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Hearing Aid Delay Effects on Neural Phase Locking

Mary Zhou,[1] Roksana Soleimanpour,[2] Aakriti Mahajan,[2] and Samira Anderson[1,3]

Objectives: This study was designed to examine the effects of hearing aid delay on the neural representation of the temporal envelope. It was hypothesized that the comb-filter effect would disrupt neural phase locking, and that shorter hearing aid delays would minimize this effect.

Design: Twenty-one participants, ages 50 years and older, with bilateral mild-to-moderate sensorineural hearing loss were recruited through print advertisements in local senior newspapers. They were fitted with three different sets of hearing aids with average processing delays that ranged from 0.5 to 7 msec. Envelope-following responses (EFRs) were recorded to a 50-msec /da/ syllable presented through a speaker placed 1 meter in front of the participants while they wore the three sets of hearing aids with open tips. Phase-locking factor (PLF) and stimulus-to-response (STR) correlations were calculated from these recordings.

Results: Recordings obtained while wearing hearing aids with a 0.5-msec processing delay showed higher PLF and STR correlations compared with those with either 5-msec or 7-msec delays. No differences were noted between recordings of hearing aids with 5-msec and 7-msec delays. The degree of difference between hearing aids was greater for individuals who had milder degrees of hearing loss.

Conclusions: Hearing aid processing delays disrupt phase locking due to mixing of processed and unprocessed sounds in the ear canal when using open domes. Given previous work showing that better phase locking correlates with better speech-in-noise performance, consideration should be given to reducing hearing aid processing delay in the design of hearing aid algorithms.

Key words: Comb-filter effect, Delay time, Envelope-following response, Hearing aid, Phase locking.

Abbreviations: EFR = Envelope-following response; PTA = pure-tone average; STR = stimulus-to-response correlation; PLF = phase-locking factor.

(Ear & Hearing 2023;XX;00–00)

INTRODUCTION

Hearing aid algorithm designs have changed considerably in the last few decades, and MarkeTrakX survey results indicated a satisfaction rate of 83% among current hearing aid users (reviewed in Picou 2020). The survey also revealed that among hearing aid users, sound quality and performance were the top contributing factors to hearing aid satisfaction (Picou 2020). Nevertheless, among individuals who return their hearing aids for credit (approximately 20%), there is a much lower hearing aid satisfaction rate of roughly 50%, and the factors that contribute to this dissatisfaction are not currently well understood. Individuals who stopped wearing their hearing aids report that the aids provided little benefit and poor sound quality, and that they experienced difficulty understanding speech in background noise while wearing them (McCormack et al. 2013). For these reasons, innovations that improve hearing aid sound quality may increase their rates of use and adoption, and thus reduce potential long-term effects of auditory deprivation on social well-being and cognitive health.

The implementation of digital signal processing in hearing aids has made it possible to meet the goals of increased audibility while maintaining comfortable listening levels; however, these algorithms naturally distort the signal to some extent and may affect speech intelligibility and sound quality (Arehart et al. 2013; Kowalewski et al. 2018). In addition to these device-related disturbances, the older listener's auditory system distorts the incoming signal due to age-related degeneration (Anderson et al. 2012; Billings et al. 2015; Presacco et al. 2016; Roque, Karawani, et al. 2019), thereby increasing the speech understanding difficulties experienced by older listeners in challenging environments. Anderson et al. (2020) demonstrated that many older listeners show partial compensation for spectrally distorted speech stimuli because these listeners had minimal effects of vocoding on the perception of temporal contrasts in speech stimuli. Because of their ability to compensate for some degree of signal distortion, behavioral testing may not be sensitive enough to detect subtle distortions imposed by signal processing algorithms (Micula et al. 2021). Yet, the presence of these distortions may tax listeners over time in their everyday environments. Therefore, an objective test of hearing aid distortion effects on neural processing may provide information to maximize the quality of the signal and lead to improved hearing aid satisfaction.

One source of distortion is the mixing of amplified signals that are processed by the hearing aid with the unamplified signals that enter the ear canal via leakage through earmold vents and especially through open domes. Open domes are widely used for individuals with mild-to-moderate hearing loss, specifically those with normal/mild hearing loss in the low frequencies who do not need large amounts of amplification (Noble et al. 1998; Alworth et al. 2010). Open domes allow sound to naturally enter the ear canal, creating a more natural sound quality while reducing the "boomy" sensation or occlusion effect that occurs with closed domes (Winkler et al. 2016). Hearing aid signal processing can delay the incoming amplified signal by up to several milliseconds, resulting in the comb-filter effect (Bramsløw 2010; Stiefenhofer 2022). Comb-filtering refers to phase cancellation that occurs when the delayed amplified sound wave mixes with the undelayed natural sound wave in the ear canal. The destructive interference of these sound waves produces a single combined sound wave that resembles the teeth of a comb (Balling et al. 2020; Stiefenhofer 2022). The perceptual sensation of the comb-filtering effect includes a diffused, unnatural, metallic sound quality that can start at delay times of 5 to 7 msec (Agnew et al. 2000; Stone et al. 2002;

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<zdoi; 10.1097/AUD.0000000000001408>

Bramsløw 2010), a noteworthy finding given that typical delay times for most premium hearing aids range from 5 to 8 msec.

