Research Article

The Output Signal-to-Noise Ratio and Speech Perception in Noise: The Effect of Multichannel and Free-Channel Hearing Aids

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Highlights

- The channel-free hearing aids are more beneficial with greater noise attenuation
- Better SNR in channel-free hearing aids, even in the presence of noise
- Subjective preference was for channel-free hearing aids

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ABSTRACT

Background and Aim: The output Signal-to-Noise Ratio (SNR) is one of the essential factors in hearing aid benefits. There is limited evidence regarding SNR improvement by the Channel-Free (CFHA) and Multi-Channel Hearing Aid (MCHA) and the speech understanding in noise through them. This study aimed to investigate the extent to which output SNR was modified by CFHA and MCHA processing and the variation in aided speech recognition abilities with a change in output SNR.

Methods: Thirty-six participants aged 50–65 years were included. A chosen CFHA and MCHA were used to obtain the output SNR and sentence recognition in noise in four different processing algorithms (linear, linear+noise reduction, WDRC, WDRC+noise reduction). Hagerman's phase inversion technique was used to measure the attenuation of noise and, in turn, to obtain the output SNR of the hearing aid.

Results: In all hearing aid processing algorithms among those with normal hearing and people with hearing loss, the output of CFHA revealed higher attenuation values than that of MCHA. There was a significant effect of the hearing aids and processing algorithms in both normal and individuals with hearing impairment on the mean SNR. Further, multiple linear regression analysis results showed that whether the hearing is channel-free or multi-channel significantly predicted speech recognition scores, while output SNR and processing algorithms did not.

Conclusion: The signal processing algorithms in CFHA had greater noise attenuation values, better output SNR, and speech recognition scores, showing an advantage over the modern MCHA among individuals with hearing impairment.

Keywords: Channel-free hearing aids; multi-channel hearing aids; sensorineural hearing loss; speech recognition

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Introduction

Many factors have been reported in the literature contributing to the hearing aid benefit for speech perception in noise, including subject-related variables (age, degree of hearing loss, cognition, duration of hearing loss, etc.), environment or speaker-related as well as factors related to hearing aids (various features, different signal processing strategies, etc.). The output Signal-to-Noise Ratio (SNR) has also been reported as an essential contributing factor to hearing aid benefit [1, 2]. Hagerman and Olofsson [3] devised a phase inversion technique to understand the effect of noise reduction and compression on hearing aid output. In this technique, output SNR is derived by measuring the difference between speech and noise output with speech and inverted phase noise output. This method has been utilized to understand the effect of various hearing aid features, such as noise reduction algorithms and compression [3-5]. Even though hearing aids amplify sounds, the SNR may not be sufficient for optimal speech understanding. Modern hearing aids have a variety of signal-processing algorithms to overcome the issues affecting speech audibility and comprehension. Among them are the Multi-Channel Hearing Aids (MCHA) and Channel-Free Hearing Aids (CFHA), which have similar purposes to improve speech recognition but vary in signal processing. The MCHA splits the incoming information (either speech or noise) into a specific number of channels and is analysed based on the amount of modulation. Based on these, each channel is subjected to different amplification and compression, enabling independent control of gain over a narrow range of the incoming signal [6, 7]. This results in the user compromising either between listening comfort or processing all the spectral information present in the signal by choosing fast or slow-acting compression [8].

Considering concerns such as channel interaction, spectral smearing, altered temporal representation, speech intelligibility, and understanding in multi-channel systems, CFHA were initially developed. In CFHA, the speech is continuously adapted to adjust for the spectrum's shape, and it is processed at a faster rate, approximately 20,000 times per second. CFHAs have input-dependent filters applied to all signals, allowing frequency-dependent speech compression without splitting the signal. It has been demonstrated to preserve listening comfort without sacrificing the temporal and spectral properties of the input sound [9].

Improvement in SNR by hearing aids is one of the essential prerequisites for good speech recognition in the presence of noise. Despite the availability of both hearing aids, more research is needed to determine how these algorithms affect SNR and speech recognition. Studies comparing these two techniques (CFHA and MCHA) are limited to speech recognition measures, and the difference in SNR improvement in these two hearing aids that vary based on the channels has not been studied. Therefore, the current study aimed to ascertain how multichannel and free-channel hearing aid signal processing affected output signal-to-noise ratio and speech perception in noise. Additionally, the improvement in speech recognition skills with an improved signal-to-noise ratio was assessed.

