Effect of Hearing Aid Channels on Acoustic Change Complex

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Abstract

The objective of the present study was to find the effect of varying the number of channels of a hearing aid on the Acoustic change complex (ACC) in individuals with sloping hearing loss and to verify the same using the speech identification scores. The ACC was obtained from individuals with normal hearing and those with sloping sensorineural hearing loss in response to /si/ stimulus, the comparison in ACC wave forms were then made between these two groups. The ACC obtained across two, four and eight channel hearing aids with the same make and manufacturer were compared. A high frequency content speech identification test was done as a behavioral correlation measure. The results showed that ACC latencies in response to the fricative portion /s/ were delayed in individuals with sloping hearing loss. Between the two, four and eight channels, there were no significant differences found in terms of latency, amplitude, morphology or speech identification scores. The electrophysiological findings were found to poorly correlate with the behavioral speech identification measure. Therefore it was concluded that naturally produced speech tokens, representing different acoustic cues, like frication and vowel steady states can evoke distinct neural response patterns, and cortical evoked potentials elicited by /si/ stimulus can be reliably recorded in individuals with and without hearing aids. There was no significant difference in perception found between the two, four and eight channels.

Keywords: *Acoustic Change Complex (ACC), Cortical Auditory Evoked Potentials (CAEP), Hearing aids, Channels, Speech Perception.*

Introduction

The acoustic change complex (ACC) is a cortical auditory evoked potential that can be obtained from the auditory system when a time varying acoustic change occurs within the stimulus. The change can be in amplitude, spectral envelope or periodicity. Speech is one such stimulus which has multiple time varying acoustic changes, and can be used to elicit this potential (Martin & Boothroyd, 2000). The ACC being an obligatory potential depends solely on the acoustic features of the stimulus and the integrity of the central auditory system, further it does not require active involvement from the participants.

There is interest in using this potential as it can probe into supra-threshold auditory skills, such as speech sound processing. This potential may be able to dwell deeper into the perception of speech as an index of speech sound discrimination as it comprises not only a detection waveform but also one which arises due to a change in the acoustic characteristics of the signal.

Aided late latency responses can be reliably recorded according to Tremblay, Billings, Friesen, and Souza (2006), which will yield a better understanding of how device settings will affect the neural response patterns which in turn affects speech perception. After varying a device settings or a feature, if a cortical response will still mimic stimulus acoustic features, then a conclusion can be drawn about both, speech perception and how the hearing aid processes the signal.

In a sloping hearing loss, as the frequency increases the degree of hearing loss also increases (Roeser & Clark, 2007). Persons with sloping hearing loss due to a cochlear pathology can face problems while listening due to decreased audibility and dynamic range, also secondary to reduced frequency and temporal resolution. While studying the distribution of hearing loss characteristics in terms of configuration, Margolis and Saly (2008), found that sloping hearing losses dominated the distributions of configuration and that sensori-neural was the most prevalent site of lesion. Demeester et al. (2008) found the prevalence of high-frequency gently sloping and steeply sloping hearing loss to be 76% in males and 50% in females between 55 to 65 years. This means that majority of the persons with hearing loss miss the high frequency portions of speech, which are consonants. Consonants as compared to vowels were found to be approximately 30 dB lower in intensity and are hence less audible (Zeng & Turner, 1990). Hearing aids can help alleviate these issues by selective amplification, as in by increasing only the gain of higher frequency components.

Selective amplification can be accomplished through hearing aids by various means, one of which is the use of multichannel hearing aids which split the incoming signal into frequency bands. A band in a hearing aid refers to a frequency region where gain adjustments are made and a channel is where the same signal processing takes place. Individual compression circuits allow these multichannel instruments the flexibility to amplify each bandwidth of frequencies independently so as to correspond to the user's needs, preferences and their dynamic range. In addition, each channel may be set with unique

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attack and release time for compression across the frequency range as reported by Dillon (2001).

The number of channels necessary for optimal speech perception varies across studies. There are various factors involved that can dictate the amount of benefit a multichannel hearing aid can provide. For example, the configuration, age, degree of hearing loss, ability to combine temporal-envelope information and hearing aid experience (Jyoti, 2010; Rubina, 2008; Souza & Boike, 2006; Yund & Buckles, 1995). There are conflicting finding about the benefit derived from multichannel hearing aids as authors have found that these hearing aids may reduce the spectral contrast which aids in the perception of vowels (Bor, Souza & Wright, 2008).

According to Plomp (1988), this occurs when there are multiple channels with large compression ratios. However, Yund and Buckles (1995), demonstrated improvement in speech recognition from four to eight channels. Jyoti (2010) compared two, four and eight channels in listeners with sloping hearing loss and reported of better speech identification scores with increased number of channels. As there are contrasting results, electrophysiological verification of the benefits or detrimental effects of having multiple channels has been explored and behavioral speech identification scores were also obtained. Therefore, conclusions can be drawn on the effect of multi-channel hearing aids on the central auditory system processing of speech.