To investigate the effects of comb-filtering on perception, studies have evaluated the effects of delay times using self-reported ratings. Groth et al. (2004) recruited individuals with normal hearing and those with mild hearing loss to rate the level of disturbance resulting from varying processing delays. The test devices were coupled to the ear with tubing that had an inner diameter of 0.86 mm and were fitted to a silicone ear tip that was 6 mm in diameter and had four vents on the flange. Each group's response to disturbances from signal processing delays of 2 msec, 4 msec, and 10 msec was assessed while listening to music, running speech, and their own voice. The listeners with normal hearing had higher disturbance ratings in the following conditions: (1) their own voice for all delay time comparisons, (2) music for the 10-msec versus 4-msec and 10-msec versus 2-msec comparisons, and (3) running speech for the 10-msec versus 2-msec comparison. The listeners with hearing loss had higher disturbance ratings for their own voice for the 10-msec versus 2-msec comparison, but no differences in disturbance ratings were noted for music or running speech. Balling et al. (2020) interviewed individuals with normal hearing and with mild-to-moderate hearing loss after wearing hearing aids with algorithms that produced a minimal 0.5-msec delay and after wearing hearing aids with a 2.5-msec delay; they found that most individuals preferred the hearing aid with the shorter delay time. Overall, these results suggest that shorter delay times may reduce perceived distortion and improve the overall listening experience.

The envelope-following response (EFR) can provide an objective measure of the auditory system's representation of signal fidelity and effects of signal distortion (Krizman et al. 2019; Anderson et al. 2020). The EFR reveals evidence of decreased phase locking in older listeners (Anderson et al. 2012; Clinard et al. 2015; Roque, Karawani, et al. 2019); therefore, processing algorithms that disrupt the signal phase of the auditory signal may be observable in the EFR. Previous studies have evaluated the effects of amplification on the EFR (Easwar et al. 2015; Jenkins et al. 2018), but there are limited studies using the EFR to evaluate hearing aid algorithms. One previous study recorded EFRs to a 50-msec/da/ syllable in individuals wearing three aids that differed in processing time and found that the magnitude of the fundamental frequency was highest for the hearing aid that had the shortest delay time (Slugocki et al. 2020).

The purpose of the present study was to examine the effects of open fittings on subcortical processing of the temporal speech envelope when hearing aid processing delay times were varied. We fit middle-aged to older listeners with three sets of hearing aids with open domes with average processing delays ranging from 0.5 msec to 7 msec. Two measures were used to assess hearing aid processing effects on the EFR: stimulus-to-response (STR) correlation and phase locking factor (PLF). Better synchrony between the direct unamplified and processed amplified sound is hypothesized to preserve the signal's natural envelope, which can be observed through the STR correlation (Roque, Gaskins, et al. 2019), an objective measure of morphology. A high STR correlation is important because a more robust temporal envelope correlates with better speech-in-noise performance in older adults (Anderson et al. 2013). The neural encoding of the fundamental frequency (F_0) of the/da/ syllable is also associated with speech-in-noise performance in younger (Song et al. 2011) and older (Anderson et al. 2011) listeners. The PLF quantifies the consistency of phase locking from trial to trial. We hypothesize that shorter processing delays will result in higher STR correlations and increased phase locking in subcortical responses to the vowel/a/ presented within the context of a speech syllable.

MATERIALS AND METHODS

Participants

Twenty-one participants were recruited who were ≥50 years old (mean: 73.1, SD: 6.9, males = 7) through advertisements in local senior newspapers. Audiometric criteria included a bilateral, symmetric (within 10 dB at any frequency) mild-to-moderate sensorineural hearing loss between 125 and 4000 Hz, including interoctave frequencies, no air-bone gaps >10 dB at any frequency, and no history of middle ear or neurological dysfunction. The hearing losses were within the recommended fitting range of open domes with pure-tone averages (500, 1000, and 2000 Hz) between 25dB HL and 50dB HL. Audiologic threshold testing was completed using an Interacoustics

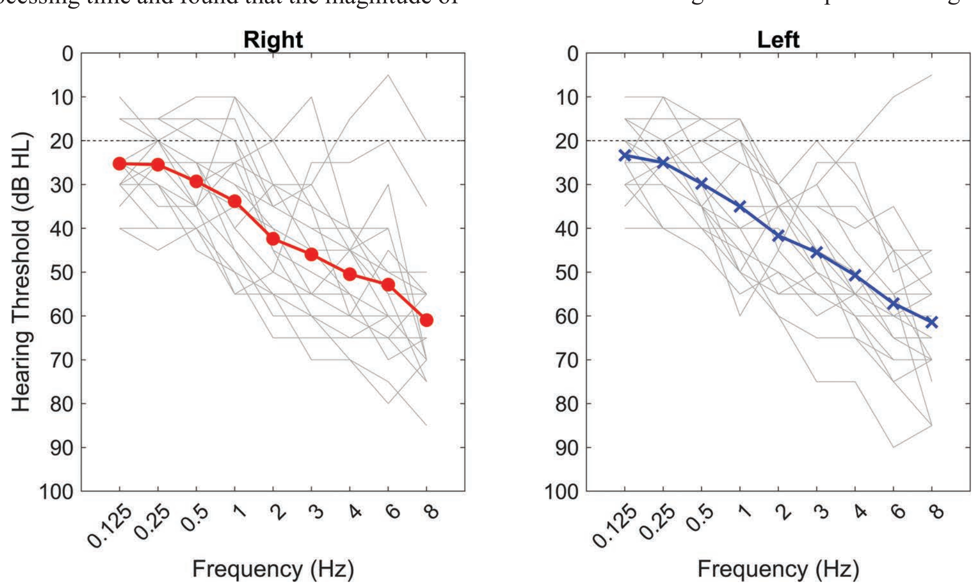

Fig. 1. Individual hearing thresholds (thin gray lines) and mean thresholds (red, right ear; blue, left ear) are displayed.

Affinity 2.0 audiometer (Interacoustics, Eden Prairie, MN) with 3A insert earphones. Figure 1 displays individual and mean audiograms for the right and left ears. Immittance testing was completed using the Interacoustics Titan tympanometer; tympanometry and acoustic reflex thresholds elicited at 500, 1000, and 2000 Hz confirmed the presence of normal middle ear function. In addition, all participants were native speakers of American English and had normal cognitive function. All subjects had scores on the Montreal Cognitive Assessment of ≥22 to screen for mild cognitive impairment (Nasreddine et al. 2005). A criterion of 22 was used to accommodate for decreased function associated with hearing loss (Dupuis et al. 2015; Saunders et al. 2018). This study was approved by the University's institutional review board and participants were compensated for their time.