Methods

Participants

The study involved 36 participants aged between 50 to 65 years. The participants were all right-handed and native Kannada speakers. The normal hearing group included 18 adults (mean=59 years) with air and bone conduction thresholds less than 25 dB HL between 250 and 8000 Hz and speech recognition scores of ≥90% in quiet. The hearing-impaired group comprised 18 adults (mean=62 years) with bilateral mild to moderately severe sloping sensorineural hearing impairment. It was confirmed through pure tone thresholds of ≤30 dB HL at 0.25 kHz to ≤4 kHz and ≥55 to ≤65 dB HL from 4 kHz to 8 kHz and the speech identification scores of >70% in quiet.

Stimuli

Kannada sentences developed by Geetha et al. [10] were recorded by a native Kannada female speaker for both acoustic (for recording hearing aid output) and behavioural testing. Two lists of ten sentences each were used for this purpose. To simulate the properties of realistic hearing situations and for the hearing aid to detect the presence of noise, a 6-talker babble (3-male and female talkers) was used. The speech and babble long-term average speech spectra were within 3 dB between 160–8000 Hz.

Hearing aids

Oticon Opn S3 (24 channels, 116 dB maximum output, 54 dB maximum gain) and Bernafone Viron 5 (channel-free processing, 115 dB maximum output, 54 dB maximum gain) receiver in the canal hearing aids were used in the present study. Both the hearing aids are from Demant A/S, with similar compression settings (attack time=2 ms and release times=20 ms) but differ in how the signal is processed and have different noise reduction strategies. The output of each hearing aid was recorded with Noise Reduction (NR) algorithms ON and OFF (to the maximum level), with Linear (LIN) and Wide Dynamic Range Compression (WDRC) settings independently, in order to account for these variations.

The hearing aids were coupled to the participant's ears with suitable ear domes. Regardless of the group (normal hearing or hearing-impaired), the hearing aids were programmed to NAL-NL2 [6] targets for all the participants using individual threshold levels and real-ear measures. Fine-tuning was done only if required to match the target and the real ear responses. Fine-tuning was not done compulsorily in all the participants. However, as normal-hearing individuals were also included in the study, loudness discomfort levels were measured as a part of ethical considerations, to make sure they were not presented with unusually loud stimulus. Frequency lowering, directional microphones, feedback manager, and automatic adaptation manager were disabled in both hearing aids. Fine-tuning was done (if necessary) to match the target and real ear responses, especially in the 250–3000 Hz region. The difference in output Sound Pressure Level (SPL) between the two devices was within 5 dB in all the hearing aid modes. The selected hearing aids had options to vary between linear and compression methods. Four memories were created for acoustic and behavioural measures (linear, linear+noise reduction, WDRC, WDRC+noise reduction) in both hearing aids.

Procedure

According to the ANSI S3.1., 1999 (R 2013) guidelines, a double-walled, sound-treated room was used for the behavioural and acoustical measures. After briefly outlining the procedure, informed consent was initially obtained from each participant. The research is compliant with ethical standards according to the 1964 Helsinki Declaration.

Behavioural testing - sentence reception threshold in noise

The speech stimuli and the recorded babble were presented through a speaker placed on the better ear side with the poorer ear blocked with foam inserts. The head of the participant was at one metre away from the speaker.

The procedure described by Miller et al. [1] was used to obtain SRTn for individual participants separately. SRTn was defined as the SNR of the stimuli an individual participant requires to correspond to 50% correct performance. The speech level was fixed at 50 dB HL. The SNR of the stimuli were varied by varying the babble level individually for each participant to correspond to 50% correct responses. To obtain SRTn, the testing started at an SNR of +15dB. Two sentences were presented at each SNR and if the participant repeated the sentences correctly, the SNR was decreased in 5 dB steps. This procedure was followed until the participant starts to miss half of the words in two consecutive sentences presented. The lowest SNR at which the participant could repeat 50% of the words in the sentences presented was further fine-tuned by varying the SNR in 1-dB steps to obtain the SRTn.

