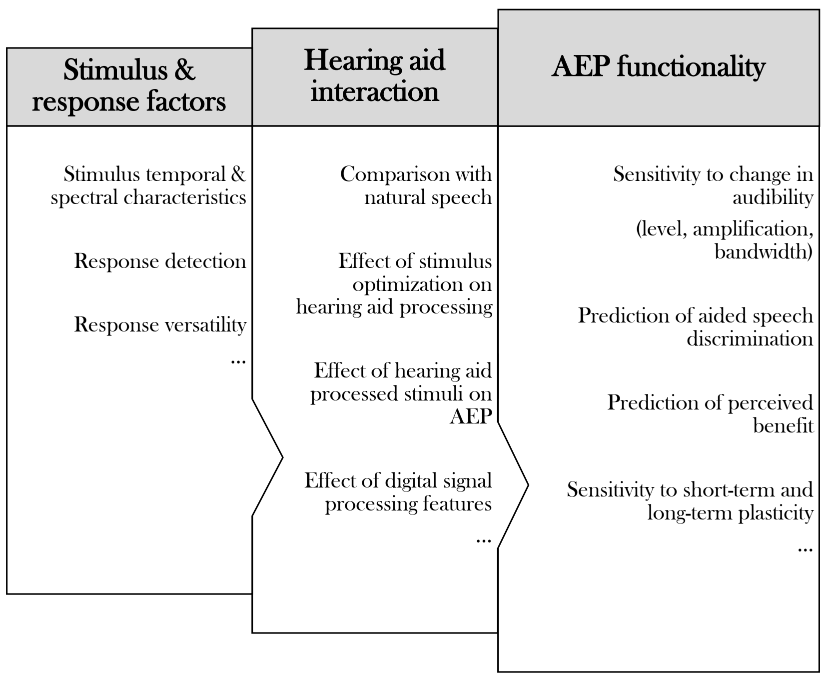

Figure 1-1: Framework for evaluation of AEPs for use as an objective aided outcome measure.

The first category entails stimulus factors, response factors and stimulus-response contingency factors. Stimulus choice, characteristics and presentation are contingent upon the nature of the responses. The nature of responses also dictates response robustness, generator sites, response analysis methods, statistical detection of responses, types of hearing disorders that it can be used to evaluate and more. The next main category of factors to consider is hearing aid interaction for the AEP stimuli. This may include analysis of hearing aid or stimulus variables that may affect representation of hearing aid function during testing or that may influence presumed stimulus-response relationships. The final category of factors is functionality of AEPs. This may include the ability of an AEP to represent changes in audibility due to level, bandwidth, and amplification. In addition, prediction of other aided outcome measures such as speech discrimination and perceived benefit using questionnaires, and sensitivity to short-term and long-term plasticity changes could also be of interest. The three main categories of factors (Figure 1-1) guided the rationale and structure of projects in Chapters 2 to 7. The remainder of this section provides an overview comparison of AEPs across framework factors that are within the scope of the projects in this thesis.

Category I: Stimulus and response factors

Stimulus

Several reasons support the use of speech stimuli. Firstly, speech-evoked AEPs are likely to have higher face and ecological validity compared to non-speech AEPs because the auditory environment of humans is dominated by speech sounds. Secondly, the auditory system is better adapted to frequently occurring stimuli such as speech by the neuro-developmental process of pruning (review by Picton & Taylor, 2007). Thirdly, external validity is likely improved because non-linearity in the auditory system makes it challenging to predict responses to speech from responses to non-speech stimuli such as tones or clicks (review: Palmer & Shamma, 2004). The fourth reason relates to hearing aid processing (further discussed under Category II: Hearing aid interaction).

AEP protocols have used naturally spoken speech stimuli (e.g., EFRs: Choi et al., 2013, CAEPs: Golding, Purdy, Sharma, & Dillon, 2006; Tremblay, Billings, Friesen, & Souza, 2006) as well as synthetic speech stimuli (e.g., cABR: see review by Skoe & Kraus, 2010, CAEP: Ceponiené, Alku, Westerfield, Torki, & Townsend, 2005, FFR: Krishnan, 2002), some of which are meaningful words (e.g., EFR: Choi et al., 2013, CAEP: Wunderlich, Cone-Wesson, & Shepherd, 2006). Naturally spoken speech stimuli for aided CAEP protocols have been extracted from either running speech (e.g, Golding et al., 2007, 2006) or isolated utterances (e.g., Tremblay, Billings, et al., 2006; Tremblay, Kalstein, Billings, & Souza, 2006). Characteristics of speech stimuli may vary by context due to co-articulation and therefore knowing the effect of stimulus source on stimulus and response characteristics could guide stimulus selection. The effect of word position of phonemes used as CAEP stimuli is evaluated in Chapter 2.

Response characteristics

As briefly discussed before, transient and steady-state responses reflect different processes underlying neural representation of sound. Transient AEPs indicate transmission of a stimulus to its generator site whereas steady-state AEPs additionally indicate how a stimulus characteristic is represented (Clinard & Tremblay, 2013; Dajani et al., 2013). CAEPs are generated from the primary and secondary auditory cortices (review by Hyde, 1997; Martin, Tremblay, & Stapells, 2007) whereas cABRs elicited by syllables (Chandrasekaran & Kraus, 2010), and EFRs elicited by vowels at the fundamental frequency of voice (f_0) have origins in the upper brainstem (Herdman et al., 2002; Purcell, John, Schneider, & Picton, 2004). Although it may be advantageous to use AEPs that reflect processing up to the higher centers of the auditory system, factors such as attention and sleep are more likely to influence such measures. AEPs belong on a continuum of exogeneity and endogeneity that parallels their latency relative to stimulus onset (Rugg & Coles, 1995). The majority of studies report that earlier occurring AEPs from the brainstem are largely exogenous or stimulus-driven and therefore unaffected by sleep and attention (ABR: Campbell & Bartoli, 1986; Deacon-Elliott, Bell, & Campbell, 1987; EFR modulation rates near 80 Hz: Purcell et al., 2004). Although later occurring CAEPs are also stimulus dependent (e.g., Agung, Purdy, McMahon, & Newall, 2006), arousal state (sleep and attention) can have significant effects on the elicited CAEP (Campbell & Colrain, 2002; Thornton, Harmer, & Lavoie, 2007). Since the population of interest for the application of aided AEPs is infants, it is likely preferable if the AEP is less affected by sleep because myogenic artifacts are problematic in awake infants. Additionally, maturation of later-occurring responses from higher centers in the auditory system have a longer trajectory (Moore & Linthicum, 2007). Another factor impacting clinical application is response versatility. Responses that require higher neural timing precision (e.g., ASSRs or ABRs) are less suitable for evaluation of individuals with ANSD relative to CAEPs (Rance, 2005).

Optimization of AEP stimulus and response analysis is an ongoing process in improving clinical usability of AEPs. Averaging responses elicited by both polarities has been recommended to reduce stimulus artifacts (Aiken & Picton, 2008; Skoe & Kraus, 2010; Small & Stapells, 2004), and to enhance EFRs (Aiken & Picton, 2008). A recent study on vowel-evoked EFRs illustrated a polarity-sensitive amplitude difference in EFRs (Aiken & Purcell, 2013) that could underestimate the best response amplitude and therefore affect detection. This questions the appropriateness of using previously suggested averaging across polarities (Aiken & Picton, 2008) and therefore requires further investigation to assess generalization of findings. The effect of polarity on vowel-evoked EFRs is investigated in Chapter 5.

In summary, the focus of Category I of factors is related to stimulus-response relationships and optimization of stimulus and response-related factors. The following category discusses stimulus-hearing aid interaction factors and optimization of stimulus parameters from a hearing aid signal processing perspective.

Category II: Hearing aid interaction

Effect of stimulus choice on hearing aid processing

Objective aided outcome measures involve measurement of AEPs while stimuli are transduced through a hearing aid. In essence, the hearing aid acts as a relay station because the elicited AEP is dependent on the hearing aid output. Choice of stimuli is therefore vital to valid aided AEP testing. Hearing aid interaction is an important factor to consider because non-linear features in hearing aids such as compression may respond differently to non-speech inputs that are not close representations of speech in terms of spectral composition, crest factor and temporal features (e.g., Henning & Bentler, 2005; Scollie & Seewald, 2002; Stelmachowicz, Lewis, Seewald, & Hawkins, 1990).

Application of objective measures in threshold estimation has led to increased emphasis on frequency specificity of stimuli, but these stimuli may not be preferred to predict hearing aid processing for speech. Since compression achieved depends on the duration of stimuli and its temporal characteristics (Stone & Moore, 1992), the compression ratio for speech is often lower than that for static pure tones (Fortune, 1997; Henning & Bentler, 2008; Stelmachowicz, Kopun, Mace, & Lewis, 1996). The aided output of non-speech weighted pure tones could differ as much as 20 dB relative to that of real speech (Scollie & Seewald, 2002). The largest discrepancies occur at the higher frequencies and at higher input levels for non-linear hearing aids

(Henning & Bentler, 2005; Stelmachowicz et al., 1996). These discrepancies are greater for high power hearing aids, typically prescribed for profound degrees of hearing loss (Scollie & Seewald, 2002), longer release times, higher numbers of channels and high compression ratios (Henning & Bentler, 2005). Additional features such as noise reduction may differentially process speech and non-speech-like inputs, and de-emphasize non-speech-like stimuli (Bentler & Chiou, 2006; Scollie & Seewald, 2002). In short, stimuli with closer resemblance to natural speech in temporal characteristics will likely ensure a better representation of hearing aid functioning for speech (experienced in everyday listening conditions) during the test.

Effect of current AEP stimulus protocols on hearing aid function

Brief stimuli such as clicks or tone bursts used for ABRs are deemed unsuitable for aided applications due to inaccuracies in the representation of hearing aid function, especially for stimuli-like speech (Beauchaine, Gorga, Reiland, & Larson, 1986; Brown, Klein, & Snydee, 1999; Frye, 1987; Gorga, Beauchaine, & Reiland, 1987; Stelmachowicz et al., 1990). Speech syllables used for cABRs are likely more suitable stimuli as they are longer than clicks and tone bursts (e.g., Anderson & Kraus, 2013). Phonemes or syllables extracted from running speech or a syllable/word have been used as CAEP stimuli (Golding et al., 2006; Tremblay, Billings, et al., 2006). Although CAEP protocols use natural speech, the segments of speech used as stimuli are not used in their natural form. CAEP protocols typically intersperse stimulus phonemes with 1–2 seconds of inter-stimulus interval. In contrast, in running speech, phonemes are mostly interspersed with other amplitude-varying phonemes. Therefore, input-level-dependent non-linear hearing aids may or may not process the same phoneme similarly in the isolated CAEP context versus running speech. A discrepancy in output levels of the same phoneme in the two contexts may imply that hearing aid function during CAEP testing is not representative of hearing aid function during natural speech. The accuracy of representation of hearing aid function during CAEP testing is investigated in Chapter 4.

Additionally, hearing aid processing may alter stimulus characteristics in one or more ways that may affect the AEP recorded (Billings, Tremblay, & Miller, 2011; Billings, Tremblay, Souza, & Binns, 2007; Jenstad, Marynewich, & Stapells, 2012; Marynewich, Jenstad, & Stapells, 2012). Hearing aid noise floor leading to poorer Signal-to-Noise Ratio (SNR) in aided conditions (Billings et al., 2011, 2007), and the increase in tone burst rise-time leading to inaccurate representation of hearing aid gain at stimulus onsets (Jenstad et al., 2012; Marynewich et al., 2012; Stapells, Marynewich, & Jenstad, 2010) have been cited as barriers to using CAEPs as a valid objective aided outcome measure. With linear processing in digital hearing aids, changes in stimulus rise-time resulted in lower gain at stimulus onset relative to the gain achieved during verification using pure tones (Jenstad et al., 2012). Studies investigating the effect of hearing aid processing on stimulus onset and CAEPs are limited to linear hearing aids. The changes caused by linear hearing aid processing may differ from stimulus changes caused by non-linear hearing aid processing. Since most hearing aids used today are non-linear (Dillon, 2012; Johnson, Cox, & Alexander, 2010), recognizing stimulus changes caused by non-linear processing and the effect of such changes on CAEPs is important and warrants further investigation. The effect of non-linear hearing aid processing on tone burst CAEPs is investigated in Chapter 3.

Speech-evoked EFRs and FFRs may prove to be advantageous because the stimulus could take the form of a sentence similar to natural running speech. EFRs to vowels embedded within naturally spoken words in sentence contexts have been recorded in clinically feasible test times (Choi et al., 2013). As seen above, stimuli presented in a sentence form with envelope modulation rates similar to that of running speech are likely favorable to represent non-linear hearing aid function for running speech encountered in everyday listening situations. However, in addition to considering hearing aid factors, we also need to consider factors in Category III (AEP functionality) to evaluate the usefulness of AEPs for objective aided outcome evaluation. The following section includes comparison of AEPs and AEP protocols in their potential to represent changes in audibility due to level, amplification and bandwidth.

Category III: AEP functionality

Sensitivity to changes in level

Sensitivity to changes in stimulus level affecting audibility is essential to reflect changes in audibility due to amplification, such as the difference between unaided and aided conditions. The relationship between stimulus level and response amplitude is well established in unaided conditions; an increase in stimulus level leads to an increase in response amplitude (e.g., CAEP: Billings et al., 2007; Lightfoot & Kennedy, 2006; Purdy, Sharma, Munro, & Morgan, 2013, ASSR/EFR: Dimitrijevic et al., 2002; Picton, Dimitrijevic, Perez-Abalo, & Van Roon, 2005, cABR/EFR: Akhoun et al., 2008).

Similar to functional gain measurements using behavioral methods, an improvement in audibility due to amplification has been measured using threshold-based paradigms (e.g., Picton et al., 1998). Lower (better) thresholds in aided conditions relative to unaided conditions indicate the improvement in an individual's audible range due to amplification. However, threshold-based measurements may not necessarily reflect hearing aid performance at conversational speech levels (McCreery, 2013; Stelmachowicz, Hoover, Lewis, & Brennan, 2002). In addition, threshold estimation protocols are more time consuming relative to evaluation of audibility at discrete levels that represent conversational speech. The accuracy of objective thresholds may also vary depending on recording parameters such as number of sweeps and recording time (e.g., Luts & Wouters, 2004; Picton et al., 2005). Apart from these factors, depending on stimulus levels, threshold measurements with non-linear hearing aids may interact with signal processing features such as expansion[2] (McCreery, 2013).

CAEPs as an objective aided outcome measure were first used in 1967 in eight infants wearing hearing aids (Rapin & Graziani, 1967). In their study, six infants showed larger and better defined aided waveforms and four infants showed lower (better) thresholds in aided conditions compared to unaided conditions, consistent with higher stimulation levels and therefore improved audibility with amplification. More recently, studies in adults and children with hearing impairment have shown similar results supporting the use of CAEP as an aided outcome measure (Carter et al., 2013; Chang, Dillon, Carter, Van Dun, & Young, 2012; Glista, Easwar, Purcell, & Scollie, 2012; Gravel, 1989; Korczak, Kurtzberg, & Stapells, 2005; Purdy et al., 2005; Van Dun et al., 2012).

However, recent literature on aided CAEPs in adults with normal hearing, demonstrated no or minimal effect of an increase in stimulus level due to amplification (Billings et al., 2007; Marynewich et al., 2012; Tremblay, Billings, et al., 2006). At equal stimulus levels, aided CAEPs were smaller in amplitude and longer in latency compared to unaided CAEPs (Billings et al., 2011; Marynewich et al., 2012). Speculated reasons are lack of improvement in acoustic SNR in aided relative to unaided conditions (Billings, Papesh, Penman, Baltzell, & Gallun, 2012; Billings et al., 2011; Billings, Tremblay, Stecker, & Tolin, 2009), and changes in

[2]Expansion is a characteristic of non-linear hearing aids in which hearing aid gain reduces as stimulus levels decrease below a specific threshold (Dillon, 2012).

stimulus rise-time due to hearing aid processing (Jenstad et al., 2012). The lack of improvement in acoustic SNR was likely due to audibility of the hearing aid noise floor and possibly amplified background noise in aided conditions. Audibility of hearing aid noise floor is an important factor to consider. However, this may be more of a concern at frequencies with better hearing. When hearing thresholds are normal or near normal, the noise floor will likely be lower due to lower hearing aid gain required at those frequencies (Lewis, Goodman, & Bentler, 2010). Therefore, the effect of hearing aid noise floor in individuals with hearing loss may vary.

ASSRs using modulated tones have shown sensitivity to amplification in threshold-based paradigms (Picton et al., 1998; Stroebel, Swanepoel, & Groenewald, 2007) as well as at discrete supra-threshold levels (Dimitrijevic, John, & Picton, 2004) in children and adults with hearing loss. Lower aided thresholds in threshold-based paradigms, as well as better detection with higher response amplitude in supra-threshold paradigms, are consistent with improved audibility due to amplification. However, representation of hearing aid function for speech using modulated tones may be uncertain due to the reasons discussed above, and therefore has led to more recent work with speech stimuli (Choi et al., 2013).

Although FFRs to individual harmonics in vowels are useful for inferring neural representation of individual harmonics (Aiken & Picton, 2008; Krishnan, 2002), scalp-recorded tonal FFRs are first observed at about 40 dB above the threshold for the stimulus (e.g., Davis & Hirsh, 1976; Marsh, Brown, & Smith, 1975; Moushegian, Rupert, & Stillman, 1973). This difference between FFR and behavioral thresholds may limit the utility of this measure in individuals with more severe degrees of hearing loss.

Sensitivity to changes in bandwidth

Since the primary goal of an aided outcome measure is to assess the appropriateness of a hearing aid fitting to listen to speech, assessing bandwidth audibility is important for two reasons: One, because speech consists of phonemes that span a wide frequency range (e.g., Boothroyd, Erickson, & Medwetsky, 1994) and two, because limited hearing aid bandwidth has been identified as a significant factor limiting a child's ability to access and benefit from aided speech (for a review, see Stelmachowicz, Pittman, Hoover, Lewis, & Moeller, 2004). Representation of bandwidth is not only comprised of frequency range but also includes frequency specificity. Since hearing loss can vary in configuration and severity, it is likely advantageous to use stimuli that represent a wide frequency range and that demonstrate frequency specificity to adequately reflect the impact of hearing loss and frequency-specific hearing aid gain.

Sensitivity to bandwidth is dependent on stimulus choice as well as AEP response characteristics. For transient AEPs, stimulus choice largely determines bandwidth sensitivity of a protocol. For example, use of the syllable /da/ in cABR protocols likely limits inference of bandwidth audibility up to about 3000 Hz. CAEP protocols have used phonemes /m/, /g/ and /t/ to represent low, mid, and high frequencies, respectively (Carter et al., 2013; Chang et al., 2012; Golding et al., 2007; Van Dun et al., 2012). Additional stimuli such as the phoneme /s/ or a 4 kHz tone burst have also been used to demonstrate changes in audible bandwidth using CAEPs (Glista et al., 2012; Zhang, Ching, Hou, Wong, & Burns, 2012).

In the case of speech-evoked EFRs, the nature of the response requires use of stimuli that exhibit periodicity. EFRs to vowels are elicited at the fundamental frequency of the voice (f_0), which is the frequency of the vowel envelope. Harmonics in a vowel are spaced at f_0 Hz apart and therefore EFRs elicited by a vowel may represent a response initiated from two neighboring harmonics in the cochlea (Aiken & Picton, 2006). However, recent studies suggest the dominance of the first formant region in the EFR elicited by a full bandwidth vowel (Choi et al., 2013; Laroche, Dajani, Prevost, & Marcoux, 2013). Since most first formant peak frequencies fall below 1 kHz, audibility based on EFRs elicited by natural vowels is likely limited to low frequencies. Therefore, stimulus modifications may be necessary to increase the suitability of speech-evoked EFRs for use as an objective aided outcome measure.

Speech-evoked FFRs are limited by the upper limit of phase locking, which is a response characteristic. In response to a vowel, FFRs can be elicited by harmonics until about 1500 Hz (Aiken & Picton, 2008; Krishnan, 2002; Moushegian et al., 1973). Although the FFR can provide evidence of encoding specific stimulus frequencies, the low-pass characteristic of the FFR limits inference of audible bandwidth beyond mid frequencies.

In conclusion, AEPs and current AEP protocols vary in their applicability, versatility and adaptability as a valid objective aided outcome measure and therefore it is likely that any *one* AEP may not emanate as a suitable measure. Considering the identified factors in the framework, the AEPs discussed above require further investigation in one or more categories to determine their appropriateness and limitations for use as an aided outcome measure. This thesis focused on two AEPs, namely CAEPs and EFRs.

1.4 Purpose of this thesis

The purpose of this thesis was to evaluate and improve the applicability of CAEPs and EFRs as objective aided outcome measures. The first three questions were based on CAEPs and the later three were based on EFRs. In the first three chapters, we evaluated stimulus and hearing aid interaction factors for CAEPs. The rationale of the first three chapters was (1) to evaluate differences in stimulus characteristics extracted from word-medial and word-initial contexts, and understand any advantages of using stimuli from one source over the other, (2) to quantify the effects of non-linear hearing aid processing on tone burst rise-time and elicited CAEPs when factors such as SNR are matched, and (3) to electroacoustically evaluate if audibility of stimulus phonemes in running speech can be inferred using common CAEP protocols.

The motive of the subsequent work was to develop a method that uses a running speech-like stimulus in an attempt to minimize hearing aid interactions and allow evaluation of the effects of amplification on audibility across low, mid and high frequency regions of speech. Collectively, the purpose of Chapters 5–7 was (1) to improve suitability of speech-evoked EFR stimuli for hearing aid evaluation by optimizing currently used stimuli (e.g., Aiken & Picton, 2006; Anderson & Kraus, 2013; Choi et al., 2013); (2) to establish appropriateness of previously recommended response analysis methods for optimized stimuli (Aiken & Picton, 2008); (3) to confirm deviations in hearing aid function for optimized stimuli (if any); and (4) to evaluate test functionality of the optimized EFR paradigm for use as an objective aided outcome measure.

1.4.1 Research questions

Research questions addressed in this thesis, stated in the order of subsequent Chapters 2–7 are:

(a) Is there an advantage of using stimulus phonemes extracted from word-initial versus word-medial positions as CAEP stimuli?

(b) What is the effect of non-linear hearing aid signal processing on tone burst CAEPs?

(c) Does the long inter-stimulus interval in CAEP test protocols affect hearing aid function for stimulus phonemes in comparison with running speech?

(d) What is the effect of stimulus polarity on vowel-evoked EFRs? What is the effect of averaging responses to opposite stimulus polarities? Does the presence of the first harmonic affect polarity sensitivity of EFRs?

(e) Is the proposed EFR test paradigm sensitive to changes in audibility due to stimulus level and bandwidth in adults with normal hearing?

(f) Is the proposed EFR test paradigm sensitive to changes in audibility due to stimulus level, bandwidth and use of amplification in adults with hearing loss?

1.5 Summary of chapters

The following chapters evaluated the applicability of CAEPs (Chapters 2–4) and EFRs (Chapters 5–7) as objective hearing aid outcome measures. Preliminary work on CAEPs prompted the proposal of a test paradigm using speech-evoked EFRs. Chapter 2 evaluated the differences in temporal and spectral characteristics of the phoneme /ʃ/ extracted from word-medial and word-initial positions, and illustrated the effect of choice of source on onset-sensitive CAEPs. Chapter 3 evaluated the effect of non-linear hearing aid processing on the rise-time of tone bursts and its effect on CAEPs. Chapter 4 evaluated the effect of stimulus presentation paradigms in CAEP protocols on non-linear hearing aid output, when hearing aids were programmed for hypothetical hearing losses. This study tested the representation of hearing aid function in CAEP test protocols compared to running speech for a sample of phonemes.

Chapter 5, 6 and 7 focused on the evaluation of an EFR test paradigm proposed for use as an objective aided outcome measure. The proposed EFR test paradigm uses a five-phoneme stimulus /susaʃi/ spoken by a male talker. This stimulus was modified to enable recording of eight individual EFRs from multiple frequency regions. Chapter 5 evaluated the incidence of polarity sensitivity of vowel-evoked EFRs and the effect of averaging responses to opposite stimulus polarities, as has been recommended previously. In this chapter, we also evaluated the presence of the first harmonic on polarity sensitivity of vowel-evoked EFRs. Chapter 6 evaluated sensitivity of the EFR protocol to changes in audibility by varying the stimulus level and bandwidth in a group of normal hearing listeners. Chapter 7 evaluated sensitivity of the EFR protocol to changes in audibility in a group of adults with sensorineural hearing loss. Audibility across conditions was additionally varied by comparing unaided responses to aided responses obtained with individually fitted hearing aids. In Chapters 6 and 7, the effect of bandwidth on EFRs was compared with the effect of bandwidth on behavioral outcome measures such as speech discrimination and sound quality rating in unaided and aided conditions, respectively.

Taken together, these studies evaluated two of the AEPs that have been proposed for use as objective hearing aid outcome measures. The later studies in this thesis describe the development and validation of a novel measure that was designed to improve sensitivity to the effects of amplification across multiple frequency regions while attempting to reduce undesirable interactions with hearing aid signal processing.

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Chapter 2

The effect of stimulus choice on cortical auditory evoked potentials (CAEP): Consideration of speech segment positioning within naturally produced speech[3]

2.1 Introduction

Recent studies have investigated the use of speech-evoked Cortical Auditory Evoked Potentials (CAEP) measures for the purpose of validation of amplification (e.g., Golding et al., 2007; Tremblay, Billings, Friesen, & Souza, 2006). Previous studies in the area of CAEPs (unaided and aided) have used brief speech stimuli excised from a variety of sources such as a standardized speech test or running speech (Golding et al., 2007; Tremblay, Friesen, Martin, & Wright, 2003). A few studies have employed nonsense syllables (e.g., /ʃi/ – Tremblay et al., 2006, 2003) and others have used speech segments from running speech (e.g ., /m/ – Golding et al., 2007; Golding, Purdy, Sharma, & Dillon, 2006; Pearce, Golding, & Dillon, 2007). There has been no clear rationale for the choice of one type of speech stimulus over

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another for use in CAEP testing, unaided or aided. Use of natural speech stimuli is likely preferred over non-speech stimuli for unaided and aided testing (e.g., Ceponiené, Alku, Westerfield, Torki, & Townsend, 2005; Scollie & Seewald, 2002). In the case of aided testing, hearing aids may respond differently to speech and non-speech stimuli due to processing non-linearity or additional algorithms such as noise reduction (Scollie & Seewald, 2002; Stelmachowicz, Kopun, Mace, & Lewis, 1996). This may result in inaccurate measurement of hearing aid performance. With natural speech stimuli, acoustic characteristics of word-initial (as in /ʃ/ in /ʃi/) and word-medial phonemes (as in /ʃ/ as in cashier) may differ. If these differences influence CAEP responses, one may be more advantageous to use as a CAEP stimulus. The advantages may include larger amplitude CAEPs that may be more readily detected and that may have better test-retest reliability.

This study quantified acoustic and temporal differences in the phoneme /ʃ/ when extracted from two different naturally occurring contexts (word-initial versus word-medial positions). The study also determined if choice of stimulus type influences the CAEP in terms of latency, amplitude and test-retest reliability.

2.2 Method

2.2.1 Stimulus recording and extraction

Two native Canadian English female speakers were instructed to repeat the speech tokens /aʃɪl/ and /ʃi/ five times each. The /ʃi/ was chosen to represent /ʃ/ in word-initial position. The /aʃɪl/ token represents /ʃ/ in the word-medial position. The latter was chosen from the University of Western Ontario Distinctive Feature Differences test (Cheesman & Jamieson, 1996), a speech recognition test that has often been used for hearing aid validation (e.g., Glista et al., 2009; Jenstad, Seewald, Cornelisse, & Shantz, 1999; Scollie, 2008; Scollie et al., 2010). Speech samples were recorded through a condenser microphone (AKG C 4000 B) suspended in a sound booth using computer software Spectraplus (version 5.0.26.0) sampled at 44100 Hz with 16-bit sampling precision. Given the similarity of the recorded repetitions, two tokens were randomly chosen from each sample set and used for further analysis. Using the software Goldwave (version 5.58), boundaries of /ʃ/ from /aʃɪl/ and /ʃi/ were marked based on amplitude-time waveform and listening. The segment was excised at the nearest zero crossing. In both cases, attempts were made to remove any vowel portions following and/or preceding /ʃ/.

2.2.2 Comparison of rise-time and spectra

Rise-time refers to the time taken to reach plateau amplitude. For the computation of rise-time, stimulus envelope was extracted using a Hilbert transform function in Spectraplus. The envelope was smoothed with a moving average window of ±100 sample points (4.5 ms). A plateau was marked visually and average plateau amplitude was obtained. To determine how long the envelope takes to reach the steady plateau amplitude (i.e. rise-time), each sample point of the envelope was divided by the average plateau amplitude and multiplied by 100 for the percentage or proportion of the plateau amplitude achieved at the given sample time. The time at which the amplitude first reached 100% was noted as the rise-time of the stimulus.

Among the eight /ʃ/ segments (two talkers x two types of tokens x two repetitions), the duration of /ʃ/ varied from 120 ms to about 250 ms. The mean duration of /ʃ/ from /aʃɪl/ was 141 ms and the mean duration of /ʃ/ from /ʃi/ was 208.3 ms. Combining all segments, rise-times of /ʃ/ from /aʃɪl/ ranged from 2 to 65 ms, mean value being 27.6 ms whereas rise-times from /ʃi/ ranged from 79 to 161 ms, mean value being 113.2 ms. One-third octave band spectra were measured for all eight samples. All samples of /ʃ/ from /aʃɪl/ showed higher energy in the low to mid frequencies compared to /ʃ/ from /ʃi/. Two /ʃ/ tokens from the same talker (one from each context) were chosen based on closely matched durations. In order to match the 150 ms duration of the shorter /ʃ/ from /aʃɪl/, the longer /ʃ/ from /ʃi/ (158 ms) was cropped at offset. The rise-time of /ʃ/ from /aʃɪl/ was 21.7 ms and from /ʃi/ was 79.1 ms. These will henceforth be referred to as the abrupt /ʃ/ and the slow /ʃ/. Comparing the envelopes of the abrupt and slow /ʃ/ revealed plateaus that overlap between 80 and 125 ms after stimulus onset. The smoothed envelopes of the two stimuli are shown in Figure 2-1.

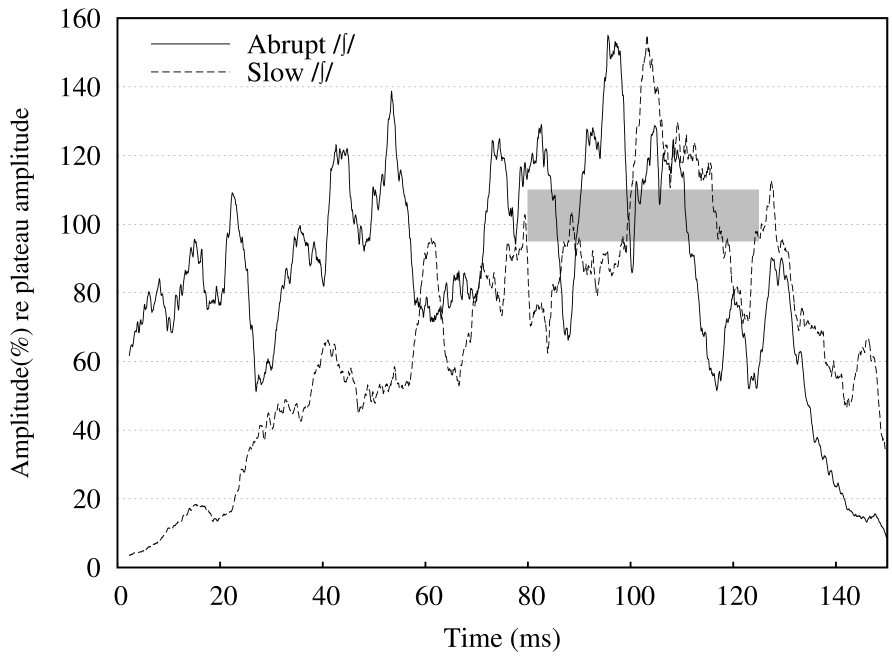

Figure 2-1: Smoothed envelope of abrupt and slow /ʃ/ (grey bar represents the plateau).

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For calibration purposes, the outputs of the insert receivers (Bio-logic broadband) were routed to an ear simulator (Brüel and Kjær type 4157 conforming to IEC 711 with microphone type 4134) and measured using Spectraplus. Since the stimuli varied in rise-times, level measurements for calibration were made across the plateau only. Figure 2-2 illustrates 1/3rd octave band spectra for both the slow and abrupt /ʃ/, presented at an overall level of 60 dB SPL. The spectra are matched in level within 4 dB for 1/3rd octave bands centered at 2.5 kHz through 10 kHz.

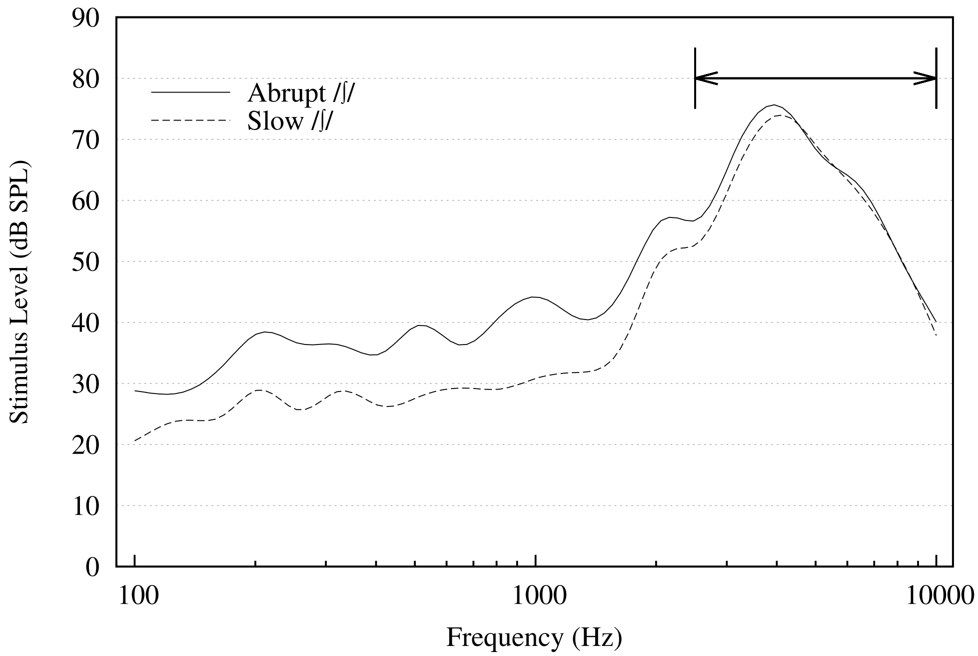

Figure 2-2: One-third octave band spectra of abrupt and slow /ʃ/. Vertical bars mark the frequency range where stimulus spectra are matched to within 4 dB.

2.2.3 CAEP Testing

Participants

The study included 16 adults (seven males and nine females) with ages between 20.3 and 29.3 years (M_{age} = 23.9 years, SD_{age} = 2.19). Inclusion criteria included passing a hearing screening at 20 dB HL using insert earphones across audiometric frequencies at octaves and inter-octaves between 250 Hz and 8 kHz, and a single peak tympanogram with peak pressure between -100 and +50 daPa. Audiometric screening was carried out using a GSI-61 audiometer and tympanograms were recorded using an Otoflex 100 middle ear analyzer. Routine otoscopic examination ruled out any contraindications such as active discharge, occluding wax or foreign bodies in the ear canal. All participants provided informed consent. Participants were excluded if they reported otological and/or neurological problems, or if their first language was not English. The study protocol was approved by the Health Sciences Research Ethics Board, The University of Western Ontario, Canada. Participants were compensated for their time.

CAEP Protocol

Testing was carried out in two sessions of 30 min each within a 7-day period. Test ear and stimulus sequence were alternated across participants. Two runs of 100 sweeps each were obtained for each stimulus condition. The participants watched a movie of their choice with subtitles and were instructed to ignore the stimuli presented (Lavoie, Hine, & Thornton, 2008; Pettigrew et al., 2004).

A single-channel ipsilateral recording (vertex to ipsilateral mastoid with ground F_{pz}) was obtained using the Natus Bio-logic Navigator Pro (version 7.0.0). Stimuli were presented at the rate of 0.49 stimuli/sec through the instrument's broadband insert receivers which have a flat frequency response between 0.1 and 10 kHz (Bio-logic Systems Corporation, 2003). The left and the right insert transducers were spectrally matched. Each recorded sweep included 350 ms of pre-stimulus baseline and 716 ms of post-stimulus activity. EEG was amplified 50000 times and digitized at the rate of 480.03 Hz. Responses were band-pass filtered between 0.1 and 100 Hz online. Artifact rejection threshold was set to ± 70 μV.

Data analysis and interpretation

Post-processing using a MATLAB script included a second order band-pass Butterworth filter (1–15 Hz). Criteria used to assess presence of N1-P2 peaks included repeatability and amplitude relative to pre-stimulus baseline activity. The examiner was blinded to the test condition during interpretation. Peaks were marked at their maximum amplitude within their expected latency regions. The N1-P2 peak-to-peak amplitude was computed as this measure may provide greater reliability and less variance across repetitions when compared to a baseline-to-peak approach (Goshorn, Marx, & Simmons, 2011). Values were averaged across the two repetitions for subsequent analysis.

Intraclass Correlation Co-efficient (ICC) was computed between repetitions as a measure of test-retest reliability (Bishop, Hardiman, Uwer, & von Suchodoletz, 2007; Bishop & McArthur, 2005; Fox, Anderson, Reid, Smith, & Bishop, 2010). ICC values range between 0 and 1, with a value of 1 indicating perfect agreement between the two repetitions (Portney & Watkins, 2008). The ICC was computed between 50 and 350 ms relative to the onset of the stimulus as this would encompass the regions of N1 and P2 (Hall, 2007). ICC was computed for responses to both stimulus types and then converted into Z scores using Fishers transform function for further group level analysis (Fox et al., 2010). Although Z scores were used for statistical analyses, only ICC values have been reported here.

Four paired t-tests were completed for the four measures N1 latency, P2 latency, N1-P2 amplitude, and Z scores of ICC. Results of the statistical analysis were interpreted with false discovery rate corrections to control for inflation of false positives (Benjamini & Hochberg, 1995).

2.3 Results

The grand CAEP waveforms averaged across all participants for the two stimulus types are shown in Figure 2-3. CAEPs elicited by the abrupt /ʃ/ are larger in amplitude with shorter peak latencies in comparison with CAEPs elicited by the slow /ʃ/. The mean latency and amplitude values for both conditions are provided in Table 2-1.

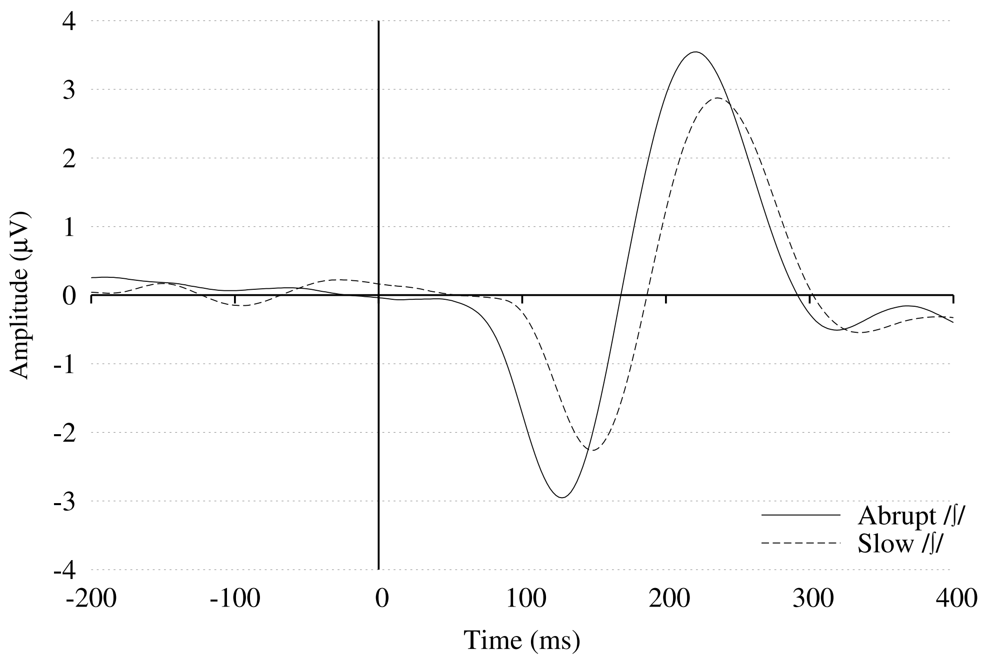

Figure 2-3: Grand waveforms averaged across all participants for abrupt and slow /ʃ/. (n = 16)

From "The effect of stimulus choice on cortical auditory evoked potentials (CAEP): Consideration of speech segment positioning within naturally produced speech," by V. Easwar, D. Glista, D. Purcell & S. Scollie, 2012, *International Journal of Audiology, 51*(12), pp. 926–931. Copyright Taylor & Francis, Inc. Reprinted with permission.

Latencies were significantly shorter and N1-P2 amplitudes were significantly larger for responses evoked with the abrupt /ʃ/, when compared to responses from the slow /ʃ/ (see Table 2-1 for *t*-test results). Differences in the ICC were not significant.

Measure	abrupt /ʃ/ M (SE)	slow /ʃ/ M (SE)	t statistic (df)	p value	Effect size
N1 latency (ms)	129.94 (2.24)	149.51 (3.23)	8.87 (15)	<0.001*	2.22
P2 latency (ms)	219.38 (3.28)	238.17 (4.56)	6.93 (15)	<0.001*	1.73
N1-P2 amplitude (μV)	6.96 (0.51)	5.77 (0.33)	-2.97 (15)	0.010*	-0.74
Test-retest reliability (ICC)	0.77 (0.03)	0.72 (0.04)	-0.99 (15)	0.338	-0.25

Table 2-1: Mean and standard error of latency, amplitude and ICC values along with results of t-tests. * indicates a significant difference.

Note. From "The effect of stimulus choice on cortical auditory evoked potentials (CAEP): Consideration of speech segment positioning within naturally produced speech," by V. Easwar, D. Glista, D. Purcell & S. Scollie, 2012, *International Journal of Audiology, 51*(12), pp. 926–931.

2.4 Discussion

2.4.1 Acoustic differences between stimuli

The word-medial /ʃ/ segments were found to be shorter in duration and to have faster rise-times compared to the word-initial /ʃ/ segments. This could reflect the effects of co-articulation in which adjacent phonemes in running speech may be shorter and may exhibit blurred phonemic boundaries, compared to phonemes articulated in isolation (Kent & Read, 2002). The onset is more gradual when the phoneme is spoken at the beginning of the utterance where the transition is from silence to speech as in the case of /ʃi/. The initiation of articulator motion could contribute to the slow onset. Spectrally, the abrupt /ʃ/ contains more energy at low and mid frequencies compared to slow /ʃ/. This could also be due to co-articulation (Kent & Read, 2002; Pickett, 1999). With respect to /ʃ/, articulation of /ʃ/ in /aʃɪl/ is not only influenced by the following vowel but also /l/ at the end. In contrast, /ʃ/ in /ʃi/ is influenced by the following vowel only. These spectral differences evident in the stimuli chosen in this study are bound to vary with different neighboring phonemes. Another factor that may contribute to the broader spectrum in the case of the abrupt /ʃ/ is a faster rise-time (Burkard, 1984).

2.4.2 Effect of stimulus on CAEP peak latencies and amplitude

The abrupt /ʃ/ elicited CAEPs with significantly larger N1-P2 amplitude and shorter peak latencies compared to the slow /ʃ/. This can be explained by the known effects of stimulus rise-time on CAEP. To our knowledge, studies that have evaluated the effects of rise-time on CAEP have only used tone bursts. This is probably because their rise-time can be precisely manipulated.

The general consensus among studies that investigated the effect of tone burst rise-time on CAEPs is that shorter rise-times lead to larger amplitudes and shorter latencies (Cody & Klass, 1968; Lamb & Graham, 1967; Onishi & Davis, 1968; Prasher, 1980; Skinner & Jones, 1968; Thomson, Goswami, & Baldeweg, 2009). This can be explained by the effect of stimulus rise-time on neural synchrony. Longer rise-time leads to increased jitter in neuron firing. The increased jitter may reflect inconsistent trigger points along the onset envelope of the tone burst (Picton, 2011, pages 335–343). Increased jitter leads to reduced neural synchrony, which results in broader peaks with lower amplitude and longer latency (Goldstein & Nelson, 1958; Hyde, 1997; Onishi & Davis, 1968). The effect of longer stimulus rise-time is more detrimental for earlier potentials, such as the compound action potential and auditory brainstem response (Arnold, 2007; Goldstein & Nelson, 1958). Even small increases in rise-time (e.g., from 2.5 ms to 5 ms) can abolish the compound action potential due to inadequate synchronous firing. In contrast, CAEPs can be elicited by longer stimulus rise-times that abolish the earlier potentials.

Another explanation can be based on detection threshold. A study that corrected

N1 latency based on behavioral detection threshold along the time course of tone burst onset ramp found that the N1 latency remained unchanged up to increases in rise-time of 100 ms. This implies that any stimulus-related activity is triggered only after the onset ramp is above the individuals threshold (Ruhm & Jansen, 1969). The effects of rise-time have also been explained by the theory of temporal integration, where detection thresholds vary as a function of stimulus duration and intensity. As rise-time is increased beyond 30 ms, the effective intensity at the onset of the stimulus is reduced (Onishi & Davis, 1968).

The results of the present study are in agreement with the literature discussed above. This implies that the use of phonemes from different sources could potentially lead to differences in the interpretation of CAEPs. For example, it is possible that stimulus choice may also affect CAEP procedures such as threshold estimation, by facilitating detection of larger amplitude CAEPs. Although larger amplitude responses can be looked at as a stimulus advantage, word-medial phonemes may have a spectral disadvantage. Co-articulatory effects and onset rate appear to contribute low frequency energy, reducing the frequency specificity of the stimulus relative to the word-initial position. This may be especially important to consider when using natural speech stimuli for the purposes of hearing aid validation. These effects were studied for the phoneme /ʃ/ in this experiment, but whether such effects generalize to other phonemes remains to be studied.

As explained above, the differences in waveforms elicited by the two variants of /ʃ/ may be attributed to the sensitivity of CAEP to onset characteristics of the stimuli. Differences in the CAEP waveforms may not be interpreted as representation of other processes (e.g., discrimination) based on the present study. Hearing aid signal processing (e.g., compression and compression attack time) may interact with the temporal properties of natural speech stimuli, but such effects have not yet been evaluated in detail. The present study evaluated the effects of rise-time in individuals with audiometric thresholds within the normal range, and provides some evidence that stimuli with a more abrupt onset may provide larger amplitude CAEPs. However, the effects of rise-time on CAEP in individuals with hearing impairment, specifically cochlear hearing loss may differ from that in individuals with normal audiometric thresholds due to recruitment (e.g., Moore, 2007).

2.4.3 Effect of stimulus on test-retest reliability

In the present study, mean ICCs of 0.77 ($SE = \pm 0.03$) and 0.72 ($SE = \pm 0.04$) were obtained for CAEPs elicited to abrupt and slow /ʃ/, respectively. Test-retest reliability of speech-evoked CAEPs assessed in terms of ICC has been previously reported in Tremblay et al. (2003). Although direct comparisons between the two studies cannot be made due to differences in stimuli (consonant vowel combinations were used in Tremblay et al., 2003), it is interesting to note that mean ICCs reported by Tremblay et al. (2003) across stimuli were higher, ranging between 0.8 and 0.9. Factors such as number of sweeps in an average and an exclusive channel for eye blink rejection used in Tremblay et al.'s study could possibly explain this difference. Specifically, the use of more sweeps and rejection of epochs with eye blinks in the Tremblay et al. study may have allowed better quality responses and more repeatable waveforms. The present study illustrates that reliable speech-evoked CAEPs can be obtained in normal hearing participants using a shorter protocol and a single-channel clinical system without eye blink rejection while maintaining reasonable test-retest reliability.

2.5 Summary and Conclusion

This study illustrated temporal and spectral differences between the same phoneme in different naturally occurring contexts and their effects on the CAEP. Phonemes extracted from running speech (word-medial position) tended to have shorter rise-times and hence elicited larger CAEPs with shorter latencies, when compared to segments extracted from a word-initial position. It is possible that the word-medial stimuli may have decreased spectral specificity due to co-articulation and onset rate. This emphasizes the need to consider the characteristics of speech stimuli when developing protocols for speech-evoked CAEP measures. Apart from the types of stimuli, this study illustrates that speech-evoked CAEPs can be reliably obtained using a single-channel clinical instrument.

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Chapter 3

Hearing aid processing changes tone burst onset: Effect on cortical auditory evoked potentials in individuals with normal audiometric thresholds[4]

3.1 Introduction

Since the 1960s, the obligatory Cortical Auditory Evoked Potential (CAEP) has been a popularly researched potential for objective validation of amplification. Recent studies have questioned the validity of using CAEP for this purpose due to the hearing aid modifying the CAEP stimulus. The areas of concern have been poor Signal-to-Noise Ratio (SNR) caused by the raised noise floor of the hearing aid (Billings, Tremblay, & Miller, 2011) and stimulus modifications such as rise-time (i.e., time taken to reach plateau amplitude; Marynewich, 2010). In addition, the gain achieved may differ between the short CAEP stimulus and the long duration sounds that are used for hearing aid verification (Marynewich, 2010; Stapells, Marynewich, & Jenstad, 2010).