Hearing Aid Fitting

The participants were fitted with premium bilateral receiver-in-the-ear hearing aids with delay times of 0.5, 5, and 7 msec from three different manufacturers. The devices were first programmed to each ear to the maximum experience level (100% gain) with default feature settings that included noise reduction, frequency lowering (following the defaults for the given hearing aids), impulse control, and wind noise reduction. Microphones were set to the manufacturer's default settings: omnidirectional for the 0.5-msec delay hearing aid and automatic for the 5-msec and 7-msec delay hearing aid. Under test conditions (see later), the microphone setting would not make any difference in the response because the hearing aids would remain in omnidirectional mode. The feedback manager was implemented, and then the devices were calibrated for open fit and verified to be appropriately programmed for each participant based on National Acoustics Laboratory-NL2 prescriptive targets (Keidser et al. 2011) for soft, average, and loud level inputs (55 dB, 65 dB, and 75 dB SPL, respectively) using the Interacoustics Affinity 2.0 real ear measurement system (Eden Prairie, MN). The differences between the targets and real ear outputs were calculated for each hearing aid fitting for frequencies from 250 to 8000 Hz at the three levels. Table 1 displays these values.

Stimulus

A 50-msec syllable/da/ was synthesized in the study by Klatt (1980). The initial 10-msec onset burst was followed by a brief 5-msec consonant-vowel transition and a 35-msec steady-state vowel. The spectral composition of the steady state vowel was as follows: fundamental frequency (F_0): 100 Hz, first formant: 556 Hz, second formant: 1435 Hz, third formant: 2517 Hz, and fourth formant: 3250 Hz.

To demonstrate the effects of delay on the stimulus, we recorded the/da/ stimulus through Knowles Electronic Manikin for Acoustic Research with the different hearing aids programmed for mild-to-moderate hearing loss. For reference, an unaided condition was also recorded. Stimuli were presented at 70 dB SPL from a single loudspeaker positioned at 1 m directly in front of Knowles Electronic Manikin for Acoustic Research in a sound attenuating booth at a 9-Hz rate over a total of 40 seconds per condition. Analysis was limited to the penultimate stimulus presentation where hearing aid processing should have stabilized. Figure 2 shows the effects of delay on the stimulus envelope. The periodicity at 100 Hz is disrupted with greater delays.

Data Recording

The /da/ stimulus was presented in alternating polarities (one polarity per sweep) at a rate of 9 Hz via Presentation software (Neurobehavioral Systems, Berkely, CA) at 70 dB SPL through a loudspeaker placed 1 m directly in front of the listener (0° azimuth) while the participants wore the programmed hearing aids (in randomized order). The EFR was recorded via Biosemi Active-Two Auditory brain stem response software (Biosemi B.V., Amsterdam) using a two-channel vertical montage with

TABLE 1. Real ear measurements for the three hearing aid delays

		250 Hz		500 Hz		1000 Hz		2000 Hz		4000 Hz		8000 Hz	
dB SPL	Ear	Output	Difference	Output	Difference	Output	Difference	Output	Difference	Output	Difference	Output	Difference
0.5-Delay													
55	Right	48.05	−1.62	48.86	−3.62	55.14	2.14	58.86	−4.90	55.00	−10.14	40.67	−11.86
	Left	48.71	−0.81	48.19	−4.33	55.48	2.10	56.38	−7.43	56.29	−9.29	41.95	−10.81
65	Right	58.10	1.95	58.33	−0.24	64.33	5.57	65.71	−2.81	63.29	−6.48	45.62	−12.29
	Left	58.57	2.67	57.95	−0.71	63.67	4.71	64.19	−4.05	65.00	−5.00	47.00	−11.14
75	Right	55.43	−9.57	60.10	−6.90	71.48	7.24	74.81	2.00	71.95	−1.48	47.86	−14.33
	Left	56.19	−7.43	59.76	−6.14	71.05	5.67	73.05	0.81	73.76	0.19	48.10	−12.67
5-msec Delay													
55	Right	48.48	−1.81	49.19	−3.71	52.57	−0.90	61.48	−2.76	56.43	−9.14	43.81	−9.24
	Left	49.10	−0.43	49.10	−3.43	50.24	−3.10	60.00	−3.81	57.67	−7.86	42.19	−10.57
65	Right	57.90	1.76	58.43	−0.10	59.71	1.00	66.81	−1.71	62.95	−6.81	49.33	−8.62
	Left	58.48	2.62	58.48	−0.19	57.81	−1.14	66.57	−1.67	66.38	−3.67	46.71	−11.48
75	Right	54.95	−10.05	59.86	−7.14	67.10	2.86	74.43	1.57	71.71	−1.76	52.33	−9.95
	Left	56.90	−8.10	60.90	−6.10	63.29	−1.05	74.76	2.52	74.24	0.62	49.48	−13.05
7-msec Delay													
55	Right	48.48	−1.19	50.14	−2.33	53.67	0.67	62.90	−0.86	55.86	−9.29	40.48	−12.05
	Left	48.71	−0.81	49.81	−2.71	54.38	1.00	61.48	−2.33	57.67	−7.90	42.62	−10.14
65	Right	57.62	1.48	58.81	0.24	60.10	1.33	67.38	−1.14	65.05	−4.71	45.29	−12.62
	Left	58.38	2.48	58.33	−0.33	61.10	2.14	66.62	−1.62	65.57	−4.43	45.81	−12.33
75	Right	55.52	−9.48	60.62	−6.38	66.86	2.62	72.81	0.10	72.29	−1.14	47.33	−14.86
	Left	56.14	−8.86	60.00	−7.00	67.86	3.52	73.24	1.05	72.62	−0.90	48.10	−14.33

The mean target outputs and mean differences between target output and actual output are displayed for hearing aids with the three delays at three levels from 250 to 8000 Hz.

