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ARTICLES

INFORMATION

EDITORIAL

The year 2020 is earmarked as International Year of Sound. In this International Year of Sound, it is a great privilege to bring out a special issue of JASI on "Acoustic analysis and speech processing for communication disorders".

Any abnormality in the speech production system, pertaining to the vocal folds / articulators, or learning environment and habits leads to speech disorders. Any defects in the hearing mechanism, congenital or acquired, will lead to hearing disorders. The umbrella of communication disorders encompasses both speech and hearing disorders. There exists a gap between the clinical requirements and current research scenario in the field of communication disorders. The manuscripts published in this special issue will help to reduce this gap and would ignite the research minds, both budding and experienced ones, to initiate further research which would finally result in enhancing the quality of life of persons with communication disorders.

After careful reviews by experts, five manuscripts were selected for this special issue. In the first manuscript, Amrin T. K., Linda Sara Babu and Arivudai Nambi P. M., have investigated the association of overshoot with contralateral suppression of Transient Evoked Oto Acoustic Emission (TEOAE) and selective attention. Results suggest the role of Medial Olivocochlear Reflex (MOCR) and selective attention in the psychophysical mechanism of overshoot.

The second manuscript, authored by Hrishitha V. M., Vibhuti Sharma, Neha Mansoor Ali, Jeslin Jose and Prashanth Prabhu, is a preliminary study to find out whether martial art training improves binaural integration. The study attempted to determine differences in binaural integration abilities using the dichotic consonant - vowel (CV) test in children with and without martial arts training. The results of the study show that the dichotic CV test scores were higher in individuals with martial arts training, in comparison to those without training.

The 3rd manuscript describes a study by Ejaz Ahemed G. P., Sai Samyuktha V., Saraswathi Thupakula and Sunil Kumar Ravi. This study is aimed to establish normative data on dysphonia severity index for Telugu speaking Indian geriatric population. Results of this study will be useful in setting up a reference for measuring the severity of dysphonia.

The 4th manuscript is an investigation by Theaja Kuriakos on the rhythmic pattern in the speech of young adults with congenital hearing impairment. Results show that speech of young adults with hearing impairment has an altered rhythmic pattern.

The last manuscript of this special issue is a scoping review of Ear-sensored EEG monitoring techniques in hearing aids. This article by Nisha K V provides a better understanding of the rationale for implementation, current trends, and future research applications of Ear-sensored EEG monitoring techniques in hearing aids.

As I conclude this overview, I would like to thank Dr. Biswajit Chakraborty, Chief Editor of JASI for providing me the opportunity to serve as the editor of this special issue. My sincere gratitude is due to all those reviewers, whose names are listed below, for their time, efforts and valuable suggestions, which helped to enhance the quality of selected manuscripts. Let me also acknowledge all the authors who contributed their manuscripts for this special issue.

I wish you a pleasant reading of this issue.

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Role of efferent suppression and selective attention in psychophysical overshoot

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ABSTRACT

Current study investigated the association of overshoot with contralateral suppression of TEOAE and selective attention.16 adults participated in the present study and their TEOAEs were measured. These measurements were performed without and with contralateral acoustic stimulation (CAS); with 2750 Hz NBN masker as CAS. The difference in TEOAE amplitude with and without CAS was calculated, to measure the magnitude of CAS suppression. Later, two methods were used to determine overshoot. Two conditions were included in the first method, where a tone was presented 2 ms and 202 ms after the onset of noise. The magnitude of overshoot was estimated by taking the difference in tone detection thresholds between the two conditions. In the second method, a 10 ms long tone and masker were concurrently presented without and with the presence of a preceding precursor and the difference in the threshold in these two conditions was taken as magnitude of overshoot. Selective attention ability was assessed using a stroop-task. Pearson's correlation analysis revealed a significant positive correlation of overshoot with TEOAE suppression and stroop reaction time. Results of the current study suggest the role of MOCR and selective attention in the psychophysical mechanism of overshoot.

1. INTRODUCTION

The olivocochlear bundle (OCB) or the descending fiber tract comprising of lateral and medial efferent neurons originate from the superior olivary complex (SOC) and innervate the cochlea[1]. The inner hair cells of the cochlea are innervated by the efferent nerve fibers of the lateral olivocochlear bundle, whereas the medial olivocochlear (MOC) efferents directly terminate at the outer hair cells (OHC). MOC efferents modify the action of the OHCs and thereby controls the gain of the "cochlear amplifier"[2]. Activation of the MOC bundle results in hyperpolarization of the OHC, resulting in inhibiting OHC electromotility and reduction in the cochlear amplifier gain[2]. Therefore MOC fibers function as a protective mechanism during intense sound exposure to the cochlea[3]. Also, investigators have shown that activation of the efferent pathway enhances the ability to discriminate speech in noise[4]. In the presence of noise, the MOC pathway helps in improving the response to transient speech signals through an unmasking effect. When background noise is present along with speech, the MOC activation can reduce the baseline rate of firing, [2] resulting in an increased dynamic range of the auditory nerve fibers[5]. The increase in the dynamic range helps to improve the ability of the auditory nerves in encoding speech signals. Studies have also shown that the MOC bundle also helps in perceptual learning of non-native contrasts. The feedback from MOCB may fine-tune the brainstem or cochlea during the process of learning[6].

The olivocochlear feedback mechanism has several implications in auditory perception, among which speech perception in noise is well known. However, recently, the role of the olivocochlear function in the psychophysical mechanism of overshoot has gained popularity[7]. The ability of human listeners to detect a tonal signal in the presence of noise depends upon the relative timing between the onset of the noise and the onset of the signal. When compared to the stimulus presented directly after the onset of noise, there is an enhanced ability to detect the tone on presenting the stimulus with a delay of few milliseconds after the noise onset. This phenomenon, known as psychophysical overshoot,[8] occurs when there is a difference in the spectrum of the masker and the signal. The psychophysical overshoot is observed in simultaneous masking wherein the only onset of the target signal is delayed with reference to the masker. However, the offset of the target signal is aligned with the offset of the masker. The magnitude of psychophysical overshoot can be as large as 15-20 dB[8]. The magnitude of overshoot is observed to be most significant when the tonal signal has a duration of a few milliseconds, and the frequency of masker is higher than the maskee[9,10].

Various hypotheses have been put forward to explain the mechanisms of overshoot, which include the "classic firing rate (CFR) adaptation"[11] and "dynamic range (DR) adaptation" via the MOCR[12]. According to the CFR adaptation hypothesis, there is a substantial increase in nerve firing throughout acoustic stimulation leading to the adaptation of neural firing. Therefore the masker effect is reduced a few milliseconds after the onset. When a signal is presented as the adaptation occurs, there is an improvement in the signal detection. CFR adaptation hypothesis predicted a 3-5 dB of improvement in threshold[13]. DR adaptation via the MOCR, explains the overshoot as a result of reduced cochlear amplifier gain. According to this hypothesis, the cochlear gain is high at masker onset, and after 100 ms duration, it decreases to a plateau[14]. The DR adaptation occurs when MOC is activated by an acoustic stimulus that is presented for a longer duration. There is a reduction in the cochlear amplifier gain leading to a decreased response to the low-level noise than compared to the response to the brief, high-level tone[12].

Reduction in cochlear gain mediated by MOCR efferent reflex may play a role in overshoot[7,15]. Nevertheless, Fletcher, de Boer and Krumbholz[16] speculated that mechanism other than efferent MOCR might be responsible for overshoot. Therefore, there is no clear cut consensus on the role of MOCR in overshoot. Studies that describe the overshoot as a function of the MOCR are based on the computational models[17-19]. However, no researches have measured the MOCR magnitude and associated it with overshoot. The suppression paradigm of TEOAE provides a means for studying the magnitude of MOCR. The relationship between overshoot and MOCR will be understood better by studying the association between the TEOAE suppression and the magnitude of overshoot. Factors like cognition may also assist in the mechanism of overshoot. Cognition, especially selective attention, may play a role in overshoot because the task of overshoot requires the participants to ignore the noise and detect the tone. However, researches have not been focused upon the role of cognitive ability in overshoot. We hypothesize that individuals with poor selective attention ability may find it difficult to selectively attend to the tone when it is presented immediately after the noise. Hence, individuals with poor selective attention ability may show a large overshoot. This hypothesis also needs to be verified. Hence the current study was taken up to investigate the association of TEOAE suppression and selective attention with the magnitude of overshoot.

2. METHOD

2.1 Participants

Sixteen participants within the age range of 18 and 25 years (mean age of 21.5 years) participated in the current study. Participants did not possess any gross otological and/or neurological problems. All the participants had a hearing threshold of ≤15 dBHL across the frequencies from 250 Hz to 8000 Hz in both the ears. Also, it was ensured that the otoscopy findings and middle ear functions are normal with 'A' type Tympanogram in both ears.

2.2 Procedures

Measurement of TEOAE and contralateral suppression : The subjects were seated comfortably in a sound attenuated room. TEOAEs were recorded using ILO software (Version 6.41.27.0). 400 sets of tone pip were presented at 60 dBpeSPL. The stimulus had a frequency of 2750 Hz and a rise-plateau-fall of 2-1-2 cycles. Responses were averaged using a linear mode of averaging. Deep insertion of the probe, along with the minimal movement of the subjects, was ensured during the recordings. A total of four TEOAEs were recorded, with and without contralateral acoustic stimulation (CAS). The noise was presented to the contralateral ear using Eartone 3A Insert earphones, which was routed through Creative sound blaster X fi USB2 sound card.NBN masker with the center frequency of 2750 Hz and intensity of 55dBSPL was presented as the contralateral acoustic stimulator. The NBN had a bandwidth of 321 Hz, which is approximately equivalent to 1 ERB (Moore & Glassberg, 1990). Initially, the baseline was recorded, and two TEOAE measurements with CAS were done. A final baseline recording then followed it. In order to eliminate the post-stimulation effect of the previously presented contralateral noise, a time gap of 15-20 seconds was provided between the second, third, and fourth TEOAE recordings.

Measurement of overshoot : Followed by the OAE measurement, each participant had to undergo psychophysical tests to estimate overshoot. The magnitude of overshoot was measured in each subject using two methods- the temporal effects in simultaneous masking leading to overshoot were evaluated in Method 1, and the effect of precursor on overshoot was evaluated in Method 2. Method 1 will be referred to as a continuous method in the following sections. In the continuous noise method, an NBN masker centered at 2750 Hz with duration of 250 ms, and a 2 kHz tone with a duration of 10 ms was used. Both masker and the tone had a 2 ms raised cosine envelope at the onset and offset. A 3 alternative forced-choice task was used in which the noise was presented at three different intervals, of which one of the noise intervals included presentation of tone 2 ms after the onset of the masker. A similar task was done with the tone presented 202 ms after the noise onset. The schematic representation of the test stimulus is given below.