The approximate SRTn of each participant was considered as the input SNR for the hearing aid test conditions. The order of presentation of the sentences, processing algorithms (LIN+NR, WDRC, WDRC+NR), and hearing aids (MCHA and CFHA) were randomised and counterbalanced across participants. In each hearing aid condition, the listener heard two sentence lists. The average was taken between lists in each condition. Noise (babble) levels that yielded SRTn for 50 dB HL speech for individual participants across all hearing aid processing algorithms were recorded and used for recording the hearing aid output for acoustical testing.

Acoustical testing-modified Hagerman's phase inversion technique

In various settings, the phase-inversion technique developed by Hagerman and Olofsson [3] and refined by Souza et al. [11] was utilized to separate speech and babble from the output of the hearing. This was later used to obtain output SNR. Sentences and babble were presented in other conditions, which vary in phase only:

1. Speech and noise in original phase (+Sp +N)
2. Original-phase noise with phase-inverted speech

(–Sp +N)

3. Speech in original phase with phase-inverted noise (+Sp –N)

4. Phase inverted speech combined with noise (–Sp –N)

Recording of hearing aid output

Real ear measurements were not taken since it is difficult to hold the participant's head still for an extended period using Hagerman's phase inversion technique, which takes multiple recordings to get a stable amplitude. Following the SRTn measurement of each participant, each hearing aid was coupled to a 2 cc coupler and placed at the same height and distance as the patient's head. A condenser microphone (B-2 PRO, Behringer, Germany) was attached to the terminating portion of the coupler, and a laptop having Adobe software in order to measure the hearing aid output. Four recordings were made for each hearing aid condition: +Sp +N, –Sp +N, +Sp –N, –Sp –N.

Comparing the average levels with and without the Noise Reduction (NR) algorithm activated allowed for measuring the gain reduction. The average gain reduction of both the aids was recorded. Also, the hearing aid output was recorded in linear and non-linear modes to understand the effect of non-linear signal processing on output SNR. Thus, the hearing aids were programmed to four processing algorithms: Linear, Linear+NR, Wide Dynamic Range Compression (WDRC), and WDRC+NR.

Extraction of speech and noise signals

An error term resulting from internal noise and distortion due to the processing in the hearing aid will be present at the output along with the speech and noise signals. The error was quantified by adding phase inverted speech and noise: error=(+Sp +N)+(–Sp –N). The output signals were superimposed based on the below equations in MATLAB (The Mathworks Inc) to extract the speech and noise levels.

1) Extracted noise=(+Sp +N)+(–Sp +N)

2) Extracted speech=(+Sp +N)+(+Sp –N)

To consider the error to be negligible, it should be below the softest level of the extracted speech or the noise signal. Hence, measuring the attenuation obtained is one technique to evaluate the difference between the error and the recovered signal of interest. "Attenuation" is the term used to describe the difference that was computed between the error level and the speech or noise level. Attenuation values of 20 dB or more have been demonstrated in prior research employing similar techniques to provide enough separation of the extracted signals [12-14]. Hence, the recordings having a minimum difference of 20 dB, or more were considered reliable and included in the analyses otherwise they were subjected to reanalyses. Further, if a recording did not pass the 20 dB criterion even after re-extraction, the recording was discarded and was not considered for analyses.

Results

The efficacy of the modified Hagerman's phase inversion technique was measured in each condition. The speech and noise attenuation values were initially analysed after error subtraction in normal hearing individuals and individuals with hearing impairment. By comparing the values to the 20 dB criteria suggested in the previous literature [12-14], the effectiveness of the Hagerman's technique was evaluated for each specific situation. Mean attenuation values for speech and noise in both groups using different hearing aids and listening conditions were calculated (Tables 1 and 2). Except for the linear condition for the CFHA, the rest met the attenuation criterion.

Effect of the hearing aids and processing strategies on the output signal-to-noise ratio

Based on the SRTn of the participants, different SNR were used for recording the hearing aid output. For each data analysis, this variability was removed by comparing the retrieved speech and noise levels to those for the linear condition (i.e., LIN). The mean change in SNR (re: LIN) at the hearing aid output was measured using Hagerman's phase inversion technique to ascertain whether the processing of the hearing aid affects the SNR from the input to the output (Table 3).

