International Journal of Audiology 2012; 51: 906–913

informa
healthcare

Original Article

Preferred delay and phase-frequency response of open-canal hearing aids with music at low insertion gain

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Abstract

Objective: Preferences between low delays and phase-frequency responses of behind-the-ear, open-canal hearing aids were investigated with acoustic conditions deemed sensitive to delay effects by normal-hearing listeners. *Design:* Hearing aids with the following selectable delay and phase response options were fitted at low insertion gain: (1) 1.4 ms delay, minimum phase; (2) 3.4 ms delay, minimum phase; and (3) 3.4 ms delay, linear phase. Blind paired comparisons were made between processing options and between each option and a muted hearing-aid output with two music stimuli. The three alternative forced choice responses were "Slightly prefer", "Prefer", or "Strongly prefer". *Study sample:* Twelve hearing-impaired musicians. *Results:* At the 3.4-ms delay, the minimum-phase response was significantly preferred to the linear-phase response for one music sample and vice-versa for the other sample with a sign test ($p < 0.04$) but not a Wilcoxon signed rank test that accounted for the low preference strength. Preferences between all other processing conditions were not significant. *Conclusions:* In acoustic conditions sensitive to delay effects, delays of 1.4 or 3.4 ms were either not detected or no less preferable than no delayed aided signal. It is unclear whether different phase-frequency responses may be preferred with different music stimuli.

Key Words: Hearing aid; open canal; delay; phase; music

Hearing aids introduce a delay between the input of sound at the microphone of the device and the output of the processed sound at the receiver. Hearing-aid delay is defined as the sum of all delays inherent in the analogue and digital signal-path circuits and processing. Hearing-aid delays are typically less than 10 ms with digital aids (Dillon et al, 2003) and around 0.5 ms with analogue aids (Frye, 2001). Due to the venting characteristics of earmolds or custom hearing aids, unaided and delayed versions of the same sound can be combined in the ear canal. When a sound is combined with a delayed version of itself, there is constructive interference at frequencies equal to N/T (N = 0, 1, 2, etc.; T = delay in seconds), where the delay is an integer multiple of the period, and destructive interference at frequencies equal to (N + 0.5)/T, where the delay is an odd multiple of 0.5 periods. The resultant comb-filtering effect adds spectral ripples to the combined sound's spectrum. When the unaided and delayed sounds have the same amplitude, the spectral ripple peaks and valleys are maximal at + 6 dB and $-\infty$ dB, respectively, relative to the unaided sound's spectrum. The peak heights and valley depths progressively reduce towards 0 dB as the amplitude difference between the unaided and delayed sound increases.

The effects of frequency-invariant delay have been extensively investigated with occluded fittings. In this case, the aid user's own voice reaches the ear canal first via bone conduction and later via the hearing aid. Several studies have reported mean disturbance ratings for this scenario with a 7-point scale where 1 = "Not at all disturbing", 4 = "Disturbing", and 7 = "Highly disturbing" (no labels for other numbers). In general, a Disturbing delay ranged from approximately 20 to 40 ms for normal- and impaired-hearing participants (Stone & Moore, 1999, 2002, 2005). At more typical delays for digital hearing aids, there was some level of disturbance with mean ratings of around 2 for delays of 6–7 ms with normal hearing (Stone & Moore, 1999, 2002) and means of approximately 2–3 at the lowest evaluated delay of 13 ms with impaired hearing (Stone & Moore, 2005). Agnew & Thornton (2000) found that for a group of normal-hearing participants listening to their own-voice stimuli, the mean just noticeable and objectionable added frequency-invariant digital processor delay was 4.1 and 14.3 ms, respectively. Some types of digital signal processing add a delay that varies with frequency. The variation of digital filter delay with frequency was investigated by Stone & Moore (2003), who found that a 9-ms increase in delay below 2.2 kHz was significantly more disturbing than a frequency-invariant filter delay.