The P1-N1-P2 neural response patterns are heavily influenced by the acoustic content of the evoking signal and hearing aids may alter the acoustic content which will degrade speech perception. For example, the perception through hearing aids can blur the boundary between the aperiodic noise of consonants and the onset of voiced vowels, making these transitions less distinct (Stelmachowicz, Kopun, Mace, Lewis & Nittrouer, 1995).

To understand the interaction between the digitally amplified signal and its neural representation, physiologic detection of CV transitions was studied in a group of individuals who were first time users of hearing aids. The results of this study can be used as a counseling tool which can help clients make informed decisions and understand why higher or lower number of channels are required to suit their need.

The objectives of this study were to, one, compare ACC in individuals with normal hearing sensitivity to those with sloping cochlear hearing loss. Second, to compare the aided and unaided ACC in individuals with sloping cochlear hearing loss. Third, to compare the performance across two, four and eight channels of hearing aid. Fourth, to correlate speech identification score and the ACC obtained across different number of channels.

Method

Participants

Two groups, the control and the clinical group were included in the study. In the control group16 individuals (20 ears; equal number of male and female ears) in the age range of 25 to59 years were included. The criteria for inclusion comprised normal hearing sensitivity (air conduction and bone conduction thresholds less than 20 dB HL across frequency range from 250 Hz to 8 kHz) in both ears with normal middle ear status (defined as peak pressure between -100 and +60 daPa, and admittance between 0.3 and 1.60 cc). They were also required to be native Kannada speakers.

In the clinical group, 10 native Kannada speakers (11 ears; 3 females and 7 males) in the age range of 25 to59 years were included. The criteria for selection included air conduction thresholds which increased by 5 to 12 dB per octave and speech identification score of >55% in the test ear.

Instrumentation and Test Environment

A calibrated double channel audiometer, GSI- 61 was used to estimate the pure tone thresholds and to obtain speech identification scores. The GSI Tympstar (version-2) middle ear analyzer was used for immittance measurements. For electrophysiological recording a 2-channel diagnostic auditory evoked potential measuring instrument, Bio-Logic Navigator Pro (version 7.0.0) was used to record the ACC waveform.

To present the speech stimulus for ACC in aided and unaided condition a calibrated dB Technologies-160 free field speaker was used which had a frequency response range of 50 Hz to 20,000 Hz and maximum sound pressure level of 99dB SPL.

Fonix 7000 hearing aid test system was used for electroacoustic measurement of hearing aids and for real ear measurement.

The recording of the test stimuli and the audiological testing were done in acoustically treated rooms, with the noise levels within permissible limits according to ANSI S3.1, 1991. Pure tone audiometry was done in a double room suite, and the ACC acquisition in a single room set-up.

Test Procedure

The procedure started with taking a detailed case history probing into any history of ear related pathologies. Pure tone thresholds were obtained in octave intervals between 250 Hz to 8000 Hz for air conduction and between 250 Hz and 4000 Hz for bone conduction using the modified Hughson-Westlake procedure (Carhart & Jerger, 1959). Tympanometry and reflexometry were

Table 1: Stimulus and Acquisition parameters of ACC

Stimulus parameters		Acquisition parameters	
Stimuli	/si/	Mode of stimulation	Ipsi
Duration of stimulus	250 ms	Electrode montage	Cz, M1/M2 of test ear, ground at Fz
Intensity	65 dB SPL		
Polarity	Alternating	Filter setting	0.1-30 Hz.
Transducer	Loudspeaker	Analysis window	535 ms.
Mode of presentation	Free Field	No. of channels	Single
		Amplification	25,000
		Repetition rate	1.1 per second
		Number of sweeps	150
		No. of repetitions	2

Table 2: Acoustic features of the /si/ stimulus

Acoustic Features	Values
Total Duration	250 ms
Fricative Duration	143 ms
Vowel Duration	107 ms
Fricative Center Frequency	Energy spread from 2 to 5 kHz
Fundamental frequency - Vowel	75 Hz
First Formant - Vowel	1059 Hz
Second Formant - Vowel	2557 Hz

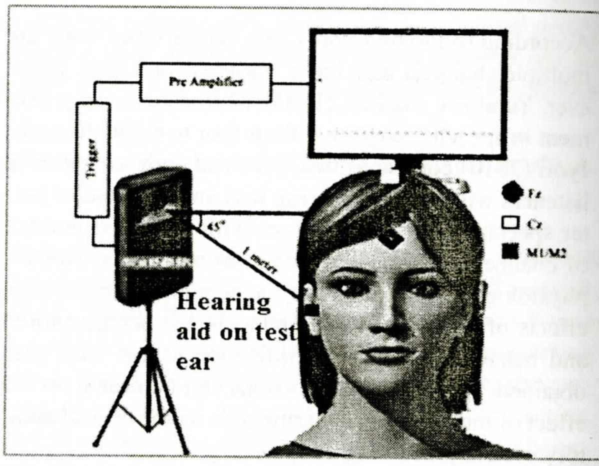

Figure 2: Set-up for aided ACC measurement - Electrode and speaker placement.

done to exclude individuals with middle ear pathology. To accomplish the same, a group of 50 individuals were administered the diagnostic protocol mentioned above, after which 16 of them (20 ears) were included in the study. The method involved 4 phases.