A two-way, repeated-measures ANOVA was carried out within each group to see if there were any changes in SNR (re: LIN) across the hearing aids (MCHA and CFHA) and types of processing (LIN+NR, WDRC, WDRC+NR). Within the normal hearing group, there was a significant main effect of the hearing aid ($F_{(2,17)}=14.6$, $p<0.001$) and type of processing ($F_{(2,17)}=160.882$, $p<0.001$). However, there was no significant interaction effect of

Table 1. Comparison of the noise attenuation values (in dB) of the two hearing aids between hearing-impaired individuals and normal hearing individuals

The hearing aid processing algorithm	Normal hearing		Hearing-impaired	
	CFHA	MCHA	CFHA	MCHA
Linear	27.55	24.03	29.60	25.33
Linear+NR	28.06	27.53	30.05	30.91
WDRC	27.81	25.78	31.03	31.06
WDRC+NR	28.50	28.79	30.54	30.51

CFHA; channel free hearing aids, MCHA; multi channel hearing aids, NR; noise reduction, WDRC; wide dynamic range compression

Table 2. Comparison of the speech attenuation values (in dB) of the two hearing aids between hearing-impaired individuals and normal hearing individuals

The hearing aid processing algorithm	Normal hearing		Hearing-impaired	
	CFHA	MCHA	CFHA	MCHA
Linear	28.99	24.08	27.47	22.80
Linear+NR	29.50	28.49	28.09	29.21
WDRC	29.25	26.29	28.08	29.01
WDRC+NR	29.47	28.50	28.09	28.05

CFHA; channel free hearing aids, MCHA; multi channel hearing aids, NR; noise reduction, WDRC; wide dynamic range compression

Table 3. Mean change in signal-to-noise ratio (re: linear) at the hearing aid output

The hearing aid processing algorithm	Normal hearing		Hearing-impaired	
	MCHA	CFHA	MCHA	CFHA
Linear+NR	0.25	0.17	0.23	–0.09
WDRC	–0.44	–0.70	–0.63	–0.89
WDRC+NR	–0.53	–1.57	–0.81	–1.70

MCHA; multi channel hearing aids, CFHA; channel free hearing aids, NR; noise reduction, WDRC; wide dynamic range compression

the hearing aids and types of processing ($F_{(2,17)}$=56.17, p>0.05). Additionally, pairwise comparisons (Table 4) with a Bonferroni adjustment for multiple comparisons revealed that there were significant variations in mean SNR (re: LIN) between MCHA and CFHA (p=0.001) as well as between all the types of processing (p<0.001). CFHA. The results of a two-way repeated-measures ANOVA similarly revealed a significant main effect of the hearing aid ($F_{(2,17)}$=62.13, p<0.001) and type of processing ($F_{(2,17)}$=83.12, p<0.001) but no significant interaction effect between the two ($F_{(2,17)}$=26.879, p>0.05) among those with hearing impairment.

The results of pairwise comparisons (Table 4) showed that the difference in mean SNR (re: LIN) was statistically significant between the two hearing aids ($p<0.001$). Further, the pairwise comparison was significant between all processing algorithms ($p<0.001$), except between WDRC and WDRC+NR.

Effect of hearing aid types and processing strategies on the speech recognition scores

The speech recognition scores for each group and hearing aid processing algorithms are shown in Table 5. The mean score of the individuals with normal hearing (MCHA and CFHA, 59.18% and 62.68% respectively) was slightly higher than individuals with hearing impairment (MCHA and CFHA, 50.42% and 56.56% respectively). A repeated measures ANOVA among the individuals with normal hearing revealed no significant main effect of the hearing aids ($F_{(2,17)}=28.4$; $p>0.05$) and listening conditions ($F_{(2,17)}=33.1$; $p>0.05$), as well as no significant interaction effect ($F_{(2,17)}=41.3$; $p>0.05$). A significant effect of hearing aid type ($F_{(2,17)}=12.61$; $p<0.05$) was obtained for individuals with hearing impairment, but there was no significant effect of listening conditions and no interaction effect (Table 6).