Comparatively few studies have investigated the effect of delay with open-canal devices, where unaided sound from the air-conduction path can freely mix with the delayed version from the hearing aid. Linear-gain fittings have mainly been used to provide sensitive test conditions to isolate the effects of delay from other effects. Table 1 provides a summary of relevant data from previous studies of delay with open-canal fittings. Stone et al (2008) investigated the disturbance of hearing-aid delay on running speech with computer

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(Received 22 January 2012; accepted 5 June 2012)

ISSN 1499-2027 print/ISSN 1708-8186 online © 2012 British Society of Audiology, International Society of Audiology, and Nordic Audiological Society
DOI: 10.3109/14992027.2012.701020

Abbreviations

ADRO	Adaptive dynamic range optimization
BTE	Behind the ear
FIR	Finite impulse response
LD	Low delay
ULD	Ultra low delay
ULD +2ms	ULD plus 2-ms delay line

simulations. In their first two experiments, an open-canal aid was simulated by adding a copy of a computer speech file to itself with the desired delay and linear gain. The combined stimuli were scaled to remove level differences across gains, and presented to normal-hearing participants via headphones. With a flat gain of 0 dB at all frequencies (i.e. worst-case comb-filtering effect), mean disturbance ratings were approximately 2–3, 4 (Disturbing) and 4–5 at hearing-aid delays of 1, 3, and 5 ms, respectively, and decreased to 1–2 with increasing gain (less gain required with less delay). With more typical open-canal fittings (0 dB insertion gain at 1 or 2 kHz, falling as rapidly as possible at lower frequencies and rising at higher frequencies) mean disturbance ratings reduced to approximately 2–2.5, 2.5–3, and 3–3.5 at hearing-aid delays of 2, 4, and 6 ms, respectively. While there was still a level of disturbance with more typical fittings, the use of simulations, headphones, and normal-hearing participants was probably ideal for the detection of delay effects.

Groth and Søndergaard (2004) investigated the disturbance of electronic circuit delay with open-canal, behind-the-ear (BTE) casings connected to a real-time, digital equalizer that amplified and delayed the audio by 2, 4, or 10 ms. In addition to running speech, own-voice and music stimuli were also used. The insertion gain was set to 10 dB from 1–6 kHz on a Knowles Electronics Manikin for Acoustic Research and steeply rolled off at lower frequencies. The disturbance of delay was rated with the same 7-point scale as Stone & Moore (1999, 2002, 2003, 2005) and Stone et al (2008). With running speech, the normal-hearing participants' median disturbance ratings of 1 (Not at all disturbing), 1, and 2 for the 2, 4, and 10-ms circuit delays, respectively, were substantially lower than those reported by Stone et al (2008) with similar simulated hearing-aid delays and fittings. Disturbance ratings could be greater with the own-voice and music stimuli (median = 3 for the 10-ms circuit delay). In comparison, the impaired-hearing group rated all three delays as not at all disturbing (median = 1), except for the 10-ms circuit delay with the own-voice stimulus (median = 2). Thus, these studies showed that delay could be less disturbing with mild hearing loss or real-time, ear-level devices compared with normal hearing and simulations, and more disturbing with stimuli other than running speech. While Groth and Søndergaard (2004) showed that two of the lower delays of current digital hearing aids (2 and 4 ms) were not at all disturbing with mild hearing loss and ear-level devices, this was with a flat 10 dB insertion gain above 1 kHz, two speech stimuli, and only one non-speech stimulus.

In comparison with studies that used disturbance ratings, Bramsløw (2010) used a paired-comparison methodology to investigate preferred delays (5–10 ms) based on sound quality judgements with critical stimuli for detecting comb-filtering effects. The stimuli were critical in that they were familiar but not annoying and had energy at many frequencies to better reveal comb-filter notches. After informal listening tests the following critical stimuli were selected: waves on a beach and soft rain on a porch. Additionally, speech and own-voice stimuli were included as plausibly critical reference sounds.

Table 1. Summary of relevant data from previous studies of delay with open-canal fittings. The first two studies used a 7-point rating scale of disturbance, where 1 = "Not at all disturbing", 4 = "Disturbing", and 7 = "Highly disturbing" (labels were not assigned to other values).