Phase 1: Acquisition of ACC in Control Group

The neural representation of a fricative /s/, its transition to a vowel, and perception of the vowel /i/ can be studied through the ACC. If these elements are perceived at the cortical level, two distinct waveforms will be observed, one in response to the consonant and the other for the vowel.

To learn the effect of sloping sensorineural hearing loss, the clinical group's unaided ACC waveforms were compared to that of the control group; the differences in latency, amplitude and morphology between the two responses gave information on the effect of sloping sensori-neural hearing loss on ACC. The inclusion of a control group can prevent extrinsic variables from influencing results, as both groups comprised native Kannada speakers, with no experience in listening through an amplification device. Additionally, cognitive abili-

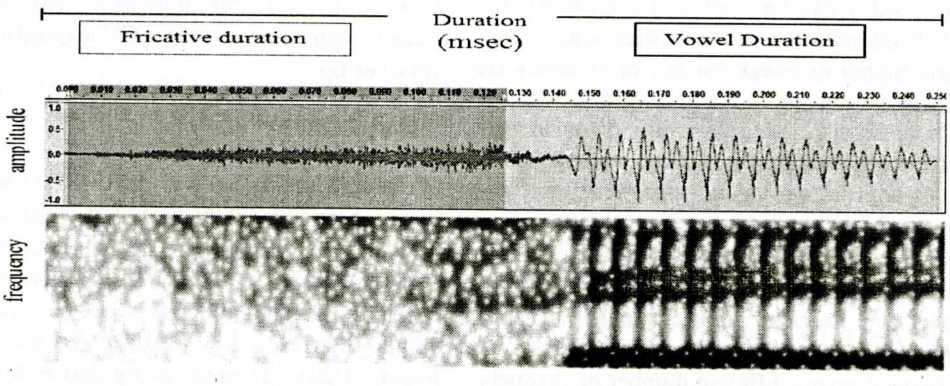

Figure 1: Wave form (top) and spectrogram(bottom) of stimulus /si/.

Table 3: Electro-acoustic measurement of the three hearing aids

Electro-acoustic measure	2 Channel	4 Channel	8 channel
Output Sound Pressure Level (OSPL) at 2100 Hz	130.7 dB	129.8 dB	131.3 dB
HFA- FOG	53.2 dB	49.8 dB	59.3 dB
Reference Test Gain	47.1 dB	46.4 dB	46.3 dB
Frequency Range	200 - 5785 Hz	200 - 5787 Hz	200 - 6212 Hz
EIN	12. 4 dB	10.1 dB	16.0 dB
THD			
500 Hz	2.0%	1.2%	0.5%
800 Hz	1.4%	1.3%	0.4%
1600 Hz	0.1%	0.1%	0.2%

Note: THD = Total Harmonic Distortion, EIN = Equivalent Input Noise, HFA-FOG = High Frequency Average-Full OnGain

ties that might influence the processing speech stimulus at the level of the auditory cortex can be taken into consideration. The stimulus and the acquisition parameters are detailed in Table 1.

While acquiring the ACC, participants were seated comfortably on a reclining chair. The electrode site was cleaned and gold disc electrodes were placed at the test site. The inter electrode impedance was maintained to be less than 2 kΩ and was monitored throughout the recording. The participants were instructed to ignore the stimulus and minimize head movements. A total of 300 sweeps were acquired for each participant. If the waveform was not found to replicate, a third recording was done.

*Stimulus Preparation:*To elicit ACC, /si/ stimulus was naturally produced and recorded using Adobe Audition (version 1.5) at a sampling rate of 48,000 with 16 bit resolution. A dB Technologies-160 free field speaker was used to present the /si/ stimulus for recording ACC for both the aided and unaided conditions. The output of the speaker was calibrated using a Larson-Davis System-824 sound level meter to be presented at a level of 65 dB SPL. The speaker was positioned at a distance of one meter at 45^o azimuth. With the PrattC software (version 5.1.31) the acoustic features of the /si/ stimulus was analyzed and is shown in Table 2, and the stimulus waveform and spectrogram are shown in Figure 1 and Figure 2 respectively.

Phase 2: Hearing aid fitting for the clinical group

Electro-acoustic Measurement: Two, four and eight channel hearing aids of the same make and manufacturer were utilized for the study. The electro-acoustic measurement with the hearing aid in test setting [ANSI S3.22-2003; (2 cc coupler)] was performed as shown in Table 3.

Hearing aid Programming: The NOAH 3 software using the Connexx platform, connected to HiPro was used for programming the hearing aid. The client would be seated with the probe microphone and the programming system placed in the same room to simultaneously match the target while programming the hearing aid. The hearing aids were in omni-directional mode with enabled compression circuits, and compression ratio set according to NAL-NL1. At an input of 65 dB SPL, gain provided across 250 Hz to 6500 Hz was verified. Of the four programmable memories in each hearing aid, only one that is the speech in quiet mode was activated, also the volume control was disabled.