Variation in speech recognition scores with change in signal-to-noise ratio

The difference between the change in SNR and the change in speech recognition (re: LIN) was assessed. A multiple linear regression analysis was carried out within each group to determine whether changes in speech recognition (related to LIN) could be anticipated. SNR (relative to LIN; continuous variable), hearing aid type (dummy variable), and processing methods (dummy variable) were the three primary predictors in the initial model. The second model incorporated predictors of each subject (coded as dummy variables) to assess any changes to significant predictors when variance within

Table 4. Post-hoc comparisons for output signal-to-noise ratio across hearing aids and different processing algorithms for the two groups

Normal hearing group			p
	CFHA	MCHA	$<0.001^{***}$
CFHA	LIN+NR	WDRC	$<0.001^{***}$
	LIN+NR	WDRC+NR	$<0.001^{***}$
	WDRC	WDRC+NR	$<0.001^{***}$
MCHA	LIN+NR	WDRC	$<0.001^{***}$
	LIN+NR	WDRC+NR	$<0.001^{***}$
	WDRC	WDRC+NR	$<0.001^{***}$
Hearing-impaired group			
	CFHA	MCHA	$<0.001^{***}$
CFHA	LIN+NR	WDRC	$<0.001^{***}$
	LIN+NR	WDRC+NR	$<0.001^{***}$
	WDRC	WDRC+NR	0.840
MCHA	LIN+NR	WDRC	$<0.001^{***}$
	LIN+NR	WDRC+NR	$<0.001^{***}$
	WDRC	WDRC+NR	0.562

CFHA; channel free hearing aids, MCHA; multichannel hearing aids, LIN; linear, NR; noise reduction, WDRC; wide dynamic range compression

*** $p<0.001$

subjects was considered in the model. Among normal-hearing individuals, the first model did not reveal either the processing strategies, SNR, or type of hearing aid as a significant predictor. This model explained 8.6% of the variation as a whole. Additionally, when the individual participants were included in the model, there were minor variations in the importance of the condition's ability to predict a change in speech recognition.

Table 5. The mean percentage correct scores for different hearing aid processing algorithms obtained by the two groups

The hearing aid processing algorithm	Normal hearing		Hearing-impaired	
	MCHA (%)	CFHA (%)	MCHA (%)	CFHA (%)
Linear	56.12	58.40	52.25	54.78
Linear+NR	58.35	61.61	48.91	54.92
WDRC	60.78	63.21	51.53	60.48
WDRC+NR	61.49	67.53	48.99	56.05
Mean of all the conditions	59.18	62.68	50.42	56.56

MCHA; multi-channel hearing aids, CFHA; channel free hearing aids, NR; noise reduction, WDRC; wide dynamic range compression

Table 6. Post-hoc comparisons for speech recognition scores across hearing aids and different processing algorithms

Normal Hearing group			p
	CFHA	**MCHA**	0.750
CFHA	**LIN+NR**	**WDRC**	0.342
	LIN+NR	**WDRC+NR**	0.621
	WDRC	**WDRC+NR**	0.428
MCHA	**LIN+NR**	**WDRC**	0.834
	LIN+NR	**WDRC+NR**	0.423
	WDRC	**WDRC+NR**	0.562
Hearing-impaired group			
	CFHA	**MCHA**	< .05*
CFHA	**LIN+NR**	**WDRC**	0.523
	LIN+NR	**WDRC+NR**	0.876
	WDRC	**WDRC+NR**	0.321
MCHA	**LIN+NR**	**WDRC**	0.071
	LIN+NR	**WDRC+NR**	0.213
	WDRC	**WDRC+NR**	0.080

CFHA; channel free hearing aids, MCHA; multichannel hearing aids, LIN; linear, NR; noise reduction, WDRC; wide dynamic range compression
*** p<0.001

Nevertheless, SNR had no discernible impact on the hearing aid output. This model accounts for the entire range of speech recognition for people with normal hearing.

Further, among the individuals with hearing impairment, the first model revealed that only the type of hearing aid (CFHA and MCHA) significantly predicted speech recognition (t=1.321; p=0.005), whereas the SNR and processing strategies did not. The total variance in speech recognition scores explained by this model was only 8%. Further, when individual subjects were added as predictors and a new second model was created, only the type of hearing aid was found to be a significant predictor (p=0.006), whereas the rest remained insignificant. This second model explained a total of 22.4% of the variation in speech recognition.