	Stone et al (2008)	*Groth & Søndergaard (2004)*	*Bramsløw (2010)*
Hearing	Normal	Normal & impaired	Normal & impaired
Delay type	Simulated, frequency-invariant, hearing-aid delay	Frequency-invariant equalizer delay	Frequency-invariant, signal path delay
Fittings	Simulated open (headphones), linear insertion gain Exp. 1: Flat 0 dB gain Exp. 2: 12–24 dB gain above 1–2 kHz	Open, BTE casings, linear insertion gain (10 dB from 1–6 kHz)	Open, BTE aids, linear insertion gain (approximately 10 dB above 1–2 kHz) High pass filter above 0.1, 1.25 or 2.2 kHz
Stimuli	Speech	Speech, own voice, music	Speech, own voice, waves, rain
Findings	Mean disturbance ratings Exp. 1: 2–3 for 1 ms 4 for 3 ms 4–5 for 5 ms Exp. 2: 2.0–2.5 for 2 ms 2.5–3.0 for 4 ms 3.0–3.5 for 6 ms	Median disturbance ratings (normal/impaired) Own voice: 1/1 for 2 ms 2/1 for 4 ms 3/2 for 10 ms Speech: 1/1 for 2 ms 1/1 for 4 ms 2/1 for 10 ms Music: 1/1 for 2 ms 1/1 for 4 ms 3/1 for 10 ms	Paired comparisons of sound quality No preference for 5, 7, or 10 ms 2.2 kHz preferred to other filters, except for: • speech (IH) • own voice (IH) 1.25 kHz preferred to 0.1 kHz filter for: • speech (NH) • waves (NH & IH)

Exp. = experiment, NH = normal hearing, IH = impaired hearing.

The hearing aids were set to a flat, linear, 10-dB insertion gain on a head-and-torso simulator with occluded ear canals (all other features turned off). When the canals were opened, the gain gradually rolled off below 2 kHz with maximum comb-filtering ripple depth around 1 kHz. Thus, the fitting was similar to those of Groth and Søndergaard (2004) and Experiment 2 of Stone et al (2008). Bramsløw (2010) also investigated the effect of reducing comb-filtering ripples by removing the low-gain frequencies with a high-pass filter (below 0.1, 1.25, or 2.2 kHz). The non-speech, spectrally rich sounds tended to be the most critical stimuli for detecting comb-filtering effects, and the normal-hearing group was more sensitive to comb filtering than the impaired-hearing group. Signal-path delays of 5, 7, and 10 ms gave similar sound quality, and removing low-frequency comb-filter ripples was preferred by both groups. Whether this also applies to the lower delays of current hearing aids (below 5 ms) remained an open question.

Therefore, the literature suggests that people with mild hearing loss are generally less sensitive to delay in open-canal aids than people with normal hearing, and some non-speech sounds may be more critical for delay effects than running speech (Groth & Søndergaard, 2004; Bramsløw, 2010). However, while Bramsløw (2010) found no main effect of signal-path delays from 5–10 ms, Groth and Søndergaard (2004) found small, significant increases in disturbance from 4 to 10 ms with own-voice (normal and impaired hearing) and music (normal hearing) stimuli. Furthermore, these two studies used a flat insertion gain of approximately 10 dB above 1–2 kHz with ear-level devices, while simulations have shown that with less gain, disturbance increases and there may be some level of disturbance below 4–5 ms (Stone et al, 2008). The literature has not investigated delays below 4–5 ms with such lower gains and open ear-level devices, so it is unclear how little delay is required to avoid audible and/or undesirable delay effects in these conditions. Less attention has been directed towards the variation of delay with frequency (i.e. phase-frequency response) from different signal-path processing methods, such as finite impulse response (FIR) and infinite impulse response digital filters. One study showed that a 9-ms increase in digital filter delay below 2.2 kHz was significantly more disturbing than no increase with occluded ear canals (Stone & Moore, 2003), while an open-canal study did not clearly separate the disturbance of across-frequency delay from the effects of simulations of fast-acting compression and hearing loss (Stone et al, 2008).

Thus, there is scope for further investigation into the effects of delay with open-canal hearing aids, especially at lower delays and with stimuli that are critical in the sense that they are highly sensitive to the occurrence of comb filtering (this definition of ‘critical stimuli’ will be used for the current study). As such, the current study aimed to investigate whether preferences exist: (1) Between two of the lower hearing-aid delays of current devices (nominally 1.4 and 3.4 ms, as measured with a click stimulus); (2) Between the two phase-frequency responses of the signal-path, FIR filter designs that resulted in these hearing-aid delays; and (3) Between having such a delayed signal or no hearing-aid signal at low insertion gain. Following a similar rationale to Bramsløw (2010), the investigation was performed in sensitive test conditions and with critical stimuli for comb-filtering effects. The assumption was that if delay differences were not important in these sensitive test conditions, then they may not be important in less sensitive conditions. The sensitive test conditions were as follows: (1) Paired comparisons of processing differences to determine preferences; (2) The unaided and aided sounds had similar amplitudes to acoustically maximize the comb-filtering effect over a wide bandwidth (0.75–6 kHz); (3) Two music stimuli that were critical for comb-filtering effects were selected, rather than randomly select stimuli that may not be susceptible to these effects; (4) Musicians were recruited, since music is an important signal for them; and (5) Participants who were candidates for open-fit hearing aids were recruited, since they may be more sensitive to delay effects than participants with more severe hearing loss.