NAL-NL1 fitting formula was used to prescribe the gain across different channels and accordingly the hearing aid was programmed to match the target gain at acclimatization level two. To specify the amplification targets for speech and the maximum output necessary to provide loudness comfort, audibility of speech, and speech intelligibility, this prescriptive formula was used.

To quantify the stimulus level at the output of the hearing aid, a probe-tube microphone system was placed in the ear canal, and sound level measurements were made with the hearing aid. Computation of hearing aid delay was done to remove the effect of the hearing aid processing delay while analyzing the latency. Processing delay of the 2 - channel hearing aid was 1.5 ms, and the 4 and 8 channel aids had 1.8 ms and 3.3 ms delays respectively. However, as the delays were very small compared to the standard deviation of the ACC, it was not taken into consideration during final analysis.

Phase 3: Acquisition of aided ACC for clinical group

The participant was comfortably seated on a recliner in the sound treated room. The electrode placement was as mentioned in Table 1 and depicted in Figure 3. Instrumentation in the single sound treated room used is shown in Figure4. Aided ACC waveforms were acquired, with the hearing aid placed on the test ear. The behind-the-ear hearing aid had to be placed in such a manner, that it caused no interference or artifacts as the inverting electrode was also placed on the test ear mastoid. Next, the speaker was positioned so that the hearing aid microphone, relative to the speaker was at an angle of 45^o. A pre-stimulus electroencephalographic recording was first observed, and once the trace was found to stabilize the test recording was initialized. To obtain a baseline of the neurological activity unrelated to the stimulus in the waveform, a prestimulus recording was done for 100 ms. The waveforms were then analysed.

Analysis of the waveforms: The three-aided ACC waveforms (ACC obtained from 2, 4 and 8 channel) were compared based on their respective latency, amplitude and morphology. First, the waveforms were visually judged to be replicable after which peak latency and amplitude measures were made. The waveforms were analyzed by two experts in the electrophysiological measurement of auditory evoked potentials. Absolute amplitude and latency were chosen at the most negative or positive point or halfway point of a broad peak was considered.

Secondly, the morphology was rated based on three point rating scale. A score of 0 indicates poor morphology, 1 and 2 indicates average and good wave form morphology respectively. The data was tabulated for N1, P2, P1' and N1'.

Phase 4: Obtaining the speech identification score in the clinical group

Aided speech identification scores were obtained using the High Frequency Kannada Speech Identification Test (Mascarenhas, 2002). Each channel of the hearing aid was evaluated with one of the four lists consisting of 25-words each; an unaided SIS score was also obtained.

This word list had been developed exclusively for high frequency sloping sensorineural hearing loss. In an acoustically shielded room, the voice of a native Kannada female speaker was recorded. A female voice was used as it has a higher fundamental frequency, and can better tap into perception of high frequency phonemes. Speaker effects were eliminated with the use of words which were recorded using Adobe Audition (version 1.5) with 16 bit resolution. To prevent familiarization, three word list of the original test were utilized along with a fourth, that was developed through randomization of the words in the first list. Prior to each list, a 1000 Hz calibration tone was recorded in each word list, and was used to adjust the VU meter of the audiometer to zero. The words were presented at 65 dB SPL through a diagnostic 2-channel audiometer at 0^{0}azimuth.

The results obtained with the above procedure was subjected to statistical analysis performed using the Statistical Package for Social Sciences (SPSS) (version 16).

Results and Discussion

Comparison of Unaided Scores between Clinical and Control Group

The first objective was to compare the ACC of participants with sloping hearing loss (clinical group) with the participants having hearing sensitivity within normal limits (control group). With a descriptive analysis of the control and clinical group, the mean and the standard deviation values could be obtained and is shown in Table 4. The N1-P2 complex was generated in response to perception of the fricative /s/ and the N1'-P2' complex in response to the vowel /i/.

To study the outcome Mann-Whitney U test was done for comparison of the two groups. The test revealed statistically significant differences between latency of P2 [$/Z/ = -02.405$, $p < 0.05$] which was significantly shorter in the control group. The amplitude of N1' [$/Z/ = -0.377$, $p < 0.001$], was significantly larger in the control group. Although the mean latency of the N1' peak for the clinical group is shorter, it was not found to be significant on statistical analysis.

A noticeable feature depicted in Table 4, which compares the mean latencies of the control and clinical group is the absence of the N1 peak in the clinical group participants. The N1 latency and amplitude values could not be considered for comparison as none of the clinical group participants had this peak present in their waveforms. As the N1 reflects detection of the stimuli at the cortical level (Martin, Tremblay & Stapells, 2007), the absence of the same in the sloping cochlear hearing loss individuals may have arose due to three main reasons. The first reason is the inaudibility of the fricative portion of the /si/ stimulus. From a visual analysis of the /s/ portion shown in Figure 1, it is evident that the fricative portion has lower amplitude than the vowel portion. The first N1-P2 complex is generated in response to the fricative portion of the stimulus and the second N1'-P2' complex is generated in response to the transition from fricative to the vowel. The transition portion is audible to those with sloping cochlear hearing loss as evidenced by the presence of the N1'-P2' complex, but the initial /s/ portion is not audible.