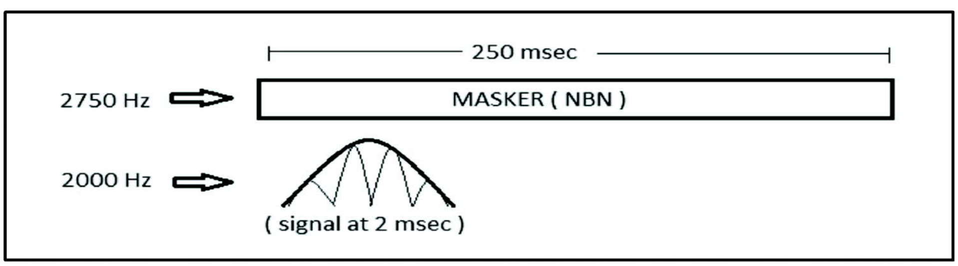

Fig. 1. Schematic representation of the signal being presented 2 ms after masker onset

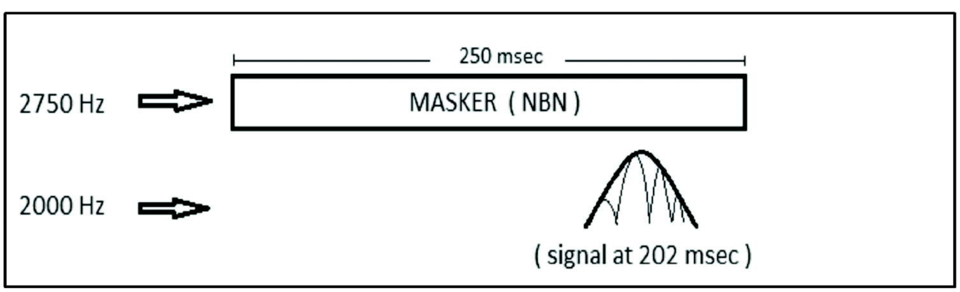

Fig. 2. Schematic representation of the signal being presented 202 ms after masker onset

The subjects were instructed to correctly identify the noise interval in which the tone was present, and the responses were recorded. Masker was presented at a fixed intensity of 55 dBSPL, and the level of the tone was adaptively varied with the 2-down 1-up psychophysical procedure[20]. The intensity was adjusted in the ratio step size of 1.1.Midpoints of the last six reversals out of 7 reversals were geometrically averaged to get the thresholds. The detection threshold for tone was estimated as a function of onset delay of the tone with reference to masker onset. The magnitude of the overshoot was estimated as the difference in the threshold at 2 ms and 202 ms stimulus onset delay. Overshoot estimated with this method will be referred to as "overshoot- continuous".

Method 2 will be hereafter referred to as the precursor method. The tonal detection threshold in the presence of masker was estimated with and without the presence of precursor. Masker and tone spectrum was similar to that used in the continuous noise method. Both masker and tone had a duration of 10 ms with 2 ms raised cosine onset/ offset envelope. Masker was presented at a fixed intensity of 55 dBSPL, and the level of the tone was varied with the Adaptive 2-down 1- up psychophysical procedure, similar to that of the continuous method. The spectrum of the precursor was also similar to the masker. However, the duration of the precursor was 250 ms with 2 ms raised cosine onset/ offset envelope. Overshoot was calculated by considering the difference in thresholds obtained through two different conditions. In condition one, the masker and the tone were presented simultaneously, and in condition two, a precursor was presented prior to the simultaneously presented masker and the tone (Figure 3). The precursor preceded the masker, and there was no time delay between the offset of the precursor and the onset of the masker.

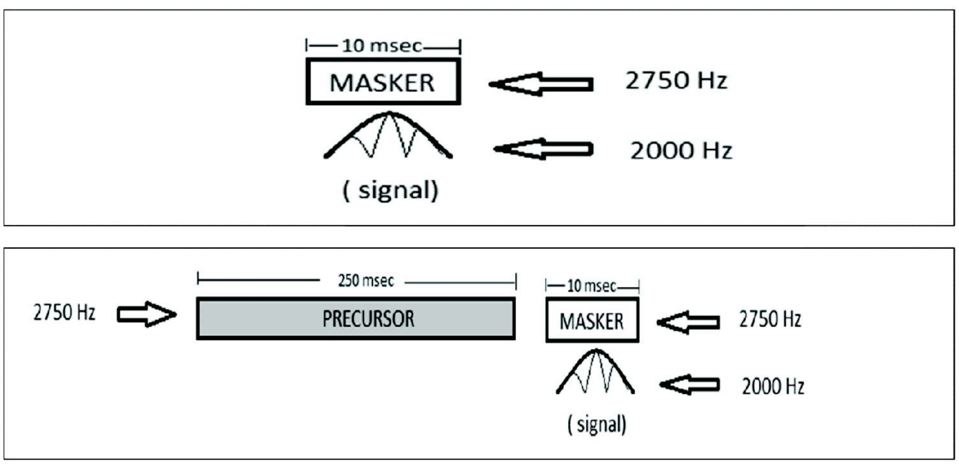

Fig. 3. Schematic representation of masker and signal presented simultaneously (Top panel) and precursor presented prior to masker and signal onset (Bottom panel)

To estimate the tonal detection threshold in both the conditions, the masker and the tone was simultaneously gated on/off. Subjects were instructed to correctly identify the noise interval that had the tone and the responses were recorded. The magnitude of overshoot was taken as the difference in threshold with the presentation of a precursor and without precursor. The magnitude of overshoot obtained from this precursor method will be referred to as "overshoot- precursor", in the following sections.

Measurement of selective attention : Stroop task was used to assess selective attention ability. With the help of custom written script in PsychoPy2 was used to perform stroop task. Stimuli used for the task were masculine and feminine voices saying the word 'male' or 'female', that were presented through ER3A insert ear phones which was connected to a Creative sound blaster X-fi USB 2 sound card, that was in turn connected to the computer. Irrespective of the word that was heard, the subjects were asked to press the left arrow in the computer when they hear the female voice and to press the right arrow for the male voice. The subjects were presented with 20 test tokens. Tokens were presented at most comfortable level.

3. RESULTS AND DISCUSSION

TEOAE responses were analyzed offline using customized software. FFT analysis was performed on the averaged TEOAE response. Amplitude of TEOAE response was estimated in the frequency range of 2789 Hz - 2911 Hz, as this frequency range corresponds to 1 ERB centered at 2750 Hz. The suppression magnitude was calculated by finding the difference of TEOAE amplitude with and without suppressor. Stroop reaction time was calculated for 20 tokens. Reaction times for correctly identified tokens were averaged to get a single estimate of reaction time for each subject. The mean and standard deviation of stroop reaction time was 0.90 seconds and 0.16 seconds, respectively.

Shapiro - Wilk test of normality was done to investigate the normal distribution nature of the data sets. Results revealed that the TEOAE suppression ($W=0.860$, $p>0.05$), overshoot-precursor ($W=0.606$, $p>0.05$), overshoot- continuous ($W=0.265$, $p>0.05$) and the stroop reaction time ($w=0.953$, $p=0.577$) were normally distributed. The mean and standard deviation scores for TEOAE suppression is 2.035 ± 0.172. Figure 4 represents the mean and standard deviation scores for magnitude of overshoot obtained in continuous and precursor method.

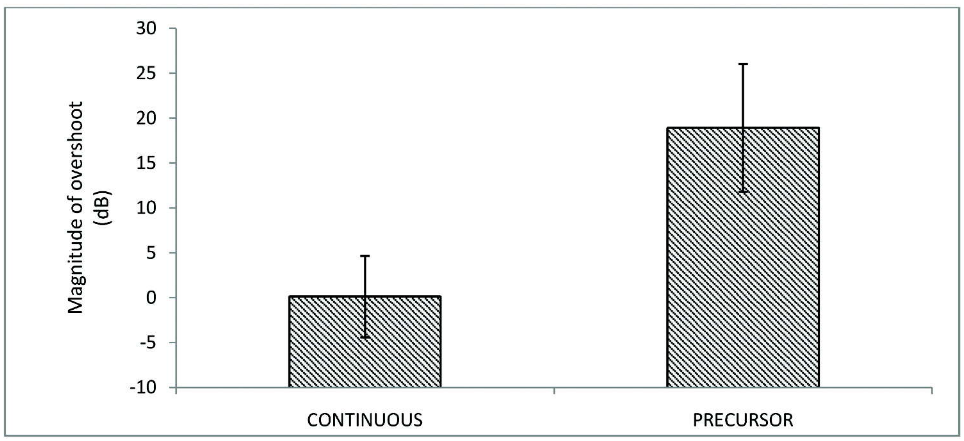

Fig. 4. Bar graph showing magnitude of overshoot estimated using continuous and precursor method. Error bars represent ±1 standard deviation.

Paired sample t-test was used to check the difference between the data obtained from overshoot precursor and continuous overshoot method. Results revealed that the magnitude of overshoot estimated using both the methods were significantly different ($t=-10.651$, $p=0.00$). Overshoot obtained in the overshoot precursor condition was significantly higher than that of overshoot continuous condition.

Pearson's correlation analysis was done to investigate the association between the magnitude of overshoot and the magnitude of suppression of TEOAEs and stroop reaction time. Analysis revealed a significant positive correlation between suppression magnitude and magnitude of overshoot obtained from the precursor method ($r=0.5513$, $p=0.013$).Whereas results did not show any significant correlation between the suppression magnitude and magnitude of overshoot obtained without the precursor ($r=0.212$, $p=0.215$).

Analysis also revealed a significant positive correlation between the stroop reaction time and overshoot estimated using the precursor method ($r=0.476$, $p=0.036$), which is depicted in figure 6. Linear Regression Analysis indicated that 23% of the variance in overshoot could be explained by variance in the stroop reaction time. There was no significant correlation observed between stroop reaction time and overshoot estimated using continuous method ($r=-0.11$, $p=0.35$).

One of the objectives of the current study was to investigate the association between TEOAE suppression and the magnitude of overshoot. A significant correlation was observed between suppression

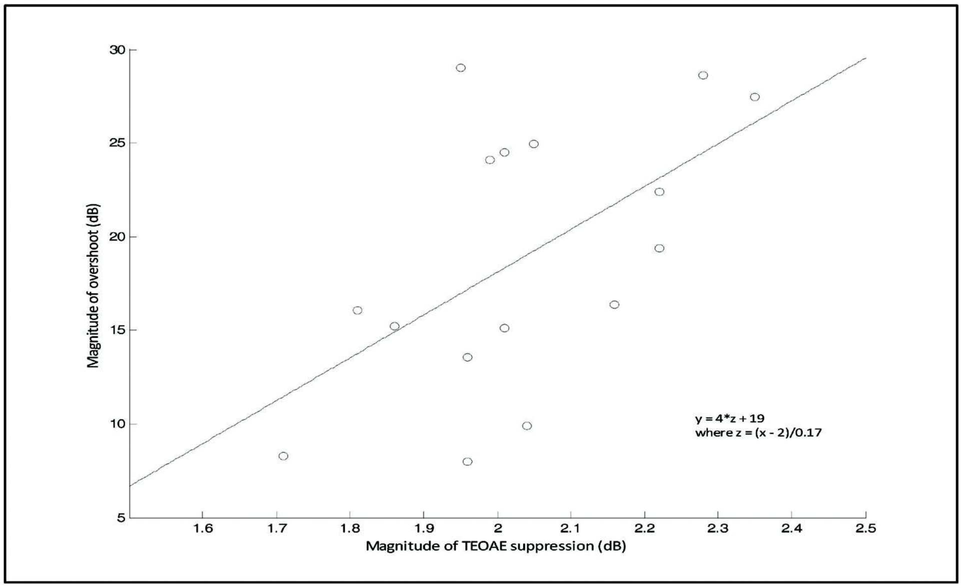

Fig. 5. Scatter plot depicting the correlation between magnitude of TEOAE suppression to that of overshoot