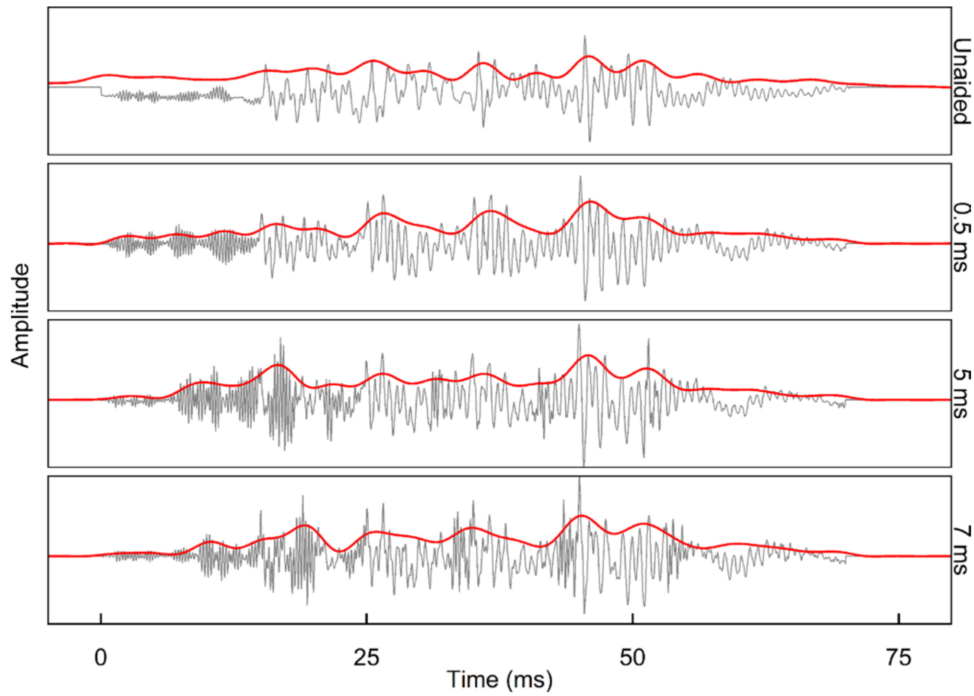

Fig. 2. The stimulus waveforms and envelopes are displayed for four recordings of the/da/ stimulus through KEMAR: unaided and hearing aids with a 0.5 msec, 5 msec, or 7 msec delay programmed to a mild-to-moderate hearing loss. KEMAR indicates Knowles Electronic Manikin for Acoustic Research.

two electrodes placed on the forehead (Driven Right Leg/ Common Mode Sense), one on each earlobe (references), and one on Cz (active) with offsets <40 μV at a sampling rate of 16,384 Hz with an online filter of 100 to 3000 Hz (highpass: first-order 6 dB/octave, lowpass: fifth-order cascaded integrator-comb). This online filter has the effect of reducing amplitude for a 100-Hz stimulus but also reduces noise in the recording. A minimum of 8000 artifact-free sweeps (4000 per polarity) was recorded for each set of hearing aids. During the recording, subjects were seated in a recliner and watched a muted movie of their choosing with subtitles to facilitate a relaxed but awake state.

Data Analysis

The raw data files were imported into MATLAB, v21b (Mathworks, Natuck, MA) using custom scripts. The artifact-free sweeps (≤30 μV) were bandpass filtered from 70 to 2000 Hz using a zero-phase Butterworth filter. To maximize the response to the temporal envelope, an averaged response of the two polarities was obtained for the first 8000 accepted sweeps.

Stimulus-to-Response Correlation (STR) • The stimulus envelope was extracted from the response stimulus and bandpass was filtered from 70 to 2000 Hz. A cross-correlation was performed using the XCORR function in MATLAB by shifting the stimulus waveform in time relative to the response waveform until a maximum correlation *r* value was found between the stimulus and region of the response.

Phase-Locking Factor (PLF) • Responses were decomposed from 50 to 8000 Hz using Morlet wavelets (Tallon-Baudry et al. 1996), using a procedure identical to that employed in previous studies (Jenkins et al. 2018; Roque, Karawani, et al. 2019). Individual values for the F_0 of 100 Hz (20 Hz bin) were calculated for the response region corresponding to the stimulus vowel (25 to 60 msec). Note that the online recording filter will reduce amplitude of the response but improves the signal-to-noise ratio.

Statistical Analysis

Linear mixed-effects models were implemented via the lme4 package (Bates et al. 2015) and assessed using the lmerTest package (Kuznetsova et al. 2017) in R (version 4.2.1) to separately evaluate effects of delay time on PLF and STR correlations. Hearing loss (quantified as the pure-tone average (PTA) from 500 to 4000 Hz) and age were included in the models to determine if these factors modulated the effects of delay time. Both models included the same fixed effects of PTA, age, and delay time that were treated as continuous variables. Individual participants were coded as random effects. A backward stepwise elimination of fixed and random effects was used to determine which (if any) random/fixed effects best accounted for participants' STR and PLF values.

RESULTS

Real Ear Measurement

A multivariate analysis of variance was performed to determine if the real ear fittings differed between hearing aids using the different values as the dependent variable, and no differences were found between the hearing aids ($F[2, 40] = 0.034$, $p = 0.97$). Figure 3 displays means and standard deviations of the differences in target and obtained output values.