Three signal-path processing options were compared: Ultra low delay (ULD) processing (minimum-phase FIR filter, 1.4-ms hearing-aid delay), Low delay (LD) processing (linear-phase FIR filter, 3.4-ms hearing-aid delay), and ULD processing followed by a 2-ms delay line (minimum-phase FIR filter, 3.4-ms hearing-aid delay). These processing options were blindly compared in a first experiment, and compared with a muted hearing-aid output signal in a second experiment to address the following research questions.

1. With a minimum-phase FIR filter design, is a hearing-aid delay of 1.4 ms (ULD) or 3.4 ms (ULD + 2ms) preferred?
2. With a hearing-aid delay of 3.4 ms, is a minimum-phase (ULD + 2ms) or linear-phase (LD) FIR filter design preferred?
3. Is ULD (minimum phase, 1.4 ms hearing-aid delay) or LD processing (linear phase, 3.4 ms hearing-aid delay) preferred?
4. Is a delayed or no hearing-aid signal preferred?

Methods

Hearing aids and algorithms

One pair of commercial, open-fit, BTE hearing aids that contained open-platform, digital signal processing circuits were loaded with only the algorithms that were required for this study. Figure 1 shows the signal processing architecture, which consisted of a time-domain

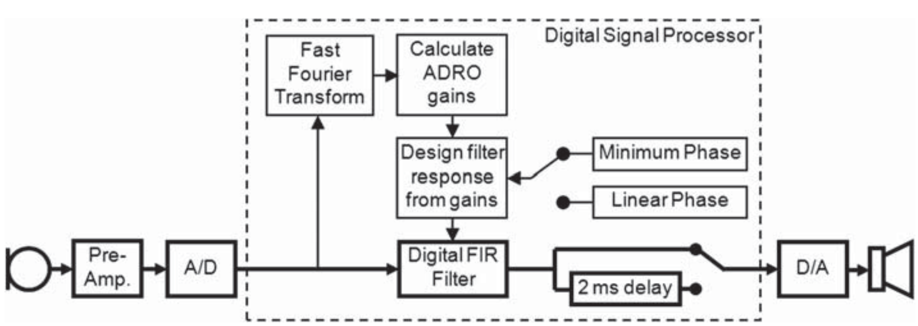

Figure 1. Hearing-aid signal processing architecture and reduced set of algorithms used in this study with selectable signal-path phase and delay settings (A/D = analogue to digital converter; D/A = digital to analogue converter).

signal path (shown in bold), and a frequency-domain processing side chain that controlled the gain-frequency response of the signal-path FIR filter. The hearing-aid circuits could handle input levels of approximately 110–115 dB SPL without clipping, and the sampling rate was 16 kHz. A fast Fourier transform computed 32-band spectral estimates from overlapping blocks of 64 samples every 2 ms, and the 32-channel adaptive dynamic range optimization (ADRO) amplification algorithm (Blamey et al, 2004, 2005) used these spectral estimates to calculate the required gain for each channel. For this study, ADRO was configured for linear (constant gain) amplification with 32-channel maximum output limiting for safety. The ADRO fitting and output-limiting gains were used to adjust the signal-path FIR filter so that it applied the desired gain-frequency response to the microphone signal. Thus, the frequency-domain side chain did not provide an additional path for the microphone signal to reach the receiver. Both aids were calibrated so that 2-cc coupler gain and output limiting threshold values entered in the fitting software were achieved to within ± 3 dB from 0.25–6 kHz (the measured values rolled off with the natural response of the transducers at other frequencies).

The FIR filter could be set to a minimum- or linear-phase design via the fitting software. The linear-phase design (LD processing) was a traditional, windowed, inverse fast Fourier transform approach that resulted in a 64-tap, symmetric, FIR filter that delayed all frequencies by 2 ms, and resulted in a nominal 3.4-ms hearing-aid delay with the study aids. The minimum-phase design (ULD processing) used the method of Dickson & Steele (2010) to produce an asymmetric, 32-tap, FIR filter optimized to introduce the lowest possible FIR filter delay, which was typically around 0–0.0625 ms, and resulted in a nominal 1.4-ms hearing-aid delay with the study aids. The gain-frequency response shaping ability of the two filter designs was effectively equivalent. A 32-sample (2 ms) delay buffer could be activated after the FIR filter for the third option of a minimum-phase filter design with a nominal 3.4-ms hearing-aid delay (i.e. ULD + 2ms). The fourth option of a muted (unaided) condition was created by setting the 2-cc coupler gain to minus infinity. By muting the aid in this way, low-level noise from the aid's output circuits was present as for the aided processing conditions, which avoided its absence acting as a cue.