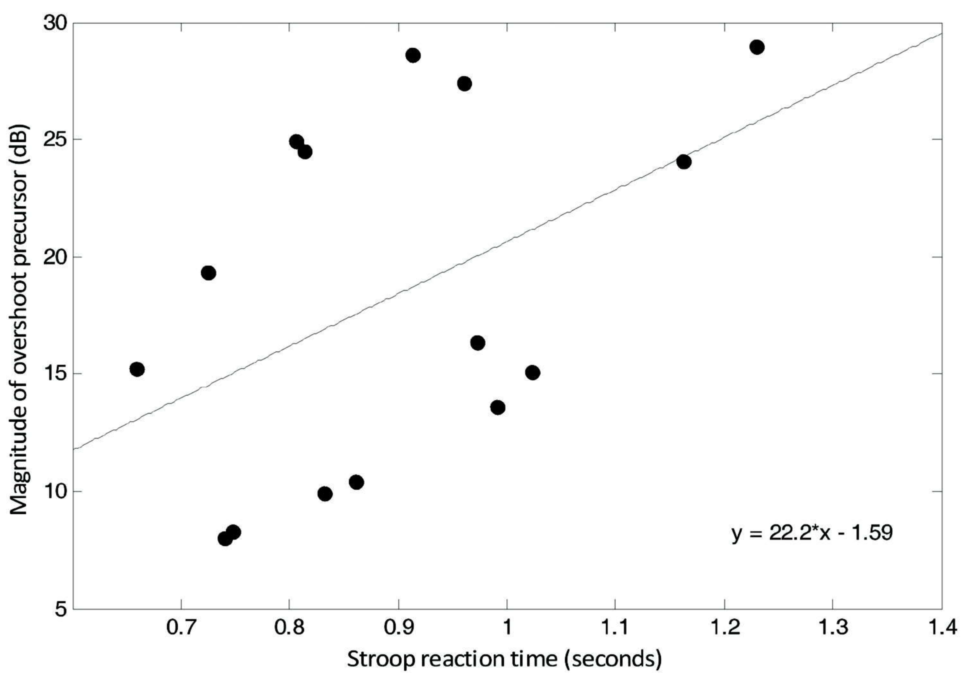

Fig. 6. Scatter plot showing the correlation between the stroop reaction time and overshoot estimated using the precursor method.

magnitude and magnitude of overshoot - precursor and not with the overshoot - continuous. Jennings, Heinz, and Strickland[7] used computational modeling to evaluate the physiological mechanism underlying the mechanism of psychophysical overshoot. In this study they have predicted the tone pip detection in the presence of noise. To predict the detection threshold, the analysis variables such as range of characteristic frequencies, auditory nerve fiber spontaneous rate (SR) pooling, and number of synapses per characteristic frequency, analysis window duration, and detection rule were used. The model could predict the magnitude of psychophysical overshoot similar to the magnitude of overshoot obtained in listening experiments. However the predicted overshoot approximated the psychophysical overshoot obtained in listening experiments only when the MOC reflex is included in the computational model. Even though the model could explain the CFR and DR adaptation in auditory nerve response, the MOC feedback did not produce overshoot when it was not included in the model.

Similarly, Chintanpalli, Jennings, Heinz and Strickland[15] conducted a study using the computational model for auditory nerve responses to evaluate the physiological mechanism resulting in overshoot. In this study, the tone-in-noise responses were predicted as a function of noise level, tone level and OHC gain. The study could predict the MOCR ability to decompress the AN fiber function as there was a decrease in OHC gain that reduced the response to noise and increased the discharge rate to the tone. This improves the tone response in noise. The studies based on models that supports the MOCR role in overshoot, is empirically backed up by the results of this current study. However, there is only one study that contradicts the role of MOCR in overshoot[16]. In this study, using two experiments the validity of the gain reduction model was investigated. Experiment 1 predicted the contribution of excitatory and suppressive masking to overshoot and the second experiment evaluated the effect of precursor on cochlear suppression using forward masking paradigm. Study findings did not show any reduction in suppression by the precursor. However, large suppression and overshoot was observed. These results contradicted the gain reduction model indicating that overshoot is caused by a mechanism other than the MOCR pathway. The investigators did not use either a computational model or physiological method to explain the mechanism of overshoot. Therefore the method used in this study may investigate suppressive masking in overshoot but not the role of MOCR in overshoot.

The overshoot estimated using precursor method in the current study was partly similar to the above said study[16]. However, results of the current study revealed a positive correlation between TEOAE suppression and overshoot. A 30% in variance overshoot has been due to suppression of TEOAE. This indicates that the reduction in the cochlear gain for masker due to the efferent feedback partly contributes to overshoot. Ipsilateral stimulation results in bilateral efferent reflex. Overshoot may be mediated by efferent reflex suppressing the cochlear response in same ear. In the current study strength of the reflex was monitored in the contralateral ear. Contralateral suppression strength was used to predict the ipsilateral reflex effect. Hence, results of this study should be cautiously interpreted. One interesting finding in the current study is that overshoot estimated using two methods were different, which gives the notion that the overshoot measured in two methods may be mediated through two different mechanisms.

Detection of target signal in background noise highly depends on the ability of the auditory system to segregate target signal to one stream and background noise to another stream[21]. Various researches show that selective attention plays an important role in this auditory stream segregation [22-24]. Short stroop reaction time indicates good selective attention ability. In the current study, individuals with good selective attention ability exhibited smaller overshoot. Whereas, individuals with poor selective attention ability exhibited larger overshoot. Individuals with poor selective attention may have problem in selectively attending to the tone when it is simultaneously presented along with the masker for short duration, resulting in poor tone detection threshold. The ability of an individual to detect the tone in the presence of a masker, might be enhanced by the precursor, which might serve as a reference for masker. This results in large overshoot in individuals with poor selective attention ability, as they find it difficult to detect the tone in noise, in the absence of a precursor. Regardless of the presence of precursor, persons with better selective attention might exhibit better tone detection in the presence of masker, by ignoring the masker. This results in smaller overshoot.

There are literatures which support the fact that selective attention can influence till the level of the cochlea,[25] lower brainstem and the auditory nerve[26-28]. There is speculation that the effect of selective attention on lower structures was mediated through MOCR. Role of MOCR in psychophysical overshoot is well documented[7,16]. Since, there is an association between strength of MOCR and selective attention ability, it is reasonable to expect that selective attention and overshoot are related.

4. CONCLUSION

The present study evaluated the association of magnitude of overshoot with contralateral suppression of TEOAE and the stroop reaction time. The TEOAE suppression correlated with overshoot-precursor indicating that the mechanism of overshoot is partly mediated by MOC activity. A positive correlation between stroop reaction time and overshoot precursor indicates that for improved tone detection in the presence of masker, a precursor is required in persons with poor selective attention.

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Does martial art training improve binaural integration? — A preliminary study

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ABSTRACT

Martial arts are combat sport which trains the body to improve the physical, psychological, tactile, and technical skills. The different senses, including auditory attention, are reported to be better in individuals with martial arts training in previous studies. Binaural integration, which involves fusing and separating information from both ears simultaneously, requires active attention, may also be enhanced in individuals with martial arts training. Thus, the study attempted to determine differences in binaural integration abilities using the Dichotic CV test in children with and without martial arts training. Binaural integration was assessed using the Dichotic CV test in children with and without martial arts training. The double correct scores were compared between the two ears. It was also attempted to correlate the double correct scores with the number of years of martial arts training. The results of the study showed that the dichotic CV test scores were higher in individuals with martial arts training compared to the control group. The results also suggested that there was a good positive correlation between dichotic scores and years of martial arts training. The results agree with previous studies, which also suggest a superior performance in individuals with martial arts training due to enhanced auditory attention. Further studies on a larger group of population are essential for better generalization of the results.

1. INTRODUCTION

The central auditory nervous system has an important role in the process of listening. The detection of the stimulus alone does not improve the act of listening; several other neurobiological processes can be measured in response to this stimulus through the electrophysiological auditory potentials recordings and the observation of physiological mechanisms involved in auditory behaviors of an individual. An essential skill in auditory processing is the binaural integration, which involves the processing of auditory information from both ears simultaneously. This ability is very important for us to be able to understand speech in quiet and in background noise[1]. Binaural integration also involves separating the sounds in both the ears and processing them individually at the level of the auditory cortex. This ability can get enhanced with improved auditory attention and superior cortical processing.

An individual with martial art training has a superior attention network in brain imaging[2]. It is also reported that similar to Yoga, martial arts can produce similar enhanced changes in attention reported in children[3]. Binaural Integration is a cognitive process involving the fusion of auditory information, which is presented binaurally. It is considered important for the process of speech perception and sound localization. Martial arts are combat sport which trains the body to improve the physical, psychological, tactile, and technical skills. For enhanced sports performance, the cognitive process should be improved. In specific, enhanced attention is particularly important in combat sports[4]. Previous studies on individuals with martial arts training report superior attention abilities, and they can modulate their attention in a better manner, which is necessary according to a situation[5, 6].

All the senses, including auditory attention, are reported to be better in individuals with martial arts training[7]. Hence, binaural integration, which involves fusing and separating information from both ears, simultaneously requires active attention and may also be enhanced in individuals with martial arts training. Binaural integration is usually assessed during dichotic tests[8]. Dichotic Consonant Vowel test is the most commonly used test for assessing the binaural integration process[8]. Considering the previous studies, binaural integration may be enhanced in children with martial arts training. Considering the dearth of studies in this area, the present study aims to determine if there are any differences in binaural integration abilities between children with and without martial arts training. Also, it is also attempted to determine if there is any correlation between binaural integration scores and the number of years of martial arts training.

2. METHOD

2.1 Participants

Thirty normal-hearing individuals between the age ranges of 9-13 years participated in the study. In total, 18 males (mean = 11.55, SD=2.12) and 12 females (mean = 10.55, SD=1.91) participated in the study. The participants were divided into two groups (15 participants each) based on who practiced martial arts-Karate with intensive training for a minimum of 2 years and those who do not practice any form of martial arts. All the participants who practiced karate used a similar method of training. The number of years of martial arts training was also noted by all the participants of the study. All the participants in the study did not have any history of otologic symptoms, metabolic ad systemic disease causing hearing loss, and did not use any ototoxic drugs. The binaural integration abilities of the participants were assessed using the dichotic CV test. All the participants had average academic performance, as reported by their school teachers.

2.2 Procedures

Air conduction and bone conduction pure tone thresholds and speech identification scores were obtained by using a two-channel diagnostic audiometer. A personal computer was connected to the audiometer to route the recorded wordlist at the presentation level of 40 dB SL. The middle ear status was examined using an otoscope to check for any tympanic abnormalities. The middle ear status of the participants was also examined using Grason Stadler Inc. Tympstar (GSI-TS) immittance meter. Both ears of the participants were tested to obtain tympanogram and acoustic reflexes with the probe tone frequency of 226 Hz. Acoustic reflexes were measured for pure tones at 500 Hz, 1000 Hz, 2000 Hz, and 4000 Hz presented to both ipsilateral and contralateral ears.

The binaural integration abilities were tested using the dichotic consonant-vowel (DCV) test. This experiment required the presentation of different sounds (consonant-vowel combination) example: pa-ta-, ba-da, etc. to the right and left ear simultaneously. The participants were instructed to write both syllables, which were presented. They were also given trials for familiarization before the test. A calibration tone was presented before the beginning of the recorded monosyllable. The recorded stimulus consisted of 30 pairs of monosyllables. Based on the response of the participants, double correct scores (DCS) that are both the syllables were correctly identified, it was scored as 1, and if one of the syllables or both the syllables were incorrectly identified, it was scored as 0. This stimulus was presented at the most comfortable level (60 dB HL) for all the participants through the supraaural headphone. It checks for the binaural integration abilities of an individual.

2.3 Statistical Analyses

The data obtained were analyzed using the Statistical Package for Social Sciences (SPSS) Version 20 (IBM Incorporation, Armonk, New York: USA). Shapiro Wilk test of normality was done to determine if the data is normally distributed. Mann Whitney U test and Pearson's correlation coefficient was determined to analyze the data.

2.4 Ethical consideration

For the study, non-invasive testing procedures adhering to conditions of the ethical approval committee of the institute were used. All the test procedures were explained to the participants/parents, and informed consent was taken from all the parents to take part in the study. The study followed the principles of the Declaration of Helsinki.