STR

Figure 4 displays the stimulus waveform and individual waveforms overlaid with the group average waveform for the three delay times. The individual waveforms for the 0.5-msec

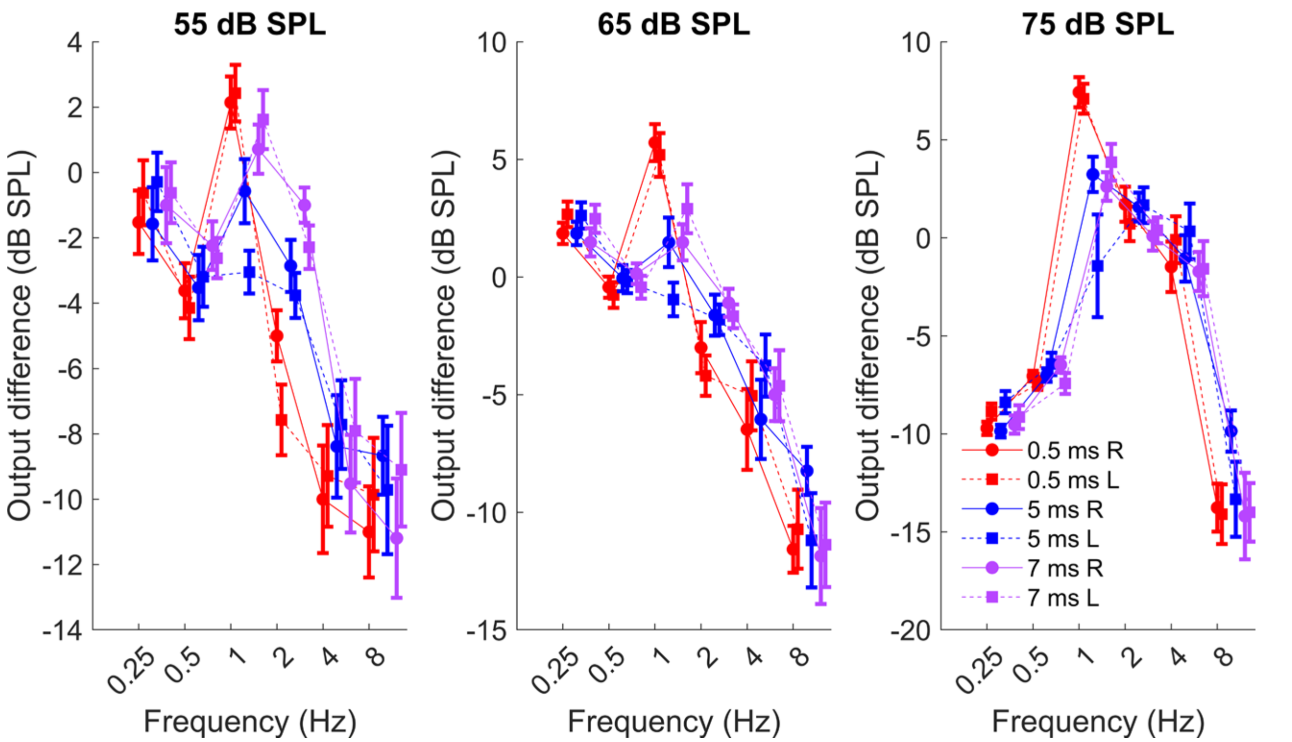

Fig. 3. The mean differences between target output and actual output are displayed for the right and left ears for hearing aids with delays of 0.5 msec (red), 5 msec (blue), and 7 msec (purple) at levels of 55, 65, and from 250 to 8000 Hz. Error bars: 1 SE.

delay time are more sharply defined than the waveforms for the 5-msec or 7-msec delay times. The backward stepwise elimination determined that the model containing only the fixed effect of delay time best accounted for variance in the data ($t(41) = -4.30$, $p < 0.001$). As shown in Figure 5, the r value was higher for the 0.5-msec delay time than either the 5-msec or 7-msec delay times, and there was no difference in the r value between the 5-msec and 7-msec delay times. Table 2 provides a summary of the model. The higher r value represents a better correlation between the acoustic stimulus and the neural response for the lower hearing aid delay.

PLF

Figure 6 displays phase locking to the 100-Hz F_0 for the three delay times. The PLF values were log transformed to account for a skew in the data. Backward stepwise elimination determined that a model containing the fixed effects of delay time ($t(40) = -4.32$, $p < 0.001$) and PTA ($t(39) = -2.21$, $p = 0.033$), as qualified by a two-way delay time × PTA interaction ($t(40) = 3.21$, $p = 0.003$), and the random effect of Participants best accounted for the variance in PLF values. As shown in Figure 7, PTA modulated the effects of delay time on the PLF, such that greater effects were noted for lower degrees of hearing loss. Table 3 provides a summary of the model. The results are consistent with the idea that longer delay times disrupt phase locking, and this effect is modulated by the extent of hearing loss.

DISCUSSION

The results confirmed our hypothesis that shorter delay times improve phase locking and response waveform morphology. STR correlation r values and PLF were higher for the 0.5-msec delay times than the longer delay times, and there were no differences between the 5-msec and 7-msec delay times. The effects of delay times were generally stronger for individuals with mild hearing loss than for individuals with greater degrees of hearing loss.

Based on these results, it appears that the comb-filter effect associated with longer delay times leads to reduced response morphology (Fig. 3) and phase locking (Fig. 5). Although behavioral effects of these delay times were not measured, we expect that higher morphology and phase locking would be factors contributing to better speech-in-noise performance based on past studies (Anderson et al. 2011, 2012, 2013; Hao et al. 2018; McClaskey et al. 2019).

Previous studies evaluated the effects of delay time on ratings of sound quality (Groth & Søndergaard 2004; Stone et al. 2008) and personal preference (Balling et al. 2020) and found qualitative benefits of shorter delay times. However, the differences between delay times were less apparent in individuals with greater degrees of hearing loss, possibly due to decreased overall phase locking with increased hearing loss, thus minimizing the potential for differences between hearing aid delays. We found that hearing loss was a factor in delay time, and that effects were somewhat weakened in listeners with greater degrees of hearing. It is possible that hearing loss reduced phase locking overall, thus minimizing the potential for differences between hearing aid delays. Figure 8 compares phase locking in an 86-year-old listener with a 53 dB PTA and a 79-year-old listener with a 38 dB PTA and shows that differences between conditions are reduced in the listener with hearing loss who also had lower phase locking overall. We note that listeners with more hearing loss are more likely to be fitted with custom earmolds or more closed domes to increase audibility without feedback, and so they would not be as affected by processing delays.