[4] A version of this chapter has been published: Easwar, V., Glista, D., Purcell, D. W., & Scollie, S. D. (2012). Hearing aid processing changes tone burst onset: Effect on cortical auditory evoked potentials in individuals with normal audiometric thresholds. *American Journal of Audiology*, *21*(1), 82-90. doi: 10.1044/1059-0889(2012/11-0039)

In recent research studies (Marynewich, 2010; Stapells et al., 2010) that quantified the effects of hearing aid processing on the stimulus, digital hearing aids were observed to alter the onset envelope and delay the rise-time whereas analog hearing aids did not, and the two types of hearing aids had varying effects on the CAEP that approximately corresponded to the lengthened rise-time. Change in stimulus rise-time is an important stimulus parameter to be considered as it has been shown to affect the CAEP peak amplitude and latency (Cody & Klass, 1968; Lamb & Graham, 1967; Onishi & Davis, 1968; Prasher, 1980; Skinner & Jones, 1968; Thomson, Goswami, & Baldeweg, 2009). The general consensus among these studies is that shorter rise-times lead to larger amplitudes and shorter latencies. This has been explained on the basis of neural synchrony. Longer rise-time increases jitter in neuron firing. The increased jitter reflects variable trigger points along the onset envelope of the tone burst (Picton, 2011, pp. 335-343). Increased jitter results in reduced neural synchrony, which leads to broader peaks with lower amplitude (Goldstein & Nelson, 1958; Onishi & Davis, 1968). The effects of rise-time have also been explained by the theory of temporal integration, where detection thresholds vary as a function of stimulus duration and intensity. As rise-time is increased beyond 30 ms, the effective intensity at the onset of the stimulus is reduced (Onishi & Davis, 1968).

Although the above-mentioned studies (Marynewich, 2010; Stapells et al., 2010) provided insight into the changes that are introduced in the stimulus envelope that could explain the differences between the unaided and aided CAEP, differences in the SNR between the two conditions were not measured (e.g., Billings et al., 2011; Billings, Tremblay, Souza, & Binns, 2007). SNR is a strong predictor of CAEP attributes, with poorer SNR leading to smaller amplitudes and longer latencies (Billings et al., 2011). The effect of SNR is seen for peaks P1, N1, P2, and N2 of the CAEP (Billings et al., 2011; Billings, Tremblay, Stecker, & Tolin, 2009). The effect of SNR is also seen at the sub-cortical level illustrated using auditory brainstem responses elicited to stimuli in noise (e.g., Russo, Nicol, Musacchia, & Kraus, 2004; Song, Skoe, Banai, & Kraus, 2011). Synchronous neural discharge is the basis of AEPs (Eggermont, 2007). The addition of noise interferes with the temporal precision of firing and therefore reduces neural synchrony (Kaplan-Neeman, Kishon-Rabin, Henkin, & Muchnik, 2006; Russo et al., 2004). Reduced neural synchrony can result in decreased peak amplitude values and increased peak latency values (Russo et al., 2004). SNR has been indicated to influence the CAEP more than the absolute stimulus level (Billings et al., 2009). The obligatory CAEP, being more stimulus-driven than the endogenous potentials such as the P300, is more sensitive to the effects of SNR (Kaplan-Neeman et al., 2006). This may be a reflection of the bottom-up process in the case of the CAEP versus a top-down process in the case of the P300 (Kaplan-Neeman et al., 2006). Hence, comparison of the effect of two stimulus conditions that vary in SNR may not truly reflect the sole effect of stimulus differences.

Therefore, it is uncertain if the CAEP is influenced by the envelope/rise-time changes in the stimuli that are caused by hearing aid processing. In order to study the exclusive effects of these stimulus changes on the CAEP, comparisons are required when the SNR is matched between the conditions being compared. In addition, the nature of hearing aid processing may vary with processing linearity of the hearing aid and stimulus level used. These factors are discussed below.

Digitally programmable analog or digital hearing aids functioning within their linear working range have been used in recent studies investigating aided CAEPs. Hearing aid technology has transformed from mostly analog linear to mostly digital Wide Dynamic Range Compression (WDRC) type of amplification (Johnson, Cox, & Alexander, 2010). WDRC hearing aids, unlike linear hearing aids, vary gain across a wide range of levels (Dillon, 2001). In the Unites States, nearly 100% of the hearing aids prescribed to children are multichannel with compression systems, and more than 90% of them use the Desired Sensation Level (DSL) prescription algorithm (Jones & Launer, 2010). This distinction is important because the effect of a compressive (non-linear) circuit will differ from that of a linear circuit as the amount of gain varies based on the input stimulus level (Dillon, 2001). Digital hearing aids with this adaptive gain scheme cause several changes in the stimulus, apart from providing gain that is shaped for a specific hearing loss. First, a delay is imposed on the signal by the digital signal processing (Kates, 2005; Schaub, 2008). Second, the time required to determine the input level and stabilize the gain for a rapid change in input level can cause a brief overshoot at the onset of the stimulus if the stimulus level is above the compression threshold (Dillon, 2001). The magnitude of the overshoot is related to the compression attack time (American National Standards Institute [ANSI], 2003). Most WDRC amplifiers have short attack times of less than 10 ms (Kuk, 2002). With a commonly used CAEP protocol (e.g., Hyde, 1997; Stapells, 2009), where the inter-stimulus interval (i.e., duration between the end of one stimulus and the beginning of the following stimulus) ranges between 1 and 2 seconds, the sudden increase in the input level due to the presence of the CAEP stimulus is likely to cause compression to act each time. This in turn would result in consistent overshoots in consecutive stimuli. Expansion, a processing stage in which gain decreases as the input level decreases, mainly catered to minimize internal noise of the hearing aid (Bray & Ghent, 2001), may or may not be activated during the inter-stimulus interval. Third, the use of a hearing aid imposes a noise floor on the signal (Agnew, 1997; Lewis, Goodman, & Bentler, 2010; Thompson, 2003), and non-linear signal processing may raise the level of the noise floor (Lewis et al., 2010; Thompson, 2003).

Some of these hearing aid signal processing considerations may also interact with the input stimulus level. Recent studies of hearing aid-evoked CAEPs were conducted in individuals with normal hearing. One such study used low stimulus levels (40 dB SPL) to avoid loud hearing aid output level (Billings et al., 2011). Low stimulus levels are atypical in validation of amplification where supra-threshold levels representing conversational levels of speech are used (e.g., Golding et al., 2007; Olsen, 1998). Typical stimulus levels for soft through loud speech inputs and for hearing aid verification range from ~55 to 75 dB SPL (Olsen, 1998). Because the stimulus level interacts with non-linear signal processing in hearing aids, lower level stimuli will receive more gain and higher level stimuli will receive less gain. For this reason, consideration of stimulus level is important if the goal of CAEP measurement is to characterize the aided response to sound.

Further research is required to determine if hearing aids introduce temporal changes in the stimulus that influences the CAEP. Hence, the aim of this study was to evaluate the effect of non-linear hearing aid processing on the onset of tone bursts with varying rise-times and consequently, the effect on tone-evoked CAEPs, when the effects of hearing aid signal processing delay and noise floor are controlled. This was evaluated in normal hearing listeners to permit comparison to previous studies. It was hypothesized that digital hearing aids with WDRC processing, unlike linear processing, would shorten the rise-time due to the occurrence of overshoot at stimulus onset. This would be expected to result in larger CAEP amplitude and shorter peak latencies.

3.2 Method

3.2.1 Stimuli

Stimuli were 1 kHz tone bursts of constant 90 ms duration with symmetrical linear rise/fall times of 7.5 ms or 20 ms. These rise-times are representative of shorter and longer rise-times that were used in past CAEP research (e.g., Beukes, Munro, & Purdy, 2009; Billings et al., 2011; Marynewich, 2010; Stapells et al., 2010) .

3.2.2 Hearing aid

In an attempt to prioritize clinical utility, we selected a hearing aid that was commonly used in clinical practice by the Ontario Infant Hearing Program, based on file review. This device was a 20-channel hearing aid that used non-linear signal processing by default. The hearing aid was programmed to match DSL v.5.0 adult targets (Scollie et al., 2005) for the standard audiogram N5, which has thresholds ranging between 65 dB HL and 80 dB HL and a three-frequency mean pure-tone average of 75 dB HL (Bisgaard, Vlaming, & Dahlquist, 2010). The N5 audiogram is one of 10 standard audiograms that were developed to represent the range of audiograms that are common in clinical practice for the purpose of standardizing hearing aid programming. The hearing aid performance and fit-to-targets were verified using the Audioscan Verifit hearing aid analyzer. The hearing aid was programmed to include an omnidirectional microphone mode with all additional features (e.g., digital noise reduction, feedback cancellation) switched off. Expansion thresholds were at hearing aid software prescribed settings. The input/output plot at stimulus frequency 1 kHz obtained using the Audioscan Verifit revealed a compression ratio of 2:1 and compression knee-point of 55 dB SPL. Attack time and release time measured using ANSI automatic gain control (2003) module in Verifit at 1 kHz were 10 and 60 ms, respectively.

3.2.3 Recording of hearing aid output

Recordings of hearing aid output used an ear simulator (Brüel and Kjær type 4157, microphone type 4134) with an ear mold simulator to which the hearing aid was connected via 25 mm of size 13 tubing (ANSI, 2003). The intention of this setup was to mimic the output of a behind-the-ear hearing aid in an average adult ear. This was set up in a Brüel and Kjær anechoic box (Box 4232) that also housed a reference microphone. The outputs of the reference and coupler microphones were captured using the Spectraplus real-time spectrum analyzer in separate channels using a sampling rate of 44100 Hz with sampling precision of 16 bits. The Spectraplus software was used to record the reference and coupler signals as .wav files for further signal analyses, including measurement of aided levels in $1/3^{rd}$ octave bands and computation of hearing aid delay.

Tone bursts (7.5 ms and 20 ms rise-times) were presented via the speaker in the anechoic box at 60 dB SPL (calculated using Root-Mean-Square [RMS] amplitude measured across the plateau). This level was above the compression threshold of the hearing aid, thereby testing the hearing aid while in a signal processing mode intended for listening to conversational-level speech (Olsen, 1998). The tone burst stimuli were presented with an inter-stimulus interval of 1910 ms similar to previous CAEP recording paradigms (e.g., Billings et al., 2011; Tremblay, Billings, Friesen, & Souza, 2006). The inter-stimulus interval was measured from the end of one tone burst to the beginning of the following tone burst. Visual inspection showed overshoots consistently occurring in consecutive tone bursts of both rise-times. For both tone burst rise-times, one of the repetitions from each recording was randomly chosen as the stimulus for the aided condition. To ensure that the recording of the hearing aid output (both stimulus and noise floor) was unaffected by the noise floor of the recording path, the internal noise of the hearing aid and that of the recording path were compared in a no-stimulus condition. The noise floor of the hearing aid measured 9 dB to 54.1 dB above the noise floor of the Brüel and Kjær anechoic box across $1/3^{rd}$ octave bands between 100 Hz and 10000 Hz, with the largest difference in the band centered at 5000 Hz. Higher $1/3^{rd}$ octave band levels of the hearing aid noise floor were measured at higher frequencies due to the sloping nature of the hearing loss and the prescribed hearing aid gain. The hearing aid provided 42.5 dB of gain for both tone bursts and introduced a delay of 7.2 ms.

3.2.4 Calibration

A Bio-logic Navigator Pro (version 7.0.0), which is a clinical diagnostic AEP measuring instrument, was used for electrophysiological recording in the present study. For calibration purposes, the outputs of the Bio-logic insert receivers were routed to an ear simulator according to ANSI (2004). Because the tone bursts had varying rise-times and the hearing aid altered the onset of the tone bursts in the aided condition, level measurements for calibration were made across the plateau only (between 35 ms and 70 ms relative to the onset of the tone burst). Plateau levels were calibrated to target presentation levels of 60 dB SPL across all test conditions.

3.2.5 Test Conditions

The protocol consisted of two conditions for tone bursts of both 7.5 ms and 20 ms rise-times:

- Aided: Stimuli for this condition were obtained from the tone bursts as recorded from the hearing aid output (Spectraplus) and were cropped for the purposes of this study using Goldwave software. These aided tone bursts included the effects of non-linear signal processing and hearing aid noise floor.

- Unaided: Stimuli for this condition were the unprocessed tone bursts with the hearing aid noise floor superimposed synthetically throughout. The hearing aid noise floor was superimposed to equalize the two conditions for the SNR (Billings et al., 2011, 2007). This was done in two steps using Goldwave software. First, the level of the unprocessed tone bursts was matched with that of the aided tone bursts. This essentially removed hearing aid gain specific to aided stimuli. Second, the recorded noise floor from the inter-stimulus interval of an aided recording was excised and was mixed with the unprocessed tone bursts.

The onset of stimuli were matched in order to remove the effects of hearing aid processing delay. The stimuli and the noise floors in both of the above conditions were spectrally matched (within 4 dB). The Bio-logic Navigator Pro allowed for custom stimuli of maximum 500 ms in duration. In an attempt to maximize the duration of the pre-stimulus noise floor, all stimuli were constructed using Goldwave software such that the tone burst occurred between 410 ms and 500 ms of the entire 500 ms duration of the custom stimulus. This essentially removed hearing aid delay from the aided stimulus.

In summary, the only differences that remained between the aided and the unaided tone bursts were any changes that were imposed on the stimulus onset/offset by the multichannel non-linear processing of the hearing aid. The onset of the hearing aid noise floor was ramped up to reach maximum amplitude at 200 ms. The purpose of this envelope ramp was to minimize the CAEP in response to the hearing aid noise floor onset. This was necessary because the Navigator Pro presents true silence between presentations of the custom stimuli. All of the stimuli were re-sampled to 48000 Hz before importing into the Navigator Pro.

3.2.6 Verification

The output of the Bio-logic insert receivers was verified for matched spectra of the stimulus and the noise floor between the aided and unaided conditions for each rise-time (Figure 3-1). The recording apparatus was the same as that used for calibration purposes. For the tone burst with 7.5 ms rise-time, the SNR obtained in the band centered at 1 kHz was 45.5 dB and 44.9 dB in the aided and unaided conditions, respectively. For the tone burst with 20 ms rise-time, the SNR was 44.0 and 43.4 dB in the aided and unaided conditions, respectively. Hence, SNR was matched within 1 dB accuracy at the stimulus frequency.

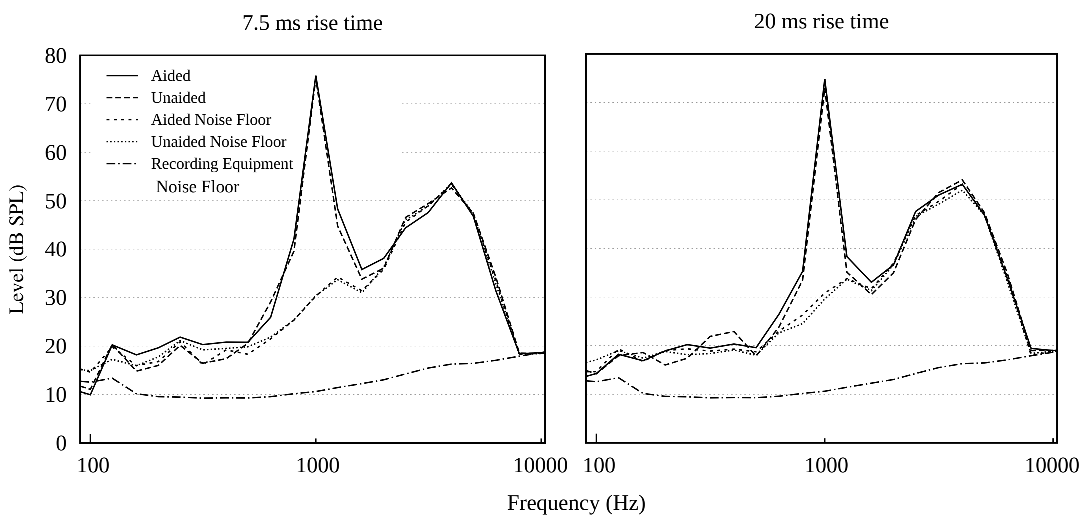

Figure 3-1: Verification of closely matched $1/3^{rd}$ octave band spectra of the aided and unaided tone bursts and noise floors for 7.5 ms and 20 ms rise-times, respectively. The stimuli were played at 80 dB SPL RMS measured across plateau to avoid inaccuracies from being close to the noise floor of the measurement system.

From "Hearing Aid Processing Changes Tone Burst Onset: Effect on Cortical Auditory Evoked Potentials in Individuals With Normal Audiometric Thresholds," by V. Easwar, D. Glista, D. Purcell & S. Scollie, 2012, *American Journal of Audiology, 21*(1), pp. 82–90. Copyright American Speech-Language-Hearing Association. Reprinted with permission.

3.2.7 Computation of tone burst rise-time

Envelopes of the stimuli (consisting of noise floor followed by the tone burst) were obtained by applying Hilbert transform in Spectraplus and were smoothed using a moving average of ±50 sample points (2.26 ms). Average plateau amplitude for the pre-defined plateau (between 35 and 70 ms relative to the onset of the tone burst) was obtained. Each sample point of the stimulus envelope was divided by the average plateau amplitude and was multiplied by 100 to obtain the percentage (proportion) of the plateau amplitude that was achieved at the given sample time. The time, relative to the tone burst onset at 410 ms, at which the amplitude first reached 100% was noted as the rise-time of the tone burst.

3.2.8 Participants

The study included 16 adults (7 males and 9 females) ranging in age between 20.3 and 29.3 years ($M_{\text{age}} = 24.2$ years; $SD_{\text{age}} = 2.1$). Eligibility criteria included passing a hearing screen at 20 dB HL using insert earphones across octave and inter-octave audiometric frequencies between 0.25 and 8 kHz (GSI-61 Audiometer) and a single peak tympanogram with peak pressure between -100 and +50 daPa (measured using the middle ear analyzer Otoflex 100). Routine otoscopic examination ruled out any contraindications such as active discharge, occluding wax, or foreign bodies in the ear canal. Participants with normal hearing were chosen to study the effects of hearing aid-processed stimuli without the influence of hearing loss (e.g., Billings et al., 2011). None of the participants reported any history of significant neurological or otological disorders. The study protocol was approved by the Health Sciences Research Ethics Board of The University of Western Ontario, Canada. Participants were compensated for their time.

3.2.9 CAEP Testing

A single-channel ipsilateral recording (vertex to ipsilateral mastoid with ground F_{pz}) was obtained using the Navigator Pro. Tone bursts were presented at the rate of 0.5 stimuli/sec, which translates to an inter-stimulus interval of 1910 ms for tone bursts of 90 ms duration. This inter-stimulus interval is the same as that used to acoustically record the aided stimuli. Each recorded electroencephalogram [EEG] sweep included 410 ms of pre-stimulus baseline (relative to tone burst onset) and 656 ms of post-stimulus activity. The EEG was amplified 50000 times and was digitized at the rate of 480.03 Hz. Responses were bandpass filtered between 0.1 Hz and 100 Hz online. The artifact rejection threshold was set to ± 70 μV.

Testing was carried out in two 1-hr sessions within a 7-day period. Test ear was alternated across participant order number. Tone burst rise-time conditions were alternated across session number for each participant. The sequence of conditions in a session was randomized. Two averages of 100 sweeps each were obtained for each stimulus condition. The participants watched a muted movie of their choice with subtitles only and were instructed to ignore the stimuli played (Lavoie, Hine, & Thornton, 2008; Pettigrew et al., 2004). Participants were given breaks when requested.

3.2.10 Response analysis and Interpretation

Post-processing using a MATLAB script included a second-order bandpass Butterworth filter (1-15 Hz). Criteria used to assess the presence of N1-P2 peaks included repeatability and relative pre-stimulus baseline activity. The examiner was blind to the test condition during interpretation. Peaks were marked at their maximum amplitude within their expected latency regions automatically using the MATLAB script. The latency regions used to identify the peaks were between 78 ms and 170 ms for N1 and between 111 ms and 280 ms for P2. This peak marking process was cross-checked manually by the examiner, and decisions were overridden where required. Corrections of 410 ms were applied to peak latencies to account for the delayed onset of the tone burst within the 500 ms duration of the entire custom stimulus. The N1-P2 peak-to-peak amplitude was computed as this measure is reported to provide greater reliability and less variance across repetitions when compared to a baseline-to-peak approach (Goshorn, Marx, & Simmons, 2011). Values were averaged across the two repetitions for subsequent analysis.

3.2.11 Statistical analysis

A Repeated Measures Analysis of Variance (RM-ANOVA) was computed for the measures N1 and P2 latencies, and N1-P2 amplitude with tone burst rise-time (levels: 7.5 ms and 20 ms) and stimulus processing condition (levels: aided and unaided) as the main factors. Results of the statistical analysis were interpreted based on an α of 5%.

3.3 Results

The envelopes of the aided and unaided stimuli for both rise-times are illustrated in Figure 3-2. The aided stimulus rose faster than the unaided stimulus for both rise-times because the hearing aid delay of $\sim$7.2 ms was taken into account while creating the stimuli. Comparing the envelopes of the aided stimuli for the two rise-times, the overshoot, relative to the plateau amplitude, was minimally higher for the 7.5 ms tone burst compared to the 20 ms tone burst. The change in rise-time was greater for the 20 ms condition compared to the 7.5 ms condition. The rise-time of the 7.5 ms aided tone burst was 5.5 ms after processing; the rise-time of the 20 ms aided tone burst was 12 ms.

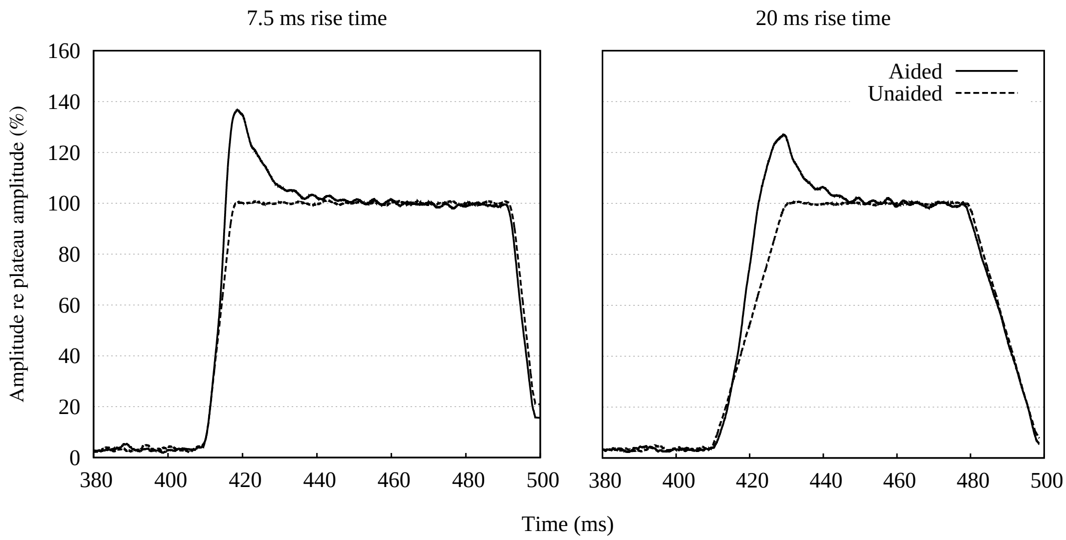

Figure 3-2: Illustration of changes in stimulus envelope due to hearing aid processing. The rise-times of the aided stimuli were 5.5 ms and 12 ms for the original tone bursts with rise-times of 7.5 ms and 20 ms, respectively.

From "Hearing Aid Processing Changes Tone Burst Onset: Effect on Cortical Auditory Evoked Potentials in Individuals With Normal Audiometric Thresholds," by V. Easwar, D. Glista, D. Purcell & S. Scollie, 2012, *American Journal of Audiology, 21*(1), pp. 82–90. Copyright American Speech-Language-Hearing Association. Reprinted with permission.

Of the 16 participants tested, P2 was judged to be absent in the aided condition for one participant and in the unaided condition for a second (different) participant. The grand waveforms averaged across all participants for each condition and rise-time are shown in Figure 3-3. Across participants, the mean within-subject test-retest difference was 10.8 ms (Standard Error [SE] = 3.1) for peak latency. For N1-P2 amplitude, the mean within-subject test-retest difference was 1.14 μV (SE = 0.21).

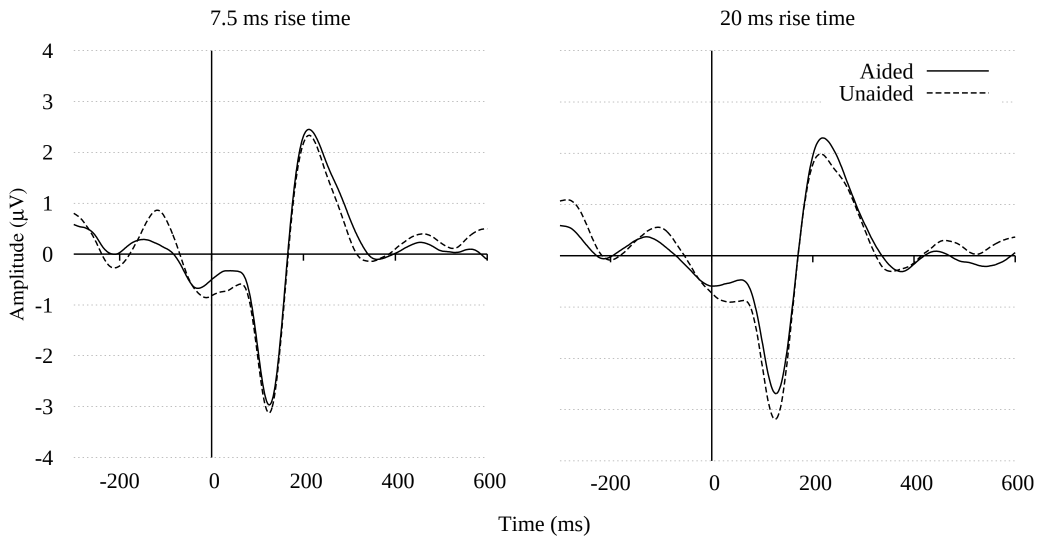

Figure 3-3: Grand response waveforms for tone bursts of 7.5 ms and 20 ms rise-times averaged across all participants.

From "Hearing Aid Processing Changes Tone Burst Onset: Effect on Cortical Auditory Evoked Potentials in Individuals With Normal Audiometric Thresholds," by V. Easwar, D. Glista, D. Purcell & S. Scollie, 2012, *American Journal of Audiology, 21*(1), pp. 82–90. Copyright American Speech-Language-Hearing Association. Reprinted with permission.

Neither tone burst rise-time (7.5 ms vs. 20 ms) nor condition (aided vs. unaided) had significant effects on N1 or P2 latencies, or N1-P2 amplitude (see Table 3-1 for mean values and Table 3-2 for RM-ANOVA results). Hence, the null hypothesis cannot be rejected.

Measure	7.5 ms rise-time				20 ms rise-time			
	Aided		Unaided		Aided		Unaided	
	M	*SE*	*M*	*SE*	*M*	*SE*	*M*	*SE*
N1 latency (ms)	125.78	2.25	125.26	1.59	128.44	2.35	127.73	2.23
P2 latency (ms)	219.05	4.78	215.77	4.09	224.43	5.69	220.68	5.88
N1-P2 amplitude (μV)	6.11	0.47	6.16	0.34	5.64	0.34	5.80	0.33

Table 3-1: Summary of means and standard errors of peak latencies and amplitude across stimulus conditions.

Note. From "Hearing Aid Processing Changes Tone Burst Onset: Effect on Cortical Auditory Evoked Potentials in Individuals With Normal Audiometric Thresholds," by V. Easwar, D. Glista, D. Purcell & S. Scollie, 2012, *American Journal of Audiology, 21*(1), pp. 82–90. Copyright American-Speech-Language-Hearing Association. Reprinted with permission.

Measure	Effect of condition		Effect of rise-time		Condition x rise-time	
	F statistic (*df*)	*p* value	*F* statistic (*df*)	*p* value	*F* statistic (*df*)	*p* value
N1 latency	0.04 (1,13)	0.850	1.80 (1,13)	.203	0.00 (1,13)	1.000
P2 latency	0.01 (1,13)	0.941	2.69 (1,13)	.125	0.02 (1,13)	0.878
N1-P2 amplitude	0.57 (1,13)	0.462	2.41 (1,13)	.144	0.27 (1,13)	0.614

Table 3-2: Summary of results of RM-ANOVA. Condition is unaided (unprocessed) or aided (hearing aid-processed stimuli), and rise-time is 7.5 ms or 20 ms.

Note. From "Hearing Aid Processing Changes Tone Burst Onset: Effect on Cortical Auditory Evoked Potentials in Individuals With Normal Audiometric Thresholds," by V. Easwar, D. Glista, D. Purcell & S. Scollie, 2012, *American Journal of Audiology, 21*(1), pp. 82–90. Copyright American-Speech-Language-Hearing Association. Reprinted with permission.

3.4 Discussion

This study investigated the influence of non-linear hearing aid processing on tone bursts and the effect of changes in tone burst onset on CAEPs when SNR, hearing aid delay and gain were carefully accounted for. Results revealed no significant effect of the change in rise-time caused by hearing aid processing on CAEP latency and amplitude measures.

3.4.1 Effect of condition: aided vs. unaided

The overshoots at the onset of tone bursts in aided stimuli (see Figure 3-2) are consistent with the description of temporal changes associated with a WDRC circuit when the stimulus level is above the compression knee-point (Dillon, 2001). No significant effect of condition was found for CAEP peak latencies and amplitude. A possible explanation for non-significant differences in the CAEP attributes could be the small magnitudes of the changes in rise-time due to hearing aid processing. The CAEP peak latency varies non-linearly with tone burst rise-time (i.e., latency increases at a slower rate than increases in rise-time). For example, at supra-threshold levels (similar to this study), for up to 20 ms increase in rise-time (10 to 20 ms or 10 to 30 ms), the latency of the peaks increases by roughly 6 ms or less (Kodera, Hink, Yamada, & Suzuki, 1979; Onishi & Davis, 1968). Hence, for changes in rise-time as small as 2 ms and 8 ms in the present study, the change expected in the peak latency would be $<$6 ms, and no significant change was found. The reduction in N1-P2 amplitude with an increase in rise-time of $<$50 ms has been reported to be non-significant (Onishi & Davis, 1968), or $\sim$0.2 μV for changes in rise-time of 15 ms (Kodera et al., 1979). No substantial amplitude changes would be expected for the small changes in rise-time observed here, and none were found. Additionally, these reported differences are smaller than the within-subject test-retest differences in peak latency and amplitude that were observed in the present study. These results suggest that the changes occurring in the tone burst onset due to hearing aid processing (with the hearing aid used in this study), may not confound CAEP measures in similar testing conditions.

Although the aided stimulus reached plateau amplitude earlier than the unaided stimulus in addition to the overshoot, the maximum increase in stimulus level (dB calculated using RMS amplitude with Spectraplus) in the first 30 ms, which is the integration time window for the CAEP (Onishi & Davis, 1968), was <2 dB. This implies that the overshoot increases the stimulus onset level by only a small amount and hence is likely to have a small level effect on the CAEP. Although this increase of 2 dB (maximum) did not significantly affect the CAEP attributes in individuals with normal audiometric thresholds in this study, individuals presenting with recruitment may show variations in this effect due to faster growth of loudness (Moore, 2007). In addition, because the effects of stimulus rise-time on the CAEP can be explained on the basis of temporal integration (Onishi & Davis, 1968), the effect of altered rise-time may influence aided CAEPs in individuals with cochlear hearing losses differently because poorer temporal processing relative to individuals with normal hearing has been documented in individuals with cochlear hearing losses (Florentine, Fastl, & Buus, 1988; Moore, 2007). The attack time in the hearing aid used in the present study was 10 ms (measured using Audioscan Verifit). Hence, it is likely that hearing aids with similar attack times will show similar changes to such stimuli.

Because the hearing aid was programmed for a hypothetical hearing loss of severe degree, it used high gain. Electroacoustic evaluation of the effect on the stimulus across varying hearing aid gain revealed that higher gain produced larger overshoots. Specifically, the same hearing aid programmed for greater degrees of hearing loss produced larger overshoots than when it was programmed for lesser degrees of hearing loss as the gain prescribed increases with poorer hearing thresholds. Therefore, the lack of effect on CAEP with this extent of overshoot can suggest a lack of effect with smaller overshoots, which would occur with less gain. In the present study, several factors such as level and SNR were artificially altered to evaluate the temporal effect of hearing aid processing on the onset of tone bursts.

Therefore, this does not mimic a typical unaided-aided condition comparison. Also, the scope of the present investigation was limited to a specific stimulus level and one hearing aid programmed for a specific hearing loss. Interactions between stimulus input level and gain applied that occur in a WDRC circuit may affect the onset of the tone burst differently, but this is beyond the scope of this study.

3.4.2 SNR

The SNR at 1 kHz in the aided condition was 44.9 dB for the 7.5 ms rise-time and 44.5 dB for the 20 ms rise-time. This is much higher than the best SNR of 22.2 dB at the stimulus peak frequency reported in Billings et al. (2011). There are several possible explanations for this difference. One reason could be differences in the noise floors of the hearing aids used. The second reason could be the higher input stimulus level that was used in the present study. The input level used in Billings et al. (2011) was 40 dB SPL, whereas it was 60 dB SPL in the current study. Assuming that the microphone is the dominant source of internal noise in the hearing aid (Agnew, 1997; Thompson et al., 2002), the SNR at the input of the amplifier will be lower when the level of the tone burst/stimulus is lower. At any instant, the gain applied to the constant noise floor and other input stimuli will be the same because they occur simultaneously, and the gain in such cases is determined only by the higher level stimulus (Wolfe, Miller, Swim, & Shafer, 2007). Therefore, it could be argued that the SNR in the Billings et al. (2011) study was probably lower due to the input level being closer to ambient noise and/or hearing aid noise floor.

3.4.3 Tone burst rise-time

Tone bursts of two rise-times were used to evaluate the effects of hearing aid processing. Tone burst rise-time, as a main variable, showed no significant effect on CAEP peak latencies or amplitude. A trend of longer peak latencies and smaller

N1-P2 amplitude with the longer rise-time can be observed (Table 3-1). This is in general agreement with literature examining the relationship between tone burst rise-time and CAEP attributes. The reader is referred to the Introduction of this paper, where the physiological basis of the effect of stimulus rise-time on the CAEP is described. In some studies, no statistical analysis was carried out (Onishi & Davis, 1968; Skinner & Jones, 1968). Kodera et al. (1979) found no significant change in N1-P2 amplitude and statistically significant increases in N1 and P2 latencies of $\sim$6 ms with an increase in rise-time from 5 ms to 20 ms. These small changes are consistent with the numerical, albeit not statistically significant, $\leq$5 ms increase that was observed in the present study.

3.5 Conclusion

This study was designed to explore the effects of stimulus modifications (specifically tone bursts) caused by non-linear hearing aid processing in individuals with normal audiometric thresholds when hearing aid noise floor, processing delay, and gain were controlled. SNR was artificially matched between the aided and unaided conditions, and the effects of processing delay and hearing aid gain were removed during stimulus preparation. This allowed examination of the specific effects of altered rise-time caused by hearing aid processing on the CAEP. The findings of this study revealed that the alterations in tone burst rise-times that were caused by the hearing aid used in this study (after controlling for gain, delay, and SNR) were not large enough to significantly influence CAEP responses in normal hearing listeners. This was illustrated using a clinically applicable hearing aid scenario with a well-controlled method. This effect still needs to be explored in individuals with hearing impairment to fully evaluate its clinical significance. These findings may only be generalized to conditions with hearing aids of similar configurations such as attack time and when tone bursts are used.

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Chapter 4

Electroacoustic comparison of hearing aid output of phonemes in running speech versus isolation: Implications for aided cortical auditory evoked potentials testing[5]

4.1 Introduction

Hearing aid validation using aided speech-evoked Auditory Evoked Potentials (AEPs) 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 non-linear, may have implications for the nature of speech stimulus used as input. The present study focused on the effect of non-linear hearing aid processing on speech stimuli used for measurement of Cortical Auditory Evoked Potentials (CAEPs).

Non-linear 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

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[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 (Dillon, 2012; Frye, 1987; Stelmachowicz, Lewis, Seewald, & Hawkins, 1990; Stone & Moore, 1992). 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 (Dillon, 1996, 2012; Fortune, 1997; Henning & Bentler, 2005, 2008; Stelmachowicz, Kopun, Mace, & Lewis, 1996). 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.

Non-linear hearing aids, being sensitive to features of the input signal, process speech or speech-like stimuli differently from non-speech stimuli (Henning & Bentler, 2005; Scollie & Seewald, 2002; Stelmachowicz et al., 1996, 1990). 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. Behavioral 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 (Bench, Kowal, & Bamford, 1979), or materials with less grammatical context such as isolated words or nonsense syllables (e.g., The Nonsense Syllable test, Dubno & Schaefer, 1992). But the speech stimuli may need to be modified for use in alternative validation methods such as aided AEPs (e.g., Gravel, 1989; Korczak, Kurtzberg, & Stapells, 2005; Purdy et al., 2005; Tremblay, Kalstein, Billings, & Souza, 2006).

Aided AEPs have used speech and non-speech stimuli (Dimitrijevic, John, & Picton, 2004; Gravel, 1989; Hecox, 1983; Korczak et al., 2005; Purdy et al., 2005; Tremblay, Kalstein, et al., 2006). 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 (Gravel, 1989; Tremblay, Billings, Friesen, & Souza, 2006; Tremblay, Kalstein, et al., 2006; Van Dun, Carter, & Dillon, 2012). Often phonemes or syllables excised from running speech or from standard speech tests have been used to record reliable CAEPs (e.g., Easwar, Glista, Purcell, & Scollie, 2012a; Golding et al., 2007; Tremblay, Friesen, Martin, & Wright, 2003). Although natural speech can be used as stimuli, CAEP testing involves presentation of these stimuli with inter-stimulus intervals. These inter-stimulus intervals usually range on the order of 1–2 seconds (e.g., Easwar et al., 2012a; Golding, Purdy, Sharma, & Dillon, 2006; Tremblay, Kalstein, et al., 2006) optimized for the latency of CAEPs and refractory periods of the cortical pyramidal neurons (Budd, Barry, Gordon, Rennie, & Michie, 1998; B. A. Martin, Tremblay, & Stapells, 2007). These stimuli are repeated 100-200 times, with constant or slightly variable inter-stimulus intervals 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 (Dillon, 2005; Hyde, 1997; Purdy et al., 2005). 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; Lightfoot & Kennedy, 2006; Tsui, Wong, & Wong, 2002). 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., Korczak et al., 2005; Purdy et al., 2005; Tremblay, Billings, et al., 2006; Tremblay, Kalstein, et al., 2006). Commercial equipment such as the HEARLab use phonemes sampled across the speech frequency range. These phonemes are presented in isolation, with over 1 second of inter-stimulus interval, at their naturally occurring levels within running speech (R. Martin, Villasenor, Dillon, Carter, & Purdy, 2008).

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 inter-stimulus interval (a silence period) whereas the same phoneme in running speech is likely to be preceded by other phonemes. Since non-linear 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 inter-stimulus interval preceding every repetition of the stimulus, non-linear hearing aids may demonstrate an overshoot at the onset of the stimulus consistent with compression circuitry (Easwar, Glista, Purcell, & Scollie, 2012b). 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/inter-stimulus interval 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 non-linear signal processing in hearing aids may attenuate the level of speech sounds immediately preceded by silence (Brennan & Souza, 2009; Plyler, Hill, & Trine, 2005).

The effects of CAEP protocols on the gain achieved while processing tone bursts have been reported recently (Jenstad, Marynewich, & Stapells, 2012; Marynewich, Jenstad, & Stapells, 2012). 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 post-stimulus 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 (Onishi & Davis, 1968). Stimulus level of the hearing aid processed tone bursts was positively related to the CAEP amplitude, with stimulus level at 30 ms post-stimulus onset being a better predictor of CAEP amplitude compared to maximum stimulus level. These reports (Jenstad et al., 2012; Marynewich et al., 2012) substantiate the need to verify output levels of CAEP stimuli across contexts, and to consider stimulus onsets. The present study therefore measured both overall level (level measured across the entire duration of the phoneme) and onset level of 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.

4.2 Method

4.2.1 Hearing aids

Ten hearing aids sampled across various manufacturers were chosen. A list of the hearing aids used is provided in Table 4-1. 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 Desired Sensation Level (DSL) v5a adult prescription targets (Scollie et al., 2005) for an N4 audiogram (Bisgaard et al., 2010). 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 (Table 4-2; Bisgaard et al., 2010). The remaining four hearing 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 (Table 4-2; Bisgaard et al., 2010). 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 non-linear characteristics of the devices, in isolation of these other aspects of hearing aid signal processing.

N4 audiogram	**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	

Table 4-1: Hearing aids

	Frequency-specific thresholds (dB HL)								
Frequency (kHz)/ Audiogram	0.25	0.5	0.75	1	1.5	2	3	4	6
N4	55	55	55	55	60	65	70	75	80
N6	75	80	85	85	90	90	95	100	100

Table 4-2: Frequency-specific thresholds of N4 and N6 standard audiograms (Bisgaard et al., 2010). The threshold at 0.75 kHz for the N6 audiogram was originally 82.5 dB HL but was rounded up to 85 dB HL to allow input into the verification system.

Note. Modified from "The effect of stimulus choice on cortical auditory evoked potentials (CAEP): Consideration of speech segment positioning within naturally produced speech," by V. Easwar, D. Purcell & S. Scollie, 2012, *International Journal of Otolaryngology Special Issue 'Hearing aids and the Brain'*, doi: 10.1155-2012-518202. Copyright Hindawi Publishing Corporation. Reused with permission.

4.2.2 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. Boundaries of these phonemes were chosen such that any transitions preceding and following these phonemes due to co-articulation were excluded. 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. 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 (Ling, 1989; Scollie et al., 2012). The last three have been commonly used in a series of aided CAEP studies (e.g., Chang, Dillon, Carter, Van Dun, & Young, 2012; Golding et al., 2007; Van Dun et al., 2012) and are also a part of the stimulus choices available in the HEARLab. A silent interval of 1125 ms preceding each phoneme was created using sound editing software Goldwave (version 5.58). This was to simulate a CAEP stimulus presentation protocol where the inter-stimulus interval usually ranges between one and two seconds.

4.2.3 Recording apparatus

Recordings of hearing aid output used a click-on coupler Brüel and Kjær type 4946 (conforming to IEC 60126 fitted with microphone type 4192) with an earplug simulator. The hearing aid was connected via 25 mm of size 13 tubing (American National Standards Institute [ANSI], 2003). This was set up in a Brüel and Kjær 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 (version 5.0.26.0) in separate channels using a sampling rate of 44100 Hz with 16-bit sampling precision. Spectraplus software was used to record the reference and coupler signals as .wav files for further signal analyses.

4.2.4 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 (Olsen, 1998). 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 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 inter-stimulus interval of 1125 ms) were presented during any single recording.

4.2.5 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, inter-repetition variability was observed to be high in the first few repetitions. This is likely related to non-linear signal processing in the hearing aids but these effects were not formally evaluated in this study.

4.2.6 Analyses

Individual Repeated Measures Analysis of Variance (RM-ANOVA) were completed for overall phoneme level and onset phoneme level using SPSS (version 16). Context (running speech and isolation), level (55, 65, and 75 dB SPL), and phoneme were the three independent/within-subject factors. 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, false discovery rate method was used for multiple comparisons (Benjamini & Hochberg, 1995).

4.3 Results

4.3.1 Difference in overall phoneme level across contexts

The RM-ANOVA revealed a significant effect of context ($F(1, 9) = 10.84$, $p = 0.009$, $\eta^2_{partial} = 0.55$), input level ($F(1.02, 9.14) = 853.53$, $p < 0.001$, $\eta^2_{partial} = 0.99$) and phoneme ($F(2.05, 18.46) = 85.69$, $p < 0.001$, $\eta^2_{partial} = 0.91$). Two-way interactions between input level and context ($F(1.61, 14.47) = 9.141$, $p = 0.004$, $\eta^2_{partial} = 0.50$), phoneme and context ($F(2.93, 26.39) = 3.06$, $p = 0.047$, $\eta^2_{partial} = 0.25$), and input level and phoneme ($F(2.88, 25.88) = 5.29$, $p = 0.006$, $\eta^2_{partial} = 0.37$) were also significant. The three-way interaction between input level, context, and phoneme was not significant ($F(2.97, 26.76) = 1.42$, $p = 0.259$, $\eta^2_{partial} = 0.14$). Collapsed across phonemes, 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

4-1A and Table 4-3 for group means). In summary, the difference between contexts reduced as input level increased.

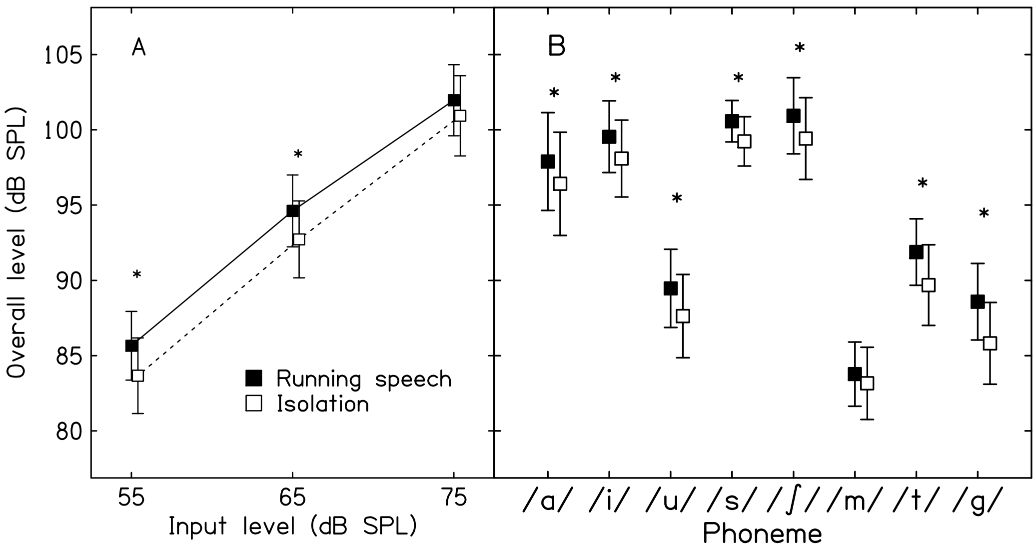

Figure 4-1: Subplot A presents variation of overall phoneme level in running speech and isolation context across input levels. Subplot B presents the same across phonemes. Error bars represent ±1 *SE*. * indicates a statistically significant difference in paired contrasts. The symbols have been offset slightly to improve clarity.

Modified from "The effect of stimulus choice on cortical auditory evoked potentials (CAEP): Consideration of speech segment positioning within naturally produced speech," by V. Easwar, D. Purcell & S. Scollie, 2012, *International Journal of Otolaryngology Special Issue 'Hearing aids and the Brain'*, doi: 10.1155-2012-518202. Copyright Hindawi Publishing Corporation. Reused with permission.

Collapsed across levels, paired contrasts comparing overall phoneme levels between contexts for each phoneme showed significant differences for all phonemes except /m/ (see Figure 4-1B and Table 4-4 for group means). All phonemes except /m/ were higher in level when they occurred in running speech compared to when they occurred in isolation.

	Input level	Running speech M[dB SPL], SE[dB]	Isolation M[dB SPL], SE[dB]	t statistic, df	p value	Critical p value
Overall level	55	85.66, 2.28	83.64, 2.51	4.45, 9	0.001*	0.017
	65	94.61, 2.38	92.72, 2.55	3.80, 9	0.004*	0.033
	75	101.97, 2.36	100.9, 2.66	1.17, 9	0.121	0.050
Onset level	55	86.35, 2.27	85.96, 2.31	4.23, 9	0.002*	0.017
	65	95.18, 2.35	94.93, 2.38	2.74, 9	0.023*	0.033
	75	102.48, 2.41	102.43, 2.44	0.45, 9	0.653	0.050

Table 4-3: Results of post-hoc analyses for level-context interaction. * indicates a significant difference. (n = 10)

Note. Modified from "The effect of stimulus choice on cortical auditory evoked potentials (CAEP): Consideration of speech segment positioning within naturally produced speech," by V. Easwar, D. Purcell & S. Scollie, 2012, *International Journal of Otolaryngology Special Issue 'Hearing aids and the Brain'*, doi: 10.1155-2012-518202. Copyright Hindawi Publishing Corporation. Reused with permission.

	Phoneme	Running speech M[dB SPL], SE[dB]	Isolation M[dB SPL], SE[dB]	t statistic, df	p value	Critical p value
Overall level	/a/	97.89, 3.25	96.41, 3.43	3.19, 9	0.011*	0.025
	/i/	99.54, 2.38	98.09, 2.55	2.86, 9	0.019*	0.038
	/u/	89.46, 2.59	87.62, 2.77	4.23, 9	0.002*	0.006
	/s/	100.57, 1.37	99.23, 1.64	3.51, 9	0.007*	0.019
	/ʃ/	100.93, 2.52	99.42, 2.71	3.98, 9	0.003*	0.013
	/m/	83.77, 2.13	83.16, 2.39	0.95, 9	0.365	0.050
	/t/	91.88, 2.20	89.69, 2.67	2.43, 9	0.038*	0.044
	/g/	88.57, 2.54	85.82, 2.73	2.45, 9	0.011*	0.031
Onset level	/a/	97.09, 3.41	95.69, 3.67	2.38, 9	0.042	0.042
	/i/	98.77, 2.45	96.85, 2.64	2.59, 9	0.029	0.025
	/u/	91.37, 2.60	89.34, 2.91	3.14, 9	0.012*	0.017
	/s/	99.16, 1.44	96.97, 2.14	2.49, 9	0.034	0.033
	/ʃ/	99.08, 2.24	96.63, 2.59	4.16, 9	0.002*	0.008
	/m/	82.54, 2.23	82.07, 2.48	0.63, 9	0.542	0.050

Table 4-4: Results of post-hoc analyses for phoneme-context interaction. * indicates a significant difference. (n = 10)

Note. Modified from “The effect of stimulus choice on cortical auditory evoked potentials (CAEP): Consideration of speech segment positioning within naturally produced speech,” by V. Easwar, D. Purcell & S. Scollie, 2012, *International Journal of Otolaryngology Special Issue ‘Hearing aids and the Brain’*, doi: 10.1155-2012-518202. Copyright Hindawi Publishing Corporation. Reused with permission.