The P2 peak has an earlier latency in the control participants when compared to the clinical group, but, the P2 amplitude was similar for both the groups. This could be because the control group participants could perceive the initial fricative /s/, but, due to reduced audibility there was a delay in latency for the clinical group. There were no differences between the two groups in the amplitude of the P2 peak in the present study, and is comparable to results found by Wall, Dalebout, Davidson and Fox (1991). These results have also been confirmed in literature by Oates, Kurtzberg and Stapells (2002),

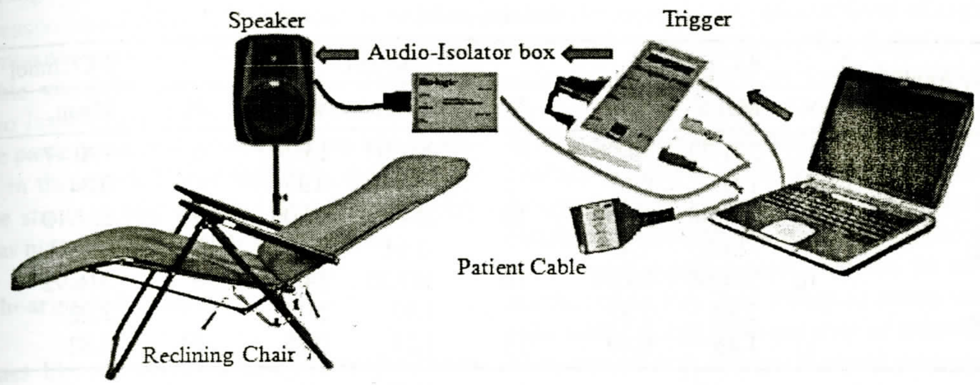

Figure 3: Equipment set-up for ACC measurement.

Table 4: Unaided parameters obtained for the control and clinical groups

Parameter Analyzed	CONTROL			CLINICAL		
	N	Mean	SD	N	Mean	SD
N1 latency (ms)	11	127.05	15.34	-	-	-
N1 amplitude (μV)		-01.35	.925	-	-	-
P2 latency (ms)	20	179.74	26.59	6	201.07	11.68
P2 amplitude (μV)		1.610	.873		1.30	1.05
N1' latency (ms)	20	256.98	11.77	8	244.89	20.50
N1' amplitude (μV)		-2.84	1.20		-2.79	1.73
P2' latency (ms)	20	316.15	20.03	9	317.93	25.93
P2' amplitude (μV)		1.06	1.14		2.67	1.27
Waveform Morphology	20	1.55	0.51	11	1.40	0.843

Note:Lat = Latency (ms), Amp = Amplitude (μv), Morph = Morphology

who found latency measures to be a more sensitive indicator of the effect of decreased audibility than a response strength measure such as amplitude.

Similar findings have been reported by Polen (1984) who found significantly longer latencies for P2 in listeners with hearing loss compared to listeners with normal hearing. The author stated that this could be because this peak is sensitive to stimulus features and a reduction in audibility of these features causes the peak to diminish.

Although the frication portion was audible to the control group, nine of the twenty control group participants did not have the N1 peak in their waveform. Therefore, there could be features inherent to the stimulus

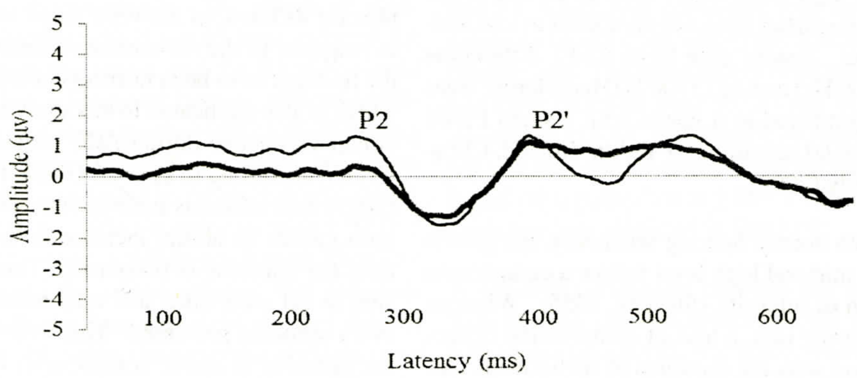

Figure 4: Grand mean ACC waveform of participants with hearing sensitivity within normal limits (thin) and sloping sensorineural hearing loss (thick).

Table 5: Latency, amplitude and morphology (mean and standard deviations for two, four and eight channel hearing aids)

Parameter Analyzed	8-Channel			4-Channel			2-Channel		
	N	Mean	SD	N	Mean	SD	N	Mean	SD
P2 Lat	8	201.72	19.00	8	209.04	13.24	6	190.47	17.21
P2 Amp		1.00	0.67		1.49	0.93		0.84	0.69
N1' Lat	10	251.48	14.62	10	254.63	19.03	9	251.10	17.54
N1' Amp		-2.64	1.49		-2.61	1.10		-2.78	1.50
P2' Lat	10	314.00	20.58	10	317.33	24.20	9	316.92	27.80
P2' Amp		2.30	1.37		1.97	1.44		2.22	1.29
Morph	11	1.45	0.69	11	1.28	0.79	11	0.82	0.75

as in how the vowel and consonant pair interact or the way the central auditory system responds to high frequency stimuli that may have caused the N1 peak to be absent. To explain further, the second reason considers the masking phenomenon; individuals with cochlear pathology are more prone to the effects of upward spread of masking due to wider auditory filters, which in turn results in reduced frequency resolution (Moore, 1998).