3. RESULTS AND DISCUSSION

A descriptive statistical analysis was done for the collected data, and the median and inter-quartile range of and double correct scores were determined. The results showed that the scores were better for individuals with martial arts training compared to the control group. The median and inter-quartile range of dichotic CV scores obtained in both groups is shown in Figure 1.

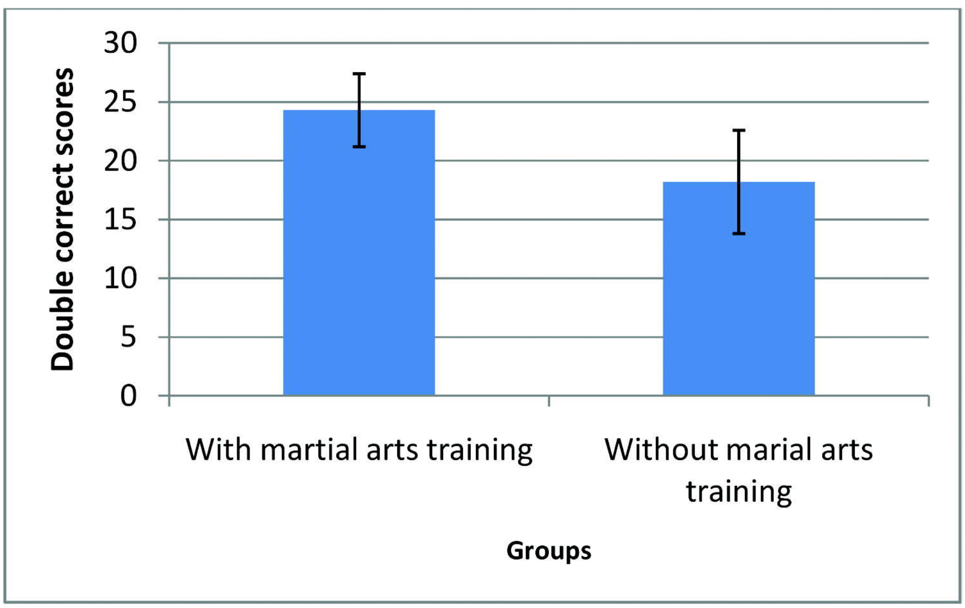

Fig. 1. Median and inter-quartile range of double correct scores (DCS) obtained for both the groups.

Shapiro Wilk's test of normality was done to determine the normality of the data. The result of the test of normality shows that the data was not normally distributed ($p<0.05$). Hence, non-parametric inferential statistics were done. Mann Whitney U tests were done to determine if there is any significant difference in double correct scores (DCS) between the two groups. The results of the study showed that the double correct scores were significantly better ($p<0.05$) for individuals with martial arts training compared to the control group. The effect size for the data was calculated using the formula $Z/\sqrt{N}$, and it was found to be 0.76, which suggests a strong difference between the groups. The result of our study shows that practicing karate significantly improved binaural integration abilities.

It was also attempted to determine if there is any correlation between the number of years of martial arts training and double correct scores. The scatter plot of the correlation is shown in Figure 2.

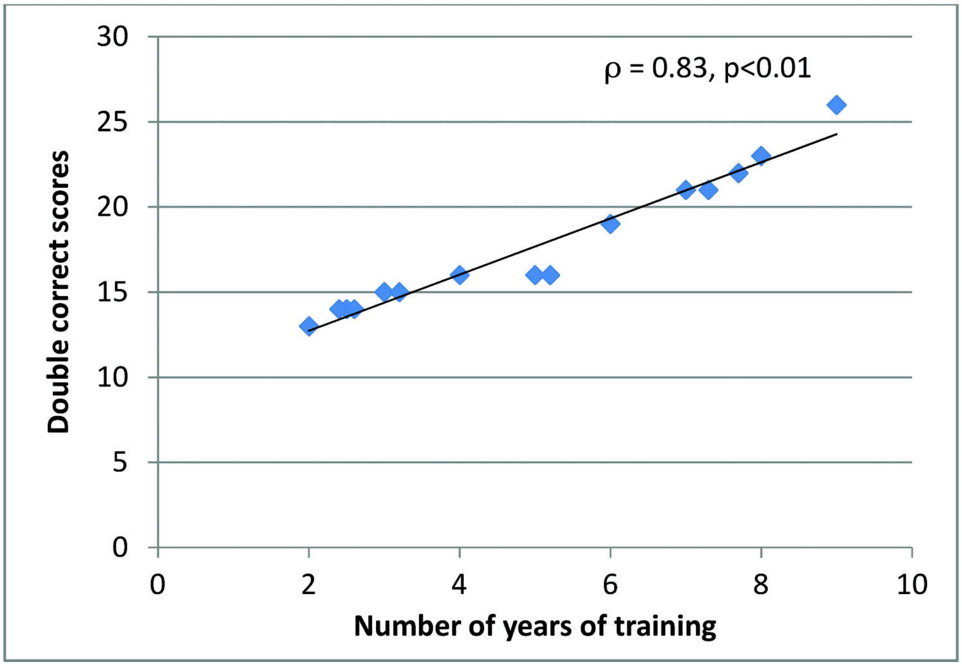

Fig. 2. Scatter plot of correlation between double corrects cores and a number of years of training.

Spearman's rank correlation showed that there was a good positive correlation ($\rho = 0.83$, $p<0.01$) between the double correct scores and the number of years of martial arts training. The results suggest that binaural integration abilities improve with martial arts training.

The results of the study showed that binaural integration abilities were enhanced in children with martial arts training. The results agree with previous studies, which also report superior attention in individuals with martial arts training[2,3]. It is well reported in the literature that enhanced cortical, cognitive, and, most importantly, attention-related abilities play an important role in combat sports[9]. It is also well known that athletes, including those who practice martial arts, have enhanced attention to extract and identify important information in their surroundings[5,9,10]. Thus, individuals who practice martial arts maybe can use their attention resources to a greater extent[6]. Hence, the improved binaural integration scores could be attributed to enhanced auditory attention to binaural information and better cortical processing.

The results of the study also demonstrated that the scores improved with the number of years of martial arts training. It is also well reported that constant training causes neuro-plastic changes in the auditory cortex through magnetic resonance imaging studies[11,12]. Brain imaging studies have demonstrated a superior attention-based neural network in individuals with martial arts training[2,13,14]. In specific, enhanced attention is particularly important in combat sports[4,15]. Thus, the enhanced auditory attention and auditory plasticity of the cortical areas due to training could explain the improved scores obtained in the present study. Further studies are essential in this area for better generalization of the findings.

4. CONCLUSION

The study attempted to determine differences in binaural integration abilities in children with and without martial arts training. The results of the study showed that the dichotic CV test scores were higher in individuals with martial arts training compared to the control group. Also, the scores improved with an increase in the number of years of martial arts training. The results are by previous studies, which also suggest a superior performance in individuals with martial arts training due to enhanced auditory attention. The enriched neuro-plastic changes in the auditory cortex with martial arts training could have resulted in an increase in scores with the number of years of training. Further studies on a larger group of population are essential for better generalization of the results.

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Normative data on dysphonia severity index for Telugu speaking Indian geriatric population

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ABSTRACT

The dysphonia severity index (DSI) has been the most widely used objective measure to quantify the severity of dysphonia based on maximum phonation time, highest fundamental frequency, lowest intensity and percent jitter (jitter%) and has been validated across normal and clinical population. Aging process can affect the overall voice quality which can be identified both perceptually and acoustically, thereby necessitating developing normative data for the objective indices. Although there are studies on establishing normative data in children and young adults in Indian context, there are no studies on geriatric population and hence, the present study was carried out with an aim to establish preliminary normative data for Telugu speaking Indian geriatric population. The present study included 20 males and 20 females in the age range of 61-89 years with G0 rating on GRBAS scale. Maximum phonation time (MPT), highest fundamental frequency (F_0-high), lowest intensity (I-low) and jitter % were measured by using Praat Software version 6.1.15 and DSI was calculated using the formula 0.13 × MPT + 0.0053 × F0-high - 0.26 × I-low - 1.18 × Jitter (%) + 12.4. The results revealed lower DSI scores in geriatric population compared to children and young adults in Indian context. The lower DSI scores were attributed to decrease in the maximum phonation time and reduced phonation frequency range among geriatric population in both males and females due to structural and physiological changes. Significant differences were observed between males and females on MPT, F_0-highbut no gender difference was found for DSI, lowest intensity, and jitter %. The results of the present study provided a preliminary normative data for DSI for Telugu speaking Indian geriatric population and can be used as reference for measuring the severity of dysphonia.

1. INTRODUCTION

The production of voice in humans involves various anatomical and physiological changes in several speech subsystems including that of respiratory and phonatory/laryngeal subsystems. Subtle variations in anatomical structures or physiological changes do result in variation in voice characteristics. Pitch, loudness, and quality are the parameters of voice through which an individual's voice is described and factors such as age, gender, *etc*. are known to affect these parameters. Although major changes in voice are reported during puberty, minor changes in voice are reported with an increase in age due to the structural and physiological changes such as atrophy of muscles in the laryngeal system, stiffened cartilages, reduced lung capacities, hormonal changes, etc. These anatomical and physiological changes in older adults result in variations in pitch, loudness, and quality of voice in the elderly which are measured through perceptual and acoustic measures.

The voice characteristics of the elderly population have been studied extensively by using perceptual and acoustic measures. The past research studies using perceptual scales have reported that elderly voice is majorly characterized by lowered pitch levels,[1,2] reduced loudness and change in the quality of voice with increased hoarseness,[3,4] breathiness,[5,6] tremors, and pitch breaks which were attributed to changes in the mass of vocal folds, hormonal variations during post-menopause, and increased laryngeal tension[2,3,7]. The acoustic analysis of elderly voice also reported reduced fundamental frequency,[8,9,10,11] reduced intensity, increase in jitter and shimmer,[7,8,12,13,14,15] and increase in noise to harmonics ratio[12,15,16]. In Indian context, a study by Sebastian *et al.*[17] reported higher fundamental frequency in males and lower fundamental frequency in females among geriatrics (60-80 years) compared to younger adults (18-25 years). However, the study did not report a difference in percentage of jitter and shimmer between older and young adults.

The major objectives of clinical voice examination are to identify the pathophysiology of the underlying condition and to understand its effects on voice parameters through subjective and objective measures. The perceptual evaluation of voice is carried out by using various subjective rating scales such as GRBAS,[18] Consensus Auditory-Perceptual Evaluation of Voice (CAPE-V),[19] *etc* by the qualified speech-language pathologists. The objective evaluation of voice includes acoustic and aerodynamic analysis of patients with voice disorders through sophisticated instruments and softwares to measure the frequency, intensity, and perturbation measures. The results of several studies have reported a good correlation between perceptual evaluation and various acoustic parameters. Although each of these acoustic parameters does provide valuable information regarding the functioning of the laryngeal and respiratory system, these parameters do not provide an objective index to quantify the severity of dysphonia. To overcome this limitation, Wuyts *et al.*[20] developed the Dysphonia Severity Index as an objective measure to quantify the severity of dysphonia.

Dysphonia Severity Index (DSI)[20] is developed based on a multiparameter approach based on the weighted combination of the highest fundamental frequency (F0-high), lowest intensity (I-low), maximum phonation time (MPT, in seconds) and jitter percentage. The DSI is calculated using the formula DSI = 0.13 × MPT + 0.0053 × F0-high - 0.26 × I-low - 1.18 × Jitter % + 12.4. A DSI value of -5 is considered as severe aphonia and a value of 5 is considered as a normal voice based on a multivariate analysis of 387 subjects. Strong correlation was reported between DSI and the GRBAS scale,[20] CAPE-V,[21] Voice Handicap Index[22] and VRQOL[23] in both normal subjects and subjects with voice disorders.