Discussion

Using Hagerman's phase-inversion technique, the findings of the current study revealed that CFHA greatly improved speech and noise attenuation; however, the level of improvement varied depending on the type of processing applied. This led to better SNR while using CFHA compared to MCHA.

Effect of hearing aids and processing strategies on the output signal-to-noise ratio

According to the results of the current investigation, the SNR variations brought about by hearing aid processing were between 0.25 and –0.53 dB SPL for people with normal hearing and between 0.23 and –0.81 dB SPL for people with hearing loss when compared to the linear condition for MCHA. CFHA exhibited higher changes in SNR; for the normal hearing group, the output change ranged between 0.17 to –1.57, and for the individuals with hearing impairment, it varied between –0.09 to –1.70. This better attenuation provided by the CFHA might be due to the hearing aid features that can alter the acoustic properties of a speech signal. Although multichannel compression systems are less at risk of background noise than single-channel compression systems [15], the channel-free system tends to be superior to the multichannel compression system. This may be due to the ability of channel-free processing to detect and process the wideband signal while offering different compression ratios for different frequencies [16]. Additionally, it has been demonstrated that the input signal can be processed in parallel by continually gauging the sound pressure level of the input signal and modifying the gain of the hearing aid in accordance with the obtained data. This prevents the hearing aid from splitting the input signal into various bands and assigning gains individually for each band in which the signal is processed. This explains the significantly better attenuation, and SNR obtained using CFHA than MCHA in the current study. Hence, in the present study, the change in SNR was slightly higher for CFHA than MCHA, regardless of combining different types of processing.

In addition, the SNR changes were found to vary between normal hearing- and hearing-impaired individuals. The mean change in SNR was found to be greater among individuals with hearing impairment than that was seen in normal hearing individuals. This could be attributed to the differences in the compression characteristics while programming the hearing aids between the two groups. Further, it was also found that when compared across various types of processing, the mean change in SNR was significantly better for WDRC than the other two processing algorithms (LIN+NR and WDRC+NR). Although by convention, the WDRC or LIN strategy and the NR strategy must work as an additional advantage to improve SNR, the present investigation has discovered conflicting outcomes. In the current study, WDRC has been shown to have a more significant effect on output SNR than when combined with the NR. This might be explained by the impact of the multi-talker speech babble employed as noise in the current investigation. Previous research has demonstrated that the NR algorithms are not fully effective when speech babble is used as noise because the inherent temporal modulations present in the speech that gets retained in the babble make it challenging for the algorithm to distinguish between speech and noise [3, 4]. An exception is the lack of significant difference in SNR change between WDRC and WDRC+NR in individuals with hearing impairment. In a study by Brons et al. [4], it was demonstrated that the WDRC processing partially offset the SNR improvement brought about by NR processing, reducing the benefits of NR when paired with WDRC. A similar offset effect of the output SNR by the combination would have been attributed to the lack of significant difference between WDRC and WDRC+NR observed in the current study.

Hence, the modifications to SNR in the present study were majorly dependent on the hearing aids and processing algorithm, with the WDRC of CFHA showing the most significant effect on SNR.

Effect of hearing aids and processing strategies on the speech recognition scores

The speech recognition scores were better among individuals with normal hearing than those with hearing impairment. This was seen irrespective of the hearing aids and processing type. This is in accordance with the fact that speech recognition is affected in individuals with hearing loss, and the changes in the ear's physiological functioning affect the hearing mechanism.

Reorienting to one of the objectives of the current study, it was discovered that only those with hearing loss had considerably higher scores using CFHA than MCHA. Previous reports have demonstrated that multichannel hearing aids divide the incoming signal into various channels and assign compression ratios based on the hearing loss for that band. They suggest it would have led to spectral smearing and reduced spectral contrast, resulting in loss of spectral shape of the speech and poor speech recognition scores [16, 17]. However, channel-free hearing aids process the incoming signal as a whole. Gains are adjusted continuously by monitoring the input sound pressure level, retaining the spectral contrast and temporal envelope for better recognition [18]. Hence, this would have led to better speech recognition using CFHA than MCHA. However, it is also noteworthy that CFHA was also found to have spectral distortions to an extent but were found to be negligible compared to MCHA [19, 20]. Hence, it can be hypothesized from the results of the present study and those in the literature that the additional temporal and spectral distortion in MCHA would have resulted in poorer speech recognition using the same when compared to CFHA.