4.3.2 Difference in phoneme onset level across contexts

The RM-ANOVA revealed a significant effect of context ($F(1, 9) = 7.41$, $p = 0.024$, $\eta^2_{partial} = 0.45$), input level ($F(1.05, 9.44) = 846.94$, $p < 0.001$, $\eta^2_{partial} = 0.99$) and phoneme ($F(1.78, 16.04) = 52.85$, $p < 0.001$, $\eta^2_{partial} = 0.85$). Two-way interactions between input level and context ($F(1.20, 10.81) = 17.72$, $p = 0.001$, $\eta^2_{partial} = 0.66$), and phoneme and context ($F(3.45, 31.09) = 3.96$, $p = 0.013$, $\eta^2_{partial} = 0.31$) were significant. Interaction between input level and phoneme ($F(2.06, 18.56) =$

1.49, $p = 0.250$, $\eta^2_{partial} = 0.14$) and the three-way interaction between input level, context, and phoneme were not significant ($F(3.25, 29.25) = 0.87$, $p = 0.473$, $\eta^2_{partial} = 0.09$). Collapsed across phonemes, paired contrasts between phoneme 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 4-2A and Table 4-3 for group means). Similar to overall phoneme level, the difference between contexts reduced with increasing input level.

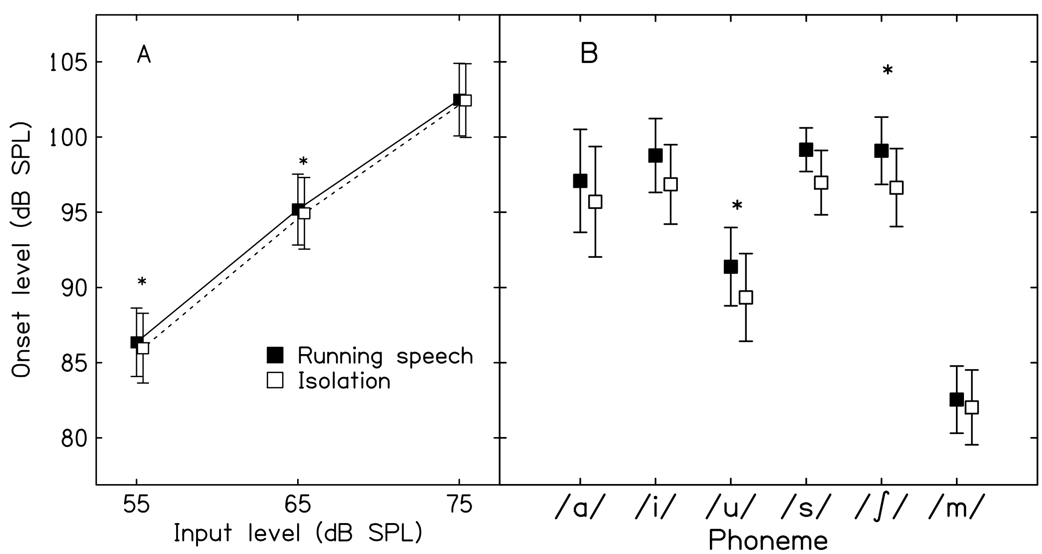

Figure 4-2: Subplot A presents variation of phoneme onset level in running speech and isolation context across input levels. Subplot B presents the same across phonemes. Error bars represent ±1 *SE*. * indicates a statistically significant difference in paired contrasts. The symbols have been offset slightly to improve clarity.

Paired contrasts comparing phoneme onset levels between contexts for each phoneme revealed no significant differences for all phonemes except /ʃ/ and /u/ (see

Figure 4-2B and Table 4-4 for group means). Averaged across levels, phonemes /ʃ/ and /u/ were higher in onset level when they occurred in running speech compared to when they occurred in isolation.

4.3.3 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 4-5A and 4-5B 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.

	Hearing aid	Phoneme /a/	/i/	/u/	/s/	/ʃ/	/m/	/t/	/g/
(A)	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	-0.37
	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

	Hearing aid	Phoneme /a/	/i/	/u/	/s/	/ʃ/	/m/
(B)	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

Table 4-5: (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). (n = 10)

Note. Modified from "The effect of stimulus choice on cortical auditory evoked potentials (CAEP): Consideration of speech segment positioning within naturally produced speech," by V. Easwar, D. Purcell & S. Scollie, 2012, *International Journal of Otolaryngology Special Issue 'Hearing aids and the Brain'*, doi: 10.1155-2012-518202. Copyright Hindawi Publishing Corporation. Reused with permission.

The proportion of difference values greater than ±3 and ±5 dB are presented in Table 4-6 for both overall phoneme levels and phoneme onset levels at each input level. Pooled across both directions of differences and input levels, about 24% of the overall phoneme levels (total of 240 observations across three levels, 10 hearing aids and eight phonemes) 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 measurements, 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.

	Input level	>3 dB	<-3 dB	>5 dB	<-5 dB
Overall level	55	9.58	0.42	–	2.92
	65	9.17	–	–	2.08
	75	4.58	0.83	–	2.50
Onset level	55	12.78	–	7.22	–
	65	12.22	1.11	3.33	–
	75	6.11	1.11	1.67	0.56

Table 4-6: Proportion of observations (%) showing differences greater than 3 or 5 dB (positive value indicates higher output levels of phoneme in running speech).

Note. Modified from "The effect of stimulus choice on cortical auditory evoked potentials (CAEP): Consideration of speech segment positioning within naturally produced speech," by V. Easwar, D. Purcell & S. Scollie, 2012, *International Journal of Otolaryngology Special Issue 'Hearing aids and the Brain'*, doi: 10.1155-2012-518202. Copyright Hindawi Publishing Corporation. Reused with permission.

4.4 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 recent reports (Jenstad et al., 2012; Marynewich et al., 2012). 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 (Jenstad et al., 2012; Marynewich et al., 2012) were set to function linearly, unlike the hearing aids used in the present study. Another study that used a non-linear hearing aid to study the effect of hearing aid processing on tone burst onsets while comparing it with unaided conditions (Easwar et al., 2012b), reported 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. 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 non-linear 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 non-linear signal processing in hearing aids (Brennan & Souza, 2009; Plyler et al., 2005).

4.4.1 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 (Dillon, 1996, 2012). Compression limiting restricts the maximum output level by using a very high or infinite compression ratio. 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 between contexts varied across phonemes. We did not perform a direct comparison across phonemes because the individual phonemes occur at different levels within running speech. Compression, being a level-dependent non-linear 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 4-5A 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 non-linearity (Brennan & Souza, 2009; Dillon, 2012; Plyler et al., 2005). However, this study did not systematically evaluate 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.

4.4.2 Inter-hearing aid variability

Differences in Tables 4-5A and 4-5B illustrate that individual hearing aids may amplify individual phonemes differently, even though they were set to produce similar gain for long-duration signals. This may reflect the different non-linear signal processing strategies employed by manufactures. Differences across hearing aid manufacturers were also reported by Jenstad et al. (2012). 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, Keidser, O'Brien, & Silberstein, 2003). The finding that hearing aids show individual variability makes it challenging to predict the nature of differences on a case-by-case basis in clinical practice.

4.4.3 Implications for aided CAEP testing

CAEPs are level-dependent (Billings, Tremblay, Souza, & Binns, 2007; Chang et al., 2012; Golding, Dillon, Seymour, & Carter, 2009; Van Dun et al., 2012). 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 under-estimate audibility when phonemes are presented in isolation. These data indicate that under-estimation 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 over-estimation 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 inter-stimulus interval and one naturally occurring preceding context per phoneme, generalization to other instances and variation across durations or levels of phonemes may require further investigation. 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 may occur, along with considerable inter-hearing aid variability. In about a fourth of the observations, 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. A few of the differences exceeded that of an audiometric step size and therefore may be clinically important.

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Chapter 5

Sensitivity of envelope following responses to vowel polarity[6]

5.1 Introduction

In electrophysiological measurements, averaging responses to opposite polarities of the stimulus has been recommended for reducing or eliminating stimulus artifact (Akhoun et al., 2008; Arnold, 2007; Small & Stapells, 2004). Averaging responses to opposite polarities has also been used to enhance Envelope Following Responses (EFR) by reducing contamination by the Frequency Following Response (FFR) and cochlear microphonic (Aiken & Picton, 2008). The EFR is a response phase-locked to the stimulus envelope which is minimally affected by an inversion of stimulus polarity and hence preserved when responses to opposite polarities are added (Aiken & Picton, 2008; Gockel, Carlyon, Mehta, & Plack, 2011; Greenberg, Marsh, Brown, & Smith, 1987; Small & Stapells, 2005). The FFR is a neural response phase-locked to the stimulus fine structure/spectral characteristics which is sensitive to inversion of the stimulus (Aiken & Picton, 2008; Chimento & Schreiner, 1990; Huis in't Veld, Osterhammel, & Terkildsen, 1977; Krishnan, 2002). Likewise, the cochlear microphonic is a pre-synaptic response generated from the outer hair cells that also

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closely follows the polarity of the stimulus (Chimento & Schreiner, 1990; Huis in't Veld et al., 1977). Therefore, averaging responses to alternate polarities to enhance EFRs is based on evidence that demonstrates sensitivity of the FFR and cochlear microphonic to stimulus polarity as well as general insensitivity of the EFR to stimulus polarity, preserving it in the averaged alternate polarity response.

Only a few studies report and evaluate differences in EFRs due to stimulus polarity. The stimuli used in these studies are either amplitude-modulated (AM) tones or vowels. In a study that used AM tones, a small but statistically significant difference was noted in the response amplitude (mean difference of ∼6 nV) and phase (mean difference of ∼13 degrees; Small & Stapells, 2005). In an earlier study that used synthetic vowel stimuli (Greenberg, 1980, pp. 122–125), polarity-sensitive amplitude differences were observed but not quantified. In a more recent study that used natural vowel stimuli (/a/ and /i/), significant and marginally larger differences (mean of absolute differences was ∼17 nV) were observed in EFR response amplitudes to opposite stimulus polarities (Aiken & Purcell, 2013). Although EFR amplitudes to opposite polarities were significantly different in both these studies, response amplitudes in conditions where opposite polarities were averaged did not vary significantly from the responses where only the same polarity was averaged (Aiken & Purcell, 2013). This is further supported by a more recent study that illustrated no significant differences in response amplitudes at the fundamental frequency (for the stimulus /da/) between condensation, rarefaction and alternating polarity stimulus conditions (Kumar, Bhat, D'Costa, Srivastava, & Kalaiah, 2013). These results show that EFRs elicited by AM tones and certain vowels may be sensitive to polarity but only to a degree that does not significantly affect response amplitude estimates at the group level when responses to opposite polarities are averaged.

However, it is important to note that individual variations exist in the degree of polarity sensitivity. In Aiken and Purcell (2013), describing data from nine participants, differences in response amplitudes observed due to polarity in any individual ranged from a minimum of 0.4 to a maximum of 72 nV. The differences varied across individuals and across stimuli within individuals. At the group level, averaging responses to opposite polarities did not affect the EFR amplitudes. However, it is likely that averaging responses to opposite polarities may under-estimate the maximum response amplitude possible in those individuals exhibiting large amplitude differences due to polarity. The extent of polarity-sensitive amplitude differences may also affect response detection, as statistical detection tests (for example, the F-test) are typically based on the individual's response averaged across polarity. Since individual variations are of significance for clinical measures, our first aim was to study the incidence and degree of polarity-sensitive amplitude differences in EFRs recorded on a large number of individuals in Experiment I.

Greenberg (1980, chapter 7) was the first to report polarity sensitivity for vowel stimuli. Figures 7.13 and 7.14 (Greenberg, 1980, chapter 7) illustrate that the EFR at the modulation frequency (the fundamental frequency of the voice, f_0) was not completely canceled when responses to opposite polarities of synthetic vowels were subtracted from each other. This provides evidence for differences in EFR characteristics in response to opposite stimulus polarities that, if otherwise equal, would have completely canceled in the subtracted average or the – – average shown in Aiken and Picton (2008; Panel L, Figure 2; the – – average is obtained by subtracting responses to opposite stimulus polarities). The residual EFR varied across different vowels and Greenberg's three participants. A suspected reason for the residual EFR activity in the subtracted response was the acoustic envelope asymmetry in the eliciting stimulus (Greenberg, 1980; Skoe & Kraus, 2010).

Another reason that could contribute to the polarity sensitivity of EFRs to vowel stimuli is the influence of the FFR. Unlike AM tones, the frequency at which the EFR is elicited is also the frequency of the first voice harmonic (h1). Since we know from previous studies that the FFR is sensitive to stimulus polarity, it is possible that polarity sensitivity of the EFR is a compound effect of both the FFR and EFR elicited at the same frequency. The FFR may add constructively or destructively with the EFR depending on the stimulus polarity. Hence, this study evaluated the influence of h1 (possibly eliciting an h1 FFR) by comparing polarity-sensitive differences in EFR amplitude elicited by vowels with and without h1.

The above discussion pertains to vowels elicited by the full bandwidth of a vowel, natural or synthetic. EFRs elicited by vowels are thought to be dominated by harmonics in the region of the first formant (F1), relative to the second (F2) and higher formants (F2+; Choi, Purcell, Coyne, & Aiken, 2013; Laroche, Dajani, Prevost, & Marcoux, 2013). The first seven to eight harmonics are spectrally resolved, meaning, the auditory filters respond to a single harmonic of the complex signal and hence the output of these filters is a sinusoid (Oxenham, 2008; Oxenham, Micheyl, & Keebler, 2009). But the higher order harmonics are unresolved, meaning, the auditory filters process more than one harmonic of the complex signal and hence the output is an interaction between individual harmonics (Oxenham, 2008). The F1 of English vowels are mostly below 1 kHz and the F2 are above 1 kHz (Hillenbrand, Getty, Clark, & Wheeler, 1995). Therefore, for f_0 ranging between 100 and 140 Hz (male talker), it is likely that the EFRs from the region of F1 are generated from harmonics that are spectrally resolved (the corresponding harmonic would be 700 to 980 Hz respectively) whereas EFRs generated from the region of F2 or F2+ are generated from harmonics that are spectrally unresolved (Laroche et al., 2013). In the literature, there is evidence to suggest that characteristics of EFRs elicited by lower frequency carriers or resolved harmonics vary from EFRs elicited by higher frequency carriers or unresolved harmonics. Data from Greenberg et al. (1987) illustrates that at low intensities, the modulation depth, systematically varied by altering the onset phase of individual harmonics, had no effect on the EFRs generated by a low frequency harmonic complex. However, the EFRs generated by a high frequency harmonic complex systematically reduced in amplitude with a reduction in modulation depth. This is similar to findings from Krishnan and Plack (2011), where the onset phase only affected EFRs generated from the region of unresolved harmonics. As well, the effects of noise on EFRs generated from the F1 region varies from those generated from regions of higher formants (Laroche et al., 2013). EFRs generated from the region of unresolved harmonics were more susceptible to noise compared to EFRs generated from resolved harmonics at equal stimulus levels and signal-to-noise ratios. This suggests that generation mechanisms and stimulus-response relationships likely vary for EFRs from the two regions. Hence, our second aim of this study was to evaluate polarity sensitivity of EFRs generated from the regions of the first and higher formants individually. Differentiating the two regions also helps to evaluate the contribution of h1, since it is present only in the F1 band.

In summary, this paper evaluated the incidence of polarity-sensitive differences and explored causes that may contribute to this phenomenon. In Experiment I, we evaluated the incidence and degree of polarity-sensitive amplitude differences in EFRs elicited by a naturally produced vowel in a group of 39 young adults. In Experiment II, we evaluated polarity-sensitive amplitude differences in EFRs across three different vowels as well as unresolved and resolved spectral regions of each vowel. Experiment II also tested the hypothesis that the presence of h1 contributes to the polarity sensitivity of EFRs.

5.2 Experiment I: Incidence and degree of polarity-sensitive amplitude differences in EFRs elicited by vowel /ε/

5.2.1 Method

Participants

Thirty-nine participants (24 females, 15 males) with ages ranging between 17 and 29 years (M_{age} = 22 years, SD_{age} = 3.35) were recruited from Western University and the city of London, Ontario. Participants reported learning English as their first language in Canada, predominantly in Ontario. Audiometric thresholds were determined using a bracketing approach (10 dB-down, 5 dB-up) under TDH-39 headphones at octave frequencies between 250 and 4000 Hz using a Madsen Itera audiometer. With the exception of one participant who had a slightly elevated threshold (35 dB HL) at 2 kHz in the left ear, thresholds of the remaining participants were lower than 20 dB HL. The participant with threshold >20 dB HL, was not excluded because of the suprathreshold stimulus presentation level and the repeated measures design used in this experiment. Participants reported no neurological, speech, hearing, or language impairments and provided informed consent. Routine otoscopy ruled out any contraindications such as occluding wax, discharge or foreign bodies in the ear. Participants were compensated for their time. The study protocol was approved by the Health Sciences Research Ethics Board, Western University, Canada.

Stimulus

The stimulus was the English vowel /ε/ in the word 'head' produced by a 28-year old male who had completed most of his schooling in Western Canada and Ontario

(Figure 5-1A). Each sweep consisted of the recorded token in its original polarity (Polarity A) followed by the same stimulus in the opposite polarity (Polarity B). Polarity B was obtained by multiplying the stimulus waveform in Polarity A by the factor -1. The duration of the original stimulus was 437 ms (vowel duration = 227 ms) and each sweep, with both polarities was approximately 6 seconds in duration. This included three other tokens (not shown in the figure) recorded for other applications, and hence data from these conditions were not included in the present study.

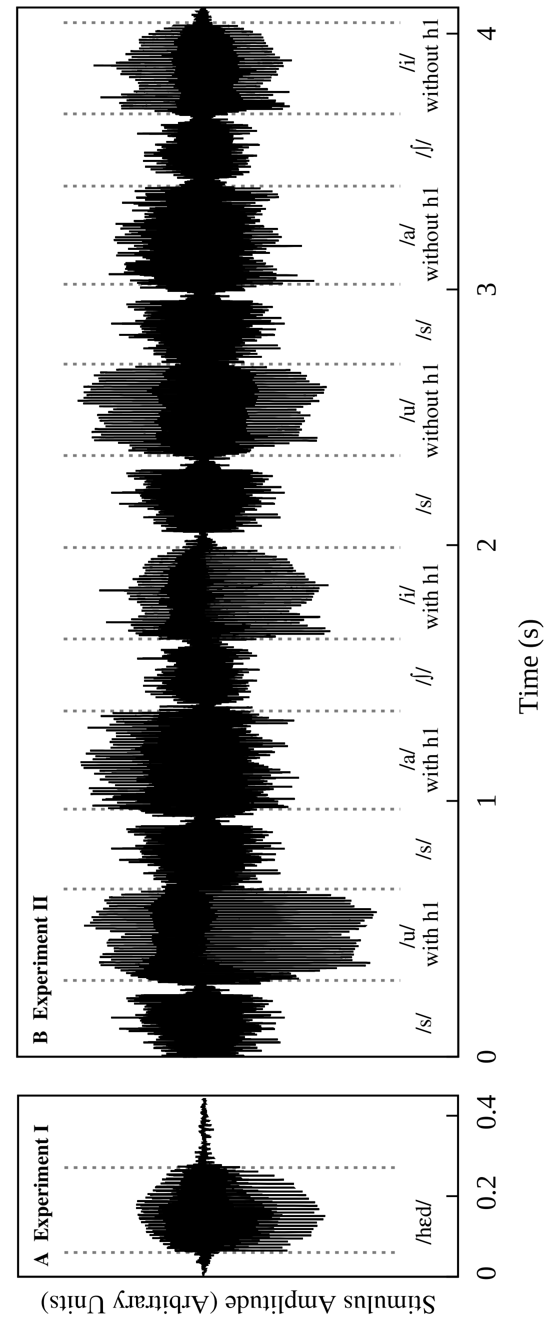

Figure 5-1: Amplitude-time waveform of stimuli used in Experiment I (panel A) and Experiment II (panel B). The vertical grey dashed lines indicate the analysis windows of each vowel.

Stimulus presentation and response recording

Stimulus presentation was controlled by software developed using LabVIEW (version 8.5; National Instruments, Austin, TX, USA). Digital to analog conversion of the stimulus and analog to digital conversion of the electro-encephalogram (EEG) were carried out using a National Instruments PCI-6289 M-series acquisition card. The stimulus was presented at 32000 samples per second with 16-bit resolution and the responses were recorded at 8000 samples per second with 18-bit resolution. The stimulus was presented using an Etymotic ER-2 insert earphone at a level of 80 dB SPL for 500 sweeps. The total recording time was approximately 55 minutes. Stimulus level was calibrated in an ear simulator (Type 4157) using a Brüel and Kjær Type 2250 sound level meter. The level was measured in flat weighted L_{eq} while the stimulus was presented for 60 seconds.

EEG was recorded using three disposable Medi-Trace Ag/AgCl electrodes. The non-inverting electrode was placed at the vertex, the inverting electrode was placed at the posterior mid-line of the neck just below the hairline, and the common was placed on the collarbone. Electrode impedances, measured using an F-EZM5 Grass impedance meter at 30 Hz, were maintained below 5 kΩ, with inter-electrode differences under 2 kΩ. A Grass LP511 EEG amplifier band-pass filtered the input EEG between 3 and 3000 Hz, and applied a gain of 50000. An additional gain of two was applied by the PCI-6289 card making a total gain of 100000.

The stimulus was presented monaurally to the left ear coupled with a foam tip of appropriate size. To minimize stimulus artifact, the ER-2 and the recording electrode leads were placed as far apart as possible. Participants were seated in a reclined chair in an electromagnetically shielded sound booth. A rolled towel was placed under their neck to help reduce neck tension and a blanket was provided for comfort. The lights were switched off and participants were encouraged to sleep during the recording.

To check for stimulus artifacts, the system was set up as described above with 10 individuals fitted with electrodes. The stimulus from the ER-2 was routed to a Zwislocki coupler on the participant's chest while the EEG was measured from the individual. The false positive rate obtained was equal to the assumed alpha for response analysis (5%).

EFR analysis and detection

Response analysis was carried out offline using MATLAB (version 7.11.0 [R2010b]; MathWorks, Natick, MA, USA). Each sweep was divided into four epochs of approximately 1.5 s each and a noise metric was computed for each epoch. The noise metric was the average EEG amplitude in each epoch between ∼80 and 120 Hz. Any epoch that exceeded the mean noise metric plus two standard deviations was discarded prior to averaging. EFRs to vowels were analyzed between pre-selected boundaries that were chosen such that the ramp-in and ramp-out sections at the beginning and the end of the vowel were excluded.

EFRs were analyzed in three conditions namely, Polarity A, Polarity B and Polarity AB average. The third condition refers to vector averaging responses to opposite Polarities A and B (the + − average in Aiken & Picton, 2008). This condition was included to determine if the averaging of opposite polarities affected response amplitude estimates. EFRs to the three conditions were analyzed individually using a Fourier analyzer (FA; Choi et al. 2013). Reference cosine and sine sinusoids were generated using the instantaneous f_0 frequency in the f_0 track. An estimated brainstem processing delay of 10 ms was used to correct for delay in the response (Aiken & Picton, 2008; Choi et al., 2013). The delay-corrected EEG was multiplied with the reference sinusoids to create real and imaginary components of the EFR, which were further averaged across the vowel duration to obtain a single complex number for each vowel. Each complex number provided an estimate of EFR amplitude and phase.

To determine if an EFR was detected, background EEG noise was estimated using five frequency tracks on either side of the response track at f_0. These noise tracks were adjacent to the response track and were separated (in Hz) by integral multiples of the reciprocal of the vowel duration (1/0.211 seconds = 4.74 Hz; Choi et al., 2013). Noise estimates from the 10 noise tracks were averaged to provide a single value for use in an F-test. For an EFR to be detected, the ratio of the EFR amplitude at the response frequency to the average noise estimate needed to exceed the critical F ratio (2, 20 degrees of freedom) of 1.86 at an α of 0.05. However, it was noted that certain participants had a significant detection in one polarity only. Since this may be an effect of polarity sensitivity, a participant's data was included for statistical analysis based on a significant detection in *either* Polarity A or B, or both. To avoid multiple comparison bias, the detection of EFRs was made more stringent using a Bonferroni correction (critical p value = 0.025). In cases where this criterion was not met in one polarity, the response amplitude estimated by the FA, although not significantly higher than the noise estimate, was used for statistical analyses as the best estimate of a small amplitude EFR buried in the noise. This estimate, albeit not independent of the noise level, was considered acceptable as this estimate may represent a small amplitude EFR. Using this estimate would be a more conservative approach than using zero when comparing amplitudes between conditions.

Statistical analysis

All statistical analyses in this study were completed using SPSS (version 21; IBM, Armonk, NY, USA). As described above, a participant's data was included for statistical analyses only if he/she had a significant detection in either Polarity A or B, or both (Bonferroni corrected). For comparison of EFR response amplitudes across polarity conditions, a one-way Repeated Measures Analysis of Variance (RM-ANOVA) was carried out with polarity as the within-subject factor. Polarity A, Polarity B and Polarity AB average were the three levels of the within-subject factor. For all statistical analyses completed in this study, Greenhouse-Geisser corrected degrees of freedom were used for interpretation. Statistical analyses were interpreted at an α of 0.05.

5.2.2 Results

Of the 39 participants, 15 did not have a significant EFR detection in either polarities A or B and hence were excluded from statistical analysis. We suspect that many participants did not have a detection due to the relatively short duration of the vowel that resulted in a wider FA bandwidth and consequently poorer signal-to-noise ratios (Choi et al., 2013). Of the remaining 24 participants included in the statistical analysis, 11 had a significant response in *either* polarity A or B whereas 13 participants had a significant response in both polarities. The number of significant detections was 20 and 17 in Polarity A and B, respectively. All 13 participants who had EFRs detected in both polarities, had a significant detection in the Polarity AB average. Of the 11 participants who had a significant detection in one polarity, two did not have a significant detection in the Polarity AB average.

Results of the RM-ANOVA indicated no significant effect of polarity on EFR amplitude, $F(1.03, 23.57) = 1.11$; $p = 0.305$; $\eta^2_{partial} = 0.05$. That is, the EFR amplitudes did not vary significantly between Polarity A ($M = 100.13$ nV; $SD = 38.12$), Polarity B ($M = 92.71$ nV, $SD = 36.18$) and Polarity AB average ($M = 94.08$ nV, $SD = 33.77$).

Although there were no significant differences in EFR amplitudes at the group level, the degree of polarity sensitivity varied across individuals. To examine individual variations, two histograms of amplitude changes between Polarity A and B were constructed (Figure 5-2). As with the RM-ANOVA, histograms included data from participants who had a significant EFR detection in at least one polarity. The histogram in the left panel (Figure 5-2A) plots the frequency of absolute amplitude differences between Polarity A and B. The histogram on the right panel (Figure 5-2B) plots the frequency of amplitude differences when amplitude in Polarity B was subtracted from Polarity A, thereby indicating direction of change. The first histogram of absolute differences indicates that the majority of participants exhibited <40 nV of amplitude change due to polarity. However, five participants showed differences >40 nV, with one participant presenting a difference as large as 110 nV. The second histogram also indicates that the majority of participants exhibited <±40 nV of amplitude differences between the two polarities. The minimal negative skew in the histogram suggests that Polarity A response amplitudes tended to be higher than Polarity B amplitudes.

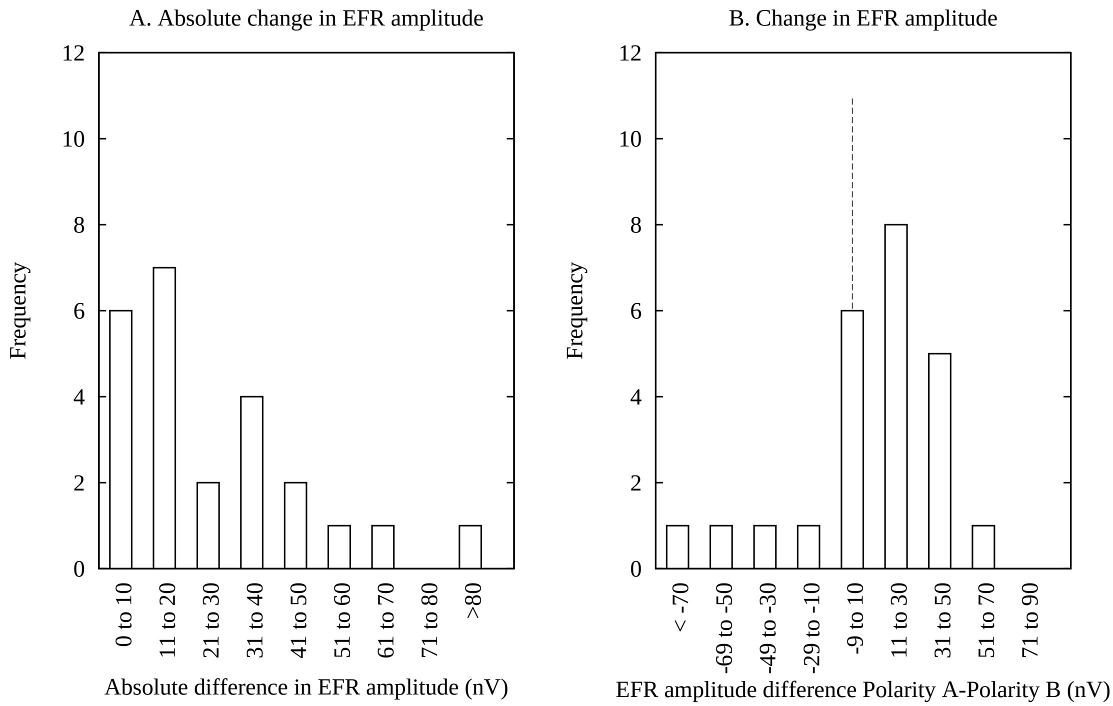

Figure 5-2: Incidence and degree of polarity-sensitive differences in EFR amplitude (n = 24). The vertical dashed line in subplot B indicates the bin containing zero/no difference in response amplitude between the two polarities.

To assess if variations in electrophysiological noise between the two polarity conditions contributed to EFR amplitude differences, a correlation analysis was completed between absolute differences in noise estimates and absolute differences in EFR amplitude estimates between Polarity A and B. Correlation between the variables was non-significant ($r(22)$ = -0.169, p = 0.431), suggesting that variations in electrophysiological noise are unlikely to have contributed to the polarity-sensitive differences in EFR amplitude. For the five participants who showed polarity-sensitive amplitude differences of >40 nV, the range of differences in noise estimates between the two polarities was between 0 and 11 nV.

In summary, Experiment I reveals that substantial polarity-sensitive amplitude differences in vowel-elicited EFRs may be observed with inter-subject variability in magnitude and direction. As noted in the Introduction, polarity sensitivity of EFRs may be influenced by factors such as the presence of h1, vowel and/or harmonics included in the stimulus. Experiment II probes this issue further using stimulus modifications to test EFRs evoked by low and high frequency harmonics of three different vowels, with and without h1.

5.3 Experiment II: Polarity-sensitive differences across vowels, with and without h1, and with low and high frequency harmonics

5.3.1 Method

Participants

Twenty participants (four males and 16 females) with ages ranging between 20 and 29 years (M_{age} = 23.4 years, SD_{age} = 2.42) were recruited from Western University. To be included in the study, all participants had to pass a hearing screen at 20 dB HL using TDH-39 headphones at octaves and inter-octaves between 250 Hz and 8000 Hz in both ears. Other screening procedures and criteria were similar to Experiment I. Five participants who participated in Experiment I also participated in Experiment II.

EFR test protocol

Stimulus

The vowels /u/ (as in 'school'), /a/ (as in 'pot') , and /i/ (as in 'see') were chosen to represent a range of F1 and F2 frequencies (see Table 5-1). These vowels were produced by a 42-year old male talker from Southwestern Ontario, whose first language was English. The vowels were interspersed with fricative sounds /s/ and /ʃ/ and hence the token produced by the male talker was /usaʃi/. One of many repetitions was chosen for further processing using Praat (Boersma, 2001) and Goldwave (version 5.58; Goldwave Inc., St. John's, NL, Canada) software based on vowel duration and flatness of f_0 contour. Voice recordings were made using a studio-grade microphone (AKG Type C 4000B) in a sound booth with software Spectraplus (version 5.0.26.0; Pioneer Hill Software LLC, Poulsbo, WA, USA) at a sampling rate of 44100 Hz and later down-sampled to 32000 Hz using Praat. During EFR recordings, the stimulus was presented repeatedly with no inter-stimulus interval, so the natural sequence was edited such that the fricative /s/ also preceded /u/, thereby making the stimulus token /susaʃi/. This modification maintained a consonant-vowel context for all vowel stimuli and avoided abrupt transitions between two repetitions of the stimulus token. The fricatives chosen for this stimulus were for other applications and hence, for the purposes of this study, only responses to the vowels were analyzed and are reported here. The duration of the final token (single polarity) was 2.05 seconds.

Stimulus vowels were individually edited using Praat. Prior to editing, the wav file was high-pass filtered at 50 Hz to reduce low frequency noise. Boundaries of phonemes were marked based on spectrograms and listening. To permit evaluation of polarity-sensitive differences in EFRs generated in the regions of low (resolved) and high (unresolved) frequency harmonics, vowels were edited such that the original f_0 was maintained in the regions of F2 and above (F2+), and a lower f_0 (original f_0–8Hz) was created in the region of F1. The original vowel was lowered in f_0 and low-pass filtered to isolate the F1 frequency band. The f_0 frequency was lowered using the 'Shift frequencies' function in Praat that shifts the f_0 by a specified amount (in Hz), while approximately maintaining the formant peak frequencies. The lower f_0 decreased the spacing between harmonics in the region of F1. The original vowel was high-pass filtered to isolate the F2+ frequency band. The F1 and F2+ bands were then combined to form the dual-f_0 vowel. The dual-f_0 structure of the processed vowels enabled recording of two simultaneous EFRs, one from the region of F2 and higher formants at the original f_0 frequency and the other, from the region of F1 at the original f_0–8 Hz. This permitted study of polarity-sensitive differences in EFRs generated by the two carriers presented simultaneously, like in naturally occurring vowels. The cut-off frequencies of the F1 and F2+ carriers were chosen close to the mid-point between the estimated F1 and F2. The carriers were filtered using very steeply sloping filter skirts in Praat. One may note that the cut-off frequencies are close to but not exactly between the 7th/8th harmonics to differentiate resolved and unresolved harmonics (Oxenham et al., 2009). However, with the cut-off frequencies chosen, it is likely that the carriers will be dominated by resolved harmonics in case of the F1 band and unresolved harmonics in case of the F2+ band. Duration, mean f_0 in each band, cut-off frequencies, and the harmonics included in each band for the three vowels, are provided in Table 5-1. Processed phonemes were concatenated using Goldwave. The average f_0 of the Experiment I talker was 10 Hz higher than the Experiment II talker.

Vowel	Duration (ms)	Band	F1 or F2 frequency (Hz)	Mean f_0 (Hz)	Cut-off frequency (Hz)	Harmonics
/u/	386	F1	266	92	615	h1 – h6
		F2+	1016	100	615	h7+
/a/	447	F1	684	88	1050	h1 – h11
		F2+	1467	96	1080	h12+
/i/	435	F1	277	89	1150	h1 – h12
		F2+	2160	97	1170	h13+

Table 5-1: Characteristics of F1 and F2+ vowel carriers

To investigate the possible influence of the presence of h1 on polarity sensitivity, another stimulus sequence was created with vowels that were high-pass filtered at 150 Hz to remove h1. This sequence was identical to the original stimulus sequence in all aspects except for the vowels (Figure 5-1B). The two stimulus sequences, one with the original dual-f_0 vowels and one with vowels without h1 were concatenated. Each sweep consisted of the stimulus sequence presented in both polarities, Polarity A followed by Polarity B. The duration of a sweep was 8.2 seconds. The stimulus was presented for 300 sweeps for a total duration of almost 41 minutes. Figure 5-1B on p. 101 shows the time-amplitude waveform of the stimulus in one polarity (Polarity A).

Stimulus presentation and response recording

Stimulus presentation and response recording were carried out using the same procedure, equipment and in the same environment described in Experiment I. The stimulus was presented at an overall level of 65 dB SPL, calibrated with flat weighted L_{eq} while the stimulus was presented for 60 seconds. Test ear was chosen randomly (11 right ears) and the recording was completed within one session for all participants.

Response analysis and detection

EFRs elicited by the F1 and F2+ carriers were estimated using an FA (described in Experiment 1) from the 300-sweep average EEG. To determine if EFRs were detected, the background EEG noise was estimated using 14 frequency tracks surrounding the response tracks (f_0 for F2+ and f_0–8 Hz for F1). Due to simultaneous presentation of F1 and F2+ carriers at f_0 frequencies 8 Hz apart, frequency tracks for noise estimates were chosen such that the response track of one band was not included as one of the noise tracks for the other. For example, the

F2+ response at f_0 Hz was excluded from the noise estimate of the F1 response at f_0–8 Hz and vice versa. One additional track on either side of the two response tracks (f_0 and f_0–8 Hz), and a track including 60 Hz were also excluded to reduce contamination that may arise from spectral leakage of energy into neighboring bands. For both the F1 and F2+ responses, there were six and eight noise tracks below and above the response track, respectively. Noise estimates from the 14 noise tracks were averaged to provide a single value used for the F-test. For an EFR to be detected, the ratio of the EFR amplitude at the response frequency to the average noise estimate needed to exceed the critical F ratio (2, 28 degrees of freedom) of 1.82 at an α of 0.05. The response estimates used for the F-test were corrected for possible overestimation of response amplitude due to noise (see Appendix of Picton, Dimitrijevic, Perez-Abalo, & Van Roon, 2005). As with Experiment I, a Bonferroni correction was applied to avoid multiple comparison bias when using the *either* polarity detection rule.

Statistical analysis

As in Experiment I, a participant's data was included for statistical analysis if he/she had a significant detection in at least one polarity (Bonferroni corrected). To evaluate polarity-sensitive differences in EFRs across the different stimulus conditions, a three-way RM-ANOVA was completed. Identical but separate analyses were completed for F1 and F2+ carriers. Since RM-ANOVA excludes individuals with even one missing data point, including F1 and F2+ carriers in the same analysis would have reduced the sample size substantially as individuals with non-significant responses were different for F1 and F2+ carriers. For the F1 and F2+ carriers, the three within-subject factors in the RM-ANOVA were polarity (A, B or AB average), vowel (/u/, /a/ or /i/) and h1 (with or without). For post-hoc analyses of significant interactions between within-subject variables, multiple paired

t-tests were completed. Critical p values were obtained using False Discovery Rate method (FDR; Benjamini & Hochberg, 1995) for multiple comparisons.

5.3.2 Pilot study: Comparison of between and within-polarity differences in EFR amplitude

Prior to exploring variability in polarity sensitivity across multiple conditions in Experiment II, we conducted a pilot study in a group of 16 young adults with normal hearing to evaluate whether changes associated with stimulus polarity could be differentiated from changes due to test-retest variability. That is, we compared between-polarity differences with within-polarity test-retest differences for vowels with h1 (similar to Aiken & Purcell, 2013). Results of the pilot study indicated that absolute differences in EFR amplitude due to polarity i.e., between-polarity differences, could be significantly greater than absolute differences in test-retest variability for certain carriers and hence are unlikely due to test-retest variations. Please refer to Appendix B for further details on the pilot study. As will be discussed below, results of the pilot study were consistent with the response patterns seen in Experiment II. Stimulus artifact checks were completed for the Experiment II stimulus. The false positive rate obtained was close to the assumed α for response analysis (see Appendix C for further details).

5.3.3 Results

Effect of noise variability on polarity-sensitive amplitude differences in EFRs

To assess if electrophysiological noise contributed substantially to polarity-sensitive differences in EFR amplitude, a correlation analysis was completed between absolute differences in noise estimates and absolute differences in EFR amplitude estimates between polarities A and B. The correlation between the two variables was found to be non-significant for both F1 ($r(110) = 0.052$; $p = 0.590$) and F2+ bands ($r(108) = -0.035$; $p = 0.718$). This suggests that variations in electrophysiological noise did not contribute substantially to the observed polarity-sensitive differences in EFR amplitude.

Individual variability in polarity-sensitive amplitude differences of EFRs

Like in Experiment I, individuals varied in their degree of change in EFR amplitude due to polarity. Figure 5-3 illustrates histograms of amplitude differences due to polarity for all three vowels, with and without h1, and for both F1 and F2+ carriers. The histogram peak of the F1 carrier of /u/ without h1 (Figure 5-3A) is shifted leftward towards the bin of zero differences compared to the with h1 condition. This suggests a decrease in polarity-sensitive amplitude differences with removal of h1. The F1 carrier of /i/ (Figure 5-3C) shows a modest trend of leftward shift in polarity-sensitive amplitude differences with removal of h1. The F2+ carrier of /i/ (Figure 5-3F) shows a trend of rightward shift in the histogram peak with removal of h1. The carriers /u/ F2+ and /a/ F1 and F2+, do not demonstrate a clear shift in histogram peak with and without h1. These results indicate that the effect of stimulus polarity on the EFR may be specific to certain vowels and formant carriers.

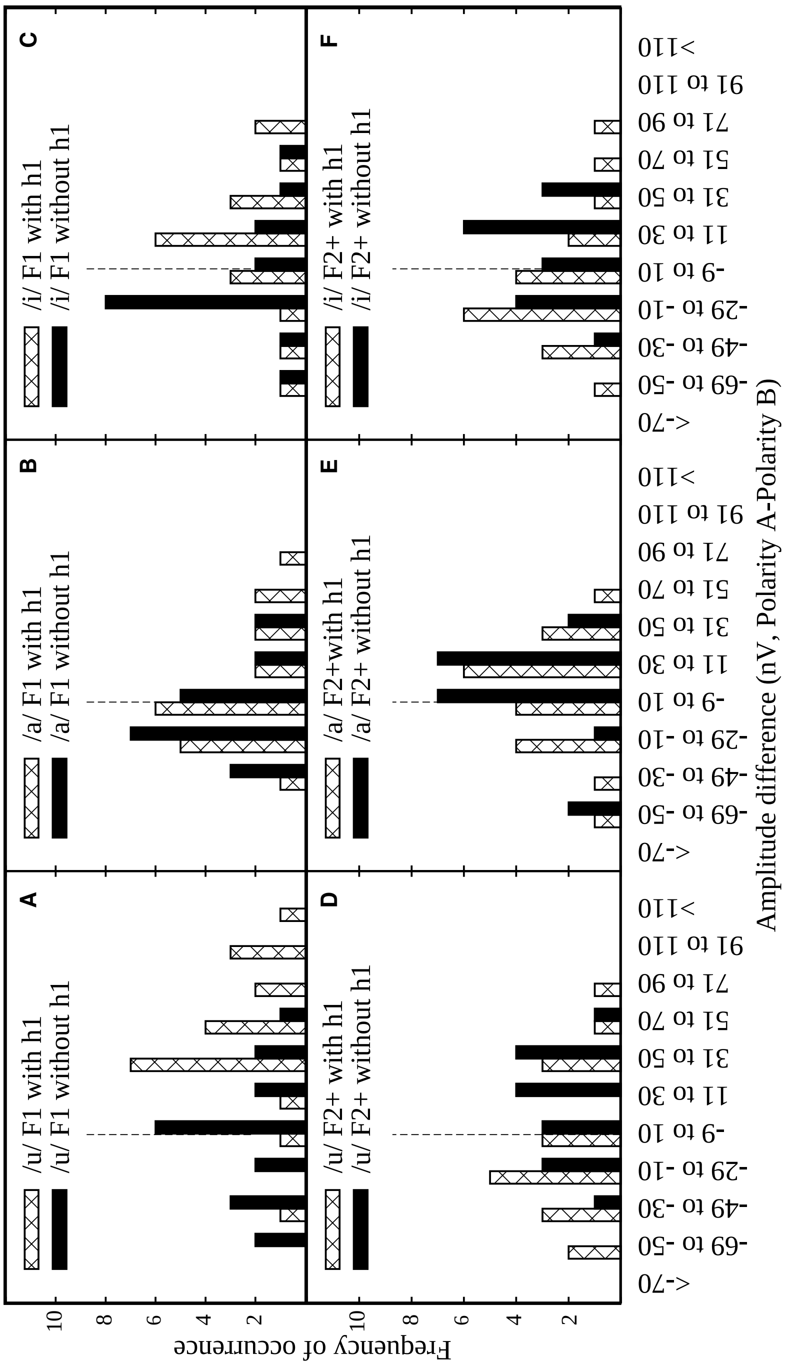

Figure 5-3: Individual variability in polarity-sensitive amplitude differences across carriers. The vertical dashed line indicates the bin containing zero/no difference in response amplitude between the two polarities.

Effect of polarity on F1 carriers

Of the 20 participants, 16 had a significant EFR detected in at least one polarity (Bonferroni corrected). Figure 5-4 illustrates mean EFR amplitudes for the two polarities and Polarity AB average for all stimuli along with the average noise estimates in each condition. RM-ANOVA indicated a significant three-way interaction between polarity, vowel and h1, $F(1.53, 23) = 13.71$, $p < 0.001$; $\eta^2_{partial} = 0.48$. In post-hoc analyses, statistically significant differences in EFR amplitude were found between Polarity A, B and Polarity AB average conditions for /u/ with h1. Polarity A amplitude (M = 147.29 nV; SD = 60.02) was significantly greater than Polarity B amplitude (M = 87.26 nV; SD = 45.91) by a mean difference of 60.04 nV, $t(15) = 7.99$, $p < 0.001$, 95% CI [44.03, 76.04] and significantly greater than Polarity AB average amplitude (M = 116.47 nV; SD = 51.41) by a mean difference of 30.83 nV, $t(15) = 8.33$, $p < 0.001$, 95% CI [22.94, 38.72]. Also, Polarity AB average amplitude was significantly larger than Polarity B amplitude by a mean difference of 29.21 nV, $t(15) = 7.44$, $p < 0.001$, 95% CI [20.84, 37.59]. The difference between Polarity A and Polarity AB average approached significance for /i/ with h1 ($t(15) = 2.59$, $p = 0.020$; critical p value = 0.011). EFR amplitudes did not vary significantly between polarity conditions for /a/ with h1 and all vowels without h1. Of the 16 participants who had a significant detection in at least one polarity, one participant did not have a significant detection in the Polarity AB average for the stimulus /u/ F1 without h1.

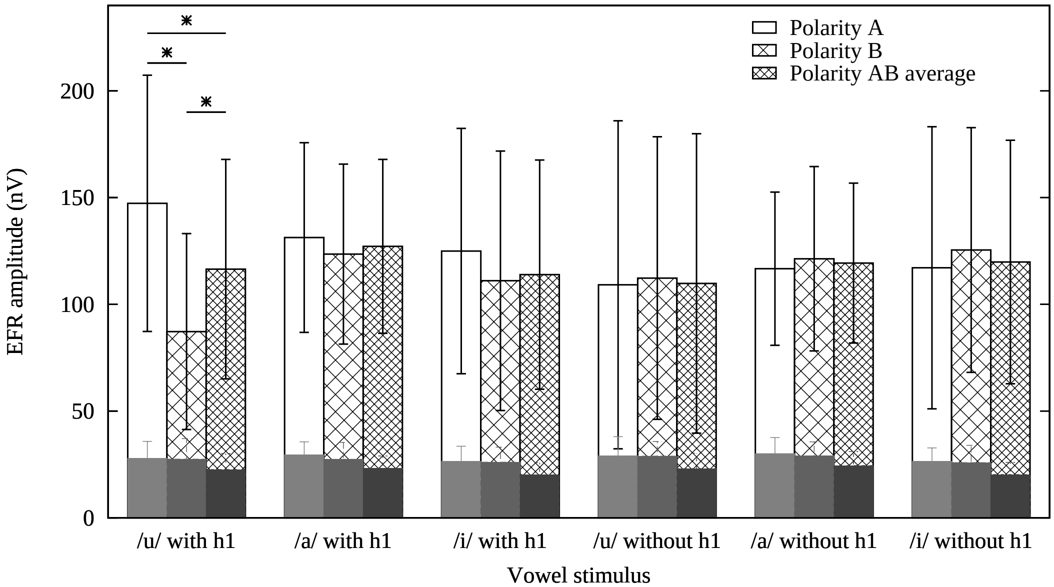

Figure 5-4: Comparison of EFR amplitudes between polarities across F1 carriers (Error bars represent ±1 *SD*; n = 16). * indicates a significant difference. Solid grey bars represent the average noise estimate in each condition.

Effect of polarity on F2+ carriers

Of the 20 participants, 14 had a significant F2+ EFR detected in at least one polarity (Bonferroni corrected). Figure 5-5 illustrates mean EFR amplitudes for the two polarities and Polarity AB average for all stimuli along with the average noise estimates in each condition. RM-ANOVA indicated a significant main effect of vowel, $F(1.41, 18.28) = 4.82$, $p = 0.031$, $\eta^2_{partial} = 0.27$. Averaged across polarity conditions and vowels with and without h1, post-hoc pairwise comparisons indicated that EFR amplitudes were significantly higher for /a/ ($M = 117.72$ nV; $SD = 38.37$) relative to /i/ ($M = 98.93$ nV; $SD = 51.06$) by a mean difference of 18.71 nV, $t(13) = 2.88$, $p = 0.013$, 95% CI [4.66, 32.76]. A main effect of polarity was not significant, $F(1.01, 13.08) = 0.65$, $p = 0.434$, $\eta^2_{partial} = 0.05$. Of the 14 participants who had a significant detection in at least one polarity, one participant did not have a significant detection in the Polarity AB average for the stimulus /u/ F2+ without h1 (this was not the same individual as for /u/ without h1, above).