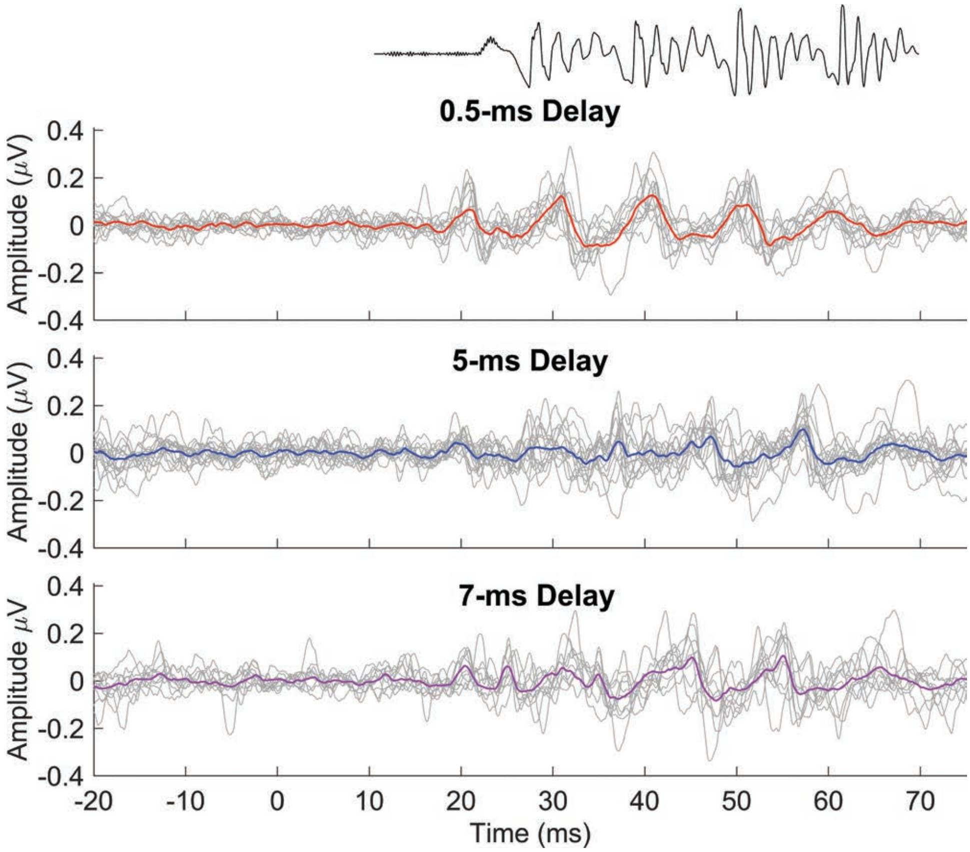

Fig. 4. The top panel displays the 50-msec /da/ waveform. The lower panels display individual response waveforms (thin gray lines) and mean average response waveforms for hearing aids with 0.5-msec (red), 5-msec (blue), and 7-msec (purple) processing delay times.

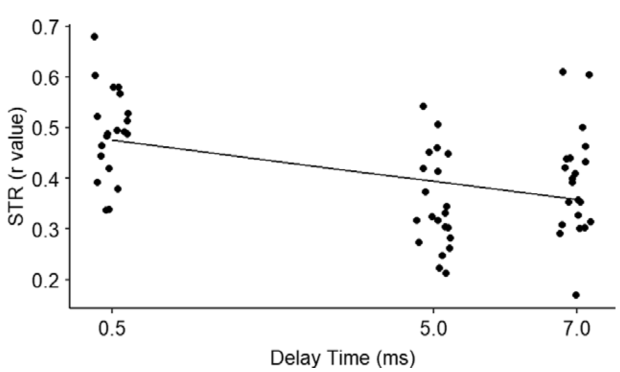

Fig. 5. Individual stimulus-to-response correlation (STR) *r* values are displayed for hearing aids with 0.5-msec, 5-msec, and 7-msec processing delay times. The diagonal line represents the significant fixed effect of delay time as fit by a LME model. LME indicates linear mixed effects.

Previous studies have demonstrated the efficacy of using the EFR to evaluate improvements in audibility with hearing aids (Easwar et al. 2015; Jenkins et al. 2018). Easwar et al. (2015) also investigated the effects of bandwidth on the EFR and behavioral measures of speech discrimination and sound quality. They low-pass filtered the speech stimulus at 1, 2, and 4 kHz and presented these filtered stimuli through individually programmed hearing aids to 21 listeners with mild-to-moderate sensorineural hearing loss. They found that increased bandwidth results in higher EFR response amplitudes, higher speech discrimination scores, and higher sound quality ratings.

TABLE 2. Delay time effects on stimulus-to-response correlations

	STR		
Predictors	Estimates	CI	*p*
Intercept	0.48	0.44 to 0.53	**<0.001**
Delay	–0.02	–0.03 to –0.01	**<0.001**
Random effects			
σ^2		0.01	
$\tau_{00\ subnum}$		0.00	
ICC		0.18	
N_{subnum}		21	
Observations		63	
Marginal R^2/ conditional R^2		0.197/0.340	

The backward stepwise elimination model demonstrated effects of delay time only on the stimulus-to-response correlations (STR).
CI, confidence inerterval; ICC, intraclass correlation coefficient; STR, stimulus-to-response correlations.
Bold font indicates significant p values.