DSI has been developed based on a multiparameter approach in which four different acoustic parameters are included which can be affected by various parameters such as age, gender, and ethnicity, etc. Although, the range of DSI is between -5 to +5, a score above 1.6 is considered as normal considering the variations in typical population due to the factors such as age, gender, and ethnicity, *etc*. This warranted more number of studies on the development of normative data across different age groups, different gender, and different ethnic groups.

Hakkesteegt, *et al.*[24] in their study on the influence of age and gender on DSI reported a DSI of 3.8 ± 1.94 and 4.3 ± 2.01 for males and females respectively in the age range of 20-79 years. Hakkesteegt *et al.*[22] have proposed a DSI cut off of 3.0 as an appropriate score to discriminate between normal and pathological voices with a sensitivity of 72% and specificity of 75% in their study on exploring the correlation between DSI and GRBAS. Awan and Enssien[25] have reported a significant effect of vocal training on DSI with a mean DSI of 6.48 ± 2.19 for trained vocalists and 4.0 ± 2.04 for untrained vocalists in the age range of 18-30 years. Schneider *et al.*[23] reported a DSI of 1.2 ± 2.4 among elderly people without voice complaints in the age range of 66-94 years. Hwang, Lee and Kim[26] have reported a DSI of 6.36 ± 2.41 and 6.92 ± 2.79 for young males and female groups of the Korean population. Aghadoost *et al.*[27] have reported a mean and standard deviation of DSI of 3.58 and 0.92 respectively for normal female teachers in their study on comparing DSI of female teachers with and without voice disorders in Iran. Nemr *et al.*[21] in their study on the correlation between DSI and CAPE-V among the Brazilian population have reported a mean DSI of 2.27 with a standard deviation of 1.36 for nondysphonic group of 42 participants. Kim *et al.*[28] in their study on Shanghainese population, reported a DSI of 5.95 ± 1.47 and 6.11 ± 1.74 for male and female young adults in the age range of 18-23 years and DSI of 4.70 ± 1.31 and 5.01 ± 1.47 for male and female children in the age range of 7-9 years. Uloza *et al.*[29] reported that a DSI threshold of 3.30 has been found to have a sensitivity of 85.8% and specificity of 83.4%. Latoszek *et al.*[30] in their study on the influence of gender and age on DSI reported a DSI score of 5.41 ± 1.78 and 6.04 ± 2.18 for males and females in the age range of 20-79 years. The dysphonia severity index is also validated among carnatic singers,[31] post thyroidectomy patients,[32] teachers with voice complaints,[27] in children with velopharyngeal dysfunction[33].

Few studies have been carried out in the Indian context on establishing normative data in children and adults. Jayakumar and Savithri[34] reported a DSI of 3.07 ± 1.15 and 3.8 ± 1.21 for Indian young male and female groups in the age range of 18-25 years. Further, the authors also have highlighted the importance of establishing DSI normative data based on geographical and ethnic groups. Pebbili *et al.*[35] reported a mean DSI value of 2.9±1.23 and 3.8 ± 1.29 for typical Indian male and female children in the age range of 8-12 years. Boominathan *et al.*[36] have reported DSI of 0.07 ± 1.1 and 0.16 ± 1.2 for elderly male and female college teachers in India. The review of literature suggested that there are variations in the DSI among children, adults, and geriatric population across geographical and ethnic groups. In Indian context, normative data is established for Kannada speaking children in the age range of 8-12 years and adults in the age range of 18-25 years. There is a great need to develop normative data for the geriatric population which would help in understanding the vocal functioning in geriatrics and in clinical use. Hence, the present study was carried out with an aim of documenting the DSI in Telugu speaking Indian geriatric population in the age range of 60-85 years.

2. METHOD

2.1 Participants

The present study included 40 Telugu native participants in the age range of 61-89 years with 20 males with a mean age of 65.60 (± 4.30) and 20 females with a mean age of 69.85 (± 9.10). A total of 57 individuals without any history of speech, language and hearing problems in the age range of 61-89 years were perceptually evaluated using GRBAS scale by two speech-language pathologists and 20 males and 20 females with a rating of G0 on GRBAS scale indicating perceptually normal voice were included as participants in the final study. Institute Ethical committee approval was obtained for the present study and written informed consent was collected from all the participants before the study.

2.2 Procedures

Praat software version 6.1.15 was used to record and analyze the voice samples in all the participants in a quiet room to measure the F_0-high, I-low, Jitter%, and maximum phonation time. The participants were made to sit comfortably and the voice samples were recorded using Praat software version 6.1.15 with a sampling frequency of 44,100 Hz. The tasks used for measuring the various parameters are described below.

F0-high and I-low : The participants were given a model by the speech-language pathologist for the task involving phonation from lowest F_0 to highest F_0 before the actual recordings. Later, participants were instructed to phonate /a:/ as softly as possible raising from the lowest pitch to highest pitch with three trails. The highest F0 and lowest intensity levels were selected for calculating DSI.

Maximum phonation time : All the participants were instructed to take a deep breath and were asked to phonate /a:/ vowel at a comfortable pitch and loudness for as long as possible for three times. The maximum MPT value from three trials was selected for calculating DSI.

Jitter% : All the participants were asked to phonate /a:/ vowel for five seconds at a comfortable pitch and loudness and Jitter% was measured for intermediate sample of three seconds by removing rise time and fall time using Praat Software.

The obtained data was entered into a Microsoft Excel Spreadsheet and the DSI was calculated for further statistical analysis using Statistical Package for Social Sciences (SPSS) version 20.0. Descriptive statistical analysis was carried out to measure the mean and standard deviation of all parameters and an independent sample t-test was used to compare the differences between male and female groups across all parameters.

3. RESULTS AND DISCUSSION

The mean and standard deviations were calculated for MPT, F_0-high, I-low, Jitter % and DSI and the results are given in Table 1 and the data is represented in Fig. 1. The mean MPT values were higher for males (14.75 ± 3.09) compared to females (11.45 ± 2.58) and the difference was statistically significant as revealed by independent sample t-test results ($p = 0.001$). The mean F0-high values were higher for females (387.95 ± 49.87) compared to males (282.24 ± 41.31) and the independent sample t-test results revealed statistically significant difference between two groups ($p = 0.00$). The mean I-low values were higher for females (50.45 ± 2.73) compared to males (50.32 ± 2.37) and the difference was statistically not significant ($p = 0.95$). The mean Jitter% values were higher for females (0.83 ± 0.52) compared to males (0.62 ± 0.42) and the difference was statistically not significant ($p = 0.167$). The DSI was higher for males (1.99 ± 0.62) compared to females (1.86 ± 0.92) and the difference was statistically not significant ($p = 0.607$).

Table 1. Mean and standard deviation of MPT, F_0-high, I-low, Jitter % and DSI.

Parameter	Males		Females		Independent t-test		
	Mean	**SD**	**Mean**	**SD**	**t**	**df**	**Sig.**
MPT	14.75	3.09	11.45	2.58	3.66	38	0.001**
F_0-high	282.24	41.31	387.95	49.87	-7.30	38	0.00**
I-low	50.32	2.37	50.45	2.73	-0.054	38	0.958
Jitter %	0.62	0.42	0.83	0.52	-1.408	38	0.167
DSI	1.99	0.62	1.86	0.92	0.519	38	0.607

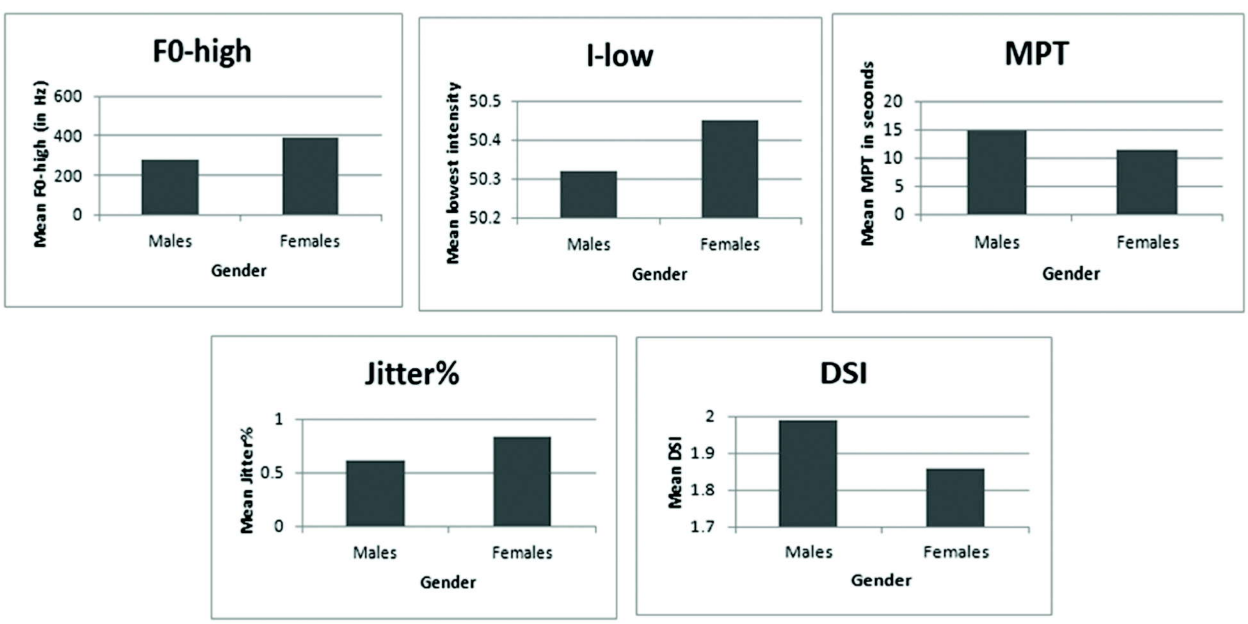

Fig. 1. Comparison between Males and Females for DSI and Parameters of DSI.

The present study was carried out on 40 participants in the age range of 61 to 89 years to obtain the preliminary normative data for dysphonia severity index (DSI) for Telugu speaking geriatric population. The DSI value obtained for Telugu speaking geriatric population was significantly lower when compared to children and younger adults in Indian context as presented in Table 2.

Table 2. Mean and Standard deviation of gender difference in DSI values for children, adults and geriatric population in Indian context.

	Geriatrics (61-89 years) Present study		**Children (8-12 years)[35]**		**Adults (18-25 years)[34]**	
	Males	**Females**	**Males**	**Females**	**Males**	**Females**
DSI	1.99 (0.62)	1.86 (0.92)	2.9 (1.23)	3.8 (1.29)	3.07 (1.15)	3.8 (1.21)

The lower values of DSI in geriatric population may be the result of decreased control of laryngeal and respiratory system among geriatrics. The maximum phonation time (MPT) of males (14.75 ± 3.09) and females (11.45 ± 2.58) among geriatric population was significantly lower than younger males (17.6 ± 5.23) and females (13.8 ± 2.80) and this may be attributed to the reduced lung volumes and lung capacities among geriatric population as reported in previous studies on effect of age on maximum phonation time and lung capacities[37,38]. The present study has also showed significant effect of gender on MPT in geriatric population which is attributed to variations in physical features between men and women[20,24,34,38,39].

In the present study, significantly higher F0-high was found for females (387.95 ± 49.87) compared to males (282.24 ± 41.31) and this is in line with the results of the previous studies who reported higher values of F0-high for females[20,24,28,34,40]. However, the phonation frequency range has reduced in geriatric population when compared to younger population as revealed by previous studies[20,24,34,36]. This decrease in phonation frequency range could be due to structural changes in vocal folds such as decreased tension or atrophy of vocal folds and is one of the major contributing factors for decrease in the overall DSI scores in geriatric population. Although the results of the present study revealed decreased for both males and females, the difference is not significant as also reported by the previous studies[20,24]. Jitter, which refers to cycle to cycle variations in frequency also, did not differ between males and females in geriatrics as reported by previous studies[13,14,20,34,36,41].