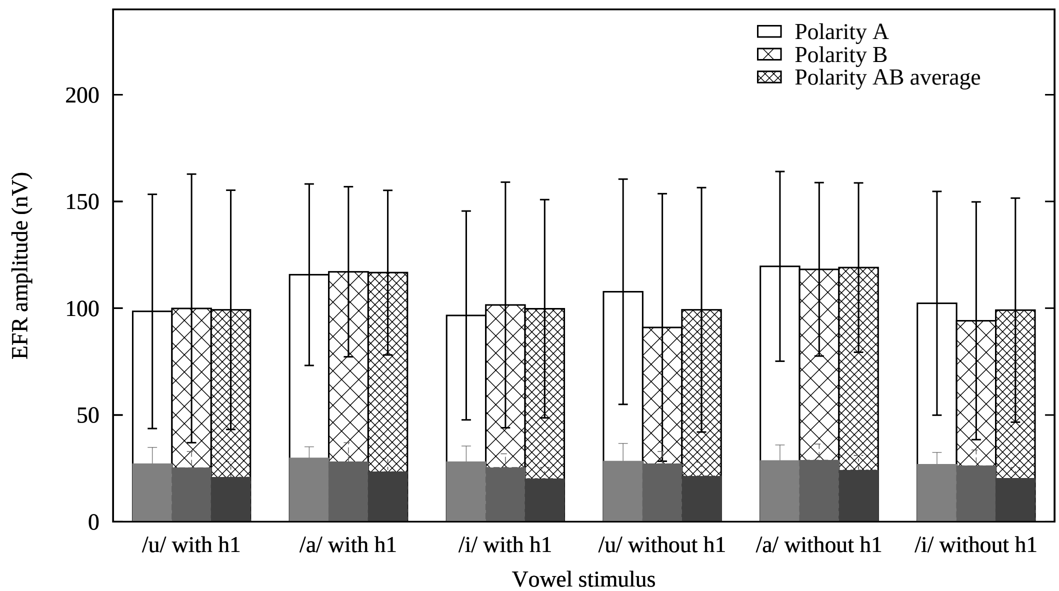

Figure 5-5: Comparison of EFR amplitudes between polarities across F2+ carriers (Error bars represent ±1 *SD*; n = 14). Solid grey bars represent the average noise estimate in each condition.

5.4 Discussion

5.4.1 Experiment I: Incidence of polarity-sensitive differences in EFR amplitude

Results of Experiment I illustrate that EFR amplitudes to vowel stimuli of opposite polarities vary across individuals. A few individuals exhibit differences over and above the expected test-retest variability and these differences are unlikely due to variations in noise. When response amplitude estimates are corrected for the influence of noise, the estimated absolute test-retest variation is 29 nV or 40% of the mean amplitude (Wilding, McKay, Baker, & Kluk, 2012). In the present experiment, the mean amplitude of Polarity A and B across 24 participants was 100.13 and 92.71 nV respectively. Using the more conservative estimate, the estimated test-retest variation in EFR amplitude would be expected to be within ∼39 nV (40% of [100.13+92.71]/2). Therefore, of the 24 participants with a significant detection in at least one polarity, 29% exhibited differences greater than that attributable to test-retest variation. As well, the polarity-sensitive differences obtained here were unexplained by variations in electrophysiological noise. Electrophysiological noise is less likely to have caused large changes in EFR amplitude across polarities partly due to the interleaved method of stimulus presentation where every sweep consisted of both the Polarity A and Polarity B stimulus. The interleaved method shares the variability in extraneous variables such as a change in participant state, electrode placement and change in electrode and inter-electrode impedances between the two test conditions being compared. The test-retest variability estimate chosen here (∼39 nV) was based on studies that evaluated variability *across* test sessions and, therefore may be higher than the variability experienced here, where the test paradigm was designed to reduce differences in extraneous variables between conditions. However, in summary, it is evident that EFRs elicited to opposite polarities of vowels vary significantly in almost 30% of individuals.

5.4.2 Experiment II: Effect of vowel, frequency of harmonics and presence of h1 on polarity sensitivity of EFRs

The aim of Experiment II was to evaluate polarity-sensitive differences in EFRs elicited by F1 (resolved harmonics) and F2+ (unresolved harmonics) bands of three different vowels, and to assess the influence of h1. Polarity-sensitive amplitude differences in EFRs were significantly greater than the test-retest variability, at least for the carrier /u/ F1 (see Appendix B). As in Experiment I, the differences due to polarity were unexplained by variations in noise estimates for both F1 and F2+ carriers, further supporting polarity sensitivity of EFRs. Results of Experiment II indicate that h1 likely contributes to polarity-sensitive differences in EFRs; a significant effect of polarity was found for F1 carriers and not F2+ carriers, and this effect disappeared when h1 was removed. This change in polarity-sensitive behavior of EFRs was specific to vowels with low frequency F1 peaks such as /u/ (statistically significant) and /i/ (approaching statistical significance) but not /a/ (see Table 5-1 for F1 frequencies).

Although statistical analysis leads to the conclusion that h1 contributes to polarity sensitivity of EFRs elicited by vowels with low F1 frequencies, this should not be interpreted as the effect of an h1 FFR alone for two reasons. First, although the vowels /u/ and /i/ have low F1 frequencies (close to h1), the stimulus magnitudes of h1 and the f_0 were highly similar across all three vowels. Hence, an FFR elicited by h1 is likely to be similar in amplitude across vowels used in this study, unless FFR amplitude is influenced by the proximity of F1, whose harmonics are considered dominant in EFR generation. In the literature, it has been common to estimate the FFR using the – – average (e.g., Aiken & Picton, 2008; Krishnan, 2002). The FFR to h1 was larger for /i/ compared to /a/ using a – – average in Aiken and Picton (2008). However, this could be explained by either a difference in level of h1 in the two vowels (visual inspection of Figures 4 and 6 in Aiken & Picton, 2008) or the residual EFR that may be present in the – – average. The residual EFR in the – – average may result from polarity-sensitive differences in EFR amplitude. An example of this is evident when Figures 1d and 3d of Aiken and Purcell (2013) are compared. Participants with larger differences in EFR amplitude due to polarity (e.g., subject 9) tend to have larger amplitude – – averages (Aiken & Purcell, 2013). Therefore, a true estimate of the FFR to h1 is difficult to obtain using the – – average when the EFR exhibits polarity-sensitive amplitude differences. The – – averages obtained at h1 are more likely to represent the FFR amplitude for /a/, because that vowel has not shown a significant effect of polarity in the present study. In the present study, for vowel /a/, only three of 20 participants had an h1 FFR significantly greater than the noise using a – – average ($M_{\text{amplitude}} = 28.29$ nV, $SD = 4.09$). In previous studies, FFR estimates at h1 for /a/ are either not reported (e.g., Krishnan, 2002) or they are not significantly different from noise estimates in neighboring bins (Aiken & Picton, 2008). Since the h1 FFR amplitude estimate in the present study is comparable to the noise estimates obtained in the present study, and since the – – average was not significantly higher than noise in the majority of participants, it is unlikely that an h1 FFR is a significant cause of the observed EFR polarity sensitivity.

The second reason suggesting that an h1 FFR may not be the only factor determining polarity sensitivity of EFRs relates to stimulus envelope characteristics. Filtering out h1 for the Experiment II stimulus not only altered its fine structure to remove the h1 FFR, but also significantly changed its envelope. To explore the contribution of an asymmetrical stimulus envelope, an envelope asymmetry index was computed using MATLAB. The envelope asymmetry index was computed for the F1 and F2+ bands separately. To evaluate the two bands individually, only-F1 and only-F2+ versions of the stimulus sequence /susaʃi/ were created. These sequences were identical in duration and structure to the original but consisted of *either* the F1 or the F2+ band. The same time windows from the beginning to the end of each vowel were used as for the analysis of the EFR (vertical grey lines in Figure 5-1B). The F1 and F2+ carriers were filtered into 21 one-third octave bands with center frequencies ranging between 79 and 8025 Hz. Each of these bands was delayed by a frequency-specific cochlear traveling wave delay estimate corresponding to its center frequency (Schoonhoven, Prijs, & Schneider, 2001) and then summed.

Further, vowels were half-wave rectified to simulate the preferential firing pattern of the auditory nerve during the rarefaction phase of the stimulus (Aiken & Picton, 2008; Lins, Picton, Picton, Champagne, & Durieux-Smith, 1995). The negative values of the stimulus waveform were retained but positive values were replaced with zeroes because oscilloscope viewings of a biphasic asymmetrical pulse confirmed that a positive deflection in the wav file was presented as condensation. The envelope of the rectified signal was then obtained using absolute values of the Hilbert transform. The peaks, approximately corresponding to each cycle of f_0, were located and an average of the peak values was calculated for each polarity of the stimulus. The absolute difference between these mean peak values obtained for the two polarities was taken as a measure of envelope asymmetry. Larger absolute differences indicated greater asymmetries of the envelope above and below the baseline. Figure 5-6A illustrates the located peaks using triangles for Polarity A and B for the F1 band of /u/ with h1. Horizontal lines represent the mean value of the located peaks for each polarity; a greater vertical distance between the two horizontal lines indicates larger envelope asymmetry. The envelope asymmetry index was normalized by dividing each of the indices by the largest index obtained, which was for /u/ F1 with h1.

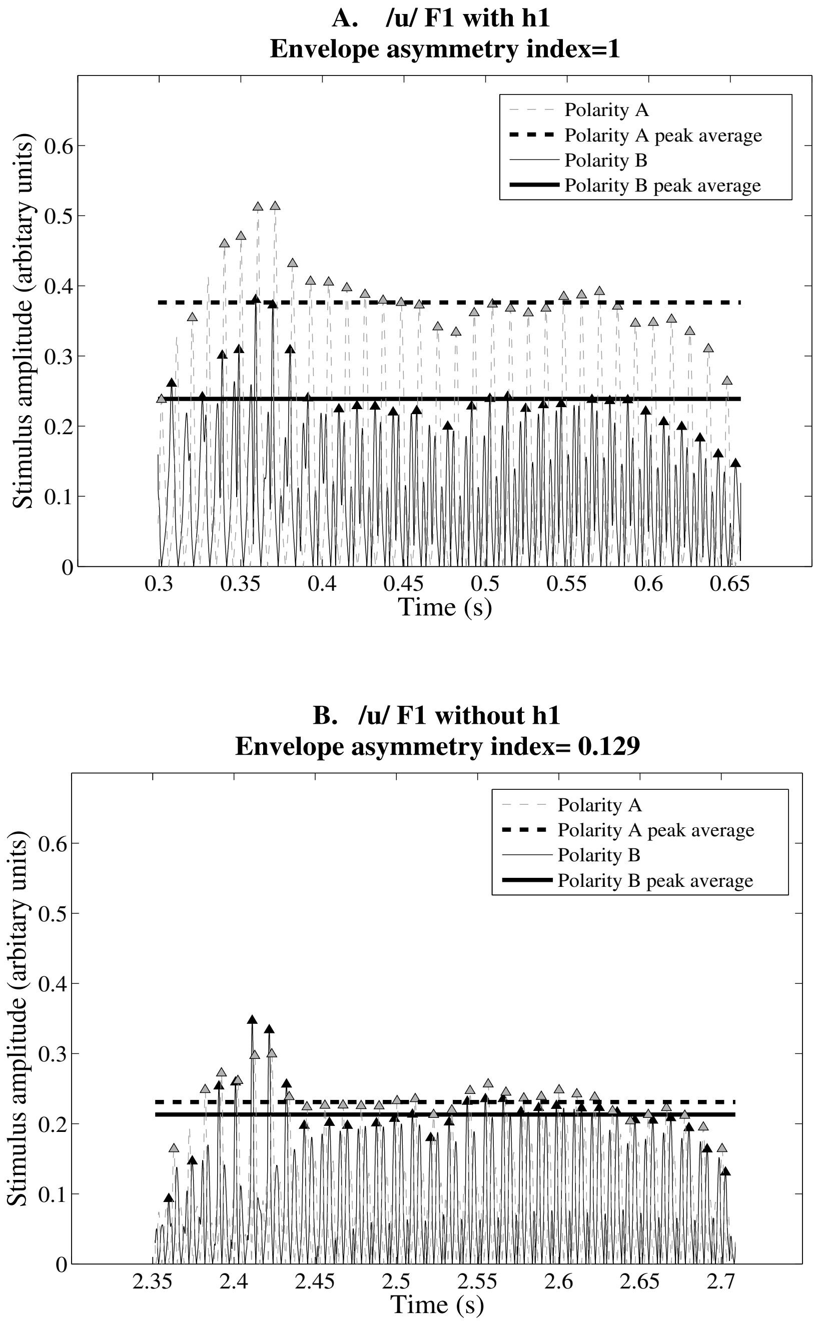

Figure 5-6: Comparison of envelope asymmetry for /u/ F1 carriers with and without h1. Triangles indicate peak values. Envelope asymmetry indices are normalized values. Time axis gives time relative to the start of a stimulus sweep.

A correlation analysis was performed between the normalized envelope asymmetry index and the mean absolute amplitude differences between polarities, averaged across all participants for each stimulus. As with the other statistical analyses in this study, participants data were included only if there was a significant detection in *either* Polarity A or B or both (Bonferroni corrected). The analysis showed a significant positive correlation between absolute polarity-sensitive differences in EFR amplitude and envelope asymmetry index ($r(10) = 0.80$, $p = 0.014$; see Figure 5-7). This suggests that differences in vowel-evoked EFRs due to stimulus polarity increase with more stimulus envelope asymmetry above and below the baseline. This is in agreement with the general consensus on the effect of stimulus intensity on response amplitudes of EFRs (Purcell & Dajani, 2008). In the case of the F1 band of /u/ with h1, the larger bottom half of the stimulus (shown fullband in Figure 5-1B), was associated with higher mean response amplitudes in Polarity A (see Figure 5-4). The bottom half of the stimulus waveform corresponds with rarefaction in Polarity A (Figure 5-6A). The F1 band of /u/ with h1 had the largest envelope asymmetry index and likely drives the correlation. However, since the differences in these stimuli were naturally occurring, all stimuli were included in the correlation analysis. Data from the present study thus confirms Greenberg's statement (1980) that envelope asymmetry leads to an effect of polarity.

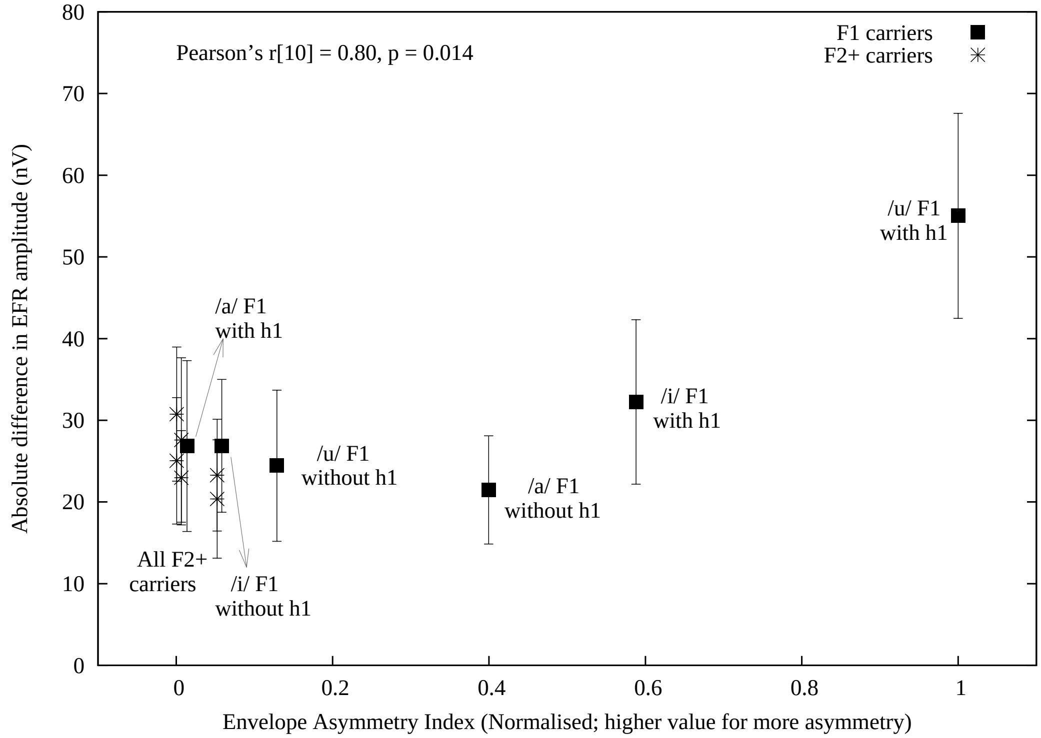

Figure 5-7: Relationship between envelope asymmetry index and absolute EFR amplitude differences due to polarity (Error bars represent 95% CI).

The current study presents converging evidence that the presence of h1 contributes to the polarity-sensitive nature of EFRs elicited by vowels. Differences in mean peak values shown in Figure 5-6 are smaller for /u/ without h1 (Figure 5-6B) relative to the original /u/, with h1 (Figure 5-6A), indicating a reduction in envelope asymmetry of the stimulus with removal of h1. This pattern parallels the results of statistical analyses where the polarity-sensitive differences in EFR amplitude were significant for the F1 band of /u/ with h1 but non-significant for the same band of /u/ without h1. The histograms in Figure 5-3A also illustrate a reduction in the frequency of large polarity-dependent amplitude differences in the case of /u/ without h1 relative to /u/ with h1. The leftward shift in the histogram peak towards zero differences suggests that the absence of h1 in /u/ reduces the frequency and degree of polarity-sensitive differences in EFR amplitude.

The present study suggests that 64.5% (R^2) of the variability in EFR amplitude due to polarity is explained by stimulus envelope asymmetry, when differences across participants are averaged for each stimulus. From Figure 5-3, we know that there is variability in the magnitude and direction of polarity-sensitive differences across individuals. Individual variability is also evident in the data reported by Aiken and Purcell (2013). Individual variability may arise from differences in the peripheral auditory system that were not explored in the current study.

5.4.3 Polarity A or B versus averaged Polarity AB

Although EFR amplitudes varied significantly in the two polarities for many individuals, detection of EFRs in the Polarity AB average was generally not affected in Experiment I and II. In all of these individuals, the average AB amplitude was between amplitude estimates for Polarity A and B, and hence EFRs varied minimally in response phase between polarities. Larger differences in response phase may have led to partial or complete cancellation in the Polarity AB average. This pattern of results is in agreement with previous studies evaluating the polarity sensitivity of EFRs (Aiken & Purcell, 2013; Small & Stapells, 2005). Based on these results, one may infer that averaging responses to opposite polarities is generally not detrimental to EFR detection, although it may under-estimate the maximum response amplitude possible with the best polarity. Averaging EFRs to opposite polarities may have less impact on detection of suprathreshold stimuli where the EFR amplitudes are well above the noise floor. In the case of smaller amplitude EFRs (for example, at lower sensation levels), averaging of two polarities could lead to a false negative, unless we are able to average more sweeps to lower noise levels. At suprathreshold levels, two out of 24 participants in Experiment I and two of 20 participants in Experiment II had a non-significant EFR in the Polarity AB average although they had a significant detection in at least one polarity. In these individuals, using the Polarity AB average resulted in a false negative due to possible polarity-sensitive differences in the vowel-evoked EFRs.

Averaging responses to opposite polarities is advantageous in reducing noise and stimulus artifacts. Noise estimates in the Polarity AB average are expected to be lower by a factor of $1/\sqrt{2}$ relative to noise estimates obtained in a single polarity due to containing double the number of sweeps (evident in Figures 5-4 and 5-5), and hence this is generally favorable for response detection. For example in Experiment II, in four of 120 instances (20 participants x six carriers), three participants had a significant detection in the Polarity AB average but did not have a significant detection in either Polarity A or B for the F1 carriers. Likewise, for the F2+ carriers, four participants had a significant detection in the Polarity AB average alone. As well, recordings are less susceptible to stimulus artifacts when opposite polarity stimuli are averaged (Aiken & Picton, 2008; Akhoun et al., 2008; Small & Stapells, 2004). However, since polarity of the stimulus does not affect perception (Sakaguchi, Arai, & Murahara, 2000), detection of EFRs in either stimulus polarity may indicate encoding of the vowel in the auditory system and this information may suffice for several applications. Therefore, using an *either* polarity detection rule is likely to be favorable in scenarios where the degree of under-estimation of response amplitude may affect response detection (for example, at lower sensation levels). Since this phenomenon varies on an individual basis, using an *either* polarity detection rule may reduce false negative outcomes. Using this detection rule, however, requires consideration of a correction for multiple comparison bias as well as careful checks to avoid stimulus artifacts.

A response at f_0 elicited by a single polarity reflects a combination of EFR, an h1 FFR and cochlear microphonic (Aiken & Picton, 2008). However, consistent with findings in the literature (Aiken & Picton, 2006), the removal of h1 had no significant effect on the EFR amplitudes (averaged across polarities and vowels, a main effect of the first harmonic was non-significant). This suggests that the contribution of FFR or cochlear microphonic elicited by h1, which are at the same frequency as f_0, was minimal. The measured EFR at f_0 is therefore derived dominantly from the envelope rather than phase-locking to the stimulus fine structure at the f_0 frequency (Aiken & Picton, 2006).

5.4.4 Effect of vowel on EFRs elicited by F2+ carriers

The main effect of vowel was significant only for F2+ carriers. The EFR amplitudes elicited by the F2+ carrier of /a/ were significantly higher than /i/. A stimulus-related factor that may contribute to higher amplitude EFRs for /a/ relative to /i/ is the level of the formant carriers. Measured in an ear simulator, the F2+ carrier of /a/ was almost 9 dB higher in RMS level relative to the F2+ carrier of /i/. An increase in stimulus level results in spread of activity to a wider area on the basilar membrane, thereby activating more neurons and leading to a higher amplitude response (Purcell & Dajani, 2008). A difference in formant levels between the vowels was expected as the F2 peak frequency is lower for /a/ compared to /i/ (see Table 5-1) and since the level of harmonics drops at -6 dB/octave (glottal source spectrum and radiation effect p. 114, Raphael, Borden, and Harris 2007).

5.5 Summary and Conclusion

Experiment I shows that polarity-sensitive differences in EFR amplitude occur in a significant proportion of individuals. For the naturally spoken vowel /ε/ used in Experiment I, almost 30% of participants presented a polarity-sensitive amplitude difference of over ∼39 nV. Results of Experiment II show that polarity-sensitive differences in EFR amplitude vary across vowels depending on the F1 peak frequency and presence of h1. EFRs elicited by carriers with low frequency resolved harmonics exhibit greater polarity-sensitive differences compared to carriers with higher frequency unresolved harmonics. Vowels with lower F1 frequencies tended to show greater polarity-sensitive differences in EFR amplitude, specifically when h1 was present. Although removal of h1 from the stimulus reduced the polarity sensitivity of the EFR, this may not be attributed solely to the contribution of an h1 FFR because removal of h1 also altered the envelope asymmetry above and below the baseline. The positive correlation between degree of polarity-sensitive amplitude differences and stimulus envelope asymmetry suggests that envelope asymmetry is a factor in the polarity sensitivity of EFRs. Envelope asymmetry explains only a part of the variability observed and there may be other factors contributing to individual variability. Since averaging EFRs to opposite polarities may reduce response amplitude and hence affect detection in some individuals, it may prove beneficial to use an '*either*' polarity detection rule with necessary precautions as discussed. Future studies that aim to isolate the contribution of an h1 FFR to the polarity sensitivity of EFRs could use stimuli that have h1 shifted in frequency by a small amount such that the envelope of the stimulus is not significantly altered.

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Chapter 6

Effect of stimulus level and bandwidth on speech-evoked envelope following responses in adults with normal hearing

6.1 Introduction

Outcome evaluation is an essential component of the hearing aid fitting process (American Academy of Audiology, 2013; Bagatto, Scollie, Hyde, & Seewald, 2010; Joint Commitee on Infant Hearing, 2007). The implementation of universal newborn hearing screening programs has significantly lowered the age of hearing loss diagnosis and thereby, of hearing aid fittings (Sininger et al., 2009). Outcome evaluation using conventional behavioral tests presents a challenge in infants diagnosed early and fitted with hearing aids (under six months of age), because reliable behavioral responses are difficult to obtain. This concern has led to an increased interest in the use of objective outcome measures. Objective outcome measures refer to the assessment of hearing acuity, with or without hearing aids, obtained directly from the infant, while requiring minimal co-operation from the infant and parent. Suggested objective measures include transient auditory evoked potentials that represent synchronous activity in response to stimulus onsets or offsets such as the cortical auditory evoked potential (CAEP; Golding et al., 2007; Purdy et al., 2005). They also include steady-state responses elicited by an ongoing periodically changing stimulus (Picton, 2011). Steady-state potentials include measures such as the envelope following response (EFR; Choi, Purcell, Coyne, & Aiken, 2013) or the frequency following response (FFR; Anderson & Kraus, 2013). An example of a combination of transient and steady-state potentials is the speech-evoked auditory brainstem response (Anderson & Kraus, 2013; Dajani, Heffernan, & Giguere, 2013). The current study proposes an objective test paradigm based on EFRs elicited by speech sounds.

Of these several proposed measures, the EFR has several advantages for use as an aided outcome measure. The EFR, which is a response phase-locked to the stimulus envelope, is generated at the voice's fundamental frequency (f_0) when elicited by a vowel (e.g., Aiken & Picton, 2006, 2008; Choi et al., 2013). The primary advantage of using EFRs as an outcome measure for hearing aid fittings is that the test stimulus can be running speech (Choi et al., 2013). Speech stimuli, or stimuli with temporal characteristics similar to running speech, are more likely to accurately elicit non-linear hearing aid function during the test, compared to pure tones and complex sounds that are spectrally and temporally different from speech (Henning & Bentler, 2005; Scollie & Seewald, 2002; Stelmachowicz, Kopun, Mace, & Lewis, 1996; Stelmachowicz, Lewis, Seewald, & Hawkins, 1990). Moreover, hearing aid function for speech stimuli in isolated contexts (e.g., CAEP test protocol with long inter-stimulus intervals), may not be similar to running speech contexts (Easwar, Purcell, & Scollie, 2012). Since one of the main goals of outcome evaluation is confirming audibility of speech sounds during conversational speech, representation of accurate hearing aid function during the test is vital to its validity. A second potential advantage of using continuous stimuli is the possible gain in test time efficiency or information-to-time ratio. Use of test paradigms that do not require inter-stimulus intervals and that have reasonably quick response detection times are likely to favor clinical feasibility (Aiken & Picton, 2006; Choi et al., 2013). The third advantage includes the use of statistical response detection methods that reduce tester bias, and increase objectivity of the test (Aiken & Picton, 2006; Choi et al., 2013; Picton, John, Dimitrijevic, & Purcell, 2003). The fourth advantage relates to the putative generator sites. Since EFRs to higher modulation rates in the range of average male f_0 are generated from the upper brainstem (Herdman et al., 2002; Purcell, John, Schneider, & Picton, 2004), they are largely exogenous or stimulus-driven responses and hence are less affected by attention or arousal state of the individual undergoing the test (Cohen, Rickards, & Clark, 1991; Purcell et al., 2004). Insensitivity to arousal state is also likely to be of clinical significance as infants could be evaluated in their natural sleep.

However, current approaches to elicit EFRs may be improved for use as an aided outcome measure. Although EFRs elicited by a vowel can be initiated by any two consecutive interacting harmonics within the cochlea (Aiken & Picton, 2006), recent studies demonstrate dominance of the first formant (F1) region (Choi et al., 2013; Laroche, Dajani, Prevost, & Marcoux, 2013). The dominance of F1 in EFRs elicited by a vowel likely limits inference of vowel representation to lower frequencies. Since the primary goal of an aided outcome measure is to assess audibility of speech sounds through a hearing aid, representation of the bandwidth of speech, and hence assessment of high frequency audibility, becomes an important factor for several reasons. One, speech consists of phonemes that span a wide frequency range (e.g., Boothroyd, Erickson, & Medwetsky, 1994). Two, several frequently occurring information-bearing segments, particularly consonants, have spectral peaks in the high frequency regions (e.g., /s/, Tobias, 1959). Audibility up to about 8 to 9 kHz is important for recognition of fricatives spoken by females and children (Stelmachowicz, Pittman, Hoover, & Lewis, 2001). Three, wider bandwidth leads to better word recognition (Gustafson and Pittman 2011), novel word learning (Pittman, 2008), and perceived sound quality (Ricketts, Dittberner, & Johnson, 2008). Four, limited hearing aid bandwidth has been identified as a significant factor affecting a child's ability to access and benefit from the aided speech signal (for a review, see Stelmachowicz, Pittman, Hoover, Lewis, & Moeller, 2004). Limited high frequency bandwidth in hearing aids is suspected to be the primary cause for the delayed acquisition of fricatives relative to low frequency sounds (Stelmachowicz et al., 2004). In summary, EFRs to vowel stimuli alone may not be informative about audibility across the entire bandwidth of speech, and therefore warrant modifications to be a more suitable aided outcome measure.

In addition to representing the bandwidth of speech, frequency specificity is also an important factor in stimulus selection. Since hearing loss varies in configuration and severity, especially in children (Pittman & Stelmachowicz, 2003), it is likely advantageous to use stimuli that represent a wide frequency range, and that demonstrate frequency specificity to adequately reflect the impact of hearing loss and amplification in different spectral regions. As well, since all commercially available hearing aids provide adjustable frequency-dependent gain (Dillon, 2012), frequency-specific information is likely to provide more specific guidance in re-evaluating hearing and/or hearing aid fittings.

This paper proposes an EFR test paradigm specifically adapted for use as a hearing aid outcome measure based on factors discussed above. The Ling 6 sounds (/m/, /u/, /a/, /i/, /ʃ/ and /s/), commonly used for aided behavioral detection tasks (Ling, 1989; Scollie et al., 2012), provide a simplified representation of a broad range of frequencies present in speech. The low to mid frequencies are represented by the vowels and the nasal /m/, while the high frequencies are represented by the fricatives. The proposed EFR test paradigm incorporates the vowels and fricatives of the Ling 6 sounds with specific modifications applied to improve the range of frequencies represented, frequency specificity, and test time efficiency while maintaining resemblance to running speech in temporal characteristics.

The utility of an outcome measure is determined by its sensitivity to change, or its responsiveness to treatment/intervention (Andresen, 2000). In the case of hearing aids or amplification aimed to improve audibility, evaluating the sensitivity to change in audibility is critical in determining the utility of a proposed outcome measure. In Experiment I, we investigated if the proposed EFR test paradigm can reliably represent changes in audibility due to stimulus level in a group of individuals with normal hearing. We hypothesized that an increase in stimulus level will result in higher amplitude EFRs, and an increased number of EFRs detected. In Experiment II, we investigated if the proposed EFR test paradigm can reliably represent changes in stimulus bandwidth. A change in bandwidth, achieved by low-pass filtering of stimuli, affects audibility of specific spectral regions. We hypothesized that an increase in the bandwidth of a stimulus carrier will result in an increase in EFR amplitude, and the number of EFRs detected.

Auditory evoked potentials in general bear no causal relationship with hearing, and the detection of an evoked potential is influenced by factors unrelated to the stimulus and hearing (e.g., myogenic noise; Elberling & Don, 2007). However, the detection of an evoked potential does confirm neural representation of one or more stimulus attributes, in at least part of the auditory pathway involved in perception (Elberling & Don, 2007; Hyde, 1997). In Experiment II, we also evaluated the relationship between neural representation of stimulus features, as inferred by the number of EFR detections and response amplitude of EFRs, and performance in behavioral measures such as speech discrimination and sound quality rating in multiple bandwidth conditions. Speech discrimination and sound quality rating are both psychophysical measures of hearing aid outcome (e.g., Jenstad et al., 2007; Parsa, Scollie, Glista, & Seelisch, 2013). Comparison of objective EFR parameters with these behavioral measures will test convergent validity of the EFR paradigm in demonstrating sensitivity to audible bandwidth. Convergent validity is a measure of the degree to which two measures, that are thought to be representing similar attributes, show similar results (Finch, Brooks, Stratford, & Mayo, 2002). Both behavioral measures vary systematically, albeit differently, as a function of stimulus bandwidth. Improvements in speech discrimination due to increasing stimulus bandwidth asymptotes at a lower frequency than sound quality rating (Cheesman & Jamieson, 1996; Füllgrabe, Baer, Stone, & Moore, 2010; Pittman, 2010; Ricketts et al., 2008; Studebaker & Sherbecoe, 1991; Studebaker, Sherbecoe, & Gilmore, 1993). The rate of growth in speech discrimination above 4 kHz is less than the rate of growth in sound quality rating in adults with normal hearing. An improvement in sound quality rating when stimulus bandwidth is increased beyond 4 kHz has been consistently demonstrated in individuals with normal hearing (Füllgrabe et al., 2010; Moore & Tan, 2003; Ricketts et al., 2008; Voran, 1997).

In summary, the present study aimed to evaluate the proposed speech-evoked EFR paradigm in adults with normal hearing. We chose adults with normal hearing for our preliminary evaluation to characterize the nature of responses in the absence of hearing loss. Experiment I evaluated the effect of stimulus level on EFR amplitudes and the number of EFR detections. Experiment II evaluated the effect of stimulus bandwidth on EFR amplitudes and the number of detections. Additionally, the experiment evaluated the relationship between changes in objective EFR measures and behavioral measures related to bandwidth.

6.2 Experiment I: Effect of stimulus level on speech-evoked EFRs

6.2.1 Method

Participants

The study included 20 adults (16 females, four males) with normal hearing whose ages ranged between 20 and 29 years ($M_{\text{age}} = 23.4$ years, $SD_{\text{age}} = 2.34$). Participants were mostly students from Western University, London, Ontario. To be eligible, participants had to pass a hearing screen at 20 dB HL, tested using insert earphones at octaves and inter-octaves between 250 Hz and 8000 Hz in both ears. Audiometric screening was carried out using a GSI-61 audiometer. Routine otoscopy completed on all participants ruled out any contraindications such as occluding wax, discharge or foreign bodies in the ear. Tympanograms obtained in the test ear using a Madsen Otoflex 100 immittance meter indicated static compliance and tympanometric peak pressure within normal limits. Volunteers with any self-disclosed neurological disorders were not included in the study. All participants reported English as their first language, and provided informed consent. The study protocol was approved by the Health Sciences Research Ethics Board, Western University, Canada. Participants were compensated for their time.

EFR stimulus

The stimulus included vowels and fricatives to represent a wide range of frequencies. The vowels /u/ (as in 'school'), /a/ (as in 'pot') , and /i/ (as in 'see') were chosen to represent a range of formant one (F1) and formant two (F2) frequencies. These vowels were produced by a 42-year old male talker from Southwestern Ontario whose first language was English. The original token produced by the male talker was /usaʃi/. One of many repetitions was chosen for further processing using softwares Praat (Boersma, 2001), Goldwave (version 5.58; GoldWave Inc., St. John's, NL, Canada) and MATLAB (version 7.11.0 [R2010b]; MathWorks, Natick, MA, USA). The best version was chosen based on phoneme duration in the case of vowels and fricatives, and in addition, flatness of f_0 contour in the case of vowels. Voice recordings were made with a studio-grade microphone (AKG Type C 4000B) in a sound booth using Spectraplus software (version 5.0.26.0; Pioneer Hill Software LLC, Poulsbo, WA, USA). These tokens were recorded at a sampling rate of 44100 Hz, and later down-sampled to 32000 Hz using Praat. During the EFR recording, since the stimulus was presented repeatedly with no inter-stimulus interval, the naturally produced sequence was edited such that the fricative /s/ also preceded /u/. The final modified stimulus sequence was /susaʃi/. This modification was made to maintain a consonant-vowel context for all vowel stimuli, and to avoid abrupt transitions between two repetitions of the stimulus sequence. The duration of a single polarity token was 2.05 seconds.

Stimulus phonemes were individually edited using Praat and MATLAB. Prior to editing, the wav file was high-pass filtered at 50 Hz to reduce low frequency noise. Boundaries of phonemes were marked based on spectrograms and listening. Fricatives /ʃ/ and /s/ were high-pass filtered at 2 and 3 kHz respectively using Praat to improve frequency specificity of these carriers. The cut-off frequencies were chosen based on the lower edge of the spectral peak in the naturally produced versions. The filtered fricatives were then 100% amplitude modulated at 93.02 Hz using MATLAB and matched in Root-Mean-Square (RMS) level before and after modulation. The durations of /ʃ/ and /s/ were 234 and 274 ms, respectively. The durations of analysis windows per iteration of /ʃ/ and /s/ were 215 and 258 ms, respectively. There were an integer number of cycles of the modulation frequency within the analysis windows of both fricatives.

The vowel spectra were modified to elicit individual responses from the first formant (F1) and the higher formants (F2+) region. To enable recording of two simultaneous EFRs from the F1 and F2+ regions, vowel spectra were edited such that the original f_0 was maintained in the F2+ region, and a lower f_0 (original f_0–8Hz) was created in the F1 region. With these dual-f_0 vowels, EFRs from the F2+ region were elicited at the original f_0 frequency, and EFRs from the F1 region were elicited at original f_0–8 Hz. In principle, this paradigm is similar to multiple Auditory Steady-State Responses (ASSR), in which the simultaneously presented Amplitude-Modulated (AM) tones are each modulated at unique modulation frequencies (Picton et al., 2003). To create the f_0–8 Hz stimulus, the original vowels were lowered in f_0 and then low-pass filtered to isolate the F1 frequency band. The lowering of f_0 was accomplished using the 'Shift frequencies' function in Praat. This function shifts the f_0 by the specified amount (in Hz) while approximately maintaining the formant peak frequencies. The original vowels were also high-pass filtered to isolate the F2+ frequency band and combined with the f_0-shifted F1 band. The level of the dual-f_0 vowel was matched with that of the originally produced vowel. The dual-f_0 structure of these processed vowels increases frequency specificity of broadband stimuli like vowels while maintaining the bandwidth of naturally occurring vowels. In addition, the use of dual-f_0 vowels is likely to increase test time efficiency by increasing the information gained in a given test time. The cut-off frequencies of the F1 and F2+ bands were chosen to be close to the mid-point between the estimated F1 and F2 peak frequencies. The duration, analysis time window, formant frequencies, mean f_0 in each band, cut-off frequencies and harmonics included in each band for the three vowels are provided in Table 6-1. Processed phonemes were concatenated using Goldwave in the same sequence as the original production with an additional /s/ preceding /u/.

Vowel	Duration (ms)	Analysis time window (ms)	Band	F1 or F2 frequency (Hz)	Mean f_0 (Hz)	Cut-off frequency (Hz)	Harmonics
/u/	386	357	F1	266	92	615	h1 – h6
			F2+	1016	100	615	h7+
/a/	447	384	F1	684	88	1050	h1 – h11
			F2+	1467	96	1080	h12+
/i/	435	356	F1	277	89	1150	h1 – h12
			F2+	2160	97	1170	h13+

Table 6-1: Characteristics of dual-f_0 vowels

The processed stimulus sequence consisted of eight EFR carriers; two fricative bands and two formant carriers from each of the three vowels. Spectra of each EFR carrier are presented in Figure 6-1. To be spectrally consistent with the International Speech Test Signal (ISTS; Holube, Fredelake, Vlaming, & Kollmeier, 2010), which is a standard hearing aid test signal, the /ʃ/ was attenuated between 2650 and 3650 Hz by 3 dB to fall within the 99th percentile of the ISTS dynamic range at matched RMS levels. Each sweep consisted of the stimulus sequence presented in both polarities. The entire duration of a sweep was 4.1045 seconds.

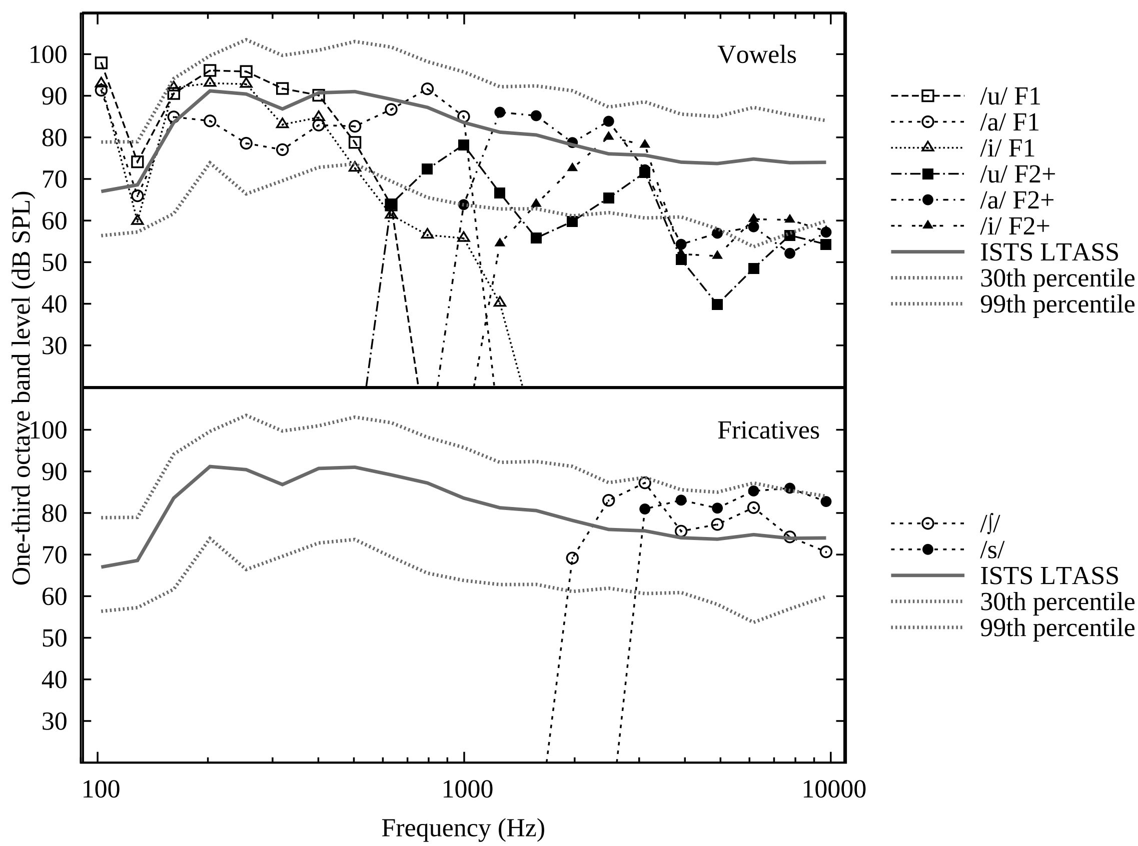

Figure 6-1: One-third octave band spectra of EFR carriers. ISTS = International speech test signal, LTASS = Long-term average speech spectrum

Each EFR elicited using the dual-f_0 vowels could be influenced by the presence of the other carrier. This has been investigated often for multiple ASSRs, in which four carriers are presented simultaneously (e.g., John, Lins, Boucher, & Picton, 1998; John, Purcell, Dimitrijevic, & Picton, 2002). When carrier frequencies within an octave apart are presented simultaneously, the presence of a low frequency carrier can enhance responses elicited by the high frequency carrier whereas the presence of a higher frequency carrier can attenuate responses elicited by a low frequency carrier (John et al., 1998). To evaluate such interaction effects, a pilot study comparing EFRs elicited by F1 and F2+ carriers presented simultaneously (using dual-f_0 vowels) and individually, was completed in 10 adults with normal hearing. Results indicated that EFRs elicited by simultaneously presented F1 and F2+ carriers were similar in amplitude to EFRs elicited by individually presented F1 and F2+ carriers (see Appendix D for details).

Stimulus presentation and response recording

Stimulus presentation was controlled by software developed using LabVIEW (version 8.5; National Instruments, Austin, TX, USA). Digital to analogue conversion of the stimulus, and analogue to digital conversion of the electroencephalogram (EEG) were carried out using a National Instruments PCI-6289 M-series acquisition card. The stimulus was presented at 32000 samples per second with 16-bit resolution and the responses were recorded at 8000 samples per second with 18-bit resolution. The stimulus was presented using an Etymotic ER-2 insert earphone coupled to a foam tip of appropriate size. Test ear was randomized.

EEG was recorded using three disposable Medi-Trace Ag/AgCl electrodes. The non-inverting electrode was placed at the vertex, the inverting electrode was placed at the posterior mid-line of the neck, just below the hairline, and the ground was placed on one of the collarbones. Electrode impedances, measured using an F-EZM5 Grass impedance meter at 30 Hz, was maintained below 5 kΩ, with inter-electrode differences under 2 kΩ. A Grass LP511 EEG amplifier band-pass filtered the input EEG between 3 and 3000 Hz, and applied a gain of 50000. An additional gain of two was applied by the PCI-6289 card, making the total gain 100000.

Test conditions

The stimulus was presented at overall levels of 50 and 65 dB SPL for 300 sweeps. The total recording time for one condition was 20.5 minutes. The stimulus level was calibrated in an ear simulator (Type 4157) using a Brüel and Kjær Type 2250 sound level meter. The level was measured in flat weighted L_{eq} while the stimulus was presented for 30 seconds. Participants were seated in a reclined chair in an electromagnetically shielded sound booth. A rolled towel was placed under their neck to help reduce neck tension and a blanket was provided for comfort. The lights were switched off and participants were encouraged to sleep during the recording.

EFR analysis and detection

Response analysis was carried out offline using MATLAB. Each sweep was divided into four epochs of ∼1.5 s each and a noise metric was calculated for each epoch. The noise metric was the average EEG amplitude in each epoch between ∼80 to 120 Hz. Any epoch that exceeded the mean noise metric plus two standard deviations was rejected prior to averaging. EFRs were analyzed between pre-selected boundaries (i.e., analysis time window) that were chosen such that the ramp-in and ramp-out sections at the beginning and the end of the phoneme were excluded.

EFRs from the 300-sweep average EEG were analyzed in the +− average (Aiken & Picton, 2008) using a Fourier analyzer (FA; Choi et al., 2013). The +− average refers to vector averaging of responses elicited by opposite stimulus polarities (Aiken & Picton, 2008). For the FA, tracks of f_0 showing changes in f_0 were obtained for the analysis window of each vowel (Choi et al., 2013). Reference cosine and sine sinusoids were generated using the instantaneous f_0 frequencies in the f_0 track. An estimate of brainstem processing delay of 10 ms was used to correct for response delay (Aiken & Picton, 2008; Choi et al., 2013). The delay-corrected EEG was multiplied with the reference sinusoids to create real and imaginary components of the EFR. The two components were further averaged to a single complex number for each response segment. Each complex number provided an estimate of EFR amplitude and phase.

To determine if EFRs elicited by vowel carriers were detected, the background EEG noise was estimated using 14 frequency tracks surrounding the response tracks at f_0 and f_0–8 Hz. Due to simultaneous presentation of F1 and F2+ carriers at f_0 frequencies 8 Hz apart, frequency tracks for noise estimates had to be chosen such that the response track of one band was not included as one of the noise tracks for the other. For both the F1 and F2+ response, there were six and eight noise tracks, below and above the response f_0 respectively. Noise estimates from the 14 noise tracks were averaged to provide a single value to be used for an F-test. For an EFR to be detected, the ratio of the EFR amplitude at the response frequency to the average noise estimate had to exceed the critical F ratio (2, 28 degrees of freedom) of 1.82 at an α of 0.05.

EFRs to fricatives were estimated using a discrete Fourier transform (DFT). Again, responses to fricatives of opposite polarities were averaged in the time domain. Additionally, responses to the two iterations of /s/ were averaged. Noise estimates were obtained from three noise bins for /ʃ/, and four noise bins for /s/, on each side of the bin containing the modulation frequency (93.02 Hz). The number of noise bins varied for the two fricatives because of varying durations. The difference in duration leads to differences in DFT bandwidths and therefore, the number of bins that could be accommodated within ∼15 Hz on each side of the modulation frequency (15 Hz was deemed sufficiently close to the response frequency). Consequently, the critical F ratio for an EFR to be detected was 1.97 and 1.91 for /ʃ/ and /s/ respectively. All EFR amplitude estimates used for the F-test were corrected for possible over-estimation of response amplitude due to noise (see Appendix of Picton, Dimitrijevic, Perez-Abalo, & Van Roon, 2005).

A stimulus artifact check was completed using a no-stimulus-to-the-ear run in 10 participants. The placement of the transducer and recording electrode leads during the no-stimulus run were similar to the main experiment. The stimulus was routed using a foam tip to a Zwislocki coupler placed on each participant's chest. Stimulus presentation level was 65 dB SPL, and recording time per participant was 20.5 minutes. The false positive rate (rate of significant detections) was 3.75%, which is close to the assumed α of 5% during response analysis.

Estimation of sensation level (SL)

Although the EFR stimulus sequence was presented at 50 and 65 dB SPL, the relative levels of each carrier varied and we therefore quantified sensation level (SL) for each carrier.

Behavioral threshold

To estimate SL of each EFR carrier during EFR recordings, the threshold of each carrier was obtained using an ER-2 insert earphone. Individual carriers were concatenated without gaps to form one-minute long tracks for each one. The fricatives used for behavioral detection were high-pass filtered like in the EFR stimulus, but not amplitude modulated. The f_0-shifted F1 bands and F2+ bands of the three vowels were presented individually. Stimulus presentation was controlled using the Child Amplification Laboratory experimental system. The stimuli were routed through an audiometer. The audiometer dial was used to seek thresholds using a 5-dB step size. Threshold was defined as the level at which two positive responses of three presentations in an ascending run was obtained.

Estimation of stimulus presentation level

The EFR stimulus sequence was presented in a looped manner through the ER-2 at a known audiometric dial level without gaps, as in the main experiments. The output of the ER-2 was recorded for 30 seconds using the same set up used for calibration, at a sampling rate of 48000 Hz and exported in the form of a wav file.

The wav file was down-sampled to 44100 Hz in Praat and individual EFR carriers were extracted. The vowels were filtered according to the cut-off frequencies specified in Table 6-1 to obtain F1 and F2+ bands. The RMS level of each carrier was obtained using Spectraplus software. The software was calibrated using a calibration tone of known level recorded by the sound level meter. Since the ER-2 is linear, the presentation level of each EFR carrier was extrapolated to test levels of 50 and 65 dB SPL. The level of /s/ was the average of the two iterations of /s/, which were very similar. The RMS level of each carrier at threshold was computed based on the audiometric dial reading at threshold. The SL was computed as the difference in RMS level at presentation level and at threshold.

Response detection time

Since responses in the main experiment were analyzed using a fixed number of sweeps, we also evaluated the times required to obtain detections in each participant. Responses were analyzed in 50-sweep increments starting with 50 sweeps. To counteract inflation of α error due to multiple comparison bias, p values for determination of response detection were corrected using an adjusted Bonferroni correction, that considers the correlation between consecutive sweeps (used in the F-test) in a running average (Choi, Purcell, & John, 2011). The adjusted critical p values for detection were 0.05 at 50 sweeps, 0.042 at 100 sweeps, 0.038 at 150 sweeps, 0.036 at 200 sweeps, 0.034 at 250 sweeps, and 0.032 at 300 sweeps. For a given carrier, the first detection at intervals of 50 sweeps was recorded as the detection time for a given carrier.