Further, comparison across the different hearing aid processing algorithms showed that among normal hearing individuals, WDRC+NR showed the highest speech recognition scores compared to other processing algorithms, irrespective of the hearing aids. However, among individuals with hearing impairment, the highest speech recognition scores were found for WDRC processing compared to the other processing algorithms. This is in accordance with the findings on SNR change in the present study. It was seen that the SNR change was significant for WDRC processing compared to other strategies in the current study. Hence, more favourable SNRs have resulted in better speech recognition scores for WDRC processing. Further, studies have also reported that WDRC processing results in more significant temporal envelope distortions, leading to poor benefits from hearing aids [1, 13]. Thus, it can be assumed that during the combined condition (WDRC+NR), the temporal envelope distortions due to WDRC would have reduced the strength of the NR algorithm in differentiating speech and noise. This could have resulted in the hearing aid identifying speech as unwanted noise and preventing primary spectral information from being processed, leading to reduced speech recognition scores. Hence, from the current study's findings and from the literature, it can be hypothesised that WDRC alone works better for individuals with hearing impairment than the combined condition (WDRC+NR).

Variation in speech recognition scores with change in signal-to-noise ratio

In the present study, none of the variables included were found to be significant predictors of speech recognition scores among the normal-hearing individuals. However, among individuals with hearing impairment, the hearing aid being channel free or multi-channel (CFHA and MCHA) was a significant predictor of speech recognition scores. However, it should also be noted that the changes in SNR or the signal processing strategies were not observed to be significant predictors of change in speech recognition among individuals with hearing impairment. Thus, the current study shows that speech is being processed differently through MCHA and CFHA, which can predict the changes in speech recognition scores. Earlier reports in the current study also pave the way for similar findings that there are differences in SNR and speech recognition using the two hearing aids. Similarly, this model in the current study also suggests that the hearing aid compression and gain influence subjective speech recognition irrespective of the strategy used and the output measured objectively. Further, the lack of ability to predict speech recognition from the change in SNR might be attributed to the contextual cues provided for recognition in the standardised sentences used as stimuli in the present study. If the response to the given stimuli had primarily depended on audibility rather than the influence of other factors like contextual cues, the change in SNR would have been a strong predictor of speech recognition scores. These cues would have concealed any advantages the difference in SNR may have brought about on speech recognition scores. Similar results were demonstrated by Miller et al [1], wherein they hypothesised that the stimuli were not sensitive enough to reflect changes in SNR on speech recognition scores. Hence, stimuli with low contextual influence must be employed to further understand the connection between the change in SNR and the relative change in speech recognition among individuals with hearing impairment.

Conclusion

It can be concluded from the findings of the present study that channel-free hearing aids have an advantage over multi-channel hearing aids among individuals with hearing impairment. In addition, the subjective speech recognition scores were found to be better using channel-free hearing aids than modern multichannel hearing aids. Consequently, these findings encourage the audiologist to envision the potential of channel-free hearing aids as a feasible alternative for people with hearing impairment, particularly in those for whom speech recognition is the predominant demand.

Limitations of the study

Current study has few limitations which if taken care in future research would give better insight about the results. Due to a lack of availability of an ear simulator or a manikin at the time of the present study, a coupler was used. Also, the scores from the better ear was obtained; however, analysing ear specific scores would have given us more information about the involvement of central auditory processing system as speech in noise was utilized.

Ethical Considerations

Compliance with ethical guidelines

The research is compliant with ethical standards according to the 1964 Helsinki Declaration. The institutional ethics committee approved the study (Ref No. SH/194/2022), and further data was collected. Informed consent before the evaluation after the briefing about the study was taken from all the participants.

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Authors' contributions

SD: Study design, drafting the manuscript, interpretation of the results, critical revision of the manuscript, statistical analysis; VK: Study design, acquisition of data, critical revision of the manuscript; RCV: Study design, interpretation of the results, critical revision of the manuscript and statistical analysis.

Conflict of interest

There is no conflict of interest to disclose

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Declaration of Generative AI and AI assisted technologies in the writing process

We have not used AI in the manuscript preparation process.

Declaration

Not applicable.

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