The vowel portion of the stimulus could have sufficiently masked the fricative causing a response to arise only from the vowel portion. Low frequency portions of speech like vowels may mask higher frequency components like frication noise (Dillon, 2001). The /i/ portion being larger in amplitude could have masked the /s/ portion of the stimulus used in this study. The third reason is the inherent lower amplitude of the cortical response when elicited from stimulus containing high frequency content such as /ʃ/ and /s/ compared to /m/, /a/, /u/ and /i/ which had predominant low frequency content as documented by Agung, Purdy, McMahon and Newall (2006). The N1 amplitude in their study was significantly smaller for the stimuli containing high frequency content. As responses evoked by a high frequency stimulus have inherent lower amplitude, an additional cochlear pathology could have resulted in an elimination of the response.

The vowel /i/ is capable of eliciting earlier latency. This is because the high front vowel such as /i/ can evoke CAEPs that have earlier latencies compared to low mid-back vowels /u/. Vowels with large F2-F1 differences such as /i/ (2̃300 Hz) and /u/ (1̃700 Hz) have larger areas of activation compared to a vowel with a small F2-F1 distance such as /a/ according to Yetkin, Roland, Christensen and Purdy (2004).

In listeners with normal hearing sensitivity, the growth of loudness at mid and high level follow a compressive power function of intensity (Stevens, 1955). Whereas, in cochlear hearing loss, a loss of compression occurs, this is consistent with the presence of recruitment. According to Buus and Florentine (2001), this growth of loudness occurs near threshold; on average for every 16 dB of hearing loss, the loudness near threshold doubled. The finding of a larger P2' amplitude in the clinical group can be explained based on the differences in the growth of loudness between the control and clinical group. The increase in the amplitude of P2 in the clinical group may be attributed to recruitment which could have occurred as a result of cochlear hearing loss. The grand mean waveforms of the control and the clinical group ACC has been depicted in Figure 5. The amplitude differences have been shown in Table 4. The second objective was to compare the unaided and aided ACC in individuals with sloping cochlear hearing loss. Comparison between the three aided conditions and the unaided condition was done. The aided conditions considered three hearing aids comprising two, four and eight channels.

Comparison of the unaided and aided conditions within the clinical group

Wilcoxon Signed Ranks Test result did not reveal any significant differences between the aided (two, four and eight channels) and unaided conditions ($p > 0.05$). Therefore, for individuals with sloping cochlear hearing loss, the performance on the ACC did not differ with amplification or without it. While studying the neural representation of amplified speech sounds, similar findings were reported by Tremblay et al., (2006). An increase in hearing aid gain did not result in an amplitude increase or latency decrease as compared with unaided ACC at the same input level. They found no significant differences between aided and unaided ACCs in response to the /si/ stimulus. Reasons suggested for the findings have been hearing aid compression effects, which is also applicable in this study as the compression circuit was active during ACC acquisition. To explain further, according to Easwar, Glista, Purcell and Scollie (2012) inter stimulus interval between one and two seconds causes an abrupt increase in the input level each time the stimulus is presented. This causes compression to act each time, and may result in overshoot for every stimulus presented. This overshoot occurs when the stimulus is above compression threshold, because the hearing aid requires time to stabilize the gain when there is a rapid change in the input level.

Another reason could be that the amount of response change (improvements) seen in the ACC demonstrated considerable variability across subjects. This could have led to insignificant findings in the current study. One of the participants in the clinical group had no visible ACC in the unaided condition or in the aided although the stimulus was audible. The finding of this subject was not included for statistical analysis.

Effect of hearing aid channels

Speech cues like transitions with dynamic frequency changes are not perceived well even when audibility is provided. Zeng and Turner (1990) stated the loss of transition cue perception which occurs in sensorineural hearing loss cannot be compensated by hearing aids. In contrast, the hearing aids in the present study could reflect the transition portion as evidenced by the prominent ACC N1'- P2'peak.

Each peak obtained in the ACC, was analyzed for latency, amplitude and morphology. The descriptive statistics has been shown below; in Table 5. To learn the effect of hearing aid channels on the central auditory system, the ACC latency, amplitude and morphology across 2, 4 and 8 channels were compared. The Friedman's test was used and the results obtained have been discussed below. No significant differences were found between the two, four and eight channels in terms of latency and amplitude; this means that the three hearing aids had equivalent performance. The same has been shown in Figure 6 and Figure 7. It has been reported that the vowel-consonant differences will be reduced due to spectral contrast reduction caused by a higher number of channels (Plomp, 1988).