Detection time for each carrier was further differentiated in two ways. In the first method, response detection time for a carrier was based on sweep duration of the token /susaʃi/. Detection time was computed by multiplying the response detection time (in sweeps) by the sweep duration (= 4.1045 s). This estimate reflects the detection time for a carrier when used in the /susaʃi/ stimulus context and therefore testing time with the current stimulus. However, this estimate does not account for the differences in carrier durations. In the second method (henceforth referred to as 'carrier recording time'), response detection time for a carrier was based on the carrier time within each sweep. In this method, detection time for each carrier was computed by multiplying the detection time for a carrier (in sweeps) by the duration of their respective analysis windows including iterations. This estimate reflects the stimulus duration required to obtain a significant detection for each carrier.

Statistical analysis

Statistical analyses in this study were completed using SPSS (version 21; IBM, Armonk, NY, USA). Results of the main analyses were interpreted at an α of 0.05. In Experiments I and II, we measured the effect of level and bandwidth on the number of EFRs detected (maximum of eight) and EFR amplitude. To compare the number of significant EFR detections at 50 and 65 dB SPL, a paired t-test was completed. To compare the effect of level on response amplitudes for all eight carriers, a 2-way Repeated-Measures Analaysis of Variance (RM-ANOVA) was completed with carrier (/u/ F1, /a/ F1, /i/ F1, /u/ F2+, /a/ F2+, /i/ F2+, /ʃ/ and /s/) and level (50 and 65 dB SPL) as the two within-subject factors. The dependent variable was response amplitude estimated by the FA or DFT, irrespective of a significant detection. Since the test protocols in Experiment I and II varied audibility of the EFR carriers, there were several carriers with non-significant EFR detections due to experimental manipulation. Use of response amplitude from significant detections alone would have resulted in a large reduction in the study sample size due to exclusion of individuals with even one non-significant EFR. Use of response amplitudes, irrespective of a significant detection, was considered acceptable as these may contain small amplitude EFRs (albeit, not independent of the noise), and hence assessment of growth of response amplitude due to experimental manipulation would remain valid. Greenhouse-Geisser corrected degrees of freedom were used for interpretation of RM-ANOVA. For post-hoc analyses, paired t-tests were completed and results were interpreted using False Discovery Rate (FDR) method for multiple comparisons (Benjamini & Hochberg, 1995).

Data exclusion

Since we used response amplitudes estimated by the FA/DFT irrespective of a significant detection, a noise criterion was applied to exclude those participants who exhibited high residual noise estimates. Non-significant response amplitudes are close to, and not independent of the noise floor. Therefore differences in noise estimates across conditions under comparison could be erroneously interpreted as changes in amplitude. To minimize the effect of variable noise estimates between conditions being compared, participants with high noise estimates in >25% of conditions were excluded. A test condition was considered ‘high noise’ if the residual noise estimated by the FA/DFT was greater than the mean plus two standard deviations of residual noise estimates computed across all 20 participants. Computed in this manner, data from one participant was excluded in Experiment II.

6.2.2 Results

Effect of SL on response amplitude

Figure 6-2 illustrates the relationship between estimated SL of EFR carriers and response amplitudes of significantly detected EFRs from 20 participants. For all eight carriers, the data show a positive relationship between estimated SL and response amplitudes. The SLs vary across carriers as the test levels were calibrated using the entire stimulus sequence /susaʃi/, and the level of each carrier in the sequence varied in the natural production. As well, hearing thresholds varied across carriers of different frequencies.

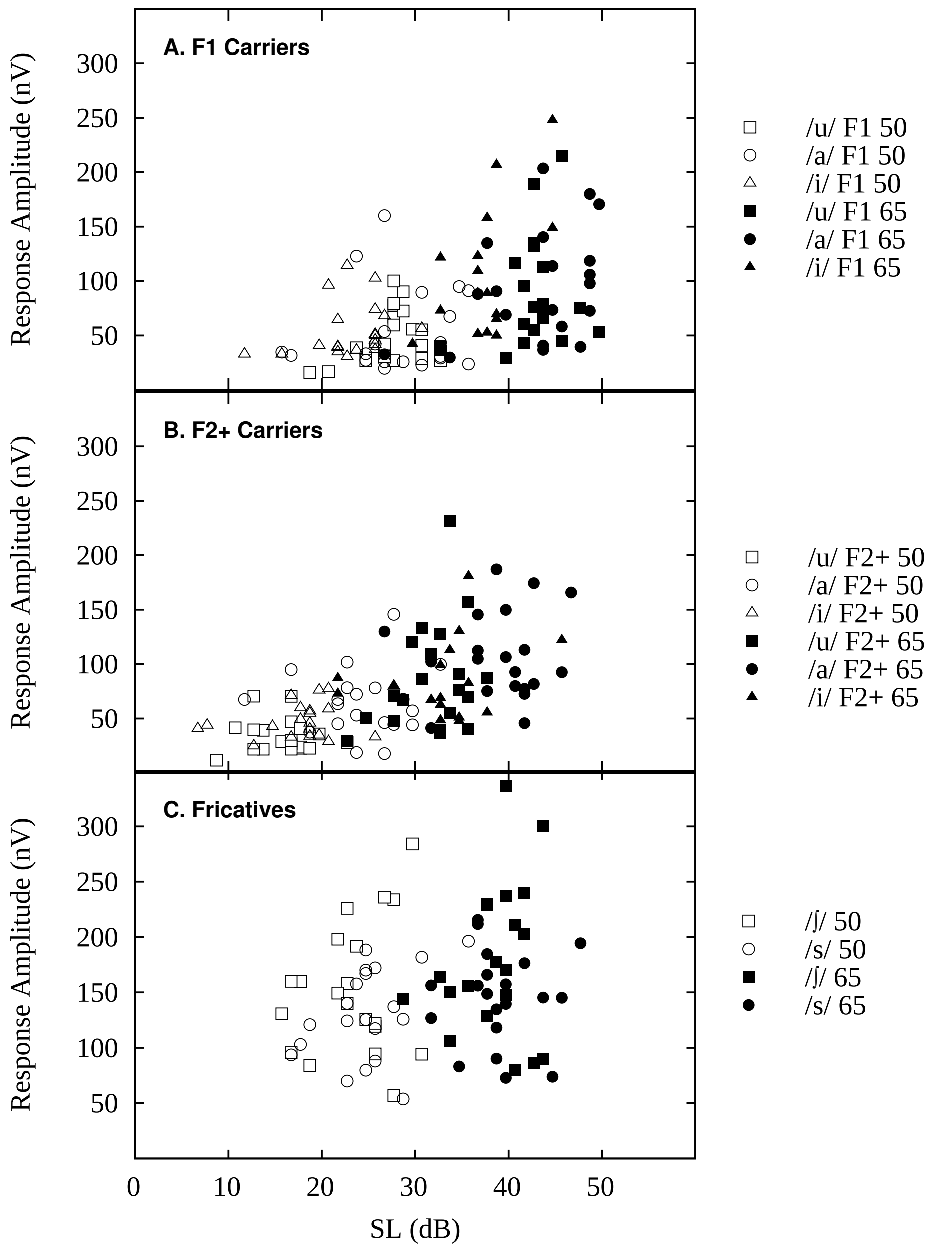

Figure 6-2: Growth of response amplitude with SL. Data include response amplitudes of EFRs that were detected from all 20 participants. Open symbols represent 50 dB SPL and filled symbols represent 65 dB SPL. Random jitter has been added to SL values to help with visualization.

Effect of level on the number of detections

Figure 6-3A illustrates the average number of significant EFRs, out of the eight recorded at each stimulus level. As expected, an increase in the number of detections with level is evident. Figure 6-3B illustrates that the increase in the average number of detections was mainly due to an increase in the number of detections for F1 and F2+ vowel carriers. EFRs to the fricatives were detected in all 20 participants at both test levels, 50 and 65 dB SPL. A paired t-test showed a significant increase in the number of detections with increase in stimulus level from 50 to 65 dB SPL ($t(19)$ = -4.92; $p < 0.001$; Cohen's $d = 1.10$). The average number of detections increased from 6.0 ($SD = 1.65$) at 50 dB SPL to 7.4 at 65 dB SPL ($SD = 0.94$; 95% CI for mean difference [0.80, 1.99]).

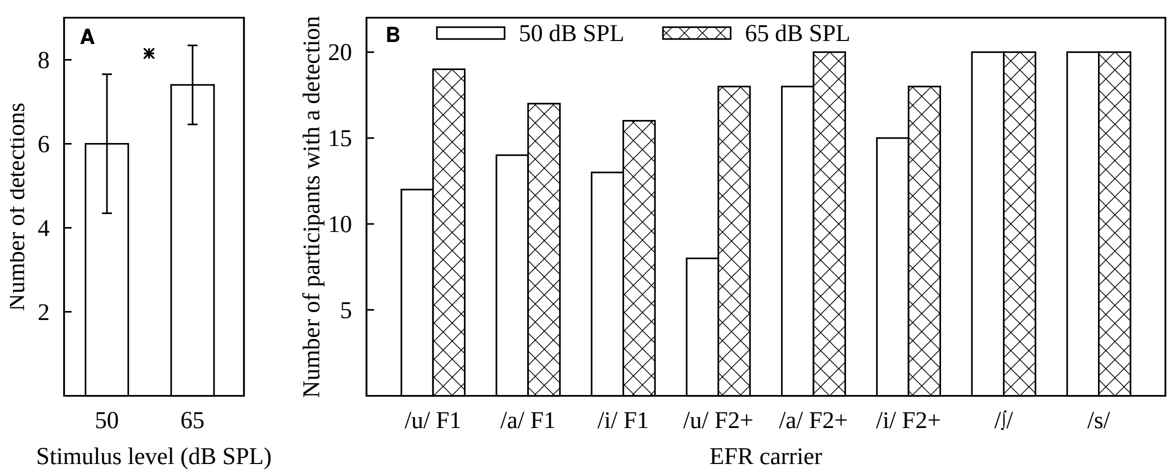

Figure 6-3: Subplot A indicates the average number of EFRs detections out of the eight recorded at each test level for 20 participants. Error bars represent ±1 *SD*. * indicates a significant difference. Subplot B illustrates the number of participants (maximum of 20) with a significant detection for each carrier at both test levels.

Effect of level on response amplitude

Figure 6-4 illustrates the effect of level on response amplitude for each carrier. The increase in average response amplitude is greater for vowel carriers compared to fricatives. The RM-ANOVA indicated a significant main effect of level ($F(1, 19) = 47.57$; $p < 0.001$; $\eta^2_{partial} = 0.72$), a significant main effect of carrier ($F(3.14, 59.69) = 46.93$; $p < 0.001$; $\eta^2_{partial} = 0.71$), and a significant interaction between level and carrier ($F(4.39, 83.40) = 2.75$; $p = 0.029$; $\eta^2_{partial} = 0.13$), suggesting a carrier-specific effect of level. Paired t-tests, comparing response amplitudes at 50 and 65 dB SPL for each carrier, indicated that there was a significant increase in response amplitude with increase in stimulus level for all carriers (see Figure 6-4). The difference in the change in amplitude across carriers could explain the significant interaction. The average change in response amplitude varied from a minimum of 14.21 nV for /ʃ/ to a maximum of 50.65 nV for /u/ F2+.

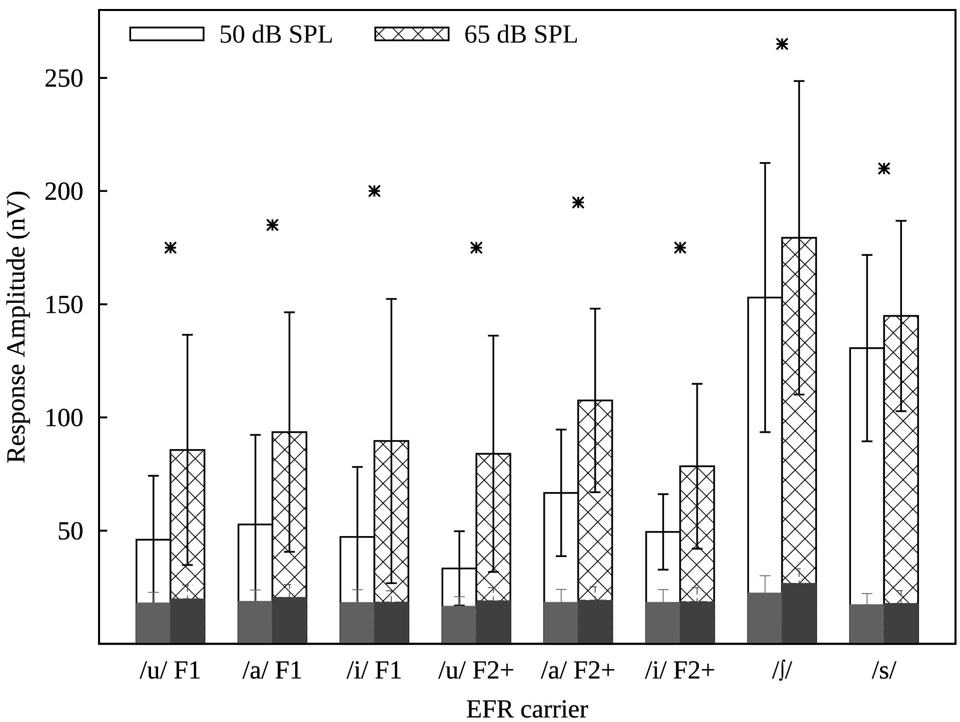

Figure 6-4: Average response amplitude across all eight carriers at both test levels. Data include response amplitudes of all EFRs (significant and non-significant detections). Error bars represent ±1 *SD*. * indicates a significant difference. Solid grey bars represent average noise estimates. (n = 20)

The rate of change of response amplitude (or slope in nV/dB) was calculated by dividing the difference in mean amplitude at 50 and 65 dB SPL by the change in stimulus level (also SL) of 15 dB. Only responses that were detected (i.e., significantly higher than the noise floor) were included in the average, and therefore the participants and the number of data points contributing to each average varied. To allow comparisons with estimates in the literature based on AM tones, the slopes were averaged across a few carriers based on their spectral ranges. The three F1 carriers represent low frequencies (average F1 frequency: 409 Hz), the two F2+ carriers /a/ and /u/ represent mid frequencies (average F2+ frequency: 1241 Hz), and F2+ of /i/ and the fricatives (peak frequencies >2 kHz) represent high frequencies. Average slopes of 2.38, 2.70 and 1.54 nV/dB were obtained for the low, mid and high frequency carriers respectively.

Response detection time

Response detection times for each carrier in terms of sweep length and carrier recording time are illustrated in Figures 6-5A and 6-5B respectively. The figure only includes detection times of responses that were detected (i.e., significantly higher than noise), hence the number of responses/participants contributing to each average varied. The figure indicates variable detection times across carriers and levels. Average detection times using the /susaʃi/ sweep length ranged from approximately 4 to 10 minutes across carriers. Detection times are lower for fricatives relative to vowel carriers, and at 65 dB SPL relative to 50 dB SPL. Figure 6-5B illustrates that mean recording times needed for a significant detection are under two minutes for all carriers. EFRs elicited by /ʃ/ required the shortest recording time for a significant detection.

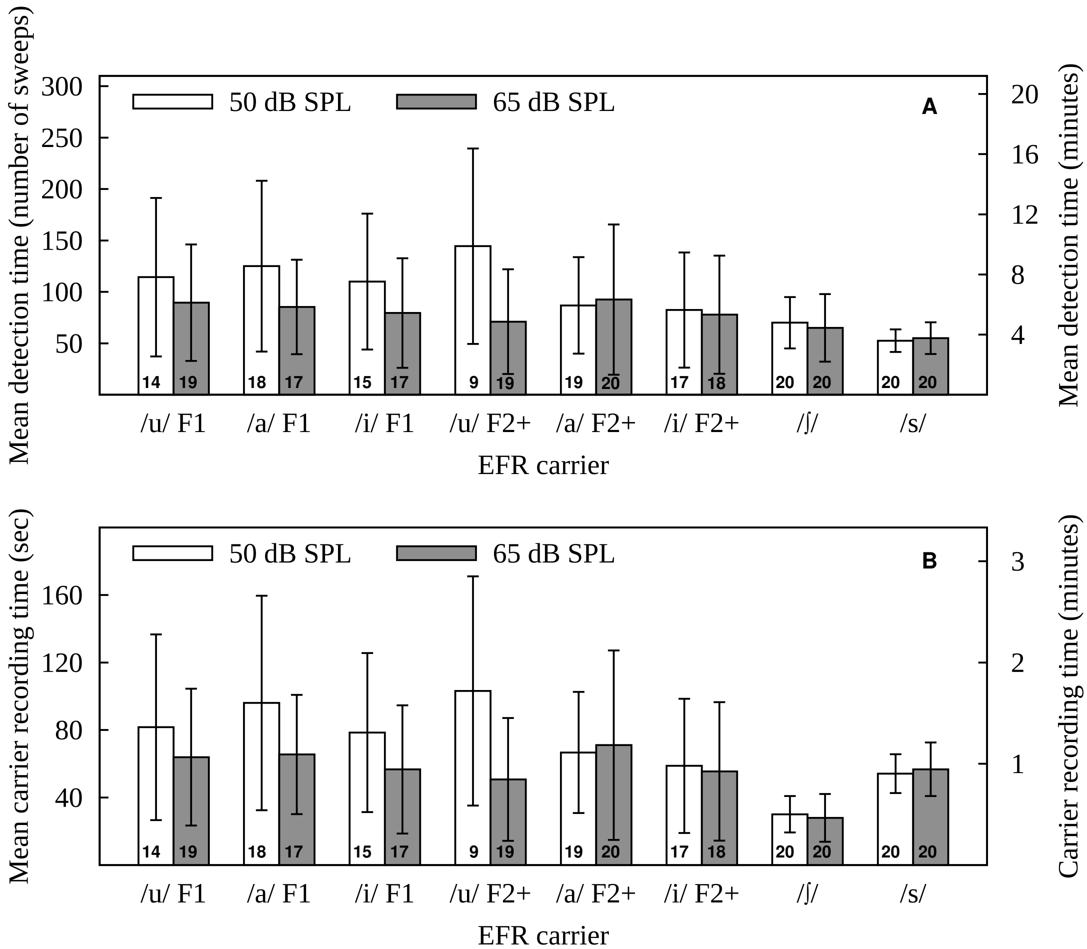

Figure 6-5: Subplot A illustrates the average response detection times in sweeps (primary y-axis) and minutes (secondary y-axis) for participants with a significant detection within 300 sweeps. Detection time in minutes was obtained by multiplying the number of sweeps required for a detection by sweep duration (= 4.1045 s), and dividing by 60. Subplot B illustrates the average carrier recording time (in seconds and minutes) required for a significant detection. Detection time was obtained by multiplying the number of sweeps required for a detection by carrier-specific analysis window lengths. The number of responses/participants contributing to each average is indicated within each bar. Error bars represent ±1 *SD*. (n = 20)

Response detectability and detection patterns

The number of participants with significant EFR detections for the eight individual carriers varied at 50 and 65 dB SPL (Figure 6-3B). Figure 6-1 and Table 6-1 illustrate that the spectra of carriers overlap in peak frequencies and frequency ranges. For example, F1 carriers of /u/ and /i/ have dominant energy under 500 Hz. If we were only interested in inferring about neural representation and hence audibility of a certain frequency region, detection of an EFR elicited by either carrier may suffice. Therefore, we also examined detectability of low, mid and high frequency regions by pooling across a few carriers within a frequency region. Similar to slope computation above, the three F1 carriers were grouped to represent low frequencies, the two F2+ carriers /a/ and /u/ were grouped to represent mid frequencies, and F2+ of /i/ and fricatives were grouped to represent high frequencies. If any one of the two or three EFRs within a frequency band was detected (using a Bonferroni correction), we inferred that the frequency band was represented at the level of the brainstem. For the low and high frequency bands with three carriers each, the Bonferroni corrected critical p value was 0.017. For the mid frequency band with two carrier frequencies, the critical p value was 0.025. The number of participants with a minimum of one detection per frequency band is presented in Figure 6-6A. It is evident that 15 or more participants had a minimum of one detection per frequency band. When individual carriers are considered (Figure 6-3), the number of participants with a detection ranged from 8 to 20 across carriers at 50 dB SPL and from 16 to 20 at 65 dB SPL. However, when frequency bands are considered (Figure 6-6), the number of participants with a detection ranged from 15 to 20 at 50 dB SPL and from 19 to 20 at 65 dB SPL. Therefore, use of alternative detection rules improve detection rates.

Stacked histograms in Figure 6-6B illustrate the distribution of participants with detections in one, two or three bands. At 65 dB SPL, the majority of participants (19/20) had a significant detection in all three frequency bands. At 50 dB SPL, a large proportion (14/20) of participants had a significant detection in all three bands, 19 of 20 participants had a significant detection in at least two bands.

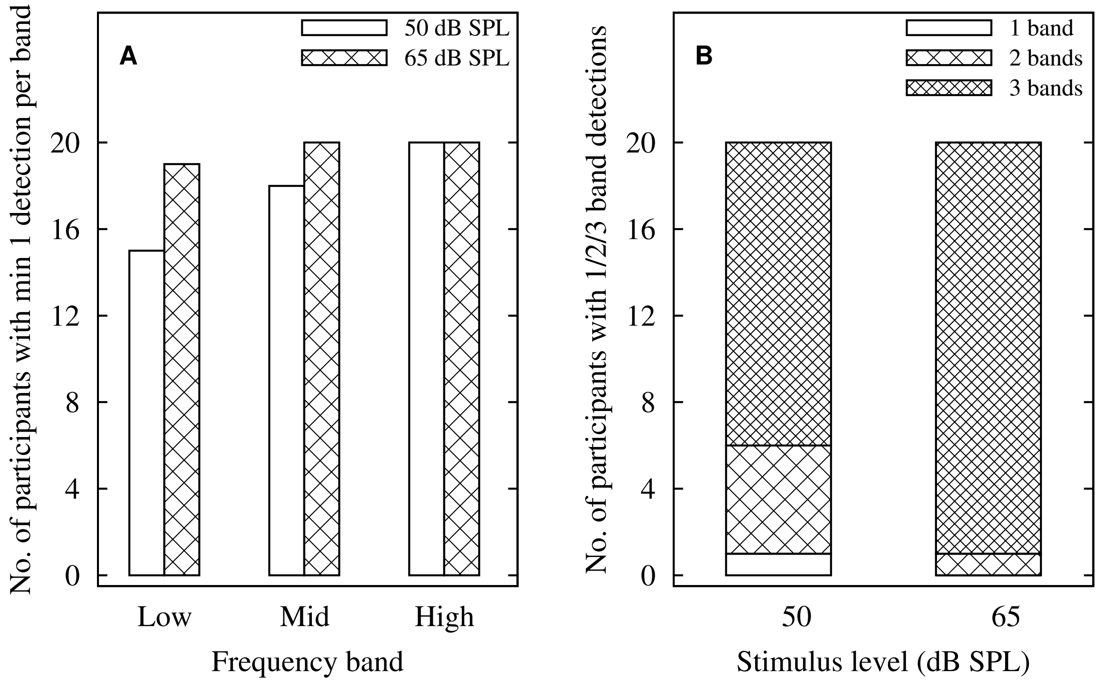

Figure 6-6: Detectability of EFRs across frequency bands. Subplot A illustrates the number of participants with a minimum of one detection in low, mid and high frequency bands. The stacked histograms in subplot B illustrates the number of participants with a detection in 1/2/3 frequency bands. (n = 20)

6.3 Experiment II: Effect of stimulus bandwidth on speech-evoked EFRs and behavioral measures

6.3.1 Method

Participants

All participants who took part in Experiment I also participated in Experiment II.

EFR protocol

Experiment II involved comparison of responses in four conditions that varied in bandwidth. The four bandwidth conditions were low-pass filtered 1kHz (LPF1k), low-pass filtered 2 kHz (LPF2k), low-pass filtered 4 kHz (LPF4k), and full bandwidth (FBW) condition. The stimulus /susaʃi/ was low-pass filtered at cut-off frequencies of 1, 2 and 4 kHz with steep filter skirts using Praat. The responses obtained for the original EFR stimulus sequence at 65 dB SPL in Experiment I, represented the FBW condition in this experiment. The low-pass filtered stimuli matched the original EFR stimulus sequence in spectral levels of their respective pass bands. The attenuator settings for the low-pass filtered conditions were the same as the FBW condition presented at 65 dB SPL, and hence, each condition varied in overall level. The recording time for each condition was 20.5 minutes and the presentation order was randomized across participants. Procedures for stimulus presentation, EFR recording, response analysis and detection were the same as in Experiment I.

Behavioral tests

The effects of bandwidth on the EFR paradigm were compared with the effects of bandwidth on behavioral measures of speech discrimination and sound quality rating.

Speech discrimination task

The University of Western Ontario Distinctive Features Differences Test (UWO DFD) was used as a measure of speech discrimination. The UWO DFD test evaluates the accuracy of consonant identification in fixed context, word-medial positions using a closed set response task paradigm (Cheesman & Jamieson, 1996). This test has been frequently used as an outcome measure of hearing aid fittings (e.g., Jenstad et al., 2007). A total of 42 items (21 consonant items spoken by two male talkers) was used in the study. Stimulus presentation, response recording and scoring was controlled by the Child Amplification Laboratory experimental system. Once the test began, a list of the 21 consonants was displayed on a computer screen. The 42 items were presented in a randomized order, and participants were required to click on the consonant they identified. Percent correct scores were calculated and converted to Rationalised Arcsine Units (RAUs; Sherbecoe & Studebaker, 2004) for statistical analysis. The results presented in figures and tables are in % correct scores.

Similar to the EFR stimulus sequence, the 42 test items in the UWO DFD test were low-pass filtered at 1, 2 and 4 kHz. The original UWO DFD stimuli represented the FBW condition. All participants underwent a practice run with FBW stimuli prior to starting the test. Order of presentation of the four bandwidth conditions was randomized across participants. Stimuli were presented at a level of 65 dB SPL, calibrated in using a concatenated list (with no gaps) of the 42 original (FBW) items used in the study. The level was measured in flat weighted L_{eq} while the stimulus was presented for 30 seconds. Attenuator settings were identical for all four bandwidth conditions. The time taken to complete this task ranged between 10 to 20 minutes.

Sound quality rating task

The MUltiple Stimulus Hidden Reference and Anchors (MUSHRA) paradigm was used to obtain sound quality ratings (International Telecommunication Union [ITU], 2003). The task was implemented using custom software described in Parsa et al. (2013). The MUSHRA task uses a paired comparison paradigm to measure sound quality, allowing for grading of multiple stimuli in comparison with each other (ITU, 2003) with good reliability (Parsa et al., 2013). In this task, a reference stimulus (the best quality), multiple test stimuli that need to be rated, and an anchor (typically of poor quality) are used. The reference and anchor stimuli act as examples for ratings on either ends of the scale provided. In the visual display on a computer screen, a copy of the reference stimulus is always indicated on the top-left corner of the screen. The participant is instructed to make a pairwise comparison of each test stimulus with the reference stimulus on the top-left corner of the screen. The participant rates each test stimulus (one of which is another copy of the reference or the hidden reference) by moving a slider along a continuous scale, arbitrarily labeled as 'bad' on one end and 'excellent' on the other. The presentation of these stimuli are controlled by the participant. The task allows the participant to listen to any stimulus any number of times, in any order, until they are comfortable with their ratings. The order of the stimuli across filter conditions, including the reference and the anchor, were randomized for each participant.

The sentence pair "Raise the sail and steer the ship northward. The cone costs five cents on Mondays" spoken by a male talker was used in this study. This pair was a subset of the four IEEE Harvard sentence pairs used in Parsa et al. (2013). Similar to the EFR stimulus and speech discrimination test, the sentence pair was low-pass filtered at 1, 2 and 4 kHz. The original sentence represented the FBW condition, and served as the reference stimulus. The sentence was low-pass filtered at 0.5 kHz to serve as the anchor. For calibration, the original (FBW) sentence pair was presented in a looped manner without silent gaps and the level was measured in flat weighted L_{eq} while the stimuli was presented for 30 seconds. The attenuator settings were identical for all four bandwidth conditions. The time taken to complete this task ranged between 4 to 10 minutes.

Statistical analysis

Individual RM-ANOVAs were completed to examine the effect of the within-subject factor bandwidth condition (four levels: LPF1k, LPF2k, LPF4k and FBW) on four dependent variables, namely, the number of EFRs detected (maximum of eight), EFR composite amplitude, speech discrimination scores (in RAUs), and sound quality rating. An additional level (anchor, low-pass filtered at 500 Hz) was added for sound quality rating. EFR composite amplitude was obtained by summing response amplitudes across all eight carriers. Composite amplitude was considered to allow incorporation of EFR amplitude as a dependent measure within the four levels of the within-subject factor in the ANOVA. Post-hoc analysis was completed using paired t-tests between all bandwidth conditions. The t-tests were interpreted using the FDR correction for multiple comparisons (Benjamini & Hochberg, 1995).

6.3.2 Results

Data from one participant were excluded from statistical analysis due to high noise estimates.

Effect of stimulus bandwidth on EFR and behavioral tasks

Figure 6-7 illustrates average change in EFR and behavioral measures with increase in bandwidth, superimposed on individual data. Composite response amplitude and behavioral scores increase with increase in bandwidth up to and beyond 4 kHz. The relative change in test scores between LPF4k and FBW is greater for sound quality rating compared to speech discrimination % correct scores. Not indicated in the figure is the mean sound quality rating for the anchor (M = 10.11; SD = 8.72).

The main effect of bandwidth was significant for all four dependent variables: EFR number of detections ($F(2.64, 47.45) = 134.08$; $p < 0.001$; $\eta^2_{partial} = 0.88$), EFR composite amplitude ($F(1.45, 26.15) = 83.51$; $p < 0.001$; $\eta^2_{partial} = 0.82$), speech discrimination scores ($F(2.35, 42.23) = 398.27$; $p < 0.001$; $\eta^2_{partial} = 0.96$), and sound quality rating ($F(2.72, 48.97) = 292.88$; $p < 0.001$; $\eta^2_{partial} = 0.94$). Significant differences due to bandwidth in post-hoc analyses are indicated in Figure 6-7. The change between LPF4k and FBW was not significant for the number of detections. In contrast, increases in composite response amplitude, speech discrimination and sound quality rating from LPF4k to FBW conditions were significant. As expected, the anchor stimulus was significantly lower than all wider bandwidth conditions, indicating that participants likely understood the MUSHRA task.

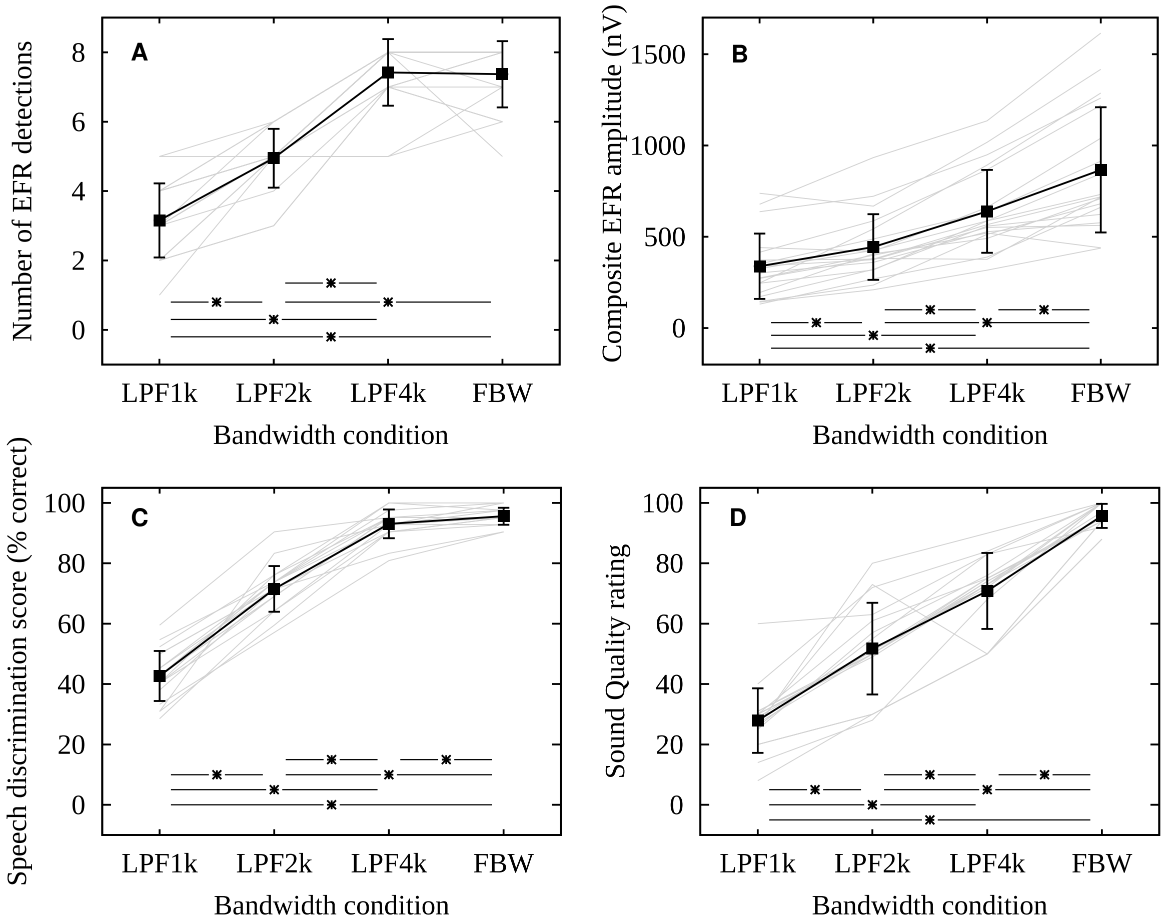

Figure 6-7: Average number of EFRs detected (maximum of 8), composite EFR amplitude (sum of eight response amplitudes), speech discrimination score and sound quality rating across bandwidth conditions are represented in filled black squares in subplots A, B, C and D, respectively. Individual data are represented in grey. Error bars represent ±1 *SD*. * indicates a significant difference. LPF = Low-pass filter, FBW = Full Bandwidth. (n = 19)

Effect of stimulus bandwidth on response amplitudes of individual EFR carriers

The analysis above indicated a significant change in EFR composite amplitude with increase in bandwidth. Since EFR carriers in the stimulus sequence were chosen to represent a range of frequency bands, the change in composite amplitude due to bandwidth may reflect changes in the response amplitudes of any one, or a combination of individual EFRs. The mean response amplitude for each carrier in multiple bandwidth conditions is illustrated in Figure 6-8. The figure suggests that the change in EFR amplitude due to bandwidth varied across carriers. To assess the effect of bandwidth condition on response amplitudes for each individual EFR carrier, a RM-ANOVA was completed with bandwidth (four levels: LPF1k, LPF2k, LPF4k and FBW) and carriers (eight levels: /u/ F1, /a/ F1, /i/ F1, /u/ F2+, /a/ F2+, /i/ F2+, /ʃ/ and /s/) as the two within-subject factors, and response amplitude as the dependent measure. The results indicated a significant main effect of bandwidth ($F(1.45, 26.15) = 83.51$; $p < 0.001$; $\eta^2_{partial} = 0.82$), a significant main effect of carrier ($F(2.49, 44.48) = 10.61$; $p < 0.001$; $\eta^2_{partial} = 0.37$), and a significant interaction between bandwidth condition and carrier ($F(5.33, 96.01) = 44.31$; $p < 0.001$; $\eta^2_{partial} = 0.71$). Post-hoc analysis included paired t-tests comparing response amplitudes across all possible pairwise combinations of bandwidth condition for each carrier. Significantly different conditions are indicated in Figure 6-8.

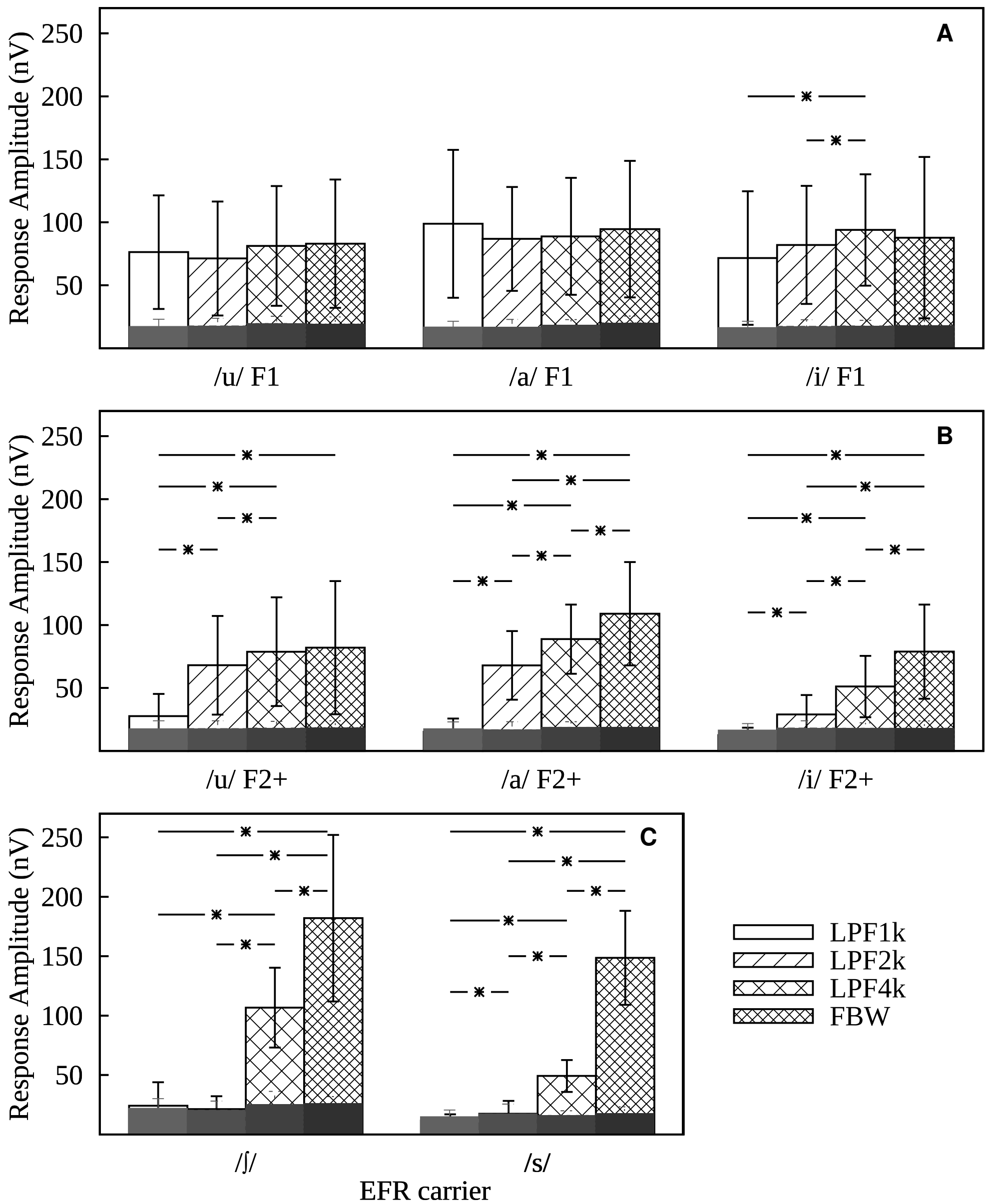

Figure 6-8: Average response amplitudes for each carrier across the four bandwidth conditions. Error bars represent ±1 *SD*. * indicates a significant difference. The solid grey bars represent average noise estimates. (n = 19)

The /u/ and /a/ F1 carriers do not show a significant change in response amplitudes across the different bandwidth conditions. This is consistent with the low frequency peaks for both carriers (see Table 6-1 and Figure 6-1). The carrier /i/ F1 increases significantly in response amplitude between LPF1k and LPF4k, as well as LPF2k and LPF4k, although the dominant spectral peak in this carrier is around 300 Hz, and the low-pass filter cut-off was 1150 Hz (see Table 6-1). All three F2+ carriers show a significant increase in response amplitude between LPF1k and LPF2k bandwidth conditions. This is consistent with the location of F2 peak frequencies, carrier spectra and cut-off frequencies used during stimulus preparation (see Table 6-1 and Figure 6-1). The /a/ and /i/ F2+ carriers show a steady increase in response amplitude up to the FBW condition, consistent with their spectra. Both fricative carriers show a substantial increase in response amplitudes between LPF2k and LPF4k, as well as LPF4k and FBW. On average, the carriers /ʃ/ and /s/ show a change of 75.28 and 99.33 nV, respectively, between LPF4k and FBW. This relatively large change likely contributes substantially to the increase in composite amplitude beyond 4 kHz (Figure 6-7B). A significant increase in amplitude was also found between LPF1k and LPF2k for the carrier /s/, but not for /ʃ/. The significant change for /s/ was unexpected because the carrier was high-pass filtered at 3 kHz.

Relationship between objective (EFR) and subjective (behavioral) measures of bandwidth

Figure 6-9 illustrates the behavioral test scores plotted against the number of EFR detections and EFR composite amplitude. The figure displays mean scores averaged across 19 participants along with individual scores. The overall positive trends suggest that an increase in bandwidth has a similar impact on behavioral as well as EFR measures. Although the asymptotic trend in the number of detections does not reflect the change in behavioral measures (Figures 6-9A and 6-9B), changes in EFR composite amplitude parallel changes in both behavioral measures beyond LPF4k (Figures 6-9C and 6-9D). A significant positive correlation was found between the number of detections and speech discrimination scores (Pearson $r = 0.87$; $t(17) = 18.36$; $p < 0.001$), as well as the number of detections and sound quality rating (Pearson $r = 0.74$; $t(17) = 14.59$; $p < 0.001$). Also, a significant positive correlation was found between EFR composite amplitude and speech discrimination scores (Pearson $r = 0.67$; $t(17) = 9.25$; $p < 0.001$), and between EFR composite amplitude and sound quality rating (Pearson $r = 0.60$; $t(17) = 10.88$; $p < 0.001$). The significant positive correlations between EFR parameters and behavioral scores support convergent validity of the EFR paradigm in demonstrating sensitivity to audible bandwidth.

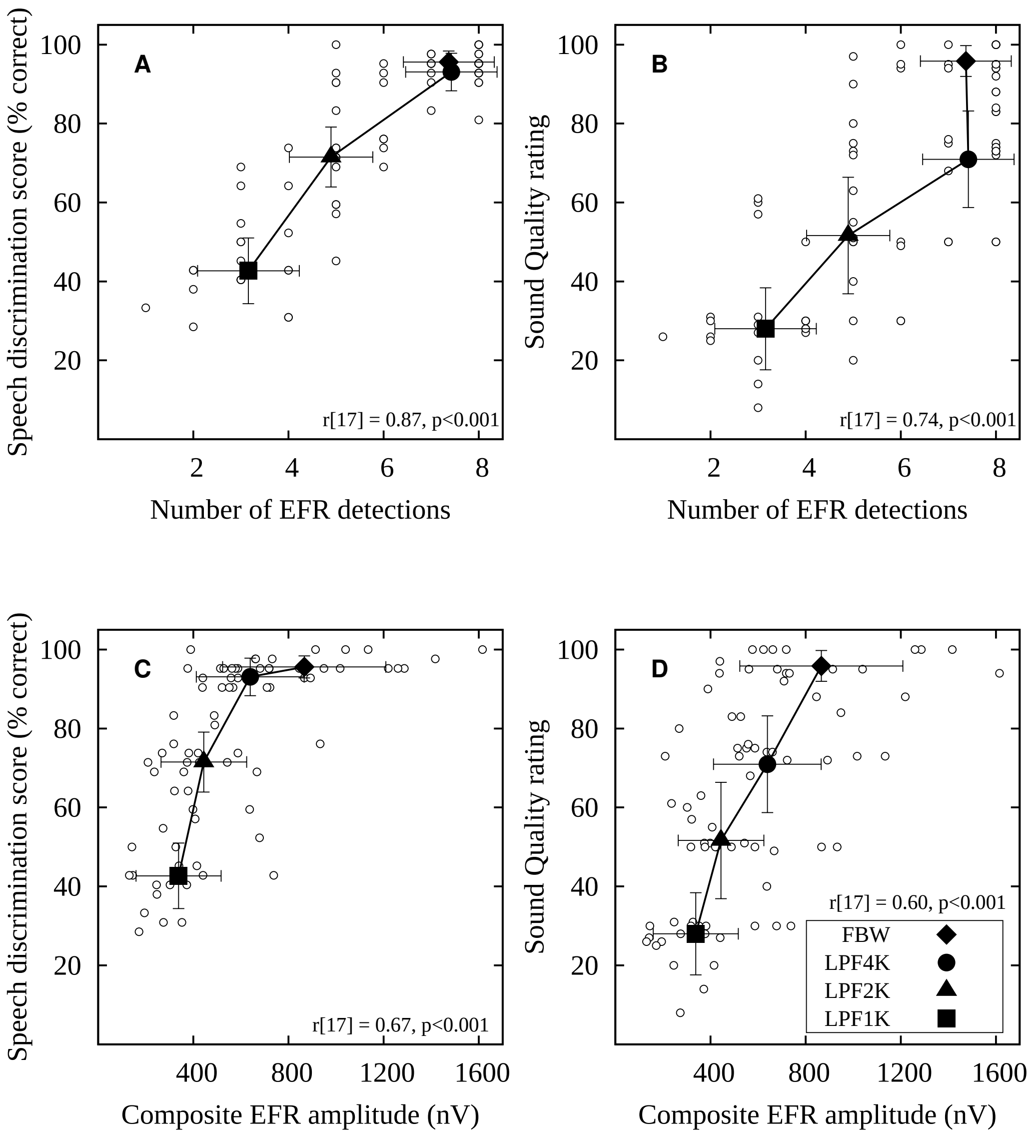

Figure 6-9: Behavioral scores plotted as a measure of EFR parameters. Subplots A and B illustrate speech discrimination scores and sound quality rating as a measure of the number of EFR detections across the four bandwidth conditions. Subplots C and D illustrate the same behavioral scores as a measure of composite EFR amplitude across the four bandwidth conditions. Black filled symbols represent mean data and open circles represent individual data. Error bars represent ±1 *SD*. (n = 19)

6.4 Discussion

The aims of Experiment I and II were to assess sensitivity of the EFR paradigm to changes in audibility due to stimulus level and bandwidth. Change in the number of EFRs detected and EFR amplitude were compared across test conditions. In general, both experiments demonstrated an increase in the number of EFRs detected and EFR amplitude with improved audibility.

6.4.1 Experiment I: Effect of level

Results indicate that the increase in test level, and thereby SL, of carriers by 15 dB, resulted in a significant increase in response amplitudes for individual EFR carriers as well as the overall number of detections at each test level. The increase in response amplitude leads to a higher probability of detection as the criteria for detection (F ratio) is based on the response amplitude relative to the noise amplitude. The level-dependent changes in the present study support previous findings that demonstrate a positive relationship between level and response amplitudes (Lins, Picton, Picton, Champagne, & Durieux-Smith, 1995; Picton et al., 2005, 2003; Vander Werff & Brown, 2005). The change in response amplitudes due to level can be explained by a combination of increased neural firing rates (Sachs & Abbas, 1974) and spread of excitation in the peripheral auditory system (Moore, 2003; Moore & Glasberg, 1987). Spread of excitation recruits more sensory cells, and thereby nerve fibers and synapses, resulting in an increase in input to the brainstem generators of EFRs (Picton et al., 2003; Purcell & Dajani, 2008).

The pattern of intensity-response amplitude slope of speech EFRs is similar to AM tones (Lins et al., 1995; Vander Werff & Brown, 2005); high frequency carriers tend to show a shallower slope than lower frequency carriers. However, the absolute magnitudes of the slopes are higher than that for AM tones. The estimated slopes for low (500 Hz), mid (1 & 2 kHz) and high (4 kHz) frequency AM tones were ~1.4, 1.3, and 0.84 respectively (Lins et al., 1995; Vander Werff & Brown, 2005). The steeper slope for this study's broadband stimuli could be due to a larger excitation area on the basilar membrane, compared to AM tones with only three frequency components. For example, in the case of vowels, it may be that the higher stimulus level allows for better interaction between many harmonics causing a larger increase in response amplitude. The difference in slope across carrier frequencies could be explained, at least in part, on the basis of excitation patterns on the basilar membrane. With increase in level, excitation patterns show more basal spread of activity towards frequencies higher than the stimulus frequency (Moore, 2003; Moore & Glasberg, 1987). Since the basal spread is likely to be greater for low frequency stimuli, the growth is probably steeper for low frequency stimuli relative to high frequency stimuli (Lins et al., 1995).

Response detection time

Response detection time based on sweep length and carrier recording time indicate lower detection times for fricatives, on average (Figures 6-5A & 6-5B). Particularly, the fricative /ʃ/ has the least carrier recording time (Figure 6-5B) likely due to higher response amplitudes relative to other carriers (see Figure 6-4). Average noise floors are variable across carriers too (see Figure 6-4), however, the differences in amplitude are larger than the differences in noise estimates. Variations in response amplitudes lead to differences in F ratios and therefore the time to reach the critical F ratio. Also, detection times tend to be longer at test levels of 50 dB SPL relative to 65 dB SPL. This can also be explained in terms of differences in response amplitudes while noise estimates are similar; shorter detection times at the higher test level are facilitated by higher amplitude responses at 65 dB SPL (see Figure 6-4). The carrier recording times for vowels appear to be longer than detection times reported by Aiken and Picton (2006), possibly due to a floor effect in the present study as responses were not analyzed under 50 sweeps.