DSI is developed based on a multiparameter approach and any variations in the parameters can affect the overall DSI scores. The present study on geriatric population in Indian context has shown a DSI of 1.99 ± 0.62 for males and 1.86 ± 0.92 for females and there was no statistically significant difference between males and female groups as reported by Hakkesteegt, *et al.*[24] and Wuyts, *et al.*[20]. Although the mean DSI score is above the normal rage (>1.6), few of the participants of the study have DSI score less than normal range falling into mild to moderate dysphonia although they all were rated as perceptually normal voice as shown on GRBAS scale. The DSI scores of the Indian geriatric male and female participants is much lower than the Indian children in the age range of 8-12 years and Indian young adults in the age range of 18-25 years as given in Table 2.

The DSI scores of the present study were similar to the study carried out by Boominathan *et al.*[36] on Indian elderly college teachers and Schneider *et al.*[23] who reported a DSI score of 1.2 ± 2.4 among elderly people without voice complaints in the age range of 66-94 years. However, the DSI scores obtained for males and females in the present study are much lower when compared to the previous studies[20,21,24,26,27,30]. The results of the present study highlight the importance of developing normative data for different age groups across different geographical and ethnic groups as reported by Jayakumar and Savithri[34]. Although, the present study has provided a preliminary normative value of DSI for Telugu speaking Indian geriatric population, the study has included only small number of participants in both groups and a broad age range of 29 years. These limitations of the present study warrant future studies on building normative data for different age and ethnic groups with large number of participants.

4. CONCLUSION

The present study investigated the gender differences on dysphonia severity index and its parameters in Telugu speaking Indian geriatric population in the age range of 61 to 89 years. A significant difference was found between males and females for MPT and F0-high values and no significant difference was found between males and females for I-low, Jitter% and DSI. Further, the overall DSI scores were found to be significantly lower than young adults and children in Indian context. The results of the present study provided a preliminary normative data on DSI among Telugu speaking Indian geriatric population which will help the speech language pathologists and otolaryngologists in clinical practice. Further it is necessary to carry out more studies to investigate the effect of aging on DSI and establish normative for different age groups in Indian context and compare to clinical population.

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Speech rhythm in young adults with congenital hearing impairment

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ABSTRACT

Rhythm refers to any patterned movement in continuous speech and is one of the suprasegmental or prosodic features. Previous studies have reported that speech rhythm of children with congenital hearing impairment deviates from normal speaking population. The present study aimed at investigating the rhythmic pattern of young adults with congenital hearing impairment and compares the same with that of normal group quantitatively. A total of 60 subjects in the age range of 17 to 22 years participated in the study. The subjects were divided into two groups: group one consisted of 30 normal subjects and group two consisted of 30 subjects with moderate to severe degree of congenital hearing impairment. All subjects were native speakers of Kannada language. About five minutes of extempore speech sample were elicited using picture cards depicting a story from all the participants and were recorded using a digital recorder. All the recorded speech samples were then analyzed using PRAAT software to get the vocalic (V) and intervocalic (IV) durations. Normalized 'Pair-wise Variability Index' (nPVI) and 'raw Pair-wise Variability Index' (rPVI) were obtained for both the groups and was compared using Independent t-test. Results revealed a statistically significant difference in the parameters of speech rhythm between the groups indicating that speech of individuals with hearing impairment has an altered rhythmic pattern

1. INTRODUCTION

Rhythm is one of the fundamental components of speech. It involves regularity, such that there is a pattern of recurrence of some particular event. It has an organizing function[1]. It allows listener to focus on semantically important parts of the language uttered by the speaker and reduces the need of sustained attention to speech input by the listener. "Rhythmic structure thus produces useful perceptual redundancy in speech by constraining the time when important articulatory events may occur"[1]. In conversational speech the rhythm is vastly more complicated, but it is clear that the timing of speech is not random.

According to Pike[2] and Abercrombie[3] every language can be classified as either stress-timed or syllable-timed. In stress-timed languages, there is near-equal intervals between stressed syllables or rhythmic feet, whereas in syllable-timed languages, there is near-equal length of successive syllables. Bloch,[4] Han[5] and Ladefoged[6] put forth a third type of rhythm called Mora-timed. In mora-timed languages, the durational difference between syllables is reported to be negligible.

Measuring the rhythm of language was started with the concept of isochrony. The preliminary attempt, by Abercrombie[7] used the average syllable duration as a method to test Rhythm Class Hypothesis. It was reported that, this method is not an efficient way in segregating languages based on the type of rhythm. Later, Pair-wise Variability Index (PVI) was developed[8] for the analysis of rhythm. Using this method, the acoustic aspects of speech rhythm can be quantified by calculating the pattern of successively occurring vocalic and intervocalic intervals. A decade later, normalized Pairwise Variability Index (nPVI) and raw Pairwise Variability Index (rPVI) were developed[9] for measuring vocalic and intervocalic durations. Vocalic duration was found out by measuring the vowel durations, and intervocalic durations was found out by measuring the duration of speech segments in between the vowels. Pairwise Variability Index for the measurement of vocalic and intervocalic duration was then computed. The level of variability of the successive measurements was expressed as Index. The rPVI is shown in equation (1).

$$rPVI = \left[\sum_{k-1}^{m-1} \left| d_k - d_{k+1} \right| / (m-1) \right] \tag{1}$$

In the equation 'm' stands for the number of intervals, vocalic or intervocalic, in the text and 'd_k' shows the duration of the k-th interval. Here, rPVI is not normalised for rate of speech. For measuring the duration of vowels a normalised version of the Pairwise Variability Index was used by Low *et al.*[8]. The equation (2) shows the nPVI

$$nPVI = 100 \times \left[\sum_{k-1}^{m-1} \left| \frac{d_k - d_{k+1}}{d_k + d_{k+1} / 2} \right| / (m-1) \right] \tag{2}$$

In the equation 'm' represents the number of items in an utterance and the duration of k^{th} item is represented by 'd'. Second equation represents that the nPVI is derived by finding out the sum of absolute difference in the duration between each successive measurement and dividing that by their mean duration. The value is then divided by the number of items in the utterance. The final value must be multiplied with 100 to convert fractional values which normalization creates.

According to this, based on the relative values of vocalic nPVI and intervocalic rPVI languages can be classified as stress timed, syllable timed and mora timed *i.e.* stressed-time language has a pattern of

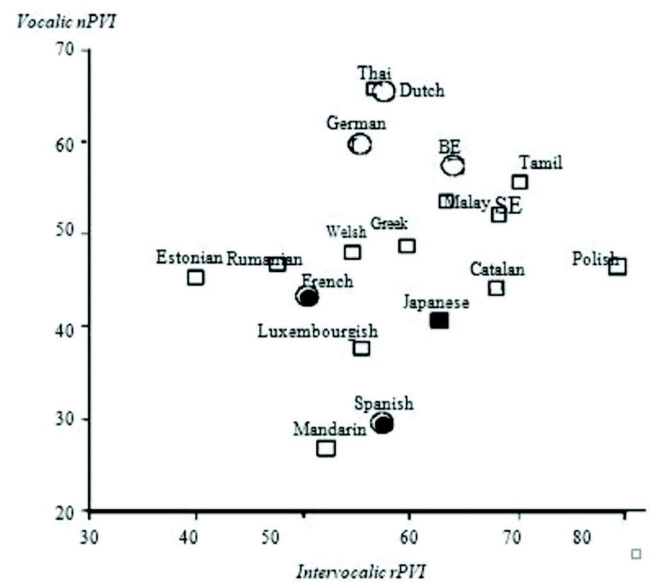

Fig. 1. Vocalic nPVI values and the intervocalic rPVI of 18 languages

high nPVI and high rPVI, low nPVI and low rPVI pattern indicates mora- time language and syllable-timed language shows a pattern of low nPVI and high rPVI. Figure 1 depicts data of prototypical examples of stress-, syllable- and mora-timing profiles: German, British English and Dutch were classified as stress-timed, Spanish and French as syllable-timed languages and Japanese a mora-timed language. Vertical axis shows vocalic nPVI values and the intervocalic rPVI values shown on the horizontal axis published by Low and Grabe[10].

Numerous researches have been done on rhythm in normal as well as in disordered population. Rhythm in twelve Indian languages, *viz.*, Bengali, Hindi, Assamese, Kashmiri, Marathi, Punjabi, Gujarathi, Tamil, Kannada, Malayalam, Oriya, and Telugu (Dravidian languages) was compared[11]. The results indicated Hindi as syllable-timed language and all other languages studied were mora-timed language. Another study[12] investigated the type of speech rhythmic characteristics in Kannada speaking normal children of aged 8 - 9 years. A range of 44.97 to 78.17 was obtained with a mean of 65.90 for intervocalic rPVI for these children and the vocalic nPVI values were in the range of 80.10 to 122.75 and the mean value was 96.06. The results thus revealed high vocalic nPVI and low intervocalic rPVI values and therefore the pattern of rhythm remains unclassified and cannot be put under any of the rhythmic classes (stress-timed, syllable-timed, mora-timed). Savithri, Jayaram, Kedarnath & Sanjay[13] studied rhythmic pattern in Kannada speaking normal adults and results revealed low rPVI and low nPVI and reported Kannada as a mora-timed language.

Speech of children with hearing impairment deviates from normal both at the segmental level and at the suprasegmental level. A few researchers have studied rhythm in individuals with hearing impairment in Kannada language. Rhythm between normal and children with hearing-impairment in the age range of 5-10 years was compared[14]. The mean rPVI and nPVI values of normal children were 15.70 and 62.49 respectively, whereas those for the hearing-impaired children the mean rPVI was 20.54 and nPVI values 67.14. The results revealed that speech rhythm in both the groups remained unclassified. The children used a simpler syllabic structure in the acquisition stage of rhythmic patterns. As there is a dearth of data related to rhythm in adult hearing impaired population, the current study investigated the speech rhythm of young adults with congenital hearing impairment and to compare it with that of the normal population.

2. METHOD

2.1 Material

Sequence of pictures depicting a story ('thirsty crow') was used to elicit speech sample.

2.2 Subjects

A total of 60 subjects in the age range of 17 to 22 years participated in the study. The subjects were grouped into two. Group one consisted of 30 normal subjects who were screened to rule out any deficits in speech, language and hearing. Group two consisted of 30 subjects with moderate to severe degree of hearing impairment. However none of the subjects in group two were regular hearing aid users.

2.3 Procedure

The subjects were tested individually. About five minutes of extempore speech sample were elicited using picture cards depicting a story from all the participants. The subjects were instructed to watch the pictures carefully and describe the pictures. Verbal prompts were given by the examiner during the period of lack of response from the subject. The speech sample elicited from the subject was audio-recorded using Sony digital voice recorder (Model No. ICD-PX333).