Verification measurements

The gain-frequency response was automatically measured by a Frye FONIX 7000 Hearing Aid Test System (0.1 dB resolution), with the ear hook connected to the 2-cc HA2 coupler with size 13 tubing. In the fitting software, the linear 2-cc coupler gains were set to −4, 2, 8, 7, 8, 1, 6, 12 and 3 dB at 0.25, 0.5, 0.75, 1, 1.5, 2, 3, 4 and 6 kHz, respectively. These 2-cc coupler gains were used because they were the average across participants' ears that achieved the desired insertion gain for the study, which was the same for all participants regardless of hearing loss (see Fitting Procedure section below). Tone sweeps at 60 dB SPL were repeated five times for each processing condition, and the aids were not physically moved between measurements. Averaged across 35 measurement frequencies from 0.5–6 kHz, the measured gain was 0.5 dB greater with the linear-phase design (LD) than with the minimum-phase design (ULD) and varied from 0.0–1.1 dB greater across frequency. This difference was probably not perceptually significant for this study, since it was lower than typical intensity difference limens for stimuli separated by short periods of silence (Turner et al, 1989).

The phase response of the FIR filter designs was measured in MATLAB® with the filter coefficients that gave the mean study fitting. Figure 2 (open symbols) shows the time difference between the positive peaks of a sinusoidal test signal, sin(2πft) from t = 0, at the filter's output relative to its input, which is the phase shift applied by the filter in units of time and is a measure of the filter delay. The filter delay was 2 ms at all frequencies with the linear-phase design, while it was around 0 ms with the minimum-phase design. Small negative phase shifts indicate where the phase was slightly advanced by the filter; not that the stimulus appeared at the filter's output before its input.

The hearing-aid delay (i.e. the filter delay plus all other delays from the microphone input to receiver output) was automatically measured by a Frye FONIX 7000 with 0.1 ms resolution. This system used a wideband click stimulus and displayed a single delay value, which will be referred to as the nominal hearing-aid delay that would be measured by a manufacturer or clinician with such equipment. The aids were set up as for the gain verification. The nominal hearing-aid delay was consistently measured as 1.4, 3.4, and 3.4 ms with the ULD, ULD + 2ms, and LD processing conditions, respectively. Differences between the hearing aid and filter delays were due to other sources of delay in the hearing-aid signal path, such as the analogue-to-digital converter.

The variation of hearing-aid delay with frequency was manually measured with the aids placed in a Brüel & Kjær Type 4232 anechoic test box and set up as described above. A Brüel & Kjær Type 4192 microphone was placed next to the aid's microphone with another inside the 2-cc coupler. The Brüel & Kjær microphones were amplified by a Brüel & Kjær pre-amplifier and sampled with a 24-bit sound card, which also drove the test box speaker. Adobe Audition 1.5 computer software simultaneously recorded the aid's input and output (16 kHz sampling rate) while presenting sequential, ramped (to avoid wideband clicks), one-second bursts of sinusoidal stimuli at 1/3rd octave frequencies from 0.1–6 kHz at 70 dB SPL. Recordings were also made with the aid and 2-cc coupler removed

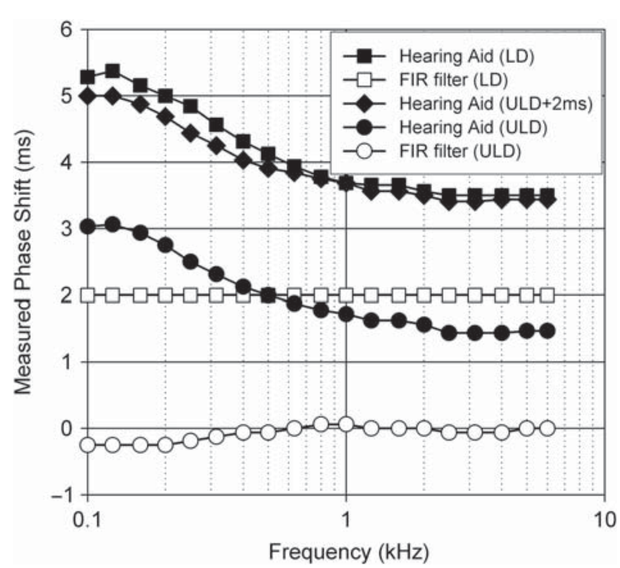

Figure 2. Measured phase shift (or phase delay) between the output and input of the FIR filter and the entire hearing aid (i.e. from microphone input to receiver output) for sinusoidal stimuli at 1/3rd octave frequencies with the ULD, ULD + 2ms, and LD signal-path processing options, which resulted in nominal hearing-aid delays of 1.4, 3.4, and 3.4 ms, respectively.

to establish the measurement system delay. Three recordings were made for each condition, and the delay between the input and corresponding output sinusoid waveform peaks was visually analysed to estimate the total phase shift (delay) imposed by the hearing-aid delay. Figure 2 (filled symbols) shows that the hearing-aid delay fell with increasing frequency for all processing conditions, and in the high frequencies it matched the delay measured by the Frye FONIX 7000 with a wideband click stimulus. Differences between the variation of hearing-aid (filled symbols) and filter (open symbols) delay with frequency were entirely due to electronic circuits external to the digital signal processor.

Participants

Twelve volunteers (six male and six female, 20 to 81 years old) agreed to participate in the study. The selection criteria were: (1) Sensorineural hearing loss; (2) Hearing thresholds of 25 dB HL or greater at audiometric frequencies above 1 kHz; and (3) Trained in reading music and participating in music performance. Two of the 12 volunteers had hearing losses greater than 80 dB HL in the right ear with a lesser degree of hearing loss in the left ear, so their right ears were excluded by inserting a foam plug. The remainder of the participants had hearing loss asymmetry no greater than 11 dB averaged across audiometric frequencies from 0.25 to 8 kHz. Figure 3 shows the mean audiogram for the 22 test ears (12 left ears and 10 right ears) with the error bars showing ±1 standard deviation. A further two musicians (one male and one female) with hearing thresholds better than 20 dB HL from 0.25 to 8 kHz were recruited as pilot participants to assist in the selection of critical music stimuli for comb-filtering effects. All participants were reimbursed for any travel expenses incurred to attend the sessions and received no other payments. The study was conducted under the approval of the Human Research and Ethics Committee of the Royal Victorian Eye and Ear Hospital for Project 08/831H.

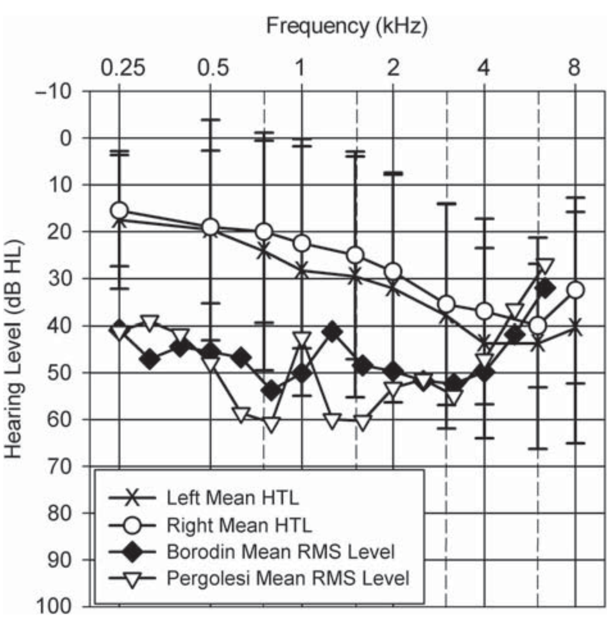

Figure 3. Mean pure-tone hearing thresholds for 22 test ears of the 12 participants (the bars show ± 1 standard deviation) and the long-term-average, one-third-octave-band, root-mean-square presentation levels (shown in dB HL units) of the two music stimuli averaged across participants.

Fitting procedure

The same pair of aids was used for all participants. The ear hook was connected to an Oticon Comfort Tip, which is a length of standard BTE plastic tubing with a 90° bend into the ear canal. This provided a wider frequency response than the alternative thin tubes. No dome was connected to the tubing so that the ear canal was as open as possible. With the aids turned off, the tubing tended to