A direct comparison with other aided protocols such as CAEPs is difficult to make due to limited data available on test time. Test time computed based on the stop criterion on 'accepted' sweeps and inter-stimulus level may range from under 2 to 4 minutes *per* carrier (Carter, Dillon, Seymour, Seeto, & Van Dun, 2013; Van Dun, Carter, & Dillon, 2012). Average carrier recording time required for a significant detection (Figure 6-5B) are comparable to these estimates. However, the effective testing time necessary using the current stimulus would likely be shorter due to simultaneous presentation of vowel carriers. For the testing conditions in the present study, the average test time necessary for a significant detection using the stimulus /susaʃi/ fell under 10 minutes for all carriers (see Figure 6-5A). It is encouraging to note the clinically feasible test times of the EFR paradigm using the stimulus /susaʃi/ to infer neural representation of low, mid and high frequency dominant carriers in the auditory system. Note that the detection times are representative only of responses that were significantly higher than the noise. Due to the use of a fixed number of sweeps, detection times do not represent responses that may have taken over 300 sweeps to be detected (e.g., responses that might have been detected at 320 sweeps are not represented).

In summary, results of Experiment I suggest that the EFR paradigm shows changes in amplitude and detection with improvement in SL. Detection times at suprathreshold levels demonstrate clinically feasible test times. Alternate scoring rules such as combining carriers within a frequency band improves detection rates. Although combining carriers within a frequency band reduces carrier-specific information, it may prove to be an advantage clinically to demonstrate that detection is possible in each band in possibly shorter test times. The test could continue for longer to obtain carrier-specific detections, if time permits.

6.4.2 Experiment II: Effect of bandwidth

Effect of bandwidth on speech discrimination and sound quality rating

The effects of bandwidth on speech discrimination scores and quality ratings are consistent with findings in the literature. As the low-pass filter cut-off is increased, the number of consonants correctly identified increases in the UWO DFD (Cheesman & Jamieson, 1996) and other tests of speech recognition (see review by Stelmachowicz et al., 2004). Similar trends are observed in other speech discrimination tests such as the NU-6 (Studebaker et al., 1993) and CID-W22 (Studebaker & Sherbecoe, 1991), both of which are meaningful monosyllabic word lists. In the present study, the increase in speech discrimination score was substantial until the LPF4k condition. There was only a small (but significant) improvement in scores with further increase in bandwidth (Figure 6-7C). This growth function is broadly consistent with patterns seen in the speech discrimination tests mentioned above, and is supported by the band importance functions of various speech tests (Pavlovic, 1994). Band importance functions represent contributions of different frequency bands to speech intelligibility: larger values indicate higher impact on speech intelligibility. These indices tend to be lower for frequency bands over 4 kHz relative to those between 1 and 4 kHz (Pavlovic, 1994), explaining the small increase in speech discrimination scores beyond LPF4k in the present study.

Sound quality ratings show a steady increase until the FBW condition (Figure 6-7D), even beyond 4 kHz where changes in speech discrimination scores tend to be small (Pittman, 2010). The improvement in sound quality rating when stimulus bandwidth is increased beyond 4 kHz has been repeatedly demonstrated in individuals with normal hearing (Füllgrabe et al., 2010; Moore & Tan, 2003; Ricketts et al., 2008; Voran, 1997).

Effect of bandwidth on EFRs

The number of detections show an increase only until the LPF4k condition however, the composite amplitude, which is the sum of amplitudes of individual carriers, continues to increase even beyond the LPF4k condition (Figures 6-7A & 6-7B). All carriers used in this study have at least some stimulus energy below 4 kHz (see Figure 6-1). This may have therefore led to the asymptotic trend in the number of detections beyond LPF4k. However, the increase observed in composite response amplitude up to the FBW condition supports sensitivity of the EFR paradigm to stimulus bandwidth even beyond 4 kHz. The effective change in stimulus bandwidth is likely to have been from 4 kHz to 10 kHz, consistent with the flat frequency response of the ER-2 transducer up to 10 kHz and the presence of significant frication energy in the 4 to 10 kHz region (Figure 6-1). The growth in response amplitude with stimulus bandwidth is likely due to concomitant increases in the area of excitation in the cochlea. Similar to the effects of level, the larger excitation area will involve recruitment of a larger number of sensory cells and neurons contributing to more robust inputs to the brainstem (Picton et al., 2003; Purcell & Dajani, 2008).

Considering carriers individually, we observe that most carriers show a change in response amplitude in the expected direction based on their spectra (Figure 6-8). Some changes that were unexplained by carrier spectra, were the effects of bandwidth on EFRs elicited by /i/ F1 and /s/. In the case of /i/ F1, response amplitude in the LPF4k condition was significantly higher than the LPF1k and LPF2k conditions by an average of 22 and 11 nV respectively. It may be that addition of higher frequencies may enhance the EFR elicited by /i/ F1, but only by a small amount. The change in response amplitude of /i/ F1 is in a direction opposite to that observed in the case of multiple ASSRs, in which multiple carrier frequencies are presented simultaneously. At moderate to high stimulus levels, addition of a high frequency tone an octave higher than the target stimulus frequency, can suppress the response generated at the target stimulus frequency (John et al., 1998, 2002; McNerney & Burkard, 2012). The interference effect is non-significant at lower levels (John et al., 1998, 2002; McNerney & Burkard, 2012). The differences in phenomena observed between the current and previous studies may be due to the stimulus levels used and/or the level of the high frequency interfering band.

In the case of /s/, the response amplitude in the LPF2k condition was significantly higher than the LPF1k condition. As suggested in the Results section, this was an unexpected finding because /s/ was high-pass filtered at 3 kHz. The changes in response amplitude of /s/ may have been due to random variations in noise or a type I error.

Stimuli used in this study are naturally occurring broadband stimuli. The goals during stimulus preparation were to represent the broad range of frequencies present in speech, as well as improve frequency specificity of naturally occurring broadband phonemes. Although we attempted to improve frequency specificity of these stimuli, they are nevertheless more broadband than stimuli commonly used for threshold estimation. Carrier-specific changes in bandwidth (Figure 6-8) suggest that the stimuli are fairly frequency specific. However, it is important to note that frequency specificity does not ensure place specificity in terms of the extent of basilar membrane activation (Picton et al., 2003).

6.5 Summary and Conclusions

The current study presented a novel test paradigm based on speech-evoked EFRs for use as an objective hearing aid outcome measure. The method uses naturally spoken speech tokens that were modified to elicit EFRs from spectral regions important for speech understanding, and speech and language development. The vowels /u/, /a/ and /i/ were modified to enable recording of two EFRs, one from the low frequency F1 region, and one from the mid to high frequency F2+ regions. Fricatives /ʃ/ and /s/ were amplitude modulated to enable recording of EFRs from the high frequency regions. The present study evaluated the sensitivity of the proposed test paradigm to changes in level in Experiment I, and changes in stimulus bandwidth in Experiment II in young adults with normal hearing. The paradigm demonstrated sensitivity to level and bandwidth in terms of the number of EFR detections and response amplitude. Additionally, the paradigm demonstrated convergent validity when compared with changes in behavioral outcome measures such as speech discrimination and sound quality rating due to bandwidth. In summary, this method may be a useful tool as an objective aided outcome measure considering its running speech-like stimulus, representation of spectral regions important for speech understanding, level and bandwidth sensitivity, and clinically feasible test times. The validity of the paradigm in individuals with hearing loss who wear hearing aids requires further investigation.

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Chapter 7

Effect of stimulus level, bandwidth and hearing aid amplification on speech-evoked envelope following responses in adults with hearing loss

7.1 Introduction

Outcome evaluation, a process of measuring the impact of intervention, is a recommended component of the hearing aid fitting process (American Academy of Audiology, 2013; Joint Commitee on Infant Hearing, 2007). The importance of outcome evaluation in young infants has received increased emphasis since the advent of universal newborn hearing screening programs. Outcome evaluation is a challenge in infants diagnosed and fitted with hearing aids by six months of age because reliable behavioral responses are difficult to obtain. To evaluate the benefit of amplification in young infants, several outcome measures have been proposed that can be placed on a continuum ranging from subjective to objective measures. Some authors have suggested the use of auditory evoked potentials as an objective outcome measure for infants (e.g., aided Cortical Auditory Evoked Potentials [CAEP], Golding et al., 2007; Purdy et al., 2005). The present study evaluated the utility of a previously proposed objective test paradigm based on speech-evoked Envelope Following Responses (EFR; Chapter 6). EFRs are neural responses phase-locked to the envelope of a periodically varying stimulus, and are elicited at the stimulus envelope frequency (Picton, John, Dimitrijevic, & Purcell, 2003). EFRs evoked by vowels are elicited at the fundamental frequency of the voice (f_0; Aiken & Picton, 2006). This speech-evoked EFR paradigm has not been previously evaluated in a clinical population and therefore the goal of the present study is to evaluate the method in a sample of adults with hearing loss as a precursor to further work in the pediatric population.

The EFR approach under investigation uses a naturally spoken speech token /susaʃi/, modified to elicit eight individual EFRs: two by each vowel and one by each fricative (Chapter 6). Each vowel elicits two simultaneous EFRs, one from the band containing the first formant (F1) and one from the band containing the higher formants (F2+). Our previous work demonstrated that this measurement is sensitive to changes in stimulus level and bandwidth in a group of young adults with normal hearing (Chapter 6). Specifically, increases in stimulus level and bandwidth led to significant increases in response amplitude and the number of EFRs detected per condition.

The speech-evoked EFR test paradigm has multiple advantages as an objective aided outcome measure. The foremost is that the stimulus resembles running speech. Assessment of hearing aid responses with running speech, rather than non-speech test signals or brief excerpts of speech, is preferred in order to avoid undesired interactions with non-linear signal processing (Easwar, Purcell, & Scollie, 2012; Scollie & Seewald, 2002; Stelmachowicz, Kopun, Mace, & Lewis, 1996). The EFR stimulus includes five phonemes that represent the spectral regions important for speech intelligibility (Ling, 1989; Scollie et al., 2012). In preparing the stimulus, the frequency specificity of each carrier was improved by high-pass filtering the fricatives and differentiating formant regions of vowels with different fundamental frequencies (Chapter 6). Responses to these modified stimuli were analyzed using statistical response detection methods to reduce tester bias (Aiken & Picton, 2006; Choi, Purcell, Coyne, & Aiken, 2013). Additionally, the proposed EFR paradigm can be reliably recorded in sleep and has shown to have clinically feasible test times (Chapter 6; Aiken & Picton, 2006; Choi et al., 2013). However, it is important to note that the present EFR approach allows inference of aided sound encoding only to the level of the brainstem (Herdman et al., 2002), unlike other proposed objective aided outcome measures such as the CAEP (e.g., Korczak, Kurtzberg, & Stapells, 2005).

Auditory Steady-State Responses (ASSRs) elicited by Amplitude-Modulated (AM) tones have been used to evaluate benefit from amplification using threshold-based paradigms (Damarla & Manjula, 2007; Picton et al., 1998; Shemesh, Attias, Magdoub, & Nageris, 2012; Stroebel, Swanepoel, & Groenewald, 2007). In children with hearing loss, frequency-specific aided ASSR thresholds were obtained within 13 to 15 dB of the aided thresholds obtained in behavioral tests (Picton et al., 1998; Stroebel et al., 2007)). Also, improvement in ASSR thresholds with a hearing aid was found to be significantly correlated with the measured hearing aid gain (Damarla & Manjula, 2007). These studies suggest that AM-ASSRs can reflect improved audibility due to amplification.

The sensitivity of speech-evoked EFRs to amplification and hearing aid settings has been illustrated previously using the syllable /da/ (Anderson & Kraus, 2013). Use of a hearing aid led to an increase in response amplitude at f_0 relative to unaided tests in sound field. Results of this case study supports the sensitivity of EFRs to improved audibility, although further evidence with a larger study sample is needed.

In the present study, we assessed the utility of the speech-evoked EFR test paradigm using the five-syllable stimulus /susaʃi/ as an objective aided outcome measure in adults with hearing loss wearing hearing aids. In Experiment I, we evaluated the sensitivity of speech-evoked EFRs to changes in audibility caused by an increase in stimulus level and use of amplification. Based on earlier AM-ASSRs studies on response-intensity relationships in unaided and aided conditions (e.g., Dimitrijevic, John, & Picton, 2004; Picton, Dimitrijevic, Perez-Abalo, & Van Roon, 2005), we hypothesized that response amplitude and the number of EFR detections will increase with an increase in stimulus level as well as the use of amplification.

As described above, the EFR test paradigm demonstrated systematic changes to varying stimulus bandwidth in adults with normal hearing tested using an ER-2 insert earphone (Chapter 6). The changes in EFR characteristics across different bandwidth conditions also paralleled the changes observed in speech discrimination scores and sound quality ratings. This lends further support for the use of a stimulus like /susaʃi/ with eight carriers to assess changes across hearing aid conditions. In Experiment II of the present study, we investigated the sensitivity of speech-evoked EFRs to changes in aided audible bandwidth. We hypothesized that an increase in stimulus bandwidth will result in an increase in EFR amplitude and the number of aided EFRs detected.

In addition, we compared changes in aided EFRs with changes in aided behavioral outcome measures such as speech discrimination and sound quality rating across varying stimulus bandwidths. The effects of stimulus bandwidth on speech discrimination and sound quality are well established in adults with hearing loss, with speech discrimination improving substantially with bandwidths to ∼3–4 kHz (Amos & Humes, 2007; Hornsby & Ricketts, 2006; Vickers, Moore, & Baer, 2001) and in smaller increments above 4 kHz (Hornsby & Ricketts, 2003, 2006; Horwitz, Ahlstrom, & Dubno, 2008). Sound quality has been shown to improve with wider bandwidth (Parsa, Scollie, Glista, & Seelisch, 2013). However, improvement beyond 4–5 kHz has demonstrated variability associated with audiometric (Moore, Füllgrabe, & Stone, 2011; Ricketts, Dittberner, & Johnson, 2008) and amplification factors (Moore, 2012; Moore et al., 2011; Moore & Sek, 2013). Current evidence suggests that individuals with mild to moderate degrees of hearing loss prefer wider bandwidths extending beyond 5 kHz when provided with suitable high frequency gain, and this may vary with experience (Moore, 2012; Moore et al., 2011; Ricketts et al., 2008). For these reasons, in the present study, we assessed behavioral measures of speech discrimination and sound quality to ascertain the relation between aided EFRs and perception of aided sound, as the stimulus bandwidth is varied.

In summary, the present study evaluated the effect of changes in audibility on EFRs in adults with hearing loss using the proposed EFR test paradigm. In Experiment I, audibility was varied by a change in stimulus level and use of amplification, and in Experiment II, audibility was varied by varying the stimulus bandwidth with low-pass filtering.

7.2 Experiment I: Effect of stimulus level and amplification on speech-evoked EFRs

7.2.1 Method

Participants

The study included 21 adults (16 males and five females) recruited locally from the city of London, ON, Canada. Their ages ranged between 60 and 79 years (M_{age} = 72 years; SD_{age} = 4.51). Routine audiometry (air conduction and bone conduction) using a GSI-61 audiometer was completed on all participants. Air conduction thresholds were obtained using ER-3A insert earphones. Participants with hearing thresholds greater than 75 dB HL between 0.25 and 6 kHz or air bone gaps greater than 15 dB HL at 0.5, 1 or 2 kHz were not included in the study. All participants were hearing aid users, with a minimum of three months of hearing aid experience in at least one ear. Routine otoscopy completed on all participants ruled out any contraindications such as occluding wax, discharge or foreign bodies in the ear. Tympanograms obtained in the test ear using a Madsen Otoflex 100 immittance meter indicated static compliance and tympanometric peak pressure within normal limits except in one participant. This participant (female, 72 years) presented a tympanogram with peak pressure of -204 daPa and static compliance of 1.54 ml with the air bone gap not exceeding 10 dB between 0.5 and 2 kHz. Since previous records also indicated consistent negative peak pressure, the participant was not excluded from the study. Peak pressure measured prior to data collection in all test sessions varied within 15 daPa of the baseline estimate. All participants reported English as their first language and provided informed consent. Volunteers with any self-disclosed neurological disorders were not included in the study. The study protocol was approved by the Health Sciences Research Ethics Board, Western University, Canada. Participants were compensated for their time.

EFR stimulus

The stimulus used was a naturally spoken speech token /susaʃi/ modified to elicit eight EFRs with carriers representing a range of frequencies (Chapter 6). The stimulus token /susaʃi/ was produced by a 42-year old male from Southwestern Ontario whose first language was English. Voice recordings were made with a studio grade microphone (AKG Type C 4000B) in a sound booth using Spectraplus software (version 5.0.26.0; Pioneer Hill Software LLC, Poulsbo, WA, USA). Stimulus phonemes from the original production were extracted and individually edited using softwares Praat (Boersma, 2001), Goldwave (version 5.58, GoldWave Inc., St. John's, NL, Canada), and MATLAB (version 7.11.0 [R2010b]; Mathworks, Natick, MA, USA). Fricatives /ʃ/ (234 ms) and /s/ (274 ms), representing the high frequencies, were high-pass filtered at 2 and 3 kHz respectively using Praat, and 100% amplitude modulated at 93.02 Hz using MATLAB. Fricatives were matched in overall Root-Mean-Square (RMS) levels before and after modulation. The vowels, representing low and mid frequencies, were modified such that the original f_0 was maintained in the regions of Formants two and higher (F2+), and a lower f_0 (original f_0–8Hz) was created in the region of Formant one (F1). The f_0 in the original vowels was lowered using the 'shift frequency' function in Praat and low-pass filtered to isolate the F1 band. The F2+ band was obtained by high-pass filtering the original vowel. The cut-off frequencies of the F1 and F2+ bands were chosen to be close to the mid-point between the estimated F1 and F2 peak frequencies. The f_0-lowered F1 band was then combined with the F2+ band to obtain dual-f_0 vowels and the levels of the dual-f_0 vowels were matched with the naturally produced original vowel. The dual-f_0 vowel structure increases frequency specificity of broadband vowels by enabling recording of two simultaneous EFRs from different spectral regions. Using Goldwave, the carrier /ʃ/ was attenuated between 2650 and 3650 Hz by 3 dB to fall within the 99^{th} percentile of the International Speech Test Signal (ISTS) at matched overall levels. The two fricatives, together with two formant carriers per vowel, formed eight EFR carriers in the stimulus sequence /susaʃi/. For further details on vowel characteristics and carrier spectra, please refer to Table 6-1 on p. 142 and Figure 6-1 on p. 143 (Chapter 6). Each sweep consisted of the stimulus /susaʃi/ presented in opposite polarities. The duration of a sweep was 4.1045 seconds.

Hearing aid and hearing aid fitting

Unitron Quantum S and Quantum HP hearing aids were chosen to fit the range of hearing losses included in the study. These hearing aids are commercially available 20-channel digital aids that can be programmed to provide individualized fittings for each participant's hearing loss. The Quantum S hearing aid is suitable for mild to moderate hearing losses whereas the Quantum HP is suitable for moderately-severe to severe hearing losses. Hearing aids were coupled to hard unvented acrylic ear molds that were custom-made for all participants. For each participant, one of the hearing aids was chosen depending on the hearing loss in the test ear. Test ear chosen was the ear with better (lower) pure-tone thresholds, or the ear with a hearing aid, if the participant only used one hearing aid. Test ear was chosen randomly when hearing loss was symmetrical and when hearing aids were worn bilaterally. The average audiogram of the test ear along with individual audiograms are provided in Figure 7-1.

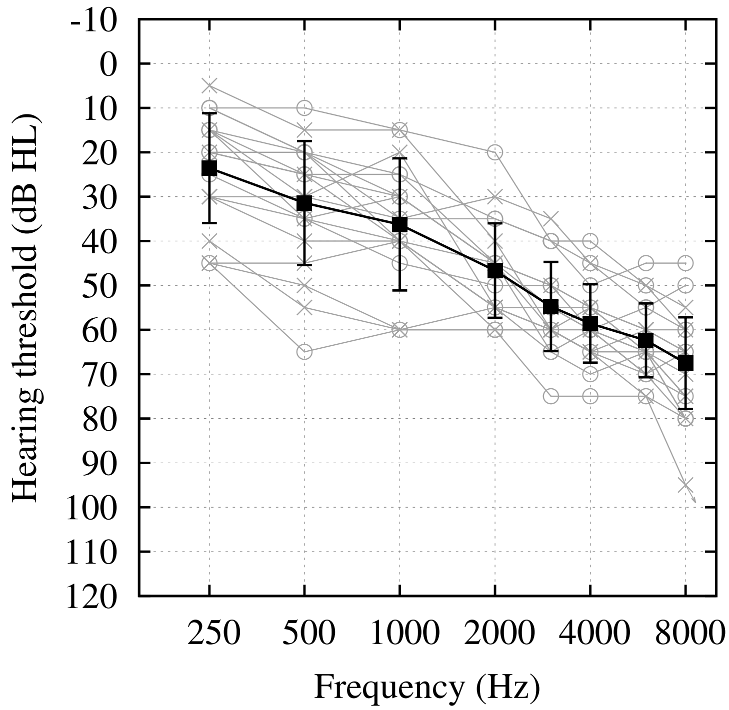

Figure 7-1: Test ear average audiogram indicated in black filled squares, superimposed on individual audiograms in grey. Open circles and crosses indicate right and left test ears respectively. Error bars represents ± 1 *SD*.

The hearing aids consisted of one program for Direct Audio Input (DAI) only. Use of DAI as the input method minimizes the potential disadvantages associated with sound field testing and is minimally affected by participant position and movement (Glista, Easwar, Purcell, & Scollie, 2012). The microphone in this program was activated in omnidirectional mode during hearing aid verification and de-activated prior to testing. The volume control and other digital signal processing features in the hearing aid such as noise reduction were de-activated. Real-ear hearing aid verification was carried out using the Speechmap feature of Audioscan RM500SL hearing aid analyzer (Audioscan, Dorchester, ON, Canada) in a sound booth for all participants. Real-ear to coupler difference was measured in the test ear using a RE770 transducer coupled to the foam tip used for audiometry. The output of the chosen hearing aid was verified to meet Desired Sensation Level (DSL) v5a targets for adults (Scollie et al., 2005) at input levels of 50 and 65 dB SPL for the ISTS. Hearing aid gain was adjusted using programming software Truefit (version 2.1.1) to achieve hearing aid output within ±5 dB of targets between 250 and 6000 Hz. The error, computed as the RMS difference between the prescribed target and measured hearing aid output from 250 to 6000 Hz, was ≤5 dB for all participants except one with 6.2 dB error. Fit-to-targets achieved is comparable to estimates from a previous study (Polonenko et al., 2010), and hence were considered an acceptable match to prescribed targets.

The Quantum S and HP, and 12 additional hearing aids were electroacoustically verified to ensure that the stimulus modifications (e.g., modulation of fricatives) did not affect hearing aid function. The original unmodified stimulus sequence, the processed stimulus sequence (the EFR stimulus), and the ISTS were presented at input levels of 55, 65 and 75 dB SPL. RMS levels of each carrier were compared between the original and modified stimulus sequence for each hearing aid. The differences between aided RMS levels of individual carriers in the original and modified stimulus sequence were within ±2 dB for all hearing aids across all three levels. Aided spectra of individual carriers were also compared with the aided dynamic range of the ISTS for each hearing aid at each level. A few carrier spectra exceeded the 99^{th} percentile. Ninety-five percent of the exceedance values fell within 2 dB of the 99^{th} percentile for one-third octave bands centered between 125 and 10000 Hz. We therefore conclude that stimulus modifications adopted for optimizing EFR stimuli do not interact significantly with hearing aid function across varying brands of hearing aids. Further details are provided in Appendix E.

Stimulus presentation and response recording

Stimulus presentation was controlled by software developed using LabVIEW (Version 8.5; National Instruments, Austin, TX, USA). Digital to analog conversion of the stimulus and analog to digital conversion of the electroencephalogram (EEG) were carried out using a National Instruments PCI-6289 M-series acquisition card. The stimulus was presented at 32000 samples per second with 16-bit resolution and the responses were recorded at 8000 samples per second with 18-bit resolution. The stimulus was presented using an Etymotic ER-2 insert earphone coupled to a foam tip of appropriate size in unaided test conditions, and through the DAI program of hearing aids in aided conditions.

EEG was recorded using three disposable Medi-Trace Ag/AgCl electrodes. The non-inverting electrode was placed at the vertex, the inverting electrode was placed at the posterior mid-line of the neck, just below the hairline, and the ground was placed on one of the collarbones. Electrode impedances, measured using an F-EZM5 GRASS impedance meter at 30 Hz, were maintained below 5 kΩ, with inter-electrode differences under 2 kΩ. A Grass LP511 EEG amplifier band-pass filtered the input EEG between 3 and 3000 Hz and applied a gain of 50000. The PCI-6289 card applied an additional gain of two making the total gain 100000.

Test conditions

The stimulus was presented at overall levels of 50 and 65 dB SPL for 300 sweeps in unaided and aided test conditions. The recording time for one condition was 20.5 minutes. Unaided and aided recordings were completed on separate days. The stimulus level was calibrated in an ear simulator (Type 4157) using a Brüel and Kjær Type 2250 sound level meter. The level was measured in flat weighted L_{eq} while the stimulus was presented for 30 seconds. For calibration purposes, the hearing aids were programmed to have zero gain across their entire bandwidth and the signals were routed through DAI. The hearing aid was coupled to an ear mold simulator via 25 mm of size 13 tubing. Frequency responses of the transducers employed in the two sound booths where data was collected were within ±3 dB of each other between 100 and 6300 Hz. During test sessions, the zero gain programming of hearing aids was replaced by the hearing aid setting prescribed for each participant.

EFR recordings were completed in an electromagnetically shielded sound booth. Participants were seated in a reclined chair and a rolled towel was placed under their neck to help reduce neck tension. A blanket was offered for comfort. Lights were switched off and participants were encouraged to sleep during the recording.

EFR analysis and detection

Response analysis was carried out offline using MATLAB and is similar to the analysis methods used in Chapter 6. Each sweep was divided into four epochs of ∼1.5 s each. For each epoch, a noise metric was computed as the average EEG amplitude between ∼80 to 120 Hz. Any epochs that exceeded the mean noise metric plus two standard deviations were rejected prior to averaging. The analysis windows for EFRs were chosen such that the ramp-in and ramp-out sections at the beginning and end of each phoneme were excluded. EFRs to vowels and fricatives from the 300 recorded sweeps were analyzed by averaging the two polarities together (i.e., +− average/vector averaging of responses elicited by opposite polarities; Aiken & Picton, 2008).

EFRs were analyzed using a Fourier Analyzer (FA; Choi et al., 2013). Tracks of f_0 showing changes in f_0 were obtained for the analysis windows of vowel carriers (Choi et al., 2013). For the FA, reference cosine and sine sinusoids were generated using the instantaneous f_0 frequency in the f_0 track. An estimate of brainstem processing delay of 10 ms was used to correct for response delay in unaided conditions (Aiken & Picton, 2008; Choi et al., 2013), and an additional hearing aid delay of 6.8 ms was used to correct for response delay in aided conditions. The delay-corrected EEG was multiplied with the reference sinusoids to create real and imaginary components of the EFR, which were further averaged to a single complex number. Each complex number provided an estimate of EFR amplitude and phase. To determine if EFRs elicited by vowel carriers were detected, the background EEG noise was estimated using 14 frequency tracks surrounding (six below and eight above) the response tracks at f_0 and f_0–8 Hz. For an EFR to be detected, the ratio of the EFR amplitude at the response frequency to the average noise estimate had to exceed the critical F ratio (2, 28 degrees of freedom) of 1.82 at an α of 0.05.

EFRs to fricatives were estimated using a Discrete Fourier Transform (DFT). Additionally, responses to the two iterations of /s/ were averaged. Noise estimates were obtained from six noise bins for /ʃ/, and eight noise bins for /s/, distributed equally on either side of the bin containing the modulation frequency (93.02 Hz). The number of noise bins varied for /ʃ/ and /s/ due to the difference in their durations and its impact on DFT frequency resolution. Consequently, the critical F ratio for an EFR to be detected was 1.97 and 1.91 for /ʃ/ and /s/, respectively. Response amplitude estimates used for F-tests were corrected for possible over-estimation of response amplitude due to noise (see Appendix of Picton et al., 2005).

Artifact check

A stimulus artifact check for the ER-2 was completed using a no-stimulus-to-the-ear run in 10 participants with normal hearing (Chapter 6). The placement of the transducer and recording electrode leads during these runs were similar to the main experiments. The stimulus was routed to a Zwislocki coupler placed on each participant's chest. The stimulus was presented at 65 dB SPL. The recording time per participant was 20.5 minutes. The false positive rate (rate of significant detections) was 3.75%, close to the assumed α of 5%.

To evaluate the possibility of stimulus artifacts with the hearing aid, a Quantum HP was programmed for one of the participants. Electrodes were suspended in water where impedances were approximately 1.7 kΩ and the stimulus was routed to an ear simulator (Type 4157) via an ear mold simulator and 25 mm of size 13 tubing. The positions of the hearing aid, electrodes and their leads were similar to EFR recordings when a participant was present. This check would reveal stimulus artifacts arising mainly from electromagnetic leakage from the hearing aid during stimulus presentation, if present (Choi et al., 2013). The false positive rate was 3%, also close to the assumed α of 5% during response analysis.

Estimation of sensation level (SL)

Audibility of EFR carriers varied across participants and conditions and therefore, Sensation Levels (SL) of carriers were estimated for each participant in unaided and aided conditions at each test level. The methods followed for threshold estimation and estimation of SL in unaided conditions were identical to the methods used in Chapter 6.

Behavioral threshold

Behavioral threshold of each EFR carrier was obtained individually using an ER-2 insert earphone. F1 and F2+ carriers of each vowel and the fricatives were concatenated without gaps to form one-minute long tracks for each carrier. Stimulus presentation was controlled using the Child Amplification Lab experimental system. The stimuli were routed through an audiometer and the audiometer dial was used to seek thresholds using a 5-dB step size. Threshold was recorded as the level at which a minimum of two positive responses in three ascending runs were obtained.

Estimation of stimulus presentation level

The EFR stimulus sequence was presented through the ER-2 at a known audiometric dial level in a looped manner without gaps as in the experiment. The output of the ER-2 was recorded for 30 seconds using the sound level meter at a sampling rate of 48000 Hz and exported in the form of a wav file. The wav file was down-sampled to 44100 Hz in Praat and the eight individual EFR carriers were extracted. The RMS level of each carrier was obtained using Spectraplus software that was calibrated using a calibration tone of known level recorded by the sound level meter. The level of each EFR carrier was extrapolated to presentation levels of 50 and 65 dB SPL. The level of /s/ was taken as the average of the two iterations of /s/, which were similar. Based on the audiometric dial reading obtained at threshold, the RMS level of each carrier at threshold was extrapolated. The SL was computed as the difference in RMS level at presentation level and at threshold.

For aided SLs, the output of each participant's hearing aid was recorded for input levels of 50 and 65 dB SPL using the sound level meter (as used for calibration). The RMS levels were obtained from the recorded wav files as described in the paragraph above. Aided SLs were computed as the difference in RMS level at aided presentation level, and at threshold obtained using the ER-2.

Response detection time

To evaluate the time required to obtain a detection, responses were analyzed in 50-sweep increments starting with 50 sweeps. To correct for inflation of false positive rates with multiple testing, adjusted Bonferroni corrections were used to account for the correlation between average sweeps (employed by the F-test) at test intervals in a running average (Choi, Purcell, & John, 2011). The adjusted critical p values for detection were 0.05 at 50 sweeps, 0.042 at 100 sweeps, 0.038 at 150 sweeps, 0.036 at 200 sweeps, 0.034 at 250 sweeps, and 0.032 at 300 sweeps. For a given carrier, detection time was defined as the least number of sweeps at these test intervals with a significant detection. This was completed only for aided conditions as the number of EFR detections were too few in the unaided condition to draw inferences.

Response detection time was further differentiated in two ways. In the first method, response detection time (in sweeps) was multiplied by the sweep duration (= 4.1045 s) to reflect the testing time necessary with the stimulus /susaʃi/. In the second method (henceforth referred to as carrier recording time), the response detection time of a carrier (in sweeps) was multiplied by the duration of the corresponding analysis window and number of iterations within one sweep. The latter method would indicate the effective stimulus time necessary for a significant detection of each carrier.

Statistical analysis

Statistical analyses in this study were completed using SPSS (version 21; IBM, Armonk, NY, USA). In Experiments I and II, we measured the effect of audibility in terms of the number of EFRs detected (maximum of eight per condition) and EFR amplitude. To evaluate the effect of amplification and level on the number of EFRs detected, a two-way Repeated Measures Analysis of Variance (RM-ANOVA) was completed with amplification (two levels: unaided and aided) and stimulus level (two levels: 50 and 65 dB SPL) as the two within-subject factors and number of EFR detections as the dependent measure. To evaluate the effect of amplification and level on response amplitude, a three-way RM-ANOVA with amplification (two levels: unaided and aided), stimulus level (two levels: 50 and 65 dB SPL) and carrier (eight levels: /u/ F1, /a/ F1, /i/ F1, /u/ F2+, /a/ F2+, /i/ F2+, /ʃ/ and

/s/) as the within-subject factors and response amplitude as the dependent measure. For all statistical analyses completed in this study, Greenhouse-Geisser corrected degrees of freedom were used for interpretation of significant effects. Results of ANOVA were interpreted at an α of 0.05. For post-hoc analyses, paired t-tests were carried out with False Discovery Rate method (FDR; Benjamini & Hochberg, 1995) for multiple comparisons.

As expected, EFRs were not detected in many conditions that had limited audibility, such as the unaided condition at 50 dB SPL. With the repeated measures design used in this study, use of response amplitudes only from significant detections would have resulted in a substantial reduction in sample size for statistical analyses. Use of response amplitudes, irrespective of a significant detection, therefore was chosen for analyses, so as to include low-amplitude values at response frequencies and facilitate assessment of growth in response amplitude due to change in audibility.

Data exclusion

Since we used response amplitudes estimated by the FA/DFT irrespective of a significant detection, a noise criterion was applied to exclude those participants who exhibited high residual noise estimates within each experiment. Differences in noise estimates across conditions could influence differences in response estimates in conditions being compared. To minimize the effect of variable noise estimates between conditions being compared within each experiment, participants with high noise estimates in >25% of conditions in Experiment I or II were excluded from the respective experiments. A test condition was considered high noise if the residual noise estimated by the FA/DFT was greater than the mean noise estimate plus two standard deviations computed across all 21 participants in each experiment. Computed in this manner, two participants in Experiment I and one in Experiment II were excluded from statistical analyses. The participants excluded from

Experiment I and II differed as the test conditions included were different in each case.

7.2.2 Results

Estimated SL across unaided and aided conditions

The mean estimated SLs of individual carriers in unaided and aided conditions at test levels of 50 and 65 dB SPL are illustrated in Figure 7-2. In unaided conditions, the average SLs of low frequency F1 carriers (subplot A) are positive and therefore indicate audibility during unaided EFR recordings. The average SLs of mid frequency carriers such as /u/ F2+ and /a/ F2+ (subplot B) indicate inaudibility at 50 dB SPL but possible audibility at 65 dB SPL. The average SLs of /i/ F2+ (subplot B) and fricatives (subplot C) indicate inaudibility during unaided EFR recordings at both 50 and 65 dB SPL. The pattern of the unaided SL distribution agrees with the sloping audiometric configurations of most participants (see Figure 7-1). In aided conditions, the average estimated SLs of all carriers are positive suggesting audibility with the use of a hearing aid.

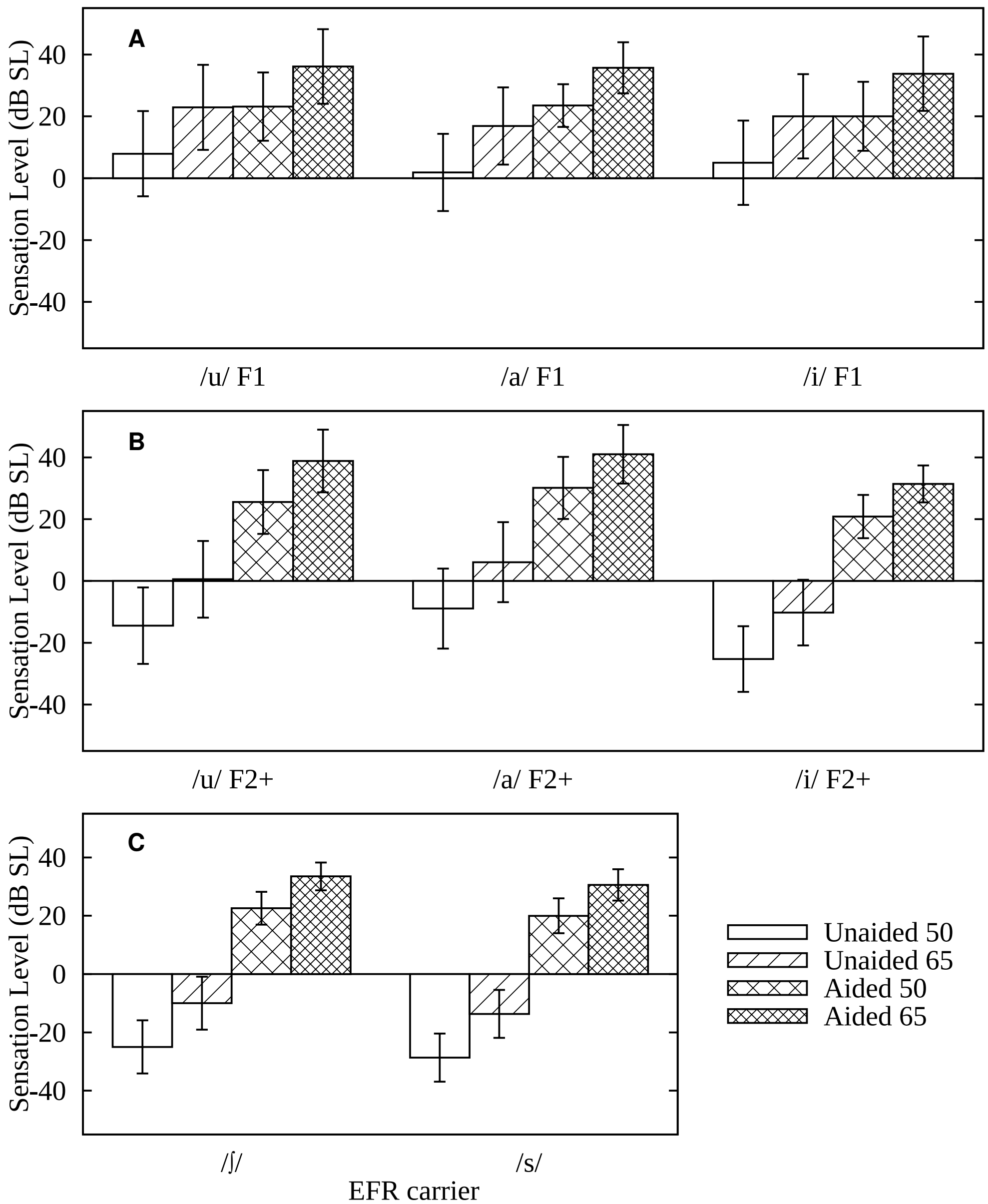

Figure 7-2: Estimated SL of carriers in unaided and aided conditions at 50 and 65 dB SPL (n = 19). F1 carriers, F2+ carriers and fricatives are represented in subplots A, B and C respectively. Error bars represent ±1 *SD*.

Effect of SL on response amplitude of detected EFRs

Figure 7-3 illustrates the spread of response amplitudes of detected EFRs across estimated SLs from all 21 participants. The following observations can be made: One, the number of detections is higher in aided conditions (right panel) compared to unaided conditions (left panel). Two, responses cluster at higher SLs in the aided condition (right panel) compared to unaided conditions (left panel), consistent with the increase in SL of carriers with use of amplification. Three, response amplitudes tend to be higher in aided conditions (right panel) compared to unaided conditions (left panel) for the majority of EFR carriers. Four, a few false positives are evident in the unaided conditions. False positive rate, estimated by the proportion of significant detections at negative estimated SLs, was 5%. This is consistent with results of the artifact checks described above.

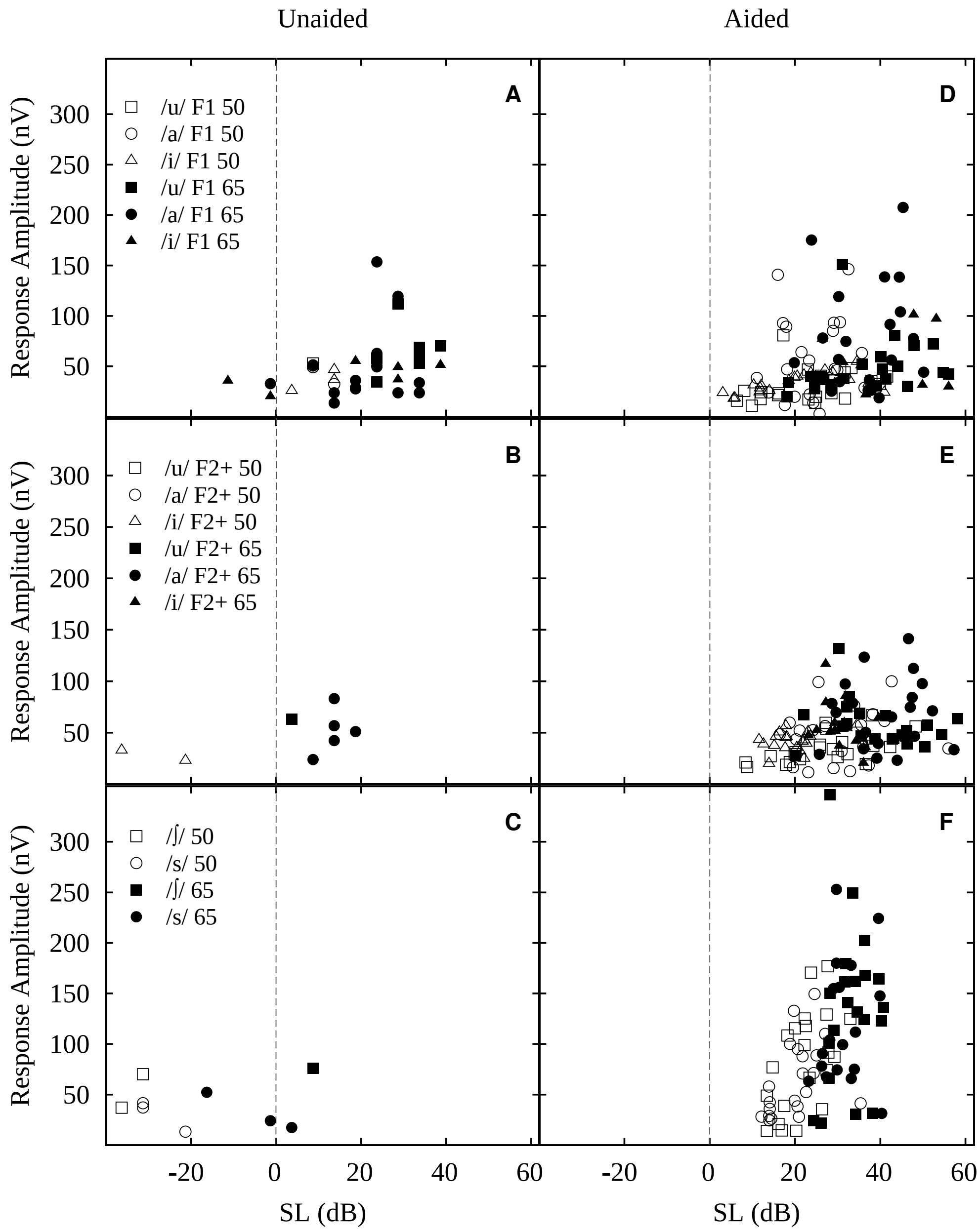

Figure 7-3: Response amplitude of detected EFRs across estimated SL of carriers (n = 21). Subplots A–C represent unaided conditions and subplots D–F represent aided conditions. Open and closed symbols represent test levels of 50 and 65 dB SPL respectively.

Effect of stimulus level and amplification on the number of EFRs detected

The average number of EFR detections (out of the eight recorded per condition) in unaided and aided conditions at 50 and 65 dB SPL are illustrated in Figure 7-4A. The number of EFRs detected are higher at 65 compared to 50 dB SPL and in aided compared to unaided conditions. The RM-ANOVA indicated a significant main effect of level ($F(1, 18) = 44.79$; $p < 0.001$; $\eta^2_{partial} = 0.71$) and a main effect of amplification ($F(1, 18) = 148.39$; $p < 0.001$; $\eta^2_{partial} = 0.89$). There was no interaction between level and amplification ($F(1, 18) = 1.56$; $p = 0.228$; $\eta^2_{partial} = 0.08$), suggesting that the effect of level was similar in unaided and aided conditions and likewise, the effect of amplification was similar at 50 and 65 dB SPL. Therefore, to assess the effect of level, a paired t-test was completed on the number of EFR detections averaged across unaided and aided conditions. On average, the number of detections increased significantly from 2.58 ($SD = 0.96$) to 3.84 ($SD = 1.38$) with the increase in test level, $t(18) = 6.69$, $p < 0.001$, 95% CI [0.87, 1.66]. To assess the effect of amplification, a paired t-test was completed on the number of EFR detections averaged across 50 and 65 dB SPL test levels. On average, the number of detections increased significantly from 1.34 ($SD = 1.08$) to 5.08 ($SD = 1.49$) with amplification, $t(18) = 12.19$, $p < 0.001$, 95% CI [3.09, 4.38].

Figures 7-4B and 7-4C illustrate the number of participants with a significant detection for each carrier in unaided and aided conditions at 50 and 65 dB SPL. The number of detections increased with the use of amplification at both test levels with fricatives showing the largest increase at both test levels.

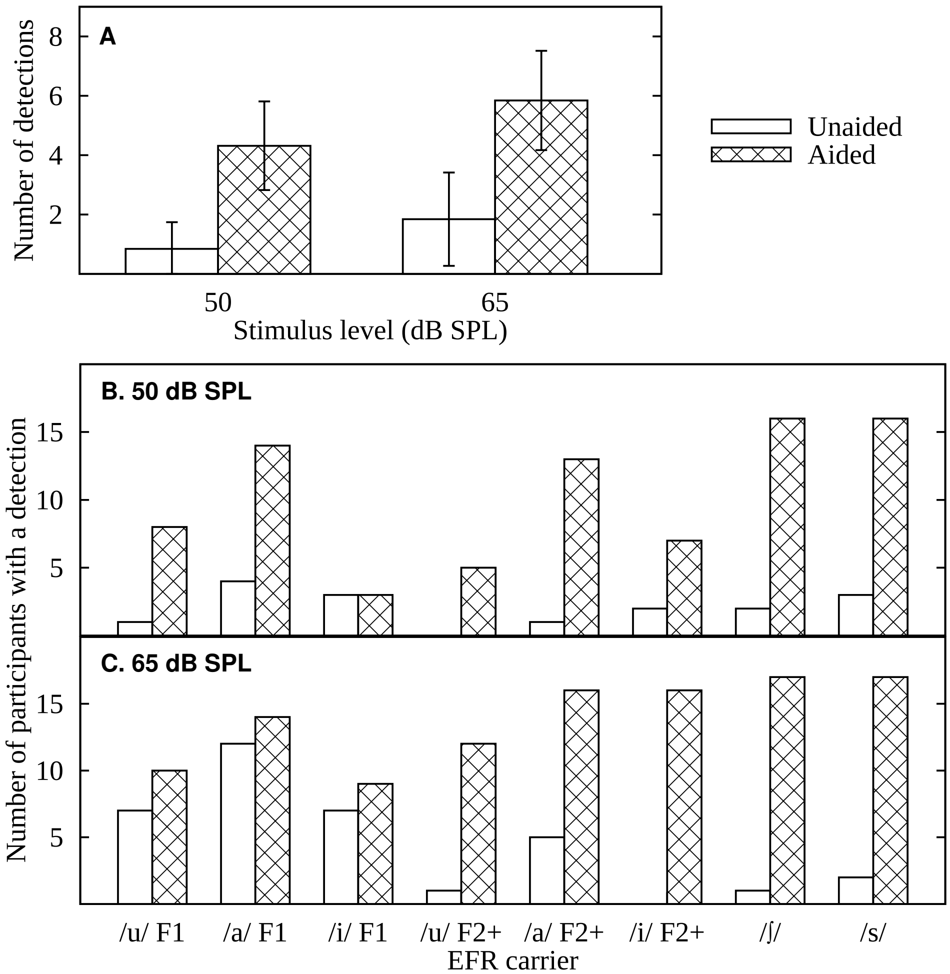

Figure 7-4: Subplot A illustrates the average number of EFRs detected, out of the eight recorded per condition in each participant. Error bars represent ±1 *SD*. Subplots B and C demonstrate the number of participants with a detection for each carrier at test levels of 50 and 65 dB SPL respectively. (n = 19)

Effect of stimulus level and amplification on response amplitude

Figure 7-5 illustrates the effect of stimulus level and amplification on response amplitude. The RM-ANOVA indicated significant main effects of all three within-subject factors: amplification ($F(1, 18) = 81.85$; $p< 0.001$; $\eta^2_{partial} = 0.82$), level ($F(1, 18) = 55.25$; $p < 0.001$; $\eta^2_{partial} = 0.75$), and carrier ($F(3.38, 60.77) = 17.64$; $p < 0.001$; $\eta^2_{partial} = 0.49$). The 2-way interaction between amplification and level ($F(1,\, 18) = 18.11$; $p < 0.001$; $\eta^2_{partial} = 0.50$), and the 2-way interaction between amplification and carrier ($F(3.22,\, 58.08) = 22.60$; $p < 0.001$; $\eta^2_{partial} = 0.56$) were significant. The interaction between level and carrier was not significant ($F(4.11,\, 74.09) = 1.07$; $p = 0.380$; $\eta^2_{partial} = 0.06$). The 3-way interaction between level, amplification and carrier was also significant ($F(4.49,\, 80.81) = 6.12$; $p < 0.001$; $\eta^2_{partial} = 0.25$). Paired t-tests were completed to assess the effect of level per carrier in unaided and aided conditions, as well as the effect of amplification per carrier at 50 and 65 dB SPL.