2.4 Analyses

The entire speech sample of each subject was transferred into PRAAT software and was displayed as a wave form. In the present the entire sample of each subject was analyzed. At the beginning of the analysis of each sample, the investigator listened to the entire sample and the unwanted pauses were identified and selected manually. These identified unwanted pauses were then removed by using the editing option

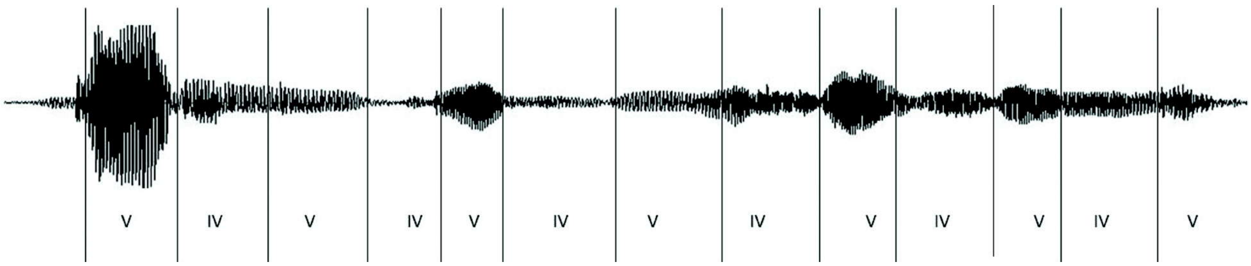

Fig. 2. Measurement of vocalic duration (V) and intervocalic duration (IV).

in the PRAAT software. This was done to obtain the accurate values of the vocalic and inter-vocalic durations for the further analysis. Vocalic interval corresponds to the duration of vowel/semivowel/diphthong. The measurement of the vocalic interval was done by measuring the time duration from the beginning of voicing of vowel/semivowel/diphthong to the end of voicing. The measurement of intervocalic duration was done by measuring the duration corresponds to the interval between two vocalic segments. Figure 2 depicts the measurement of duration of vocalic (V) and intervocalic (IV) segments.

To get the nPVI value, duration difference between successive vocalic intervals was calculated and averaged and for rPVI values the duration difference between intervocalic segments was found out and averaged. To get the measure of rhythm Pairwise Variability Index[9] was used and the formulae to get the rPVI and nPVI were given in Equations (1) and (2) of the Introductory Section. For the implementation of the equation to work out nPVI and rPVI, the Microsoft Visual Basic 6 was used.

Statistical analysis : The mean rPVI and mean nPVI values were found out for both the groups. Paired sample t-test and independent samples t-test were used to obtain significant differences within groups and between groups.

3. RESULTS AND DISCUSSION

The vocalic nPVI and intervocalic rPVI values obtained from both the groups (normal and hearing impaired) and statistical analysis was carried out using SPSS version 17 software. The mean vocalic nPVI values and intervocalic rPVI values of normal group are given in Table 1.

Table 1. Mean and SD for normal

Normal Group	Mean	Standard deviation	Sig. value
Vocalic nPVI	44.36	6.93	0.00
Intervocalic rPVI	62.35	13.39	

Table 1 reveals that the mean vocalic nPVI is less than intervocalic rPVI in normal group. Further the mean vocalic nPVI and inter-vocalic rPVI was compared using paired sample t test and the results showed a significant difference between vocalic nPVI and intervocalic rPVI within the group. *i.e.* $p<0.01$. The low vocalic nPVI and intervocalic rPVI value indicates Kannada to be considered as a mora-timed language. Speech rhythm in different languages was compared[13] and reported Kannada to be a mora-timed language, which is in consonance with the findings of the present study.

Table 2 depicts the mean and standard deviation of group 2 (Hearing impaired). Paired t-test indicated a significant difference between vocalic nPVI and intervocalic rPVI *i.e.* $p<0.01$. The mean vocalic nPVI values for hearing impaired are 55.27 with a standard deviation of 7.75 and for intervocalic rPVI is 49.45 with a standard deviation of 6.85 *i.e.* the mean intervocalic rPVI is found to be lesser than that of mean vocalic nPVI in hearing impaired group. The results thus indicated vocalic nPVI of 55.27 and intervocalic

Table 2. Mean and standard deviation for Hearing impaired

Hearing impaired group	Mean	Standard deviation	Sig. value
Vocalic nPVI	55.27	7.75	0.02
Intervocalic rPVI	49.45	6.85	

rPVI of 49.45 and is close to the rhythm of languages Welsh and Greek. The rhythm of the language Greek and Welsh is reported to be unclassified by Low and Grabe[10]. Since the present study also getting a similar vocalic nPVI and intervocalic rPVI, it can stated that the rhythmic pattern remains unclassified for hearing impaired group. Thus indicating that speech rhythm of hearing impaired individuals is deviating from the normal.

Further, the mean vocalic nPVI scores were compared between the groups using independent t-test. Results indicates a significance difference between the groups *i.e.* $p<0.01$. Mean vocalic nPVI scores was found to be greater for hearing impaired subjects compared to that of normal subjects. Similarly intervocalic rPVI scores was also compared between the groups using independent t-test. A significant difference was obtained between normal group and individuals with hearing impairment *i.e.* $p<0.01$.

The above finding shows that, there are differences in the speech rhythm of normal and hearing impaired population. The result showed a significant difference in the mean vocalic duration and intervocalic duration between normal group and hearing impaired group. This might be because of the atypical characteristics of hearing impaired speech. Individuals with severe to profound hearing impairment can have distortions in the speech which can affect the temporal aspects of their speech. These distortions can be prolongation of speech segments, and the presence of' lengthy filled and unfilled pauses. These deviant speech characteristics may be perceptually evident and it can disrupt the quality of rhythm of their speech. The higher vocalic nPVI values could be because of the more usage of vowels than the consonants in the speech output of people with hearing impairment. Susman and Hernandez[15] analysed speech of hearing impaired with respect to suprasegmentals aspects and reported when compared with the normally speaking group, the result indicated that the duration of vowels was longer before the voiced stops in subject hearing impairment than before the voiceless stops. Similar finding was reported[14] by investigating the rhythmic pattern of hearing impaired children and found that rhythmic pattern in hearing impaired children cannot be classified in any of the rhythmic classes. There are many typical characteristics of the speech of the hearing impaired which serve to differentiate it from the speech of normal children. Numerous studies have been done to identify the speech characteristics of the hearing impaired population[14-22]. These studies have identified many recurring speech deviations for individuals with hearing impairment, thus reporting that the hearing impaired speech has a nature of its own.

4. CONCLUSION

Rhythm is a component of prosody. It is an event repeated regularly over a period of time. The present study aimed to find out the rhythmic pattern of speech of subjects with hearing impairment compared with that of normal group. Results revealed a significant difference in speech rhythm between the groups indicating that hearing impaired speech has an altered rhythmic pattern. The outcome of the study has significant implication for speech language pathologist. Atypical speech rhythm of people with hearing impairment will affect the naturalness of speech. Hence to improve the speech naturalness of people with hearing impairment, the speech therapist may have to focus on training the normal rhythm during treatment program.

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Applications of Electroencephalography (EEG) in Neuro-Steered Hearing Aids: A scoping review

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ABSTRACT

Hearing aids constitute the heart of auditory rehabilitation programs. In the current scenario, hearing aids are fitted on listeners with hearing impairment using prescriptive rationales, which are based on speech characteristics. However, hearing, unlike listening, is dependent on the brain's response to sound. Hearing is based not only on the audibility of the signal, but also on how sound is biologically coded, integrated, and used by the auditory neural mechanisms and the brain. Therefore, the use of current practices in fitting hearing amplification devices based on fitting formulae might not address variability stemming from physiological and psychological processes operating at the hearing aid users' brain. Promising advancements in the field of bio-signal acquisition marked the emergence of Ear-sensored electroencephalographic (EEG) devices that have varied applications, including recording and monitoring the brain's activity. This scoping review provides a better understanding of the rationale for implementation, current trends, and future research applications of Ear-sensored EEG monitoring techniques in hearing aids.

1. INTRODUCTION

At the heart of rehabilitation programs for listeners with hearing deficits is the use of amplification devices, commonly known as hearing aids. A hearing aid is essentially a miniature public-address system that amplifies signal,[1] that otherwise remain inaudible to the hearing-impaired listeners. Depending on the degree and configuration of the hearing loss, the hearing aid performs various signal processing operations such as noise reduction, frequency-specific amplification, feedback suppression, steering directionality towards source location, etc. All these hearing aid processes are aimed at compensating the listener's hearing loss.

However, hearing, unlike listening, is a perceptual event that is dependent on the end outcome of how the brain decodes the sound information relayed through the hearing aid and auditory pathway. Hearing is based not only on the audibility of the signal, but also on how that sound is biologically coded, integrated, and used by auditory neural mechanisms and brain. As described by Tremblay *et al.*,[2] the entry-level for this complex ear-brain neural network is right at the microphone where sound signals first interact with the hearing device. The signal processing operations in the hearing aid alter the incoming signal according to the listener's characteristics (degree and type of hearing loss), and this modified signal gets encoded at subsequent stages of processing: the Ear, brainstem, midbrain, and the cortex. Each of these auditory mechanisms acts as biological codes for signal processing. Deficit at any level can have a negative impact on the resultant perceptual event. The brain is considered an essential component of auditory rehabilitation for this very reason. Yet, very little is known about how the brain contributes to the perception of amplified sound and the successful use of hearing aid amplification. The current review aims to relate the applicability of advancements in EEG technology to hearing aids.

1.1 Rationale for the application of EEG technology in hearing aids: current limitations

Electroencephalography (EEG) is a non-invasive method for recording signals from the brain. The use of EEG in studying brain activity is highly preferred owing to its high temporal resolution and reasonable spectral resolution[3]. Research evidence points at different applications of EEG signals, such as to establish hearing threshold levels,[4-6] frequency selectivity,[7,8] and loudness perception[9]. The applications of EEG can be extended to hearing aid fitting and selecting audio processing strategies[10]. Integration of EEG technology into hearing amplification devices could potentially aid in individualized hearing aid fitting and reprogramming. The use of EEG as an objective measure of hearing aid processed signals can minimize the extensive costs involved in conventional hearing aid testing and fitting expenses. Despite such promising advantages, application of EEG technology in present-day hearing aid fitting scenario is still at its infancy. The introduction of EEG in hearing aids is limited by barriers like the lack of mobility, comfort, high cost, availability of technology, and robust EEG decoding schemes.

1.2 In Ear-EEG & around the ear-EEG: promising new technologies for hearing amplification devices

Recently, a novel EEG recording approach called 'the ear-EEG' was introduced. Here, the signal is recorded from electrodes embedded in an individualized hearing aid-like earpiece placed in the ear canal[11-13] or around the Ear.[14-17] Denk *et al*.[18] compared the differences in-ear-EEG hearing device and around the ear-EEG system based on live monitoring of incoming speech on 17 normal-hearing listeners. The findings from the study showed larger SNR in the around the ear electrodes, which was significantly higher than the maximum SNR recorded for in-ear electrode. However, extending these findings to draw inferences about efficacy of in-ear and around the ear-EEG devices is limited due to various device-related (technology involved, type of electrodes used: gold plated/quick cell, type of electrolyte used) and subject-related factors (impedance, volume of the ear canal, orientation of electrodes in the ear canal).

Despite the better signal-to-noise ratios recorded for around the ear-EEG compared to the in-ear-EEG,[18] the use of latter is preferred owing to the following practical advantages:[11] (a) diminished sensitivity to interference caused by electrical equipment in the recording environment (computers, electric lights, radios), due to the reduced strength of electric fields in the ear canal and the short distances between the system leads and the amplifier; (b) conductive gel-free recordings (*e.g.*, capacitive electrode technology) without the need for skin preparation; (c) reduced interference of hair around the Ear. The studies cited in this draft focuses most on the in-ear EEG, rather than around the Ear as most reports on the former technology make use of the above discussed benefits. In-ear-EEG apart from being discrete and miniaturized, has a great potential for use in everyday listening environments and clinical practice. It has multi-folded applications in hearing assessment and rehabilitation using EEG recorded signals. The studies exploring the use of this recent technology in hearing aids are analyzed critically in the current review.