They also found that higher response amplitude was related to higher speech discrimination scores and sound quality ratings. Individuals who are having difficulty adjusting to amplification may have difficulty finding the right words to describe what they are hearing. We did not include a measure of sound quality in our experiment, but the results of Easwar et al. suggest that the EFR may be used as an objective index of sound quality.

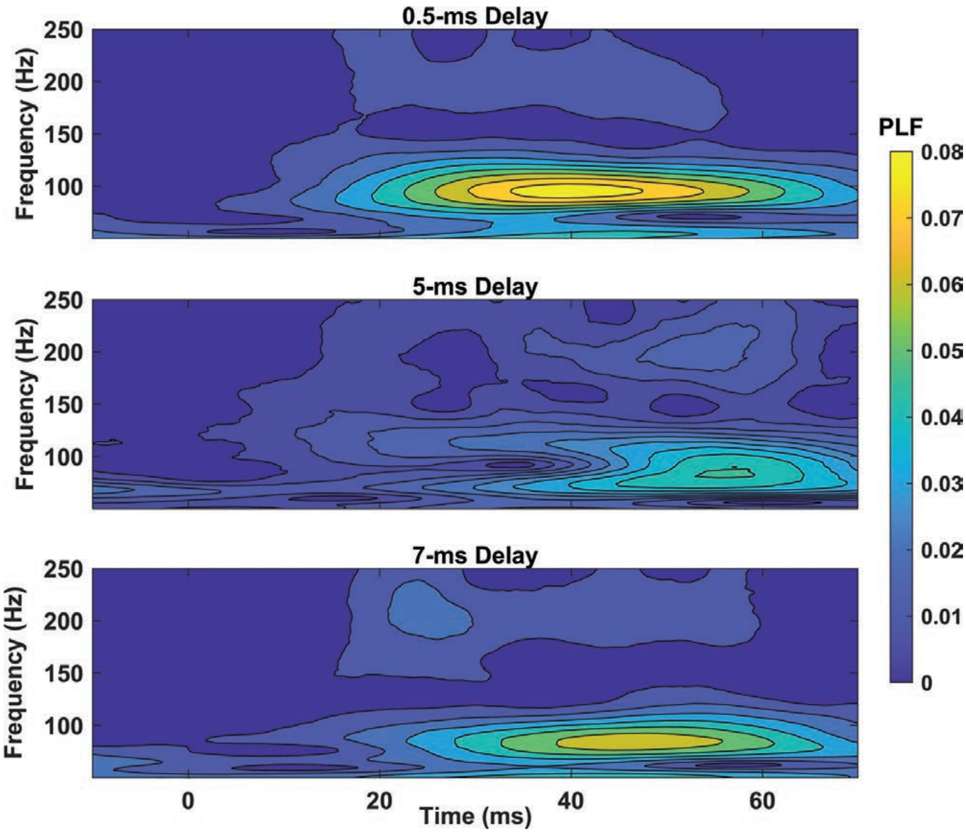

Fig. 6. Phase-locking factor (PLF) is displayed for hearing aids with 0.5-msec, 5-msec, and 7-msec processing delay times.

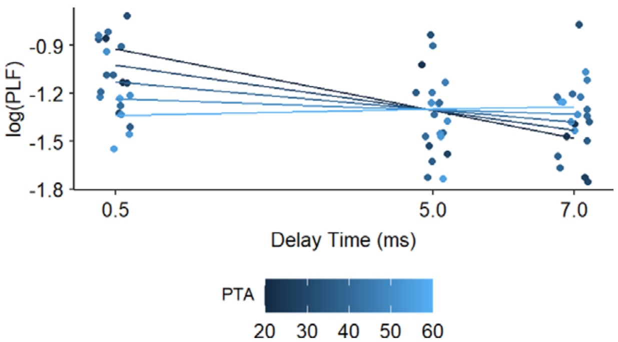

Fig. 7. Individual phase-locking factor (PLF) values are displayed for hearing aids with 0.5-msec, 5-msec, and 7-msec processing delay times. The values are color-coded according to a gradient of hearing loss in 10-dB steps from 20 to 60 dB pure-tone average hearing loss. Colored lines show how pure-tone average (PTA) affects the relationship between PLF and delay time as fit by an LME model.

TABLE 3. Delay time effects on phase-locking factor (PLF)

	log(PLF)		
Predictors	Estimates	CI	*p*
Intercept	−0.64	−1.08 to −0.20	**0.005**
Delay	−0.13	−0.20 to −0.07	**<0.001**
PTA	−0.01	−0.02 to −0.00	**0.031**
Delay * PTA	0.00	0.00 to 0.00	**0.002**
Random effects			
σ^2		0.02	
$\tau_{00\ subnum}$		0.03	
ICC		0.54	
N_{subnum}		21	
Observations		63	
Marginal R^2/Conditional R^2		0.216/0.638	

The backward stepwise elimination model demonstrated effects of delay time and pure-tone average (PTA) on phase-locking factor (PLF) in addition to a delay × PTA interaction.
CI, confidence interval; PTA, pure-tone average; PLF, phase-locking factor.
Bold font indicates significant p values.

Stone and Moore (2002) found that disturbance from hearing aid delays of up to 5 or 6 msec was tolerable for constant gain of 10 dB or more across frequency. Nevertheless, it is possible that delay-related changes to neural fidelity affect the quality of signal in ways that are not obvious to the listener. Such deleterious effects may result in increased listening effort, particularly in challenging listening environments. Previous studies examining the effects of noise and noise reduction algorithms on listening effort have shown that pupillometry and electrophysiology studies can be more sensitive to noise reduction effects than some perceptual measures (Wendt et al. 2017; Fiedler et al. 2021). A hearing aid noise reduction algorithm decreased pupil dilation but increased alpha power, both reflecting reduced listening effort, despite no significant effects of behavioral performance on a realistic listening scenario task (Fiedler et al. 2021). Distortion from the comb-filter effect may lead to an increase in sustained effort, affecting performance in everyday environments. Implementation of algorithms that reduce hearing aid distortion effects, such as reductions in hearing aid delay times, may ultimately lead to more successful hearing aid outcomes.