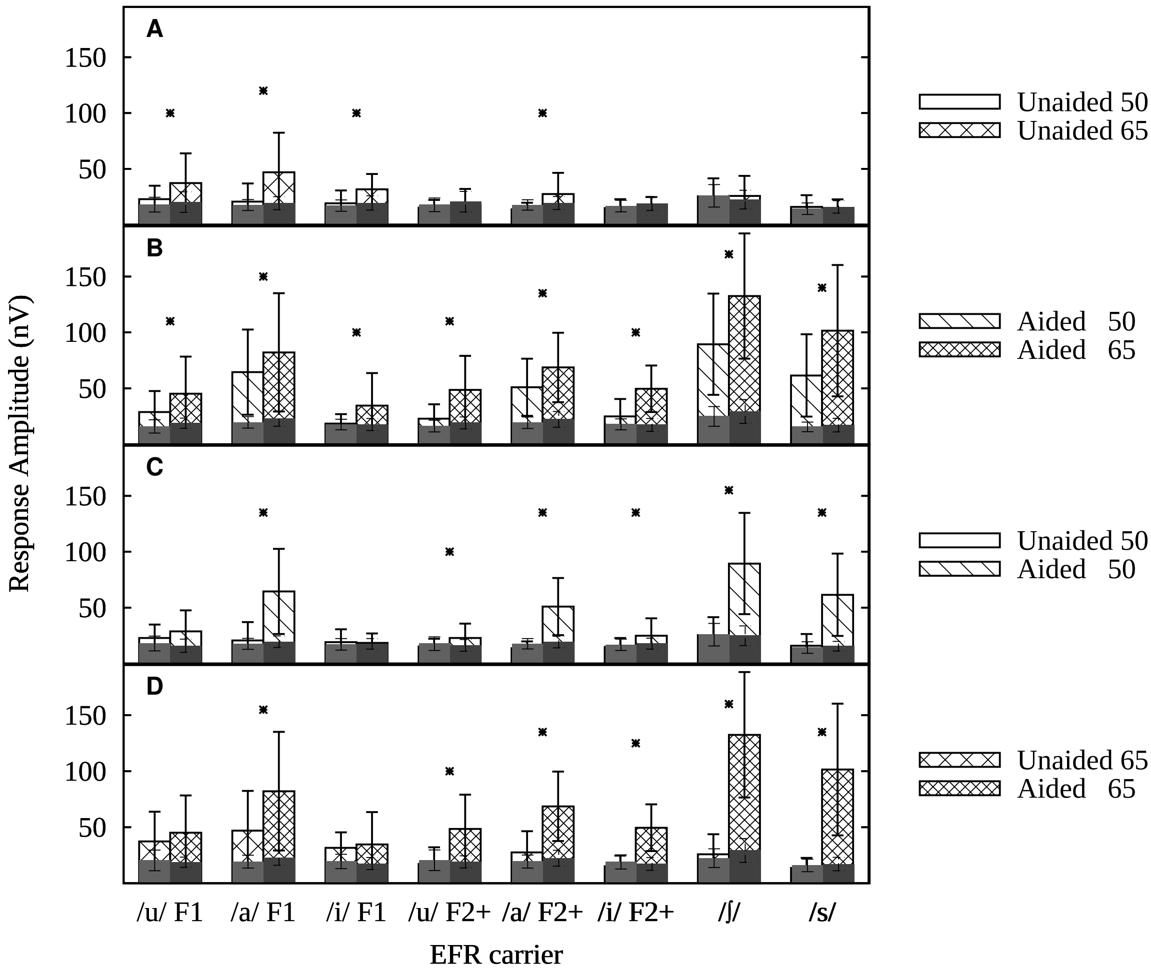

Figure 7-5: Effect of stimulus level and amplification on EFR amplitude. Subplots A and B illustrate the effect of level in unaided and aided conditions, respectively. Subplots C and D illustrate the effect of amplification at 50 and 65 dB SPL, respectively. Data presented in subplots C and D are identical to data presented in subplots A and B, but re-arranged to illustrate the effect of amplification. Error bars represent ± 1 *SD*. Solid grey bars represent average noise estimates. * indicates a significant difference. (n = 19)

Effect of level on response amplitude:

Figures 7-5A & 7-5B illustrate the mean response amplitudes at 50 and 65 dB SPL for each carrier in unaided and aided conditions respectively. In unaided conditions, the response amplitudes showed a significant increase with increase in level for the carriers /u/ F1, /a/ F1, /i/ F1 and /a/ F2+. The differing effects of level on carriers are likely due to the change in estimated SL with the increase in stimulus level (Figure 7-2). F1 carriers and /a/ F2+ carrier increase in suprathreshold SL whereas, SLs of the remaining carriers are negative at both test levels and therefore effectively demonstrate no change in audibility with the increase in stimulus level.

In aided conditions, all eight carriers demonstrated a significant increase in response amplitude, likely because the estimated SL for all carriers were suprathreshold. The rate of change in response amplitude with change in level (the slope in nV/dB) was computed in aided conditions. The slope was calculated as the average change in response amplitude of detected EFRs divided by the corresponding change in aided SL across input levels. The slopes were averaged across carriers: low (three F1 carriers; average F1 frequency: 409 Hz), mid (/u/ and /a/ F2+; average F2+ frequency: 1241 Hz), and high frequency carriers (/i/ F2+, /ʃ/ and /s/; peak frequencies >2 kHz). The slopes obtained were 3.32, 2.14 and 3.44 nV/dB for low, mid and high frequency carriers, respectively.

Effect of amplification on response amplitude:

Figures 7-5C & 7-5D illustrate mean response amplitudes in unaided and aided conditions for each carrier at 50 and 65 dB SPL respectively. At both test levels, response amplitudes for all carriers except /u/ and /i/ F1 demonstrated a significant increase in response amplitude with amplification.

The average change in response amplitude between unaided and aided EFRs plotted against the average change in estimated SL per carrier with amplification is illustrated in Figure 7-6. Change in SL was calculated as the difference between aided and unaided estimated SL, with negative SL limited to zero. The overall positive trend between change in SL and change in response amplitude suggests that the change in response amplitude was greater when the change in SL was higher. At both test levels, fricatives show larger changes in average response amplitude relative to the vowel F2+ carriers at similar changes in SL.

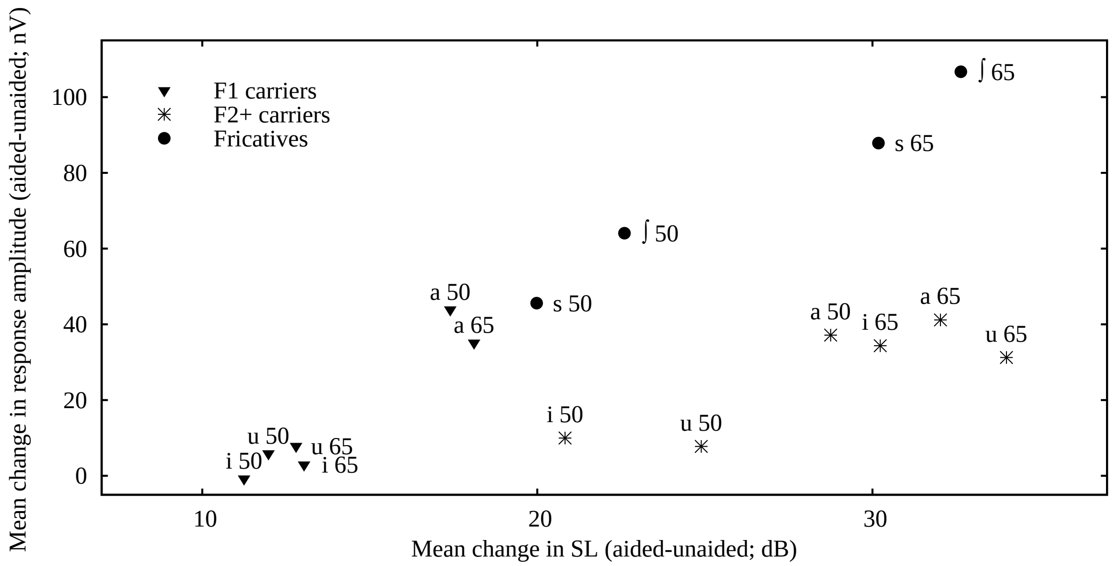

Figure 7-6: Average increase in response amplitude plotted as a function of the average increase in SL with amplification (unaided negative SL limited to zero; n = 19).

Response detection time

Response detection times for each carrier, averaged across 19 participants in the aided condition, are illustrated in Figure 7-7. The number of data points contributing to each average varied because the averages were based on detected responses only. Figure 7-7A indicates detection times in terms of sweep length and Figure 7-7B indicates detection times in terms of carrier recording time. Figure 7-7A shows that mean detection times for the individual carriers used within the stimulus /susaʃi/ range from approximately four to 15 minutes. Figure 7-7B shows that mean carrier recording times range from under one minute to approximately three minutes. EFRs elicited by the carrier /ʃ/ required the shortest recording time for a significant detection.

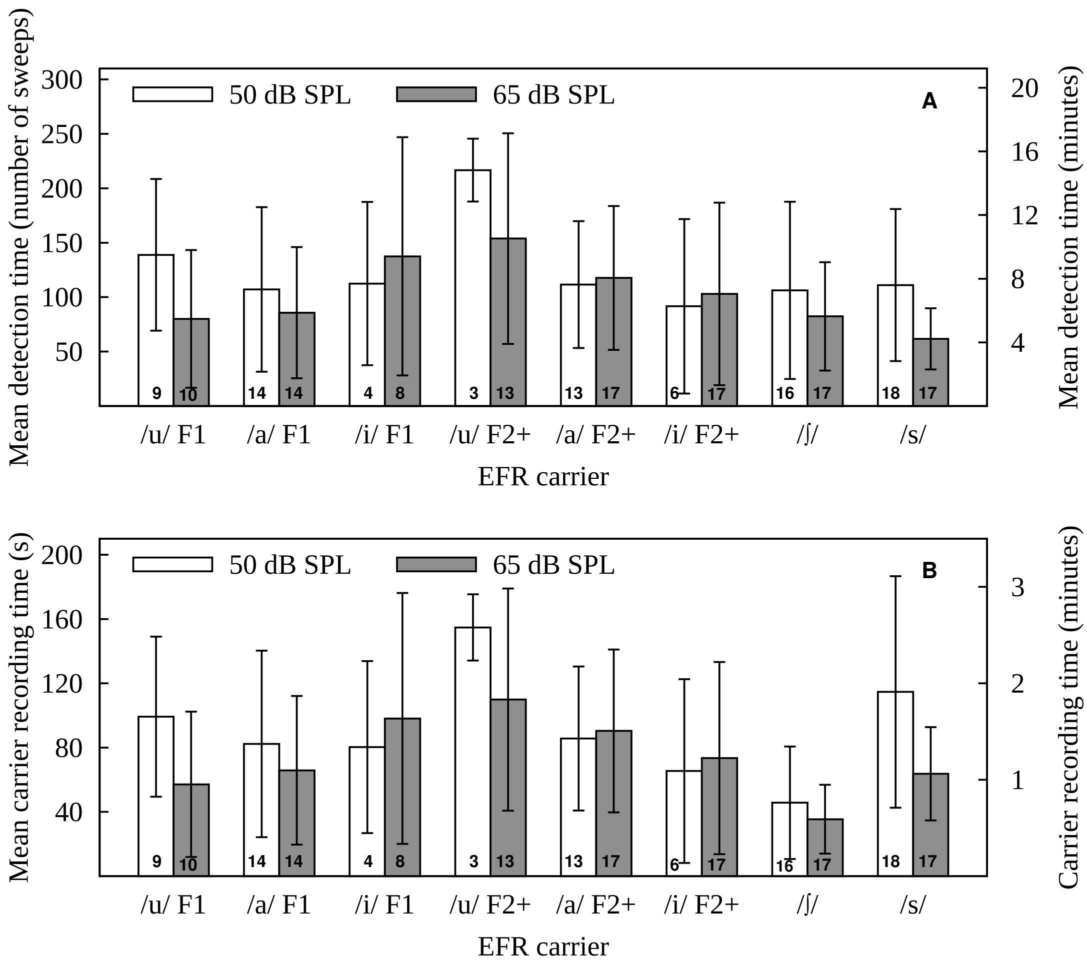

Figure 7-7: Subplot A illustrates response detection times in sweeps (primary y-axis) and minutes (secondary y-axis) in aided conditions for participants who had a significant detection within the available 300 recorded sweeps. Detection time in minutes was derived by multiplying the number of sweeps required for a detection by the sweep duration (= 4.1045 s), and dividing by 60. Subplot B shows the average carrier recording time required for a significant detection. Detection time was obtained by multiplying the number of sweeps required for a detection by carrier-specific analysis window lengths. The number of responses/participants contributing to each average is indicated within each bar. Error bars represent ±1 *SD*. (n = 19)

Response detectability and response detection patterns

Since the peak frequencies and frequency ranges of carriers overlap, we also examined the detectability of low, mid and high frequency regions by pooling across a few carriers per frequency band as a potential alternative scoring rule for clinical applications. Detection of a minimum of one carrier per frequency region indicates neural representation of amplified sound in the respective frequency band. The three F1 carriers were grouped to represent low frequencies, the two F2+ carriers /a/ and /u/ were grouped to represent mid frequencies, and F2+ of /i/ and fricatives were grouped to represent high frequencies. If any one of the two or three EFRs within a frequency band was detected, we inferred that the frequency band was represented at the level of the brainstem. Since only one of the two/three EFRs per frequency band had to be detected, Bonferroni-corrected p values were used. A critical p value of 0.017 was used for the low and high frequency bands with three carriers each and a critical p value of 0.025 for the mid frequency band with two carriers.

Figures 7-8A & 7-8B illustrate the number of participants with a minimum of one detection per frequency band in unaided and aided conditions, respectively. The detectability of F2+ carriers and fricatives are low in unaided conditions but increase substantially in aided conditions. The stacked histograms in Figures 7-8C & 7-8D illustrate the number of participants with a detection in none/one/two or all three frequency bands in unaided and aided conditions respectively. When aided, more than half the participants had a detection in all three frequency bands and more than three-quarters had a detection in a minimum of two frequency bands.

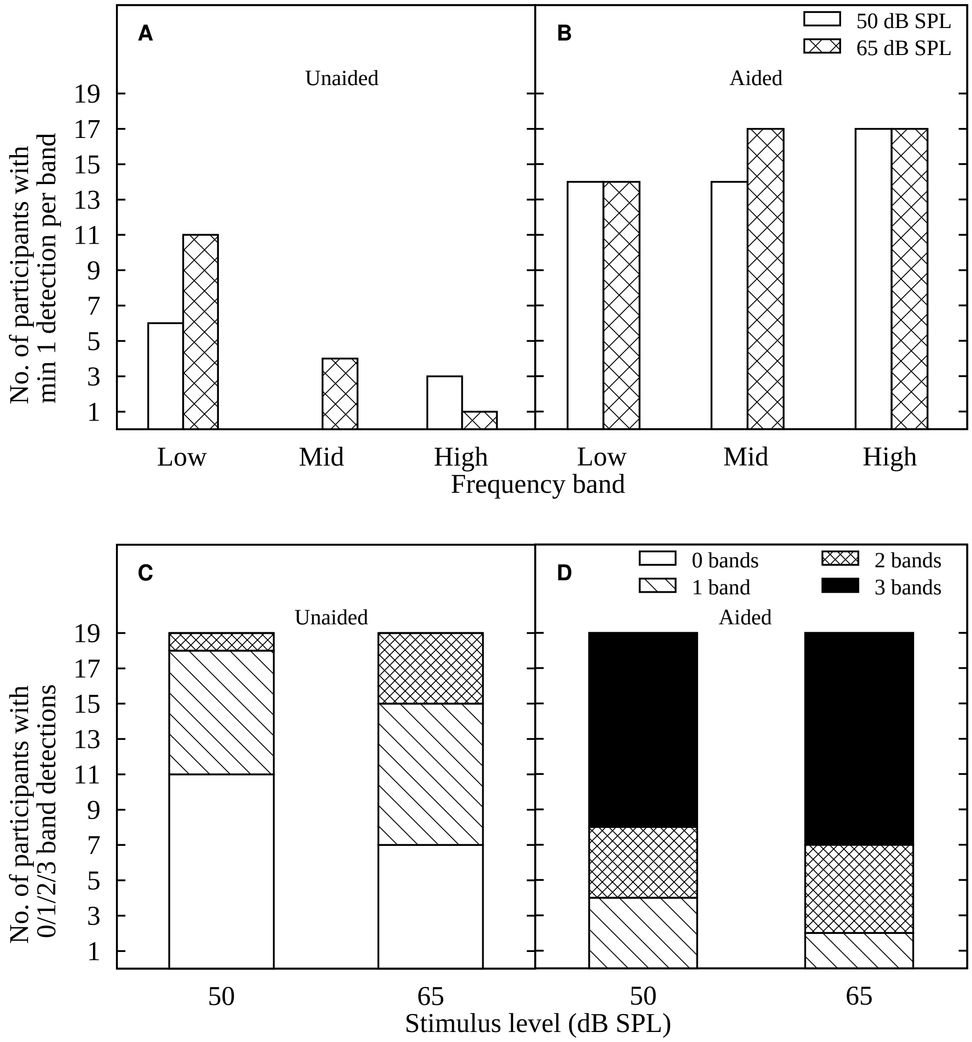

Figure 7-8: Response detection patterns in unaided and aided conditions at 50 and 65 dB test levels. Subplots A and B show the number of participants with a minimum of one significant detection per frequency band. The stacked histograms in subplots C and D represent the number of participants with at least one detection in none, 1, 2 or 3 frequency bands. (n = 19)

7.3 Experiment II: Effect of bandwidth on speech-evoked EFRs and behavioral measures

7.3.1 Method

Participants

All participants who took part in Experiment I also participated in Experiment II. In this experiment, EFR and behavioral tests were completed in aided conditions, using individually verified hearing aid fittings used in Experiment I.

EFR protocol

Aided EFR recordings were obtained in four bandwidth conditions: low-pass filtered 1 kHz (LPF1k), low-pass filtered 2 kHz (LPF2k), low-pass filtered 4 kHz (LPF4k) and full bandwidth (FBW). The 65 dB SPL aided condition from Experiment I represented the FBW condition. The stimulus /susaʃi/ was low-pass filtered at 1, 2 and 4 kHz using Praat. The attenuator settings for the low-pass filtered conditions were the same as the FBW presented at 65 dB SPL, thereby maintaining matched spectral levels in their respective pass-bands. The order of test conditions was randomized across participants. Test time per condition was 20.5 minutes for all three filter conditions. EFR recording and analysis were identical to Experiment I.

Behavioral tests

Speech discrimination and sound quality rating were evaluated aided in multiple bandwidth conditions to compare the effects of bandwidth on aided EFR responses.

Speech discrimination task

The University of Western Ontario Distinctive Features Differences test (UWO DFD) was used as a measure of aided speech discrimination (Cheesman &

Jamieson, 1996). The test evaluates the accuracy of consonant identification in fixed word-medial positions using a closed set response task and has been used as an outcome measure of hearing aid fittings (e.g., Jenstad et al., 2007). A total of 42 items (21 consonant items spoken by two male talkers) was used in the study. The 42 test items in the UWO DFD test were low-pass filtered at 1, 2 and 4 kHz. The original UWO DFD stimuli represented the FBW condition. Stimulus presentation, response recording and scoring were controlled by the Child Amplification Laboratory experimental system. Once the test began, a list of consonants was displayed on the computer screen. Participants were required to click on the consonant they heard. Percent correct scores were calculated and converted to Rationalised Arcsine Units (RAUs; Sherbecoe & Studebaker, 2004) for statistical analyses.

A practice run with FBW stimuli was provided before the test began. Presentation order of test conditions and the 42 test items were randomized across participants. Stimuli were presented at a level of 65 dB SPL, calibrated using a concatenated list of the 42 FBW items (without gaps) measured in flat-weighted L_{eq} for 30 seconds. Calibration procedures were as described in Experiment I. Like the EFR stimuli, the three low-pass filtered conditions were presented at the same attenuator level as the FBW condition. The time taken to complete this task varied between $\sim$10 to 20 minutes.

Sound quality rating task

The MUltiple Stimulus Hidden Reference and Anchors (MUSHRA) paradigm was used to obtain sound quality ratings (International Telecommunication Union [ITU], 2003). The MUSHRA task uses a paired comparison paradigm to measure sound quality allowing for grading of multiple stimuli in comparison with each other (ITU, 2003; Parsa et al., 2013). The task was implemented using custom software described in Parsa et al. (2013). In this task, a reference stimuli (the best quality), multiple test stimuli that need to be rated, and an anchor (typically of poor quality) are used. The reference and the anchor stimuli act as examples for rating on either ends of the scale provided. Participants are instructed to do a pairwise comparison of each test stimulus (one of which is a hidden reference) with the reference stimulus. The participant rates each track by moving a slider along a continuous scale, arbitrarily labeled as 'bad' on one end and 'excellent' on the other. The order of the stimuli as they appear on the screen, including a copy of the reference and the anchor, was randomized for each participant.

The stimulus used was an IEEE male-talker sentence pair "Raise the sail and steer the ship northward. The cone costs five cents on Mondays" from Parsa et al. (2013). The original stimulus represented the FBW condition and served as the reference stimulus. The stimulus was also low-pass filtered at 0.5 (anchor), 1, 2 and 4 kHz. For calibration, the original (FBW) sentence was presented in a looped manner without silent gaps and the level was measured in flat-weighted L_{eq} for 30 seconds. The same attenuator setting was used for all conditions. The time taken to complete this task varied from $\sim$4 to 10 minutes.

Statistical analysis

Individual RM-ANOVAs were completed to examine the effect of the within-subject factor bandwidth condition (four levels: LPF1k, LPF2k, LPF4k and FBW) on each of the four measures, namely, the number of EFRs detected (maximum of eight), EFR composite amplitude (sum of response amplitudes across all eight carries), speech discrimination scores (in RAUs), and sound quality rating. An additional level was included for the sound quality rating (anchor, low-pass filtered at 500 Hz). Composite amplitude was computed to allow incorporation of EFR amplitude as a dependent measure within the four levels of the within-subject variable in the

ANOVA. Post-hoc analyses were completed using paired t-tests between all possible pairs of bandwidth conditions. Results of t-tests were interpreted using FDR corrections.

7.3.2 Results

Data from one participant was excluded due to high noise estimates.

Effect of stimulus bandwidth on EFRs and behavioral tests

Figure 7-9 illustrates the average trend in EFR and behavioral measures with increase in bandwidth, superimposed on individual data. Behavioral measures increase up to the FBW condition however EFR measures asymptote at LPF4k. Mean sound quality rating for the anchor (not shown in the figure) was 12.40 (SD = 9.98).

The main effect of bandwidth was significant for all four dependent variables: EFR number of detections ($F(2.65, 50.38) = 75.86$; $p < 0.001$; $\eta^2_{partial} = 0.80$), EFR composite amplitude ($F(1.71, 32.51) = 44.28$; $p < 0.001$; $\eta^2_{partial} = 0.70$), speech discrimination scores ($F(1.94, 36.92) = 129.71$; $p < 0.001$; $\eta^2_{partial} = 0.87$), and sound quality rating ($F(2.48, 47.19) = 151.91$; $p < 0.001$; $\eta^2_{partial} = 0.89$). Results of post-hoc analyses are indicated in Figure 7-9. The number of detections showed a significant increase between LPF1k and LPF4k conditions, and a marginal increase between LPF4k and FBW conditions (approaching significance; $p = 0.07$, critical p = 0.05). Likewise, composite amplitude also demonstrated a significant increase with bandwidth up to the LPF4k condition. Speech discrimination scores as well as sound quality rating increased significantly with increase in bandwidth up to the FBW condition. As expected, sound quality ratings for the anchor was significantly lower than all wider bandwidth conditions, suggesting that participants likely understood the task.

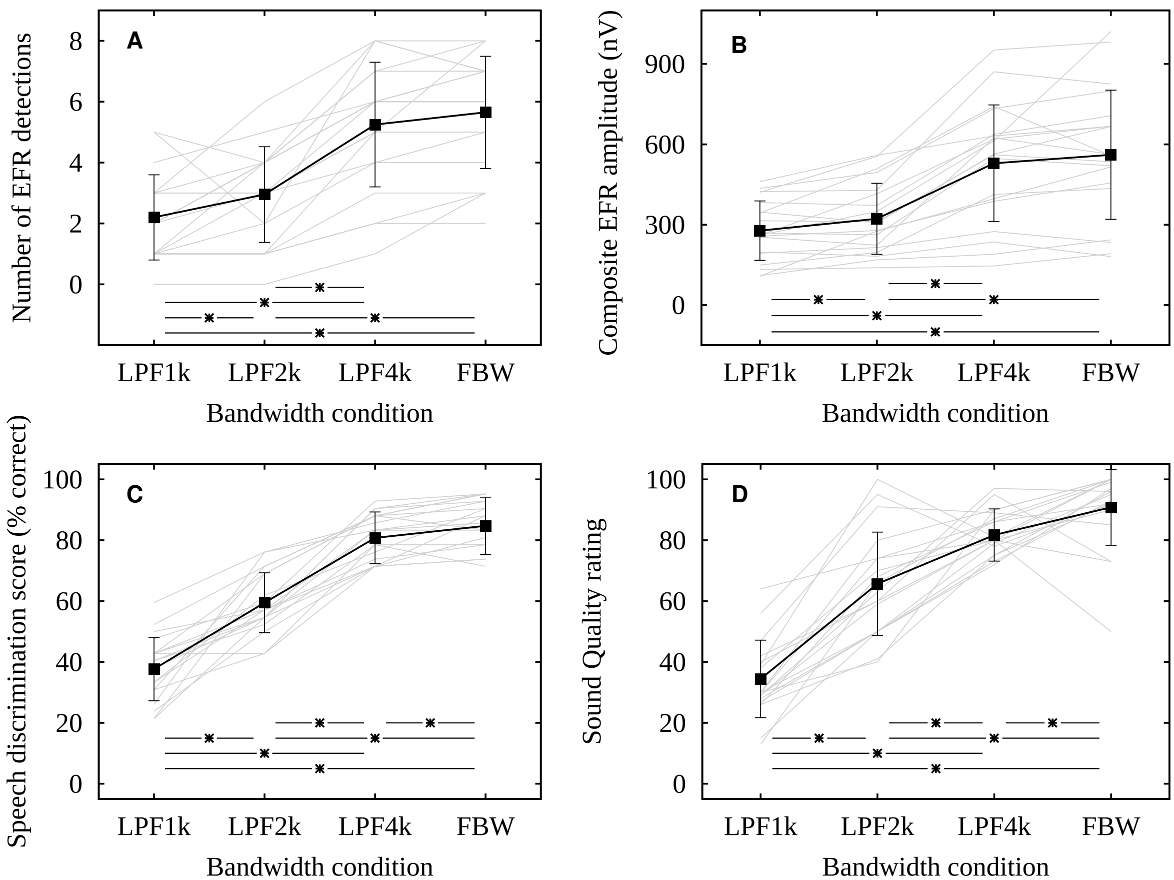

Figure 7-9: Effect of stimulus bandwidth on EFRs and behavioral measures. Subplots A, B, C and D illustrate the effect of bandwidth on the number of EFRs detected (out of the eight recorded per condition), composite response amplitude (sum of all eight EFR amplitudes), speech discrimination scores (raw scores in % correct), and sound quality rating, respectively. Mean data are represented in black and individual data are represented in grey. Error bars represent ±1 *SD*. * indicates a significant difference. (n = 20)

Effect of stimulus bandwidth on response amplitudes of individual EFR carriers

The change in composite response amplitude seen above due to bandwidth may be a result of the change in response amplitude in one or more carriers. Changes in response amplitude across bandwidth for each carrier is illustrated in Figure 7-10. To assess the effect of bandwidth condition on response amplitudes of each carrier individually, a RM-ANOVA was completed with bandwidth (four levels: LPF1k, LPF2k, LPF4k and FBW) and carrier (eight levels: /u/ F1, /a/ F1, /i/ F1, /u/ F2+, /a/ F2+, /i/ F2+, /ʃ/ and /s/) as the two within-subject factors and response amplitude as the dependent measure. Results indicated a significant main effect of bandwidth ($F(1.71, 32.51) = 44.28$; $p < 0.001$; $\eta^2_{partial} = 0.70$), a significant main effect of carrier ($F(2.85, 54.14) = 10.38$; $p < 0.001$; $\eta^2_{partial} = 0.35$), and a significant interaction between bandwidth and carrier ($F(4.17, 79.29) = 29.09$; $p < 0.001$; $\eta^2_{partial} = 0.61$), indicating a carrier-specific change due to bandwidth. Post-hoc analyses included paired t-tests comparing response amplitudes of individual carriers across all combinations of bandwidth conditions. Significantly different conditions are illustrated in Figure 7-10.

The /u/ and /a/ F1 carriers did not show a significant change in response amplitude across the different bandwidth conditions. This is consistent with the dominance of low frequency energy for both carriers. Response amplitude for /i/ F1 grows from the LPF1k to the LPF2k condition, but decreases from the LPF2k to the FBW condition despite the F1 peak of this carrier being around 300 Hz, and the low-pass filter cut-off being 1150 Hz. All F2+ carriers demonstrate an increase in response amplitude with increase in bandwidth up to FBW. However, the change between adjacent bandwidth conditions did not reach significance for all carriers. Fricatives demonstrate a significant increase between the LPF2k and the LPF4k conditions, as expected, but the increase in response amplitude between the LPF4k and FBW conditions was not significant. These results are generally suggestive of sensitivity to changes (or no changes) in aided bandwidth demonstrated at the level of individual EFR carriers.

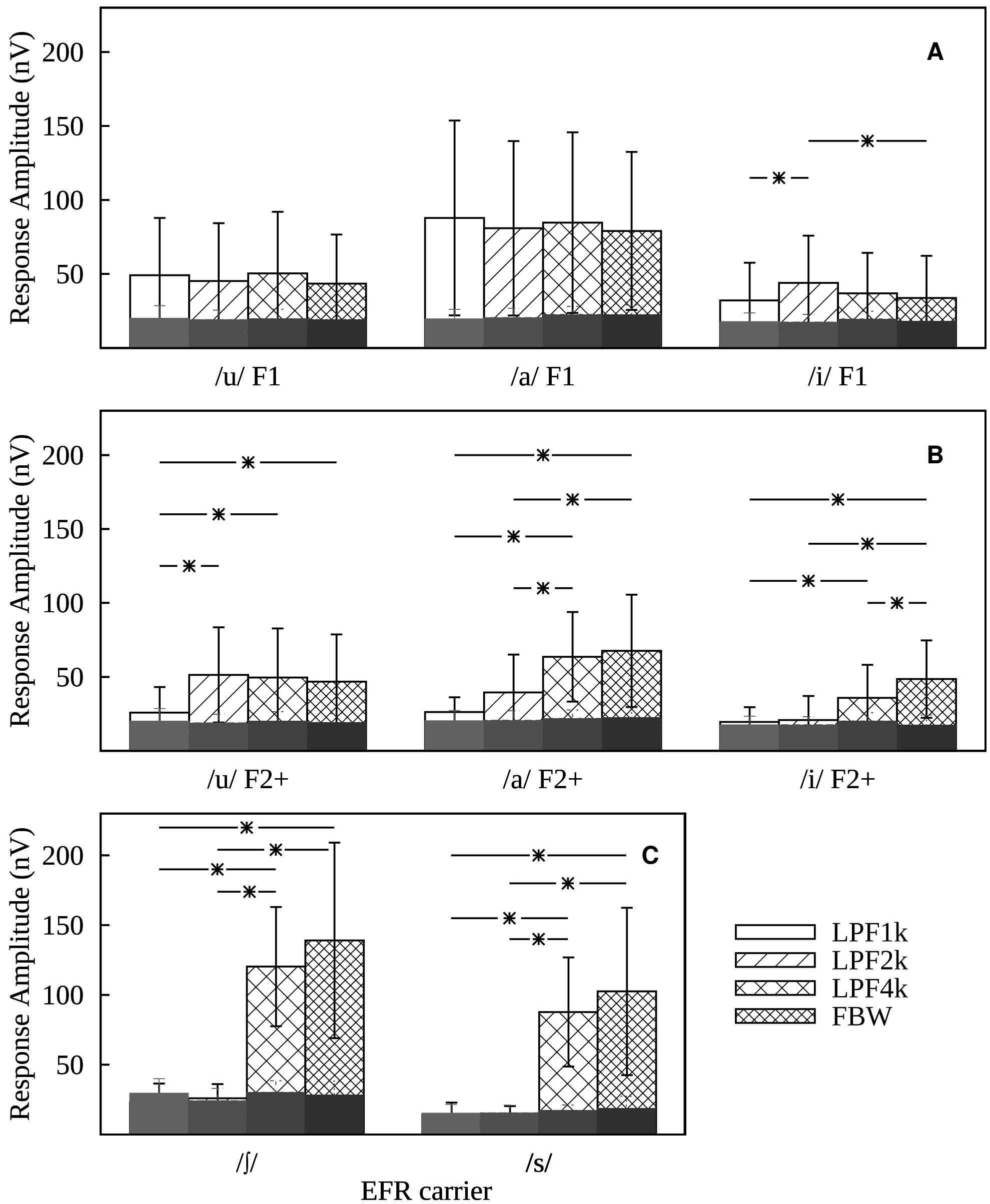

Figure 7-10: Effect of bandwidth on response amplitude for individual carriers. F1 carriers, F2+ carriers and fricatives are represented in subplots A, B and C respectively. Error bars represent ±1 *SD*. * indicates a significant difference. (n = 20)

Relationship between EFR measures and performance in behavioral tasks across bandwidth conditions

In order to describe the relation between EFR and behavioral measures, correlation analyses were performed. Figure 7-11 illustrates behavioral test scores plotted against the number of EFR detections and EFR composite amplitude of 20 participants. The overall positive trend suggests that the increase in bandwidth has a similar impact on behavioral measures of aided sounds as well as EFR measures. A significant positive correlation was found between the number of detections and speech discrimination scores (Pearson $r = 0.63$; $t(18) = 9.37$; $p < 0.001$), as well as the number of detections and sound quality rating (Pearson $r = 0.51$; $t(18) = 8.36$; $p < 0.001$). Also, a significant positive correlation was found between EFR composite amplitude and speech discrimination scores (Pearson $r = 0.56$; $t(18) = 4.76$; $p < 0.001$), and between EFR composite amplitude and sound quality rating (Pearson $r = 0.40$; $t(18) = 4.17$; $p < 0.001$). The significant positive correlations between EFR parameters and behavioral scores suggest that the EFR paradigm is sensitive to perceptually relevant changes in aided audible bandwidth. This may be classified as evidence for convergent validity of the EFR paradigm.

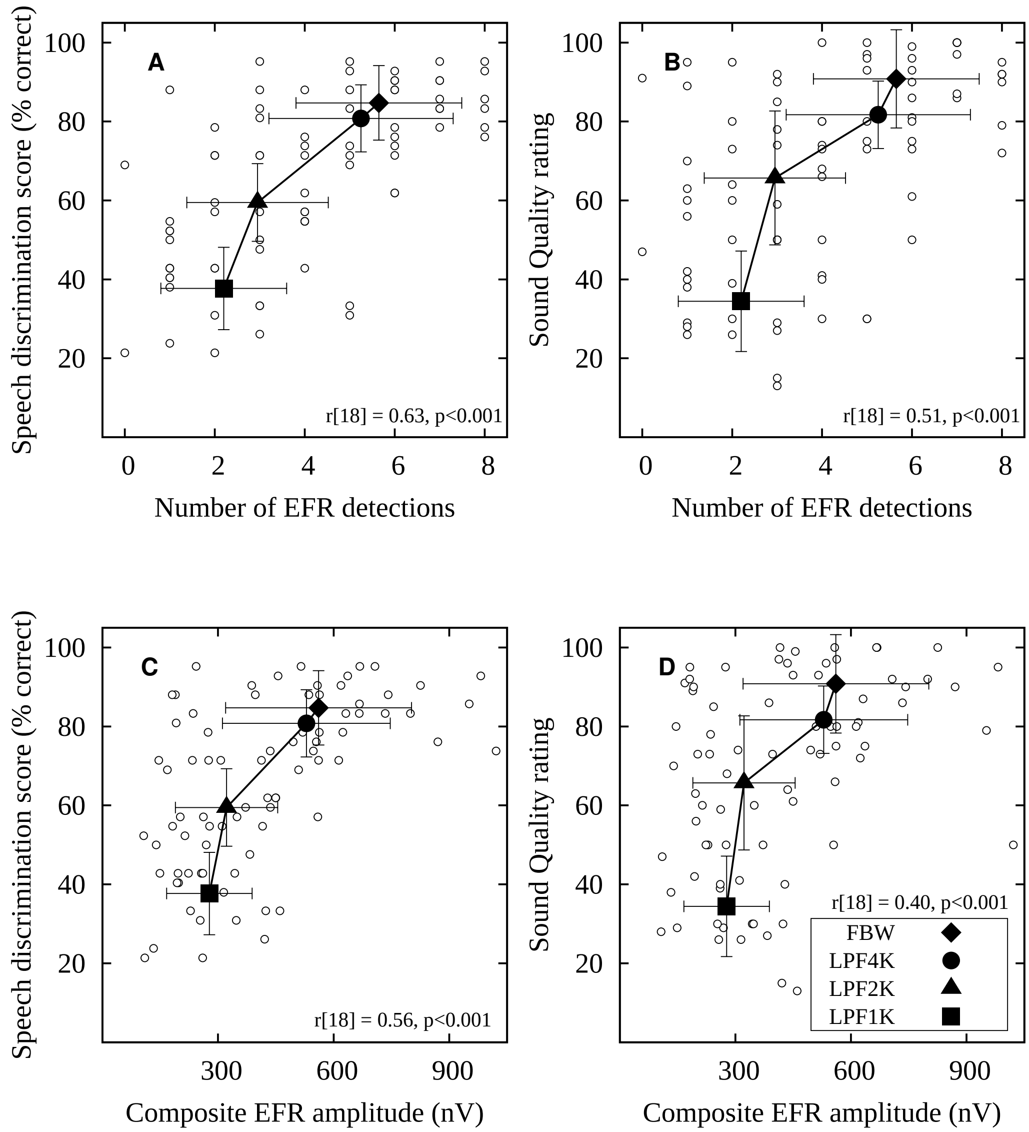

Figure 7-11: Variation in behavioral measures across EFR measures in multiple bandwidth conditions. Subplots A and B represent speech discrimination scores and sound quality rating plotted across the number of EFR detections. Subplots C and D represent the same behavioral scores plotted across composite EFR amplitude. Filled symbols represent mean data, open circles represent individual data. Error bars represent ± 1 *SD*. (n = 20)

7.4 Discussion

The aims of Experiments I and II were to assess sensitivity of the EFR paradigm to changes in audibility due to stimulus level, bandwidth and use of amplification. We measured change in the number of EFRs detected and response amplitude across multiple test conditions. In general, results of both experiments demonstrated an increase in response amplitude and a concomitant increase in the number of detected EFRs with improved audibility.

7.4.1 Experiment I: Effect of level and amplification

Number of EFRs detected

As the estimated SL of EFR stimuli increased, either by increasing test level or by adding a hearing aid, the number of EFRs detected increased. This suggests that more stimuli were represented in the brainstem at higher SLs (Figure 7-2), consistent with improved audibility. An increase in the number of responses detected with improved audibility due to amplification has also been demonstrated using tonal ASSRs (Dimitrijevic et al., 2004).

Change in EFR amplitude due to level

An increase in response amplitude with an increase in stimulus level/SL is likely due to increased neural firing rates and spread of excitation in the cochlea to involve more fibers (Moore, 2003; Picton et al., 2003; Purcell & Dajani, 2008; Sachs & Abbas, 1974). In both unaided and aided conditions, the effect of an increase in stimulus level on response amplitude was seen only for those carriers with improved audibility (Figures 7-5A, 7-5B & 7-2). This suggests that the EFR paradigm was able to reliably represent changes (or lack of changes) in audibility across carriers.

The rates of change in aided response amplitude across SL (or slope) of low and mid frequency carriers are fairly close to values obtained in adults with normal hearing (2.38, 2.70 and 1.54 nV/dB for low, mid and high frequencies respectively; Chapter 6). However, the slope obtained for high frequency carriers was much steeper than the slope obtained in adults with normal hearing, although the average amplitudes were lower in adults with hearing loss. The steeper slope may reflect physiological recruitment (Dimitrijevic et al., 2002; Picton et al., 2005), a phenomenon that refers to the steep growth in response amplitude paralleling abnormal rapid growth of loudness in individuals with sensorineural hearing loss (Picton et al., 2005). Although one of the goals of compression in hearing aids is to counteract the effects of recruitment, full normalization of the loudness function is not always observed in verified hearing aid fittings (Scollie et al., 2005).

Change in EFR amplitude due to amplification

Recall that most carriers demonstrated a significant increase in response amplitude due to amplification at both input levels, but varied in the magnitude of change (Figures 7-5C and 7-5D). The differences may be partly explained by changes in SL due to amplification (Figure 7-6). The larger changes for fricatives relative to F2+ carriers at similar SLs may be due to the nature of the responses (AM noise versus beat responses) and/or physiological recruitment (as suggested above).

The carriers that did not show a significant increase in response amplitude with amplification were /u/ and /i/ F1. Although the group average did not demonstrate a significant increase in response amplitude due to amplification, a change with amplification for these carriers was noted in a few participants. In the case of /u/ F1 at 50 dB SPL, seven of the 19 participants who did not have a detection in the unaided condition had a detection in the aided condition. Similar improvements were seen in four, two and five participants for /u/ F1 at 65, /i/ F1 at 50 and 65 dB SPL, respectively. All these participants (except one) showed an increase in response amplitude in the aided condition relative to the unaided condition. These results indicate that the EFR paradigm appears to be sensitive to changes between unaided versus aided audibility, and that changes for low-frequency carriers were not observed for all participants.

There are several possible factors that may relate to the lack of change with amplification for /u/ and /i/ F1 carriers. One, there were some participants with no significant detections in both unaided and aided conditions. These non-significant responses may have shown random fluctuations in response amplitude that, when averaged together with significant responses from the other participants, reduced the group mean change between unaided and aided conditions. The number of participants with no detections in both unaided and aided conditions was 11 and 8 for /u/ F1 at 50 and 65 dB SPL respectively, and 14 and 7 for /i/ F1 at 50 and 65 dB SPL, respectively. On a similar note, we observed some participants with a significant detection in the unaided but not in the aided condition. This detection pattern was observed in one participant for /u/ F1 at 65 dB SPL, two participants for /i/ F1 at 50 dB SPL and three participants for /i/ F1 at 65 dB SPL. These results should be interpreted considering the roles of stimulus SL and hearing aid noise floor, discussed below.

A second factor that may have contributed to the non-significant change for /u/ and /i/ F1 carriers due to amplification is the relatively small change in SL compared to the mid and high frequency carriers (Figure 7-6). This is associated with the use of clinically typical low levels of gain because participants had lesser degrees of hearing loss in this low frequency region. It is therefore not unexpected that the change between aided and unaided conditions was small for the low frequency carriers.

A third factor that may have contributed to the non-significant change with amplification is hearing aid noise floor. Studies in the aided CAEP literature on adults with normal hearing have demonstrated the sensitivity of CAEPs to change in Signal-to-Noise Ratio (SNR) rather than a change in absolute stimulus level when the stimulus is audible in both unaided and aided (real or simulated) conditions (Billings, Papesh, Penman, Baltzell, & Gallun, 2012; Billings, Tremblay, Souza, & Binns, 2007; Billings, Tremblay, Stecker, & Tolin, 2009). Therefore, although use of a hearing aid led to an increase in the absolute stimulus level relative to the participant's unaided thresholds, it is possible that audible noise floor, and consequently the reduction (or absence of change) in SNR may have affected the potential amplitude increase due to stimulus level alone. If this occurred, the functional SL during aided testing may have been lower than that estimated in this study. To investigate this factor, a measurement of the hearing aid noise floor (in DAI-only mode) was obtained using the Speech-live mode in Verifit for each participant (except one) after the hearing aid was programmed for his/her hearing loss. The measured noise floor was above or close to the pure-tone thresholds at 250 and 500 Hz for six participants, suggesting a probable audible noise floor in the frequency regions of /u/ and /i/ F1. Of these six participants, four showed changes in detection and response amplitude consistent with improved audibility for at least one carrier (/u/ or /i/ F1) at both levels. One participant showed a reduction in amplitude of 8.27 nV with amplification for /i/ F1 at 65 dB SPL, when both EFRs were significantly detected in unaided and aided conditions. Although this may reflect the effect of reduction in acoustic SNR, the magnitude of change is small. One participant had no detections in unaided and aided conditions. In summary, it appears that the likelihood of audible hearing aid noise floor had varied effects across participants.

Hearing aids in the present study were coupled with unvented molds for all participants to allow estimation of aided output levels in an ear simulator. Clinically, vented molds would likely be prescribed for participants with normal or near normal low frequency hearing thresholds to alleviate problems with the occlusion effect (Killion, 2003). The vent provides an exit pathway for low frequency sounds to escape (Dillon, 2012). Therefore, in participants with good low frequency hearing, provision of a vented mold could have reduced audibility of the hearing aid noise floor (Killion, 2003). Further research evaluating the effect of SNR using hearing aid fittings that resemble clinical fittings in individuals with hearing loss may be necessary to examine this issue further.

In summary, EFR amplitude and the number of EFRs detected were sensitive to changes in stimulus audibility. Substantial variation in SL was observed across changes in test level and the inclusion of a hearing aid. Consistent with the literature, the paradigm demonstrates SL and audibility-dependent changes in the number of EFRs detected as well as in response amplitudes of most carriers. The differential effects of audibility and amplification on F1 and F2+ carriers support the stimulus design with dual-f_0 vowels. For some participants, sensitivity of the EFR paradigm may have been limited if the stimulus had little change in SL due to amplification and/or the hearing aid noise floor counteracted the increase in absolute stimulus level. This mainly occurred for the lowest-frequency carriers /u/ and /i/ F1, which overlap with the frequency regions that may not normally receive electroacoustic amplification from the hearing aid in a clinical scenario for normal or near-normal hearing thresholds.

Response detection time

For the range of aided SLs and response amplitudes seen in the present study sample, the average detection times (Figures 7-7A & 7-7B) using the EFR paradigm are clinically feasible in duration. A direct comparison with CAEP protocols is difficult to make due to insufficient information. However, carrier recording times obtained in the present study are comparable to test times of <2 to 4 minutes *per* stimulus using CAEPs (calculated based on the number of 'accepted' sweeps and inter-stimulus interval; Carter, Dillon, Seymour, Seeto, & Van Dun, 2013; Van Dun, Carter, & Dillon, 2012). The effective testing time using the current EFR stimulus (Figure 7-7A) may still be shorter to test an equivalent number of stimuli due to multiple simultaneous carriers. Clinically feasible detections times in adults with normal hearing have been demonstrated in previous studies using speech-evoked EFRs (Aiken & Picton, 2006; Choi et al., 2013; Chapter 6). It is important to note that, due to the fixed number of sweeps, these detection times only represent EFRs that were significantly higher than the noise within 300 sweeps. As well, since we did not test under 50 sweeps, the average detection time may have over-estimated the time taken for a significant detection in some cases. The carrier recording time required to obtain a significant detection was the least for /ʃ/, likely due to higher response amplitudes in the aided conditions (Figure 7-5). The response amplitude was higher likely due to the wider bandwidth activating a larger number of neurons.

Response detectability and detection patterns

The number of participants with a significant detection is marginally improved when pooled across a few carriers with similar or overlapping frequency ranges. Pooling across carriers within a frequency band may reduce carrier-specific information but may improve test times as detection of one EFR within a frequency band may suffice to infer aided audibility for a given frequency band. Scoring rules such as these could improve clinical feasibility.

Inter-subject variability in response detectability

In comparison to adults with normal hearing (tested unaided; Chapter 6), adults in the present study had fewer significant aided EFR detections, lower response amplitudes, and increased inter-subject variability in the number of detections per condition (range: 2–7 detections at 50 dB SPL and 3–8 detections at 65 dB SPL). Lower detection rates and increased inter-subject variability in adults with hearing loss could be explained by multiple factors such as noise estimates, aided SL, and participant age.

The average noise estimate in adults with hearing loss (M = 19.41 nV, SD = 7.08) was similar to adults with normal hearing (M = 19.26 nV, SD = 6.02). However, noise estimates ranged from 7.73 to 57.03 nV in adults with hearing loss, and from 8.79 to 39.49 nV in adults with normal hearing. Since the noise estimate is the denominator of the F ratio for response detection, variability in noise estimates could increase variability in response detection. Higher average noise estimates in older adults are speculated to be due to difficulty falling asleep, possibly associated with stimulus loudness (Picton et al., 2005; Purcell, John, Schneider, & Picton, 2004). Data in the present study do not show average differences in noise estimates as large as that reported in the literature (e.g., Purcell et al., 2004), but have substantial inter-subject variability.

The second factor could be the variability in estimated aided SLs of EFR carriers, possibly due to differences in the degree and configuration of hearing losses across participants (see Figure 7-2). The third factor could be due to the effect of aging. The average age in the current study sample is 72.4 years whereas the average age of the normal hearing adults in our previous study was 23.4 years. Within the current study sample, we observed a weak trend of the effect of age on average composite response amplitude ($r(17)$ = -0.40, p = 0.088; see Figure 7-12). A decline in EFR amplitude with age at modulation frequencies around 80 Hz or higher has been previously reported (Boettcher, Poth, Mills, & Dubno, 2001; Grose, Mamo, & Hall, 2009; Leigh-Paffenroth & Fowler, 2006; Purcell et al., 2004; Skoe, Krizman, Anderson, & Kraus, 2013), and is thought to reflect reduced temporal envelope processing abilities (Grose et al., 2009; Leigh-Paffenroth & Fowler, 2006; Purcell et al., 2004). The number of neurons with best modulation frequencies above 100 Hz (Walton, Simon, & Frisina, 2002) and those participating in encoding stimulus features decrease with increasing age (Willott, Parham, & Hunter, 1988). The modulation frequencies (f_0) used in the present study ranged between 88 and 100 Hz and therefore, one of the factors influencing detection rates may be declines in response robustness due to age.

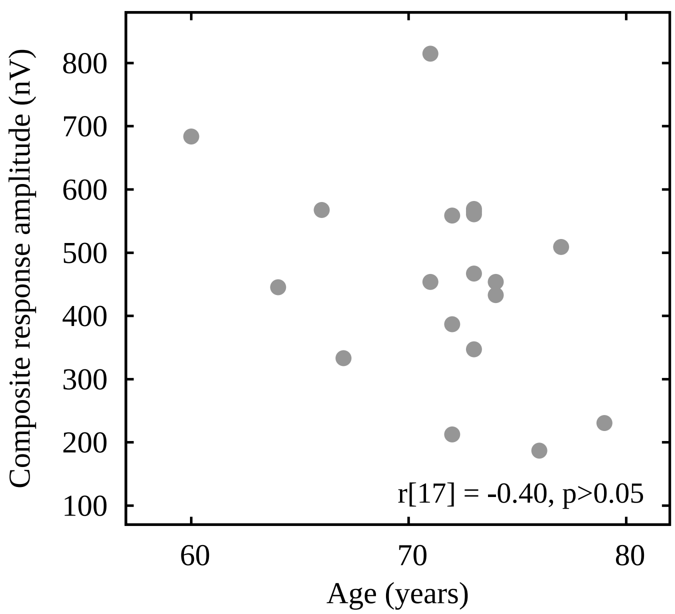

Figure 7-12: Variation in composite response amplitude across age of participants in Experiment I (n = 19).

In summary, results in Experiment I provide evidence for sensitivity of the EFR paradigm to changes in audibility either due to a change in stimulus level and/or use of amplification in clinically feasible test times. Some of the variability in response characteristics may be attributable to the current study sample and therefore further validation in other populations of interest is necessary.

7.4.2 Experiment II: Effect of bandwidth

Effect of bandwidth on speech discrimination and sound quality rating

Similar to the results seen in adults with normal hearing (Chapter 6), speech discrimination scores improved with an increase in stimulus bandwidth up to the FBW condition (Figure 7-9C & 7-9D). The increase in speech discrimination with increase in bandwidth up to a minimum of 3 kHz is consistent with the results of low-pass filtering paradigms (Amos & Humes, 2007; Hornsby & Ricketts, 2003, 2006; Horwitz et al., 2008; Vickers et al., 2001). Of note, is the significant increase in scores from the LPF4k condition to the FBW condition, although the effective increase in aided audible bandwidth between the two conditions was limited to about 6 – 7 kHz (hearing aid bandwidth). Fifteen of 20 participants showed an average improvement of 6% in speech discrimination scores with bandwidth increase beyond 4 kHz, three participants showed no change, and one participant showed a decrease of 7.1%. The small but significant increase in average benefit up to ~5.5 to 7 kHz is consistent with previous studies (Hornsby & Ricketts, 2003, 2006; Horwitz et al., 2008), and therefore lends further support for provision of high frequency amplification for the range of hearing thresholds seen in the present study.

The growth in sound quality rating with increase in bandwidth up to the FBW condition is also similar to that seen in adults with normal hearing (Chapter 6) and in adults with a similar range of hearing loss (Ricketts et al., 2008). The present study included experienced hearing aid users with the majority of pure-tone thresholds under 70 dB HL, and high frequency slope (8 kHz threshold–4 kHz threshold) of no greater than 20 dB. Experienced hearing aid users with such audiometric profiles are more likely to indicate an improvement in sound quality with increased bandwidth beyond 4 kHz (Moore, 2012; Moore et al., 2011; Moore & Sek, 2013; Ricketts et al., 2008) and therefore the results of the present study are in agreement with the literature.

Effect of bandwidth on EFRs

In Experiment II, we observed an increase in the composite EFR amplitude and the number of EFRs detected when the bandwidth of the stimulus increased. Stimuli with wider bandwidths stimulate a larger number of sensory cells and afferent neurons resulting in a stronger input to the brainstem (John, Dimitrijevic, & Picton, 2003; Purcell & Dajani, 2008). Therefore an increase in stimulus bandwidth causes an increase in response amplitude (Figures 7-9B & 7-10) that facilitates detection of EFRs (Figure 7-9A). Similar to adults with normal hearing (Chapter 6), the increase in the number of EFR detections from the LPF4k to FBW condition was non-significant (Figure 7-9A). All carriers have stimulus energy within the band below 4 kHz, and so LPF4k condition was effective in eliciting a response from all carriers. In contrast to adults with normal hearing (Chapter 6), the growth in composite response amplitude from the LPF4k to FBW condition was non-significant. The dissimilarity may be due to the difference in effective increase in bandwidth audibility between LPF4k and FBW conditions for the transducers used. An ER-2 insert earphone with a flat frequency response to 10 kHz was used in adults with normal hearing whereas hearing aids with a more restrictive bandwidth were used with the adults with hearing loss in the present study. The hearing aid bandwidth reported by the manufacturer is 100–6700 Hz for the Quantum S, and 100–6000 Hz for the Quantum HP. It is possible that that the increase from 4 kHz to the bandwidth of the hearing aids (the effective upper limit) was not large enough to cause an increase in composite EFR amplitude in many participants.

Effect of bandwidth on response amplitude for individual EFR carriers

Increasing bandwidth of the stimulus /susaʃi/ resulted in unique patterns of change in response amplitudes for F1, F2+ carriers and fricatives (Figure 7-10), suggesting that the individual carriers offer unique information on audible bandwidth. Compared to adults with normal hearing (Chapter 6), fewer adjacent bandwidth conditions were significantly different in response amplitude possibly due to broader cochlear filters in combination with higher stimulation levels due to amplification (Moore, 2007). The FBW condition demonstrated the largest response amplitude for the high frequency carriers, although the change from LPF4k to FBW condition was significant only for the F2+ carrier of /i/. These results illustrate that the EFR stimulus is sensitive to changes across filter conditions, particularly below 4 kHz.

EFRs elicited by /i/ F1 warrant further discussion. Changes in response amplitude of /i/ F1 across bandwidths (Figure 7-10) were unexplained by the carrier spectrum. The increase in response amplitude with an increase in bandwidth from the LPF1k to LPF2k condition is similar in direction to the change seen in adults with normal hearing (Chapter 6). However, in contrast to the results seen in adults with normal hearing, the response amplitude for /i/ F1 decreases with further increase in stimulus bandwidth (response amplitude increased significantly from the LPF2k to LPF4k condition in normal hearing adults). The decrease in the amplitude between the LPF2k and FBW conditions possibly reflects the attenuation of responses elicited by low frequency carriers in the presence of high frequency carriers, a phenomenon demonstrated in multiple ASSR paradigms in which up to four stimuli are presented simultaneously (John et al., 2003; John, Lins, Boucher, & Picton, 1998; John, Purcell, Dimitrijevic, & Picton, 2002; McNerney & Burkard, 2012). Based on stimulus spectra, the interaction between the F1 and F2+ bands of /i/ would be considered unlikely because of the separation between the spectral peaks being more than an octave (John et al., 2003, 1998). However, absolute stimulus levels and the aided level of the F2+ band relative to the F1 band are higher due to hearing aid gain. Higher stimulus levels lead to increased interactions, likely due to broader regions of activation in the cochlea (Ishida & Stapells, 2012; John et al., 1998, 2002; McNerney & Burkard, 2012).

7.5 Limitations

In Experiment I, the thresholds of each carrier were obtained using an ER-2 insert earphone, that differs from hearing aids in bandwidth and noise floor. Limited bandwidth in the aided condition could have possibly resulted in a marginal over-estimation of SL for fricative carriers. Also, as discussed above, potential changes in aided thresholds of low frequency carriers due to audible hearing aid noise floor may also have resulted in minor over-estimation of SL for some participants. Finally, the present study evaluated the EFR paradigm only in older adults with a specific range of hearing losses. The generalizability of these results to other populations of interest, such as infants or those with more severe degrees of hearing loss, remains unknown and warrants further investigation.

7.6 Summary and Conclusions

The present study aimed to evaluate sensitivity of the proposed EFR paradigm to stimulus level, bandwidth, and use of amplification in adults with hearing loss. In Experiment I, audibility of individual carriers was varied by use of two stimulus levels in unaided and aided conditions. The observed increase in the number of EFR detections and response amplitude for most carriers due to an increase in level and use of amplification, supports sensitivity of the EFR paradigm to changes in audibility. In Experiment II, the bandwidth of EFR carriers was varied by low-pass filtering the /susaʃi/ sequence at 1, 2 and 4 kHz in aided conditions. An increase in the number of detections as well as composite response amplitude (sum of all eight EFRs recorded per condition) with improvement in stimulus bandwidth (at least up to 4 kHz) indicates that the EFR paradigm is sensitive to changes in audible bandwidth. Bandwidth-related changes in EFRs also paralleled bandwidth-related changes in psychophysical measures of bandwidth such as speech discrimination and quality rating, further supporting bandwidth sensitivity of the EFR paradigm. Results from our electroacoustic verification confirm that the stimulus modifications adopted for stimulus optimization do not significantly interact with hearing aid function. Taken together, results from our main experiments and electroacoustic verification suggest that the proposed EFR paradigm will likely be useful in evaluating changes in audibility of low, mid and high frequency speech sounds in unaided and aided conditions in clinically feasible test times. Further investigation is necessary to understand stimulus-response relationships in aided conditions, and the lack of increase in response amplitude due to amplification for /u/ and /i/ F1 carriers in individuals with hearing loss. Further investigation is also necessary to evaluate the utility of this paradigm in children with hearing loss.

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Chapter 8

Summary of Contributions, Limitations and Conclusions

8.1 Research aims and summary of findings

The overall aim of the current work was to evaluate and optimize AEP protocols that have been proposed for use as an objective aided outcome measure. This work was based on a consolidated framework (Figure 1-1 on p. 7) of factors deemed important in the evaluation of Auditory Evoked Potentials (AEPs) as an objective aided outcome measure. This thesis presented three preliminary studies (Chapters 2–4) focusing on Cortical Auditory Evoked Potentials (CAEPs) followed by the proposal and validation of an objective aided outcome tool using speech-evoked Envelope Following Responses (EFRs). In line with the framework (Figure 1-1), the six studies investigated stimulus and response factors (Chapters 2 and 5), hearing aid interaction (Chapters 3, 4 and Appendix E), and test functionality (Chapters 6 and 7).

Results in Chapter 2 illustrated the effect of selecting stimuli from a word-initial and a word-medial position on onset-sensitive CAEPs; word-medial stimuli elicited larger amplitude CAEPs with shorter latencies (Easwar, Glista, Purcell, & Scollie, 2012a). Results in Chapter 3 illustrated the nature of non-linear hearing aid signal processing on tone-burst onsets and the lack of a significant effect of the rise-time change on CAEPs (Easwar, Glista, Purcell, & Scollie, 2012b). Results in Chapter 4 demonstrated discrepancies in the representation of hearing aid function for speech sounds in a CAEP testing context versus a natural running speech context (Easwar, Purcell, & Scollie, 2012). Findings in Chapter 4 prompted the development of a test paradigm using speech-evoked EFRs.

The proposed EFR paradigm uses a five-phoneme stimulus sequence (/susaʃi/) that resembles running speech in temporal characteristics in an attempt to minimize interactions due to non-linear hearing aid processing. The sequence consisted of dual-f_0 vowels and modulated fricatives to enable recording of EFRs from multiple frequency regions. Chapter 5 evaluated polarity sensitivity of vowel-evoked EFRs and the effect of averaging responses to opposite vowel polarities of the proposed test paradigm. Results suggest that polarity sensitivity of EFRs varies across individuals and vowel carriers, and may be influenced by the degree of stimulus envelope asymmetry above and below the baseline. Averaging vowel-evoked EFRs to opposite polarities did not however affect detection in the majority of individuals. Chapters 6 and 7 evaluated sensitivity of the EFR paradigm to changes in audibility in adults with normal hearing and hearing loss, respectively. Measures such as the number of detected EFRs and response amplitude demonstrated sensitivity to changes in audibility due to stimulus level, bandwidth and amplification. When stimulus bandwidth was varied, changes in EFR parameters correlated with changes in behavioral hearing aid outcome measures such as speech discrimination and sound quality rating. In adults with normal hearing (unaided) and hearing loss (aided), the average stimulus presentation time needed for a significant detection were under three minutes across all carriers, and the effective test time (based on the current stimulus) was under 12 minutes for the majority of carriers. Use of alternative detection rules (sections 6.2.2 on p. 156 and 7.2.2 on p. 209) within each frequency band may further reduce test times and improve clinical feasibility. Findings in Chapters 6 and 7 therefore suggest that the proposed speech-evoked EFR test paradigm may be useful for clinical applications as an objective aided outcome measure.

8.2 Strengths and contributions to Audiology

The proposed framework (Figure 1-1) for the evaluation of AEPs as an objective outcome measure brings together a comprehensive range of factors that were deemed important when evaluating an objective aided outcome measure. These factors included the optimization of unaided AEP testing along with factors important for valid aided measurement. The studies presented in Chapters 2–7 illustrate that the proposed framework is broad enough to encompass different paradigms, suggesting that it may be generalizable to other investigations. Further, the framework contributed significantly to the design of the proposed EFR paradigm. In general, the framework appears to facilitate comprehensive comparison of multiple AEPs for outcome measurement.

The contribution of the first portion of the thesis (Chapters 2–4) was extension and clarification of issues with unaided and aided CAEP measurements that have previously appeared in the literature, some prior to and some along with the work presented here. The first two studies in this thesis evaluated the effect of stimulus rise-time on CAEPs (Easwar, Glista, et al., 2012a, 2012b). Chapter 2 substantiates and extends previous findings of the effect of tone-burst rise-time (e.g., Onishi & Davis, 1968; Thomson, Goswami, & Baldeweg, 2009) to speech-evoked CAEPs. Chapter 3 provides evidence for the lack of effect of the stimulus rise-time change due to non-linear hearing aid processing after controlling for known confounding factors, and hence expands findings from previous studies that used linear processing (e.g., Jenstad, Marynewich, & Stapells, 2012; Marynewich, 2010). Chapter 4 illustrates the effect of CAEP test protocols on non-linear hearing aid processing of speech stimuli (Easwar, Purcell, & Scollie, 2012). Similar to previous studies on non-speech AEP stimuli (e.g., Brown, Klein, & Snydee, 1999; Gorga, Beauchaine, & Reiland, 1987; Jenstad et al., 2012), Chapter 4 emphasizes evaluation of AEP stimulus interactions and representation of hearing aid function during objective aided testing, even when using speech stimuli. Together, these studies and literature illustrate the importance of considering stimulus and hearing aid factors while inferring about aided audibility using AEPs.

The primary contribution of the second portion of this thesis (Chapter 5–7) was the proposal and validation of a novel objective measure that is based on a running speech stimulus. This speech-evoked EFR paradigm addresses many of the criteria identified in the framework. Use of naturally spoken vowels as EFR stimuli presented in isolation (Aiken & Picton, 2006) and in a sentence structure (Choi, Purcell, Coyne, & Aiken, 2013) have been demonstrated in previous studies. However, the EFR paradigm proposed in the current work included a wider range of phonemes (fricatives) and dual-f_0 vowels to maximize its ability to measure neural representation of low, mid and high frequencies. The use of simultaneous carriers to improve frequency specificity and reduce test times, and the use of modulated fricatives to represent the high frequencies, is a step forward in improving clinical utility and feasibility. Additionally, a consonant-vowel syllabic structure was used to include amplitude fluctuations in running speech. The stimulus modifications did not significantly affect hearing aid function and allowed valid aided measurements in clinically feasible test times. The current work thus demonstrates that speech-evoked EFRs may be adapted to improve their clinical utility, and that they may be feasible for use as an objective aided outcome measure with non-linear hearing aids.

The current work also demonstrated the sensitivity of the new paradigm by measuring carrier-specific changes due to experimental manipulations such as an increase in stimulus level, hearing aid amplification, and stimulus bandwidth. Bandwidth-related changes in EFR measures paralleled behavioral measures in unaided and aided conditions. This further supports the validity of the new paradigm, as it offers frequency-specific outcome measurement that is correlated with speech perception. This further indicates that the stimulus development was relatively successful in representing multiple frequency regions. Additionally, results from a control group (young adults with normal hearing) and a clinical group (adults with hearing loss) illustrate clinically feasible test times.

A further contribution of the second portion of the thesis (Chapter 5) was the investigation of a previously recommended EFR response analysis method. The effect of stimulus polarity on EFRs was evaluated on a large study sample using a broader range of vowel stimuli than previous studies (Aiken & Purcell, 2013). Results confirm and extend previous findings that the effects of polarity vary by vowel and by individual (Aiken & Purcell, 2013; Greenberg, 1980). In addition, the study provides further support, with additional stimuli, that EFRs elicited by vowels are dominated by responses following the envelope rather than stimulus energy at the first harmonic (Aiken & Picton, 2006).

In summary, the current work strengthens previous findings in the CAEP and EFR literature, and provides a novel objective aided outcome measure that may facilitate representation of relevant hearing aid function during aided AEP testing and evaluation of access to aided sound across a wide frequency range.

8.3 Implications

The two AEPs evaluated in this thesis vary in the type of response, stimulus characteristics and response generator sites. CAEPs are onset-sensitive transient responses signaling the arrival of a stimulus or a change in stimulus characteristic at the cortical level (American Academy of Audiology [AAA], 2013; Hyde, 1997; Korczak, Kurtzberg, & Stapells, 2005; Tremblay, Kalstein, Billings, & Souza, 2006). EFRs to the voice f_0 reflect phase-locking of neurons in the upper brainstem to the stimulus envelope (Aiken & Picton, 2006; Chandrasekaran & Kraus, 2010) and hence additionally represent the nature of sound processing (Clinard & Tremblay, 2013; Dajani, Heffernan, & Giguere, 2013). Due to the differences in generator sites, CAEPs demonstrate transmission of the stimulus to a higher level in the auditory system relative to EFRs. Although not within the scope of this thesis, the EFR approach could also allow inference of stimuli reaching the cortex by recording responses that follow the slow temporal fluctuations in speech (e.g., Aiken & Picton, 2008).

Although CAEP protocols may interact with hearing aid signal processing, possibly affecting representation of hearing aid function (Easwar, Purcell, & Scollie, 2012), CAEP-based methods are more likely to be useful in children with Auditory Neuropathy Spectrum Disorder (ANSD). About 10 to 20% of children with permanent hearing loss are diagnosed with ANSD (Bagatto et al., 2011; Berlin et al., 2010; Kirkim, Serbetcioglu, Erdag, & Ceryan, 2008). In these children, Auditory Brainstem Responses (ABRs) and Auditory Steady-State Responses (ASSR) have limited utility as a diagnostic or a prognostic measure (Rance, 2005; Rance et al., 2005). In contrast, CAEPs have shown to be prognostic markers of outcomes in children with ANSD wearing hearing aids or using cochlear implants (Cardon, Campbell, & Sharma, 2012; Rance, Cone-Wesson, Wunderlich, & Dowell, 2002;

Sharma, Cardon, Henion, & Roland, 2011). Therefore, recommendations for use of an objective aided outcome measure cannot take a one-size-fits-all approach. It is rather likely that use of objective aided outcome measures will require a multi-AEP approach to accommodate the variety of hearing disorders in an audiological caseload.

The EFR paradigm has illustrated favorable results in terms of sensitivity to changes in audibility as well as response detection times in a clinical population. However, as discussed in the limitations and suggestions for future work (below), further validation studies are essential prior to establishing its utility for clinical implementation in the population of interest. In the current work, the approach of measuring hearing aid benefit using EFRs is largely a bottom-up audibility-based approach. Although audibility is a preliminary and important factor determining hearing aid outcomes, other factors such as presence of co-morbidities and/or family involvement also have a significant impact (e.g., Bagatto et al., 2011; Moeller, 2000). Therefore using an audibility-based approach alone may be limited in explaining variability in outcomes. However, aided AEPs are likely to be invaluable in ruling out factors (e.g., audibility) that may impede typical auditory development.

Although aided EFRs have demonstrated sensitivity to changes in level, it may be more useful in determining under-amplification leading to inaudibility as opposed to over-amplification. Under-amplification leading to inaudibility is likely to lead to a non-significant detection. However, when responses are detected, response amplitude may not definitively indicate the associated sensation level because response amplitudes vary across individuals (due to factors such as volume conduction; Picton, Dimitrijevic, Perez-Abalo, & Van Roon, 2005), and also across stimuli (e.g., Aiken & Picton, 2006; Choi et al., 2013). Additionally, the change in response amplitude with sensation level also varies across stimuli (e.g., EFRs:

Vander Werff & Brown, 2005, CAEPs: Purdy, Sharma, Munro, & Morgan, 2013) and hearing disorders (e.g., recruitment in cochlear hearing losses; Picton et al., 2005). Such variability in absolute response amplitude and change in response amplitude across conditions was evident in the current work as well (Chapters 6 and 7). Therefore, it may be challenging to infer sensation level and over-amplification on an individual basis.

As well, it is important to acknowledge that AEPs are indirect measures of audibility as they reflect a part of the activity involved in perception (Elberling & Don, 2007). Since they are affected by factors unrelated to hearing, lack of a detected response does not necessarily reflect inaudibility (Elberling & Don, 2007). Nevertheless, it is hoped that prudent use of such measures will support the use of aided AEPs as early indicators of appropriateness of hearing aid fittings, confirming access to sounds. This will facilitate early referrals to other rehabilitation programs such as cochlear implantation, if necessary.

8.4 Limitations

A number of limitations in the current work needs to be acknowledged.

- The current work included evaluation of multiple factors of interest in adults although the application of objective aided outcome measures is of primary interest in the infant population. Preliminary testing of the EFR paradigm in adults as a precursor to testing in infants is valuable due to the reliable behavioral testing possible in adults. Results may have been affected by the impact of age (e.g., Purcell, John, Schneider, & Picton, 2004; Skoe, Krizman, Anderson, & Kraus, 2013) and hence generalization to infants, who may additionally show developmental trends, may be limited. Nevertheless, use of a study sample with young adults serves as a control group to evaluate factors of interest without the influence of hearing loss and/or maturation.

- Hearing aids in the current work were evaluated after de-activation of signal processing features such as noise reduction and frequency lowering. These features may be used in pediatric fittings (Bagatto, Scollie, Hyde, & Seewald, 2010) and therefore require further consideration (see framework). Features such as noise reduction may selectively reduce gain in a frequency-specific manner for noise-like sounds (for a review, see Bentler & Chiou, 2006). Noise reduction algorithms may use modulation detection paradigms to distinguish speech from non-speech sounds (Bentler & Chiou, 2006) and therefore interactions between hearing aid function and stimulus modifications adopted in the EFR paradigm require further investigation. As well, frequency lowering processing features alter the frequency composition of incoming stimuli using a variety of techniques such as frequency compression, frequency lowering or frequency translation. Some of these features may affect the harmonic structure or the spacing between consecutive harmonics in a vowel and hence may have important implications for vowel carriers. The effects of frequency lowering technologies are mainly at higher frequencies as they aim to improve high frequency audibility. The effect will however depend on the cut-off frequency beyond which frequency lowering is applied. This may have implications for vowel F2+ carriers more than fricatives and hence requires further investigation. Additionally, hearing aid technologies including compression characteristics evolve over time. The effects of context in Chapter 4 and lack of adverse effects of stimulus modifications adopted in Chapter 7 are therefore to be generalized with caution to newer technologies.

- Response analysis in Chapters 6 and 7 included averaging of responses across the two stimulus polarities. This approach was used for all responses because the majority of vowel carriers did not show a significant effect of polarity and response detection was not negatively affected in most participants (Chapter 5). In addition, comparing data across all conditions and participants in Chapters 6 and 7, the number of significant detections were higher in the two-polarity average compared to either polarity detection rule (Bonferroni corrected). Therefore, there may have been individual cases where the number of EFRs detected were underestimated using the two-polarity average in some participants.

- With the male-spoken /susaʃi/ stimulus used in Chapters 6 and 7, EFRs to fricatives were detected when the stimulus /susaʃi/ was low-pass filtered at 4 kHz. Fricatives spoken by males have more mid frequency energy compared to fricatives spoken by female talkers (Boothroyd & Medwetsky, 1992; Stelmachowicz, Pittman, Hoover, & Lewis, 2001). Therefore, predicting audibility of female-spoken fricatives based on the number of EFRs detected using the current stimulus may be limited. High-pass filtering the fricatives at a higher cut-off frequency may improve generalizability to infer audibility of fricatives at higher frequencies.

- In Chapter 7, hearing aids were coupled with unvented molds and stimuli were presented through DAI in aided conditions. Clinically, if vented molds are prescribed, use of DAI may not accurately reflect the effects of venting during everyday use. Therefore for hearing aids fittings with vents, other modes of stimulus presentation such as a speaker in sound field may be necessary.

- The EFR paradigm was evaluated in a clinical population with a homogeneous audiometric profile in terms of degree and configuration of hearing losses. Test functionality may vary with audiometric profiles and therefore results from the Chapter 7 are to be generalized with caution.

- In Chapter 7, audibility of hearing aid noise floor and improvement in signal-to-noise ratio due to amplification was not quantified. Assessment of these factors may help to evaluate their influence on the lack of a significant increase in response amplitude of low frequency carriers due to amplification.

- The interaction between F1 and F2+ EFRs during simultaneous presentation of F1 and F2+ carriers in dual-f_0 vowels was evaluated in a group of adults with normal hearing (see Appendix D), but not in aided conditions in adults with hearing loss. This could be an additional factor that may help understand individual variability in the effect of amplification on low-frequency EFRs.

8.5 Future work

Results from Chapters 6 and 7 provide favorable preliminary evidence for clinical applications of the proposed speech-evoked EFR paradigm and warrant further investigation. The following could be potential areas for future investigation:

- Effect of simultaneous and independent presentation of F1 and F2+ vowel carriers in aided conditions using the EFR test paradigm: Generalization of the lack of interaction to individuals with hearing loss tested aided requires further investigation due to the relative difference in aided F1 and F2+ stimulus levels in each vowel compared to unaided stimuli.

- Evaluation of the EFR paradigm in children and infants with normal hearing and hearing loss: Response detectability and response detection times may vary in younger children due to the maturing auditory system and hence require further validation.

- Evaluation of the EFR paradigm for different audiometric profiles: Most participants in Chapter 7 presented with mild to moderately-severe sloping sensorineural hearing loss. The distribution of hearing loss degrees and configurations may vary in children (Pittman & Stelmachowicz, 2003) and hence further evaluation is necessary to establish generalization of sensitivity to carrier-specific audibility changes in other hearing losses.

- Evaluation of the effect of acclimatization to hearing aid use: The paradigm was evaluated in experienced hearing aid users with post-lingual hearing loss. Plastic changes in the auditory system due to the presence (e.g., Anderson, Parbery-Clark, White-Schwoch, Drehobl, & Kraus, 2013), degree and duration of hearing loss (e.g., review by Gordon et al., 2011), and hearing aid use may influence EFR characteristics.

- Evaluation of the EFR paradigm in children and infants with co-morbidities or complex factors: Use of an objective aided outcome measure is likely to be informative in children with typical aided audibility who may not meet auditory milestones (Bagatto et al., 2011). Use of such measures may help rule out or narrow down potential factors leading to atypical auditory behavior.

- Evaluation of the current method to predict behavioral outcome measures of hearing aid fittings such as speech discrimination (e.g., Dimitrijevic, John, & Picton, 2004).

- Evaluation of various hearing aid signal processing schemes such as frequency lowering technologies that aim to improve audibility of specific frequency regions.

Additionally, in Chapter 4, the effect of inter-stimulus intervals on non-linear hearing aid function was assessed using clinically typical hearing aid fittings. Inter-hearing aid variability may reflect underlying dynamic compression and expansion parameters and therefore investigation of these factors may help predict possible discrepancies with certain hearing aids.

8.6 Concluding statements

Use of an objective aided outcome measure may help improve evaluation of hearing aid fittings in infants. Choice of AEPs for use as an aided outcome measure requires consideration of factors related to stimulus choice and response characteristics, hearing aid interaction and test functionality. The current work evaluated CAEPs and EFRs for use as objective aided outcome measures. Findings re-emphasize the importance of considering stimulus-nonlinear hearing aid interactions during testing. Preliminary studies led to the main contribution of this thesis which is the development of a novel test paradigm using speech-evoked EFRs. The paradigm uses a stimulus that represents speech in temporal and spectral characteristics, and has illustrated sensitivity to change in audibility due to level, bandwidth and amplification with clinically feasible test times in adults with normal hearing and hearing loss. These results are encouraging and future work is necessary to establish the utility of these measures in infants and children who wear hearing aids. Although, speech-evoked EFRs may present advantages over CAEPs in some aspects, their application may be limited in hearing disorders such as ANSD. Therefore, a multi-AEP approach is likely necessary to cater to the nature of hearing disorders typically seen by a pediatric audiologist in infant hearing programs. We envision the use of objective aided outcome measures alongside subjective outcome measures as an early indicator of access to sound in infants wearing hearing aids.

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Appendices

Appendix A: Ethics approval notices

Research Western

Use of Human Participants - Ethics Approval Notice

Principal Investigator: Dr. Susan Scollie
Review Number: 16124E
Review Level: Delegated
Approved Local Adult Participants: 16
Approved Local Minor Participants: 25
Protocol Title: Acclimatization to newly audible, high-frequency speech sounds
Department & Institution: Communication Sciences & Disorders, University of Western Ontario
Sponsor: Canadian Institutes of Health Research

Ethics Approval Date: June 10, 2011 **Expiry Date:** December 31, 2011
Documents Reviewed & Approved & Documents Received for Information:

Document Name	Comments
Revised UWO Protocol	The revision included the addition of an adult control group which will comprise of 16 participants in addition to the current total sample of 25 children. The stimuli, testing conditions and number of sessions have been revised for the adult group. An upper age limit has now been set to 35.
Increase in number of local Participants	The number of local participants has been increased to 41.
Revised Letter of Information & Consent	
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The ethics approval for this study shall remain valid until the expiry date noted above assuming timely and acceptable r to the HSREB's periodic requests for surveillance and monitoring information. If you require an updated approval noti that time you must request it using the UWO Updated Approval Request Form.

Members of the HSREB who are named as investigators in research studies, or declare a conflict of interest, do not part discussion related to, nor vote on, such studies when they are presented to the HSREB.

The Chair of the HSREB is Dr. Joseph Gilbert. The UWO HSREB is registered with the U.S. Department of Health & Services under the IRB registration number IRB 00000940.

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Western

Research Ethics

Use of Human Participants - Ethics Approval Notice

Principal Investigator: Dr. Susan Scollie
File Number:102557
Review Level: Delegated
Approved Local Adult Participants:220
Approved Local Minor Participants:60
Protocol Title: Aided Auditory Evoked Potentials
Department & Institution: Health Sciences\Communication Sciences & Disorders,Western University
Sponsor: Ontario Ministry of Research and Innovation

Ethics Approval Date: July 05, 2012 **Expiry Date:**September 30, 2014
Documents Reviewed & Approved & Documents Received for Information:

Document Name	Comments	Version Date
Western University Protocol		
Letter of Information & Consent	Normal Hearing 13 - 18 years	2012/06/19
Assent	7-12 years Normal Hearing	2012/06/19
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The ethics approval for this study shall remain valid until the expiry date noted above assuming timely and acceptable responses to the HSREB's periodic requests for surveillance and monitoring information. If you require an updated approval notice prior to that time you must request it using the University of Western Ontario Updated Approval Request Form.

Members of the HSREB who are named as investigators in research studies, or declare a conflict of interest, do not participate in discussion related to, nor vote on, such studies when they are presented to the HSREB.

The Chair of the HSREB is Dr. Joseph Gilbert. The HSREB is registered with the U.S. Department of Health & Human Services under the IRB registration number IRB 00000940.

Ethics Officer to Contact for Further Information

Janice Sutherland	Grace Kelly	Shantel Walcott

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Appendix B: Comparison of between-polarity and within-polarity differences in Envelope Following Responses (EFR) amplitude

The aim of the pilot study was to compare between-polarity differences to within-polarity differences in Envelope Following Responses (EFR) amplitude. The stimulus used in the pilot study was similar in structure to that used in Experiment II (Figure 5-1B on p. 101). The stimulus excluded vowels without h1 and instead consisted of two iterations of /susaʃi/, where all vowels had h1. Hence each sweep consisted of two iterations of the sequence /susaʃi/ in Polarity A followed by two in the opposite polarity, Polarity B. The first iteration within each polarity was considered repetition 1 and the second, repetition 2. A group of 16 young adults with normal hearing participated in the pilot study ($M_{\text{age}} = 22.75$ years; $SD_{\text{age}} = 2.21$), out of which 11 participated in Experiment II. EFRs were recorded over 300 sweeps. EFR recording, test procedures and analysis were identical to Experiment II. Each recording session was 41 minutes in duration. Similar to Experiment I and II, an individual's data were included if either of the two responses being compared was significant at $p = 0.025$ (Bonferroni correction). Therefore, data points that involved comparison of two non-significant EFR response amplitudes were excluded.

A three-way Repeated Measures Analysis of Variance (RM-ANOVA) was completed using within-subject factors of polarity (A, B), repetition (1, 2) and vowel (/u/, /a/, /i/), and response amplitudes (nV) as the dependent variable. Individual RM-ANOVA were completed for F1 and F2+ carriers. For the F1 carriers, a significant main effect of polarity ($F(1, 12) = 14.63$, $p = 0.002$, $\eta^2_{partial} = 0.55$) and a significant interaction between polarity and vowel were found ($F(1.79, 21.43) = 3.59$, $p = 0.049$, $\eta^2_{partial} = 0.23$). Repetition did not show a significant main effect or interaction. The significant interaction between within-subject factors polarity and vowel suggests that the difference in response amplitude between Polarity A and Polarity B, when collapsed across the two repetitions, varied by vowel. In contrast, none of the within-subject factors had a significant effect on EFR response amplitudes elicited by F2+ carriers.

Similar to Aiken & Purcell (2013), absolute differences in response amplitudes within each polarity and between the two polarities were also compared. Within-polarity absolute differences in amplitude represent the test-retest variability and hence the analysis below directly compares if between-polarity differences were higher than within-polarity test-retest variations. Repetitions 1 and 2 were averaged within each polarity prior to computing differences between the two polarities as the analysis above indicated no significant effect of repetition on response amplitudes. Figure A-1 illustrates within and between-polarity amplitude differences for all carriers. The average between-polarity difference is higher compared to the within-polarity difference only for /u/ F1. To compare within-polarity differences with between-polarity differences, a RM-ANOVA was completed with polarity condition (within-Polarity A, within-Polarity B and between-Polarity AB) and vowel as the within-subject factors, and absolute amplitude differences as the dependent measure. For F1, a significant main effect of polarity condition ($F(1.80, 21.63) = 4.88$, $p = 0.020$, $\eta^2_{partial} = 0.29$) and a significant interaction between vowel and polarity condition ($F(2.77, 33.22) = 3.36$, $p = 0.033$, $\eta^2_{partial} = 0.22$) were found. For post-hoc testing, paired t-tests were completed to compare differences between the polarity conditions for each vowel. FDR corrected p values were used for interpretation of significant differences. For the vowel /u/, the between-polarity absolute differences ($M = 42.17$ nV, $SD = 24.27$) were significantly higher than the within-Polarity A absolute differences ($M = 17.64$ nV, $SD = 13.91$) by a mean difference of 24.54 nV, $t(12) = 3.04$, $p = 0.010$, 95% CI

[6.94, 42.14] and likewise, the between-polarity differences were significantly higher than the within-Polarity B differences (M = 15.46, SD = 14.31) by a mean difference of 26.71 nV, $t(12) = 4.27$, $p = 0.001$, 95% CI [13.09, 40.32]. This suggests that the between-polarity differences in response amplitude were higher than the differences in response amplitude due to test-retest variability. The vowels /a/ and /i/ did not show any significant differences between polarity conditions. A similar analysis for F2+ carriers showed no significant effect of polarity condition or vowel on the absolute amplitude differences.

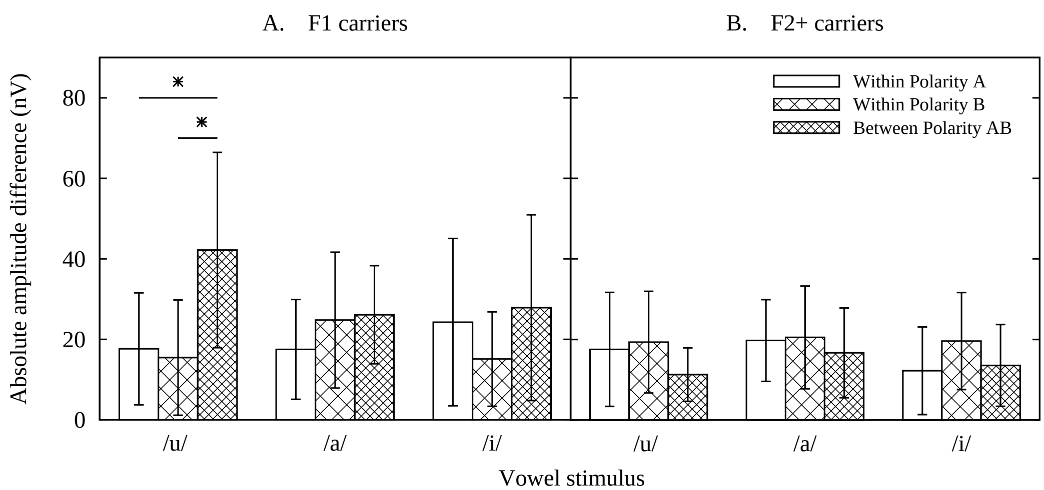

Figure A-1: Comparison of within-polarity and between-polarity absolute amplitude differences (Error bars represent ±1 SD; n[F1] = 13, n[F2+] = 10). * indicates a significant difference.

Reference:

Aiken, S. J., & Purcell, D. W. (2013). Sensitivity to stimulus polarity in speech-evoked frequency-following responses (3aPP31). In *Proceedings of Meetings on Acoustics Vol., 19* (pp. 1-5). NewYork, NY: Acoustical Society of America.

Appendix C: Stimulus artifact check for EFR stimulus

A stimulus artifact check was completed using a no-stimulus-to-the-ear run in 10 participants. The stimulus used for this check was the /susaʃi/ stimulus sequence including h1 with only one repetition of vowels in each polarity. Recording conditions and placement of the transducer and recording electrode leads were similar to the main experiments but the stimulus was routed to a Zwislocki coupler using a foam tip and was placed on each participant's chest. The recording time per participant was 20.5 minutes. The false positive rate (rate of significant detections), obtained in Polarity A and Polarity B was 6.6 and 5% respectively. Since the difference in false positive rates for the two conditions was marginal, and was close to the assumed α of 5%, we conclude that stimulus artifact was unlikely to significantly contribute to the apparent polarity-sensitive amplitude differences.

Appendix D: Comparison of simultaneous and independent presentations of F1 and F2+ vowel carriers on EFR amplitude

To investigate the effects of simultaneous presentation of F1 and F2+ carriers on EFR amplitudes, three versions of the stimulus sequence /susaʃi/ were presented in an interleaved manner. The first iteration of the sequence was the dual-f_0 version used in Experiment I and II (simultaneous condition), the second iteration of the sequence consisted of the F1 band alone (independent F1 condition), and the third iteration of the sequence consisted of the F2+ bands alone (independent F2+ condition). The spectral levels of the bands were matched in the dual and independent presentation modes. Ten participants (nine females, $M_{\text{age}} = 20.8$ years; $SD_{\text{age}} = 1.81$) took part in the pilot study. Four of the ten participants also participated in the main experiments. Participants underwent screening tests described in the main experiment to determine eligibility. EFRs were recorded over 300 sweeps. EFR recording, test procedure and analysis were identical to the main experiments. Each recording lasted for a little over an hour. An individual's data were included if either of the two responses being compared was significant at $p = 0.025$ (Bonferroni corrected). This means that data points that compared two non-significant EFR amplitudes were excluded. Due to non-significant responses, data from one participant was excluded for F1 carriers and data from another participant was excluded for F2+ carriers.

Figure A-2 illustrates the average response amplitudes of F1 and F2+ carriers in both simultaneous and independent presentation modes for all three vowels /u/, /a/ and /i/. Individual two-way RM-ANOVAs were completed for F1 and F2+ carriers. Within-subject factors were condition (simultaneous, independent) and vowel (/u/,

/a/, /i/), and the dependent variable was response amplitude. For the F1 carriers, the main effect of condition ($F(1, 8) = 0.001$; $p = 0.973$; $\eta^2_{partial} = 0.00$) and vowel ($F(1.47, 11.79) = 2.35$; $p = 0.146$; $\eta^2_{partial} = 0.23$), and the interaction between the two factors were non-significant ($F(1.33, 10.71) = 1.19$; $p = 0.318$; $\eta^2_{partial} = 0.13$). For the F2+ carriers, the main effects of condition ($F(1, 8) = 0.13$; $p = 0.728$; $\eta^2_{partial} = 0.02$), and the interaction between vowel and condition were not significant ($F(1.81, 14.49) = 2.04$; $p = 0.169$; $\eta^2_{partial} = 0.20$). The main effect of vowel was significant ($F(1.13, 9.07) = 6.29$; $p = 0.031$; $\eta^2_{partial} = 0.44$) suggesting that response amplitudes varied by vowel, when collapsed across the simultaneous and independent conditions. Post-hoc analysis was completed using paired t-tests and interpreted using FDR corrected critical p values. The F2+ response amplitude of /a/ ($M = 113.03$ nV; $SD = 29.76$) was significantly higher than /i/ ($M = 81.06$ nV; $SD = 33.34$) by a mean difference of 32.01 nV, $t(8) = 6.49$, $p < 0.001$, 95% CI [20.63, 43.39].

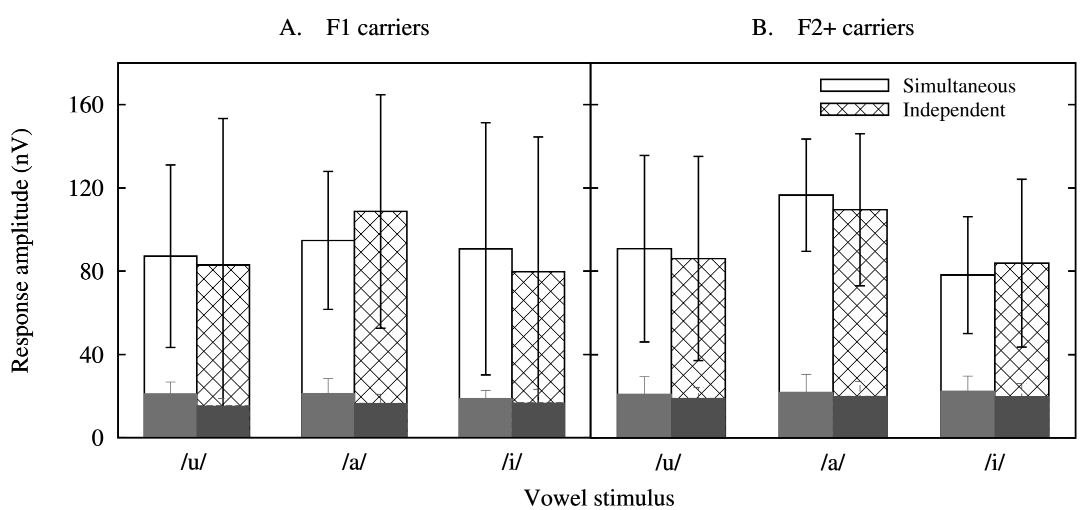

Figure A-2: Average response amplitudes of F1 and F2+ EFRs in simultaneous and independent presentation modes. Error bars represent ±1 SD. The solid grey bars represent average noise estimates. (n = 9)

Appendix E: Effect of stimulus modification on hearing aid function

To evaluate if stimulus modifications (e.g., modulation of fricatives) affect hearing aid function across a broad range of hearing aids, the difference between the EFR stimulus (modified /susaʃi/) and the original spoken stimulus (unmodified /susaʃi/) was compared in unaided and aided conditions. Fourteen hearing aids from seven manufacturers were fitted to three standard audiograms ranging from moderate to severe in overall hearing loss (N3, N4 and N5; Bisgaard, Vlaming & Dahlquist, 2010). Digital signal processing features such as noise reduction or feedback cancellation were de-activated. Hearing aid output for the unmodified and modified /susaʃi/ were recorded for stimulus presentation levels of 55, 65 and 75 dB SPL. Aided recordings were made in a click-on Brüel and Kjær coupler Type 4946 fitted with a microphone Type 4192 housed in an anechoic box (Brüel and Kjær Type 4232). Individual EFR carriers were extracted from aided recordings using Praat and the overall RMS level per carrier was measured using Spectraplus software. In unaided stimuli, the difference between carriers were within 1 dB (-0.28 to 0.73 dB) for all carriers except /a/ F2+, which was 1.16 dB higher in the modified /susaʃi/ compared to the unmodified version. Aided results indicated that the overall RMS level of all carriers in the modified and unmodified stimulus were within ± 2 dB. Averaged across all carriers, the mean difference was -0.13 dB (negative indicating a higher aided level for the EFR stimulus; $SD = 0.55$; range = -1.78 to 1.18 dB). Across the three levels and all carriers (378 observations), 19 of 27 differences $> \pm 1$ dB occurred for /a/ F2+. This is likely to have been due to stimulus modifications evident in unaided stimuli as well. Therefore in the majority of comparisons, differences between aided levels of original and modified carriers were no greater than the differences observed due to stimulus modification in the unaided signal.

To evaluate if aided spectra of the eight EFR carriers fall within the aided dynamic range (30^{th} – 99^{th} percentile) of the International Speech Test Signal (ISTS), the output of the 14 hearing aids were also recorded for the ISTS passage presented at 55, 65 and 75 dB SPL. The majority of aided carrier spectra fell within the aided ISTS dynamic range. However, one-third octave band spectra of a few carriers exceeded the 99^{th} percentile. Across all carriers and levels, 95% of the carrier octave band spectrum levels were within 2 dB of the ISTS 99^{th} percentile. We therefore conclude that stimulus modifications adopted to optimize EFR carriers does not interact significantly with hearing aid function across varying brands of hearing aids.

Reference:

Bisgaard, N., Vlaming, M. S. M. G. & Dahlquist, M. (2010). Standard Audiograms for the IEC 60118-15 Measurement Procedure. *Trends in Amplification, 14*(2), 1133-120.

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Please let me know if I could be of any further help.

Best regards,

Nada

****
Nada Ibrahim
[redacted]
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On 18 June 2014 06:29, [redacted] wrote:
> Dear Sir/ma'am,
>
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Curriculum Vitae

Name:	Vijayalakshmi Easwar
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	Graduate Thesis Research Fund Western University, 2013
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Related Work Experience:	Research Assistant Child Amplification Laboratory National Centre for Audiology Western University London, ON, Canada 2011 - present
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	Pediatric Audiologist Royal Hospital for Sick Children (Yorkhill) National Health Service Greater Glasgow & Clyde Yorkhill, Glasgow, Scotland 2009 - 2010
	Audiologist Royal Sussex County Hospital National Health Service, Brighton & Sussex University Hospitals Brighton, East Sussex, England 2008-2009
	Audiologist & Speech-Language Pathologist Nayak Hearing Health Care Clinic Bangalore, Karnataka, India 2007 - 2007

Publications:

Glista, D., Scollie, S.D., Moodie, S., & **Easwar, V.** (accepted May 27, 2014). The network of pediatric audiologists in Canada. The Ling 6(HL) Test: Typical pediatric performance data and clinical use evaluation. *Journal of American Academy of Audiology.*

Easwar, V., Boothalingam, S., Chundu, S., Manchaiah, V. K., & Ismail, S. M. (2013). Audiological Practice in India: An Internet-Based Survey of Audiologists. *Indian Journal of Otolaryngology and Head & Neck Surgery, 65*(S3), S636-S644.

Glista, D., **Easwar, V.**, Purcell, D. W., & Scollie, S. (2012). A pilot study on cortical auditory evoked potentials in children: Aided CAEPs reflect improved high-frequency audibility with frequency compression hearing aid technology. *International Journal of Otolaryngology Special Issue Hearing aids and the Brain*, doi: 10.1155-2012-982894.

Easwar, V., Purcell, D. W., & Scollie, S. D. (2012). Electroacoustic comparison of hearing aid output of phonemes in running speech versus isolation: Implications for aided cortical auditory evoked potentials testing. *International Journal of Otolaryngology Special Issue Hearing aids and the Brain*, doi: 10.1155-2012-518202.

Easwar, V., Glista, D., Purcell, D. W., & Scollie, S. D. (2012). Hearing aid processing changes tone burst onset: effect on cortical auditory evoked potentials in individuals with normal audiometric thresholds. *American Journal of Audiology, 21*(1), 82-90.

Easwar, V., Glista, D., Purcell, D. W., & Scollie, S. D. (2012). The effect of stimulus choice on cortical auditory evoked potentials (CAEP): Consideration of speech segment positioning within naturally produced speech. *International Journal of Audiology, 51*(12), 926-931.

Selected Presentations:

Easwar, V., Purcell, D.W., Aiken, S. & Scollie S, (2014, March) Use of speech-evoked envelope following responses as a hearing aid outcome measure. Podium presentation in the Student Research Forum, American Academy of Audiology AudiologyNow, Orlando, USA.

Easwar, V., Purcell, D.W., Aiken, S. & Scollie S, (2014, March) Sensitivity of speech-evoked envelope following responses To level, bandwidth, and amplification. Poster presentation in the Academy Research Conference (Hearing aids and the Brain), American Academy of Audiology AudiologyNow, Orlando, USA.

Easwar, V., Purcell, D.W., Aiken, S. & Scollie S, (2014, February) Sensitivity of speech-evoked envelope following responses to level and amplification in normal hearing and hearing impaired adults. Accepted for poster presentation at the Association for Research in Otolaryngology conference, San Diego, USA.

Easwar, V., Purcell, D.W., Aiken, S., Choi, J. & Scollie, S. (2013, June). Envelope Following Responses (EFRs) using modified speech stimuli for hearing aid evaluation. Poster presented in the International Evoked Response Audiometry Study Group, New Orleans, USA.

Easwar, V., Purcell, D.W. & Scollie, S. (2012, August). Exploration of hearing aid gain characteristics during aided cortical auditory evoked potentials testing. Poster presented in the International Hearing Aid Research Conference, Lake Tahoe, USA.