Vowels are more susceptible to this effect than consonants. Consonants in the initial position are more susceptible than those in the final position (Boothroyd, Mulhearn, Gong & Ostroff, 1996). However, the results of this study contradicts the findings of Plomp (1988) and Bor et al. (2008), as the two, four and eight channel hearing aids were not found to have significant differences in latency and amplitude. The effects of spectral contrast reduction which would have reduced the difference between the vowel and the fricative would have otherwise led to increased latency, reduced amplitude and poor morphology for the eight channel hearing aid. Among the four parameters analyzed, the waveform morphology was found to be statistically significant across the three aided conditions with $\chi^2(11) = 7.00$, p = < 0.05. Next a Wilcoxon Signed Rank test was conducted and the results showed that morphology obtained with the four and eight channel devices were equivalent. However, between the eight and two channel hearing aids there was a significant difference in morphology, /Z/ = -02.111, $p < 0.05$. The mean of the ranks in favor of a 2 channel hearing aid were 4.00, while the mean of the ranks in favor of eight channels were 4.57. From the descriptive statistics, mean value of the 8 channel hearing aid was higher than that of the 2 channel hearing aid. Therefore, it can be concluded that an 8 channel hearing aid results in better waveform morphology relative to the 2 channel hearing aid, Figure 8 compares the grand mean waveforms of the three aided conditions. Since poor waveform morphology may not have a physiological basis, and can be affected by a number of factors from sweep to sweep such as muscular artifacts, eye blink and state of arousal, it may not be consistent with changes in signal processing.

Multichannel hearing aids do not have rectangular analysis bands and compression channels, but instead the actual analysis bands and compression channels in hearing aids are non-rectangular, and may have very gradual filter slopes. The more gradual the filter slope of the channels,the compression across channels will be correlated or similar. Therefore, a hearing aid with higher number of channels may be functioning similar to a hearing aid with lower number of channels. This could be a probable reason for why no significant changes were noted across two, four and eight channels.

Another reason for no significant differences across channels could be due to the compression circuit of the hearing aid. According to Korczak, Kurtzberg and Stapells (2005), when the compression circuit is active, a ceiling effect occurs for aided responses at higher intensities. Therefore, subtle variations in signal processing brought about by an increase or decrease in the number of channels, may not be accurately reflected if other processing schemes such as compression are active.

Correlation between speech identification scores and ACC measures

Non-parametric correlations were administered to achieve the forth objective which was to compare the SIS and ACC across different number of hearing aid channels. Spearman's correlation was done, considering the peaks in the ACC waveform namely; P2, N1' and P2' which was correlated with the speech identification score obtained using two, four and eight channel hearing aids. The mean scores obtained for the two, four and eight channels were 83, 84 and 81% with standard deviations of 8.48, 10.90 and 9.78 respectively. The difference between the unaided and the aided condition approached significance at $\chi^2(8) = 7.541$, p = 0.057. The same has been depicted in Figure 9.

A positive correlation between the amplitude of P2' with the four channel device and the speech identification scores ($\rho = 0.822$, $p < 0.05$). The failure of the two and eight channel devices to show similar findings could be attributed to the inherent redundancy of words compared to a syllable. Word discrimination improves with

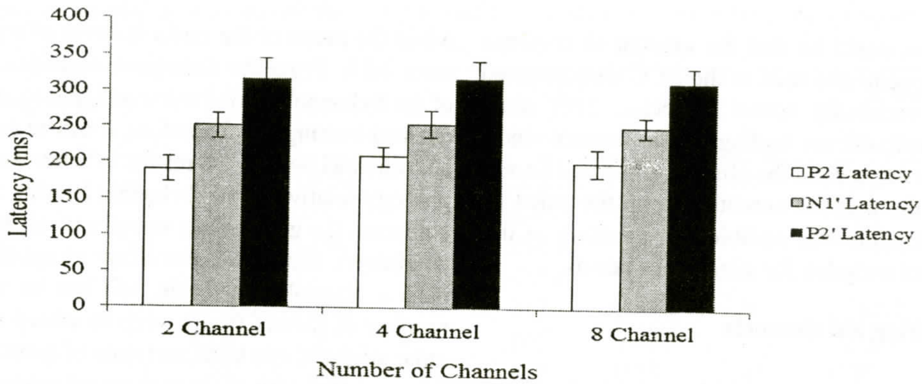

Figure 5: Mean latencies for the three aided conditions. Error bars denote one standard error around the mean.

increase in word length (Black, 1952). The speech identification material used in the present study consisted of monosyllabic words and the ACC stimulus was a single syllable /si/. The difference in redundancy between the two stimuli could have resulted in a lower correlation between the two. Subjective differences in perception through hearing aids with varying number of channels may also have be a reason. This needs to be clarified by further research in the same area.

In contrast to the present study, behavioral measures of discrimination and its correlation to CAEP have been assessed by Korczak et al., (2005). This study has found a correlation between the behavioral or discrimination score and the electrophysiological testing.