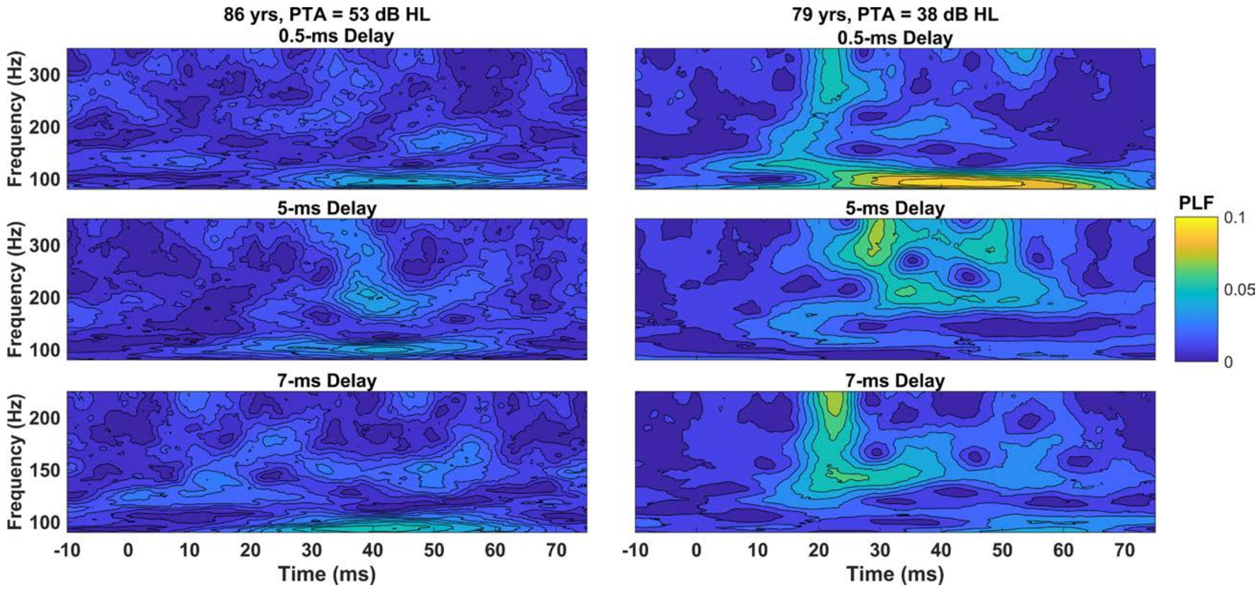

Fig. 8. Phase-locking factor (PLF) is compared between listeners with moderate (left panels) and mild (right panels) hearing loss. PTA, pure-tone average.

Incorporation of measures to evaluate aided neural processing and/or effort may help guide the design of new hearing aid algorithms and the hearing aid fitting itself.

LIMITATIONS

The effect of delay time was observed in response to a single syllable that was averaged over thousands of trials, which is not an ecological representation of the typical listening environment. It would be interesting to determine if delay effects were noted with a more ecological measure, such as the envelope tracking response. For example, aided responses could be compared in recordings to continuous speech samples to determine if delay time affects stimulus reconstruction accuracy, assuming that this measure could be performed without interference from stimulus artifacts (Bellier et al. 2015). Given that previous studies have demonstrated effects of hearing aid algorithms on listening effort, a combined electrophysiology/pupillometry protocol could add an additional dimension to understanding the effects of hearing aid algorithms on the listener's hearing loss experience.

The listeners in the study all wore hearing aids with open domes, but the delay time effects would be greatly reduced when the ear canals are occluded or partially occluded with molds or larger domes. Nevertheless, individuals with mild-to-moderate hearing loss are most likely to be fit with open domes, and are more likely to perceive algorithm-related distortions than individuals with greater degrees of hearing loss (Tan et al. 2008). Therefore, new hearing aid users with mild degrees of hearing loss may experience less than desirable hearing aid outcomes due to distortion effects associated with long delay times.

Finally, although there were no statistically significant differences between responses to hearing aids with 5-msec and 7-msec delays, Figures 5 and 6 show greater apparent effects of delay for the hearing aid with the 5-msec delay times. This result may be attributed to testing parameters: the 5-msec delay may have been more susceptible to phase cancellation because 5 msec is half of the period of the 100-Hz F_0. Therefore, the overall results may reflect the chosen testing parameters that would not be apparent in everyday listening.

Conclusions

Long delay times (≥0.5 sec) can reduce the fidelity of the neural speech signal for hearing aid fittings with open domes. Identification of factors that improve the neural fidelity of the signal may lead to better speech-in-noise performance and ultimately to better-hearing aid outcomes.

ACKNOWLEDGMENTS

We acknowledge Taylor Beinke and Hsin-Ju Lui for their assistance with data collection and Francis Kuk, Christopher Slugocki, and Laura Winthrop Balling for their feedback on the article.

M.Z. and S.A. conceived and designed the research. M.Z., S.A., and R.S. performed experiments. S.A., R.S., and A.M. analyzed data. M.Z., S.A., R.S., and A.M. interpreted the results of the experiments. S.A. and R.S. prepared figures. M.Z. and S.A. drafted the article. M.Z., S.A., R.S., and A.M. edited and revised the article. M.Z., S.A., R.S., and A.M. approved the final version of the article.

Supported, in part, by WS Audiology, W/S. The funding organization provided input in the design and conduct of the study, but did not provide input in the collection, analysis, and interpretation of the data; or in the decision to submit the article for publication.
The authors have no conflicts of interest to disclose.

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Received September 30, 2022; accepted June 15, 2023

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