2. METHOD

2.1 Review of literature

In this article, we present a comprehensive review of some studies and reviews up to May 2020, that reflect the emerging trends of EEG utilization in hearing aids with a specific focus on (a) Auditory processes identification (detection, discrimination, categorization) and (b) speech perception

2.2 Search Strategy

A systematic search was performed in the PubMed, Research gate and Google scholar, leading up to May 2020, using the following terms: neuro-steered hearing aids, EEG-based hearing aids, attention driven hearing aids, Ear centered sensing, wearable EEG, in ear-EEG and around the ear-EEG.

2.3 Study Selection

While screening titles and abstracts, any duplicates, animal studies, case reports, and articles written in languages other than English were excluded. Studies published only in abstract were also excluded. However, articles published as a part of the conference proceedings were included in the review.

3. RESULTS AND DISCUSSION

The search in PubMed, Research Gate, and Google scholar retrieved a total of 55 articles, but only 13 met the inclusion criteria and were included in the study. Out of these, two publications were not included in the study as one was in the pre-print stage, and the other was an editorial article. The remaining accepted reviews (11) evaluated the utility of EEG on at least one outcome of interest (see method). Data are summarized in Table 1.

3.1 Applications of in-ear-EEG and around the ear-EEG

As seen in Table 1, Ear-EEG (both in-ear and around the Ear)[18] can help us gain better understanding of the brain's response in terms of hearing sensitivity, auditory discrimination, and quantifying the effectiveness of source-related attention, and speech perception. As shown in Table 1, the Ear-EEG recording concept is tested using several standard EEG paradigms, benchmarked against conventional scalp EEG, and its feasibility is proven in many studies. Incorporation of such a system promises several advantages: including fixed electrode positions, user comfort, robustness to electromagnetic interference, feedback to the user, and ease of use. The inclusion of ear-EEG in monitoring brain dynamics during motion in everyday situations would be very valuable in monitoring the effects of various events on hearing aid users.

In this context, it is imperative to perform detailed comparisons between ear-EEG and regular scalp EEG, as a high correlation between them is mandatory for possible incorporation of this technology in contemporary hearing aids. Meticulous research is carried out in establishing these links (see Table 1). Extensive research is also warranted to establish such relationships not only in different listening scenarios but also in all modalities (frequency domain and time domains correlation), before the implementation of the technology in hearing aids. Although the application of EEG in hearing devices (hearing aids or cochlear implants)[16] is appealing and research into the integration of biosensors into hearing devices is underway, a feasibility evaluation should be done considering various physical (athropometric variations of Ear with gender, age and ethinicity), environmental (reverberation, attended and unattended source etc), mechanical (viscoelasticity of the ear-EEG plug, distance of electrodes) and acoustic factors (noise, music perception *etc*). Until systematic research probing into the above effects is not carried out, drawing inferences and generalization of any research finding should always be done with adequate caution. Although the full extent to which Ear-EEG can replace standard on-scalp recordings remains to be seen, it has been validated for the extraction of several key EEG features, including the AAR, ASSR, and P300 paradigms, and hence show promising implications in hearing aids.

4. FUTURE DIRECTIONS

Ear centered sensing has opened enormous possibilities for integrating bio-signals, wherein a combination of signals from different sensory modalities can be used to drive several applications, including auditory rehabilitation. The integration of physical measures (motion, temperature and moisture), behavioral measures (detection, discrimination, and comprehension of sensory stimuli), and electrophysiological measures (electroencephalography - EEG), electrocardiography - ECG), electromyography - EMG, electrooculography - EOG, and electrodermal activity- EDA), over a long duration of time will enhance understanding of psycho-physiological processes. Ear-centered sensing is hence an alluring field of interest for scientific, diagnostic, and therapeutic ventures and is predicted to have a promising role for mobile health applications in the future.

Table 1. Summary of articles reviewed.

Sl. No.	Authors	Sample size and age range	Stimuli	Measure used	Auditory process identification	Speech Perception	Implication for neuro-steered hearing aids
1.	Mirkovic et al[19].	16 NH and 15 linearly aided SNHL 62 –75y (mean age 74y)	Continuous speech stream stimuli.	EEG impulse responses (estimated using the cross-correlation between the recorded EEG signal and the temporal envelope of the signal at HA o/p)	N/A	Latency of attentional processes (P2 and N2) are is influenced by microphone settings. EEG impulse responses showed faster N1P2 responses for NH and SNHL; while larger N2 peak amplitude was noticed for the SNHI group when the directional microphone setting was activated.	The latency of evoked responses can be implied in neuro-steere HA for inferring benefi from processing strategies and acting on speaker-selection automatically
2.	Fiedler et al.[20]	8 NH subjects 23 – 49y (mean age 34.6y)	Discrete signals: dichotic oddball paradigm stimuli (pure tones 410 Hz vs. 610 Hz) Continuous: Audiobooks presented dichotically	Cross-correlation of adjacent channels	N/A	Envelope extraction was done for both acoustic and EEG waveforms. The obtained envelopes were compared using Cross-correlation in each channel, across time, which showed effective tracking (no significant difference $p>0.05$ between envelopes for all electrodes)	This study showed the feasibility of extracting relevant neural features from in-ear recorded ear-EEG, which might augment future hearing technology
3.	Mikkelsen et al. [21]	10 NH subjects 23–43y (median 30y)	ASSR: 500 Hz tone with 40 Hz amplitude modulation MMN & P300: 0.5kHz & 1kHz	Correlation between the measured and the predicted signals	Predicted EEG conveyed cortical information in ASSR, MMN and P300 paradigm. Ear-EEG yields similar performance as conventional EEG for spectrogram-based analysis and similar timing of ERP components	N/A	Reliable recording of ear-EEG is suggestive o the potential applicatio of this technology in neuro-steered hearing aids for signal detectio discrimination and categorization in both attended and unattended states.

	[11]	NH subjects Age: N/A	to close eyes suddenly ASSR: 500 Hz tone with 40 Hz amplitude modulation P300: 0.5 kHz & 1kHz	Power density and wave form morphology	spectral Coherence B/W scalp and EEG recordings for AAR Power density of 40 Hz was seen in Ear-EEG electrodes (2.5-3 V^2/Hz) and on-scalp electrodes (4-5.5 V^2/Hz) for ASSR. Comparable wave morphology, peak amplitude and latency for P300		extraction of several ke EEG features, includin the AAR, ASSR, and P300 paradigms convincingly (comparable to scalp EEG), hence can be applied in HA to drive detection (inferred fron AASR) and categorization (inferred from P300), with furthe refinement in ear-EEG.
5.	Christensen et al. [22]	15 NH subjects Age: N/A	Narrow band Chirps centered on 0.5, 1, 2 and 4 kHz	Compared the ASSR hearing thresholds recorded in ear-EEG & scalp EEG	ASSR thresholds estimated from in-ear -EEG were on average 15.0± 3.4, 9.1 ± 4.4, 12.5± 3.7, and 12.1± 2.6 dB above scalp EEG thresholds for 0.5, 1, 2, and 4 kHz, respectively	N/A	Objective hearing threshold estimation based on ear-EEG can b integrated into hearing aids, thereby allowing hearing assessment to b performed by the hearing instrument on a regular basis.
6.	Christensen et al. [6]	19 subjects diagnosed with SNHL (30 to 65 dB HL) 52 to 79 y (mean 67.3±9.6)	Narrow-band Chirps centered on 0.5, 1, 2 and 4 kHz	Compared the ASSR hearing thresholds recorded in ear-EEG & scalp EEG	Thresholds estimated using in-ear-EEG were found to be elevated at an average of 5.9, 2.3, 5.6, and 1.5 dB relative to scalp thresholds at 0.5, 1, 2, and 4 kHz, respectively. Overall the threshold detection rate for ear-EEG was 20% lower than the detection rate for scalp EEG.	N/A	Further refinement of the method is needed t optimize the ear-EEG threshold detection rat before extending its applicability for subject with SNHL.
7.	Haghighi et al. [23]	10 NH subjects 25 to 30y	Diotic presentation of auditory stimuli (stories recorded in male and female voice)	Cross-correlation coefficients between the EEG envelope and the sound source envelope for attended and unattended stimuli	N/A	Online sound source modulation detects the level of attended sound source more accurately than the unattended source, with the AUC for the former being as high as 0.8-1.0.	EEG-based attention detector can be incorporated in an online setting for hearing aid application especially for auditory stream segregation and localization.

	al. [15]	subjects mean age 24.8 y	streams consisted of lesser-known fairy tales narrated in German	method to compare EEG and i/p signal		captured neural signals that allowed the identification of the attended speaker with 69.3% accuracy, while cap-EEG signals resulted in an accuracy of 84.8%.	indicative of Ear-EEG potential in the brain-computer interface for neuro-steered hearing aids
9.	Kappel et al. [24]	12 NH subjects Mean age 30.9 ± 5.6y	White noise amplitude modulated at 40 Hz	Independent component (ICs)and source analyses for ASSR,SSVEP , MMN, and AAR	High correlation (0.76-0.85) for the ASSRs and SSVEPs between scalp- and Ear- electrode configurations	N/A	Forward models can be used to explore the ear EEG electrode sensitivity to brain sources for different ASSRs and SSVEPs, which can then be used to drive HAs based on online tracking of the corresponding ICs.
10.	Goverdov-sky et al. [12]	One NH subject 32 years	1 kHz sinusoid, amplitude-modulated with a 40 Hz sinusoidal signal.	Power spectral density (PSD) estimates for the Ear- and scalp-recordings of SSVEP	Degree of overlap was 50% between ear EGG and scalp EEG recordings	N/A	Ear-EEG device can be readily used to acquire ASSR and SSVEP, thus paving the way for wearable EEG acquisition, whose utility can be extended to HAs
11.	Nogueira et al. [16]	10 NH listeners, 10 bilateral CI users NH (mean age: 42.8, range: 24–77), CI (mean age: 67.5, range: 59–80)	Two German stories, one of which is attended and the another ignored.	Envelope tracking algorithm to compare cap-EEG and around the ear-EEG system	N/A	All NH listeners (50-60% accuracy) and 9 out of 10 CI (50-60% accuracy) achieved moderate decoding accuracies in the ear-EEG systems, which were lower to cap-recorded accuracy (70-80 in NH, 60 – 70 in CI (see figure 10 of Nogueira et al. [16])	The study proved that i is possible to decode selective attention in CI users despite the electrical modality of auditory nerve stimulation and related artifacts of the same in the recording.

Abbreviations: AAR - alpha attenuation response; ASSR-auditory steady-state response;
AUC- area under the curve; CI- Cochlear implant; IC- Independent component;
MMN – mismatch negativity; NH- Normal hearing; N/A- Not applicable; P300-positivity at 300;
SNHL – sensorineural hearing loss; SSVEP-steady-state visual evoked potential

5. CONCLUSION

Although the research on the utility of EEG in steering hearing aids has overcome many hurdles in the last decade and looks to be promising new approach, using hearing aids to improve human communication will ultimately depend on more than just brain measures. Many factors can contribute to aided speech understanding in noisy environments, including device centered (*e.g.,* directional microphones, signal processing and gain settings) and patient-centered variables (*e.g.,* age, attention, motivation, biology, personality and lifestyle). The research aimed at exploring one variable in isolation (*e.g.,* neural mechanisms underlying sound detection and discrimination) is likely to fall short when trying to optimize the many interactive stages involved in human communication.

The use of EEG to quantify and model neural mechanisms associated with the perception of amplified sound is complicated and sometimes counter-intuitive. For this reason, audiologists need to exercise caution in the interpretation of EEG evoked for amplified sounds, which cannot merely be generalized using prior published data acquired from normal hearing or unaided participants.

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INFORMATION FOR AUTHORS

ARTICLES

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SUMISSIONS

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