A reason for no significant changes seen between the number of channels and speech identification scores could be that an increase in the number of channels negatively affects the perception of diphthongs and vowels (Bor et al., 2008) and not to a large extent the perception of consonants according to Boothroyd et al., 1996. High Frequency Kannada Speech Identification Test (Mascarenhas, 2002) was used to assess the speech identification score.The high frequency content in the wordlist was contributed by consonants which were semivowels (/j/, /r/, /l/), stops (/t/, /p/, /k/), fricatives (s, /f/, /ʃ/) and the affricate /tʃ/. The consonants themselves could have sufficiently improved intelligibility without the need for depending on the vowels for perception. Therefore, only minor variations in the scores were seen across different number of channels.

Conclusion

When compared to individuals with hearing sensitivity within normal limits, those with cochlear hearing loss have delayed latencies on the ACC; amplitudes were found to be similar between the groups. For sloping cochlear hearing loss, with or without amplification the responses were similar. When the two, four and eight channel hearing aids were evaluated electrophysiologically through the ACC, the performance was equivalent. When the electrophysiological findings were correlated with the speech identification scores for each aid, ACC obtained with the four channel device was found to correlate with the SIS; but, there was no correlation found between the two and eight channel hearing aids.

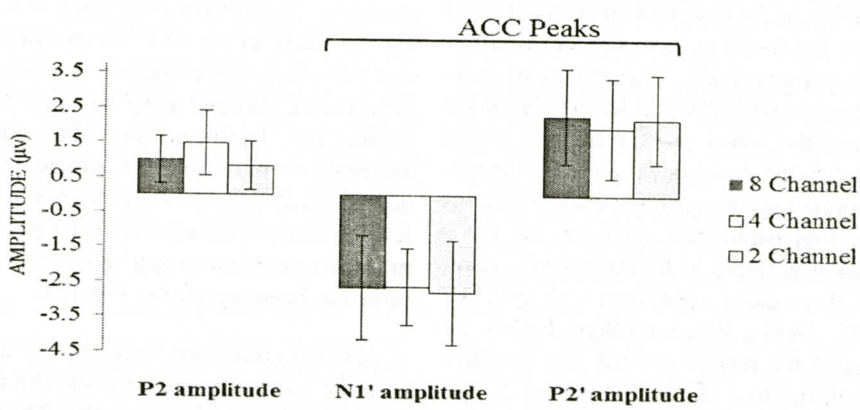

Figure 6: Mean amplitudes for the aided condition of two, four and eight channels. Error bars denote one standard error around the mean.

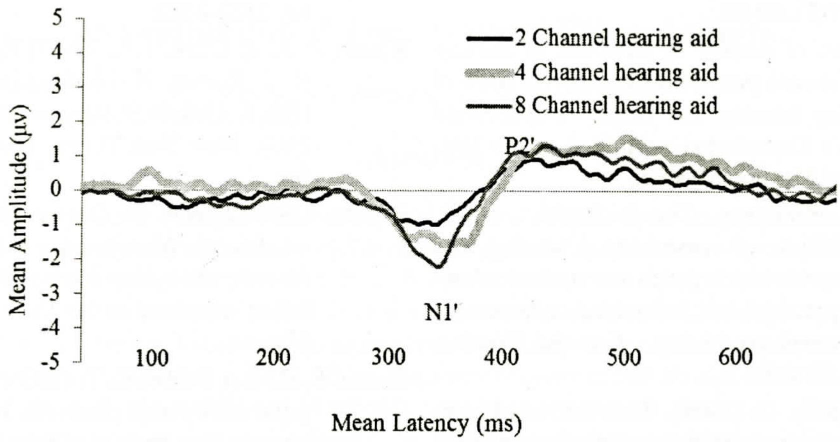

Figure 7: Grand mean waveform of the ACC obtained for the aided condition using the two, four and eight channel device.

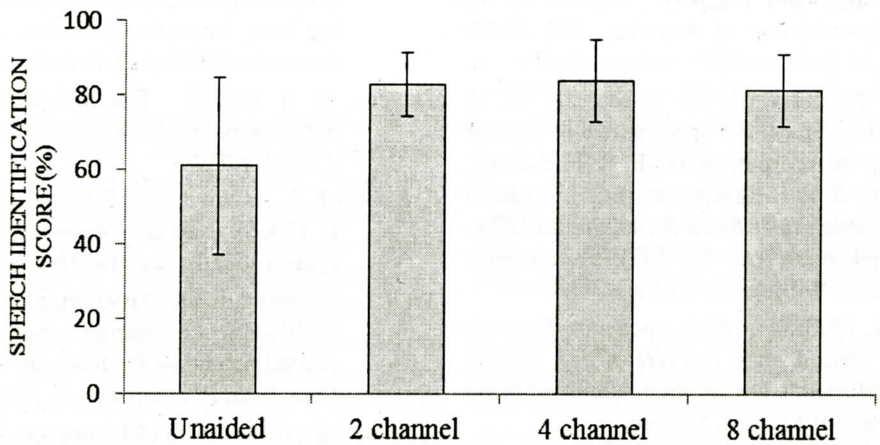

Figure 8: Mean speech identification scores obtained for different conditions.Error bars denote one standard error around the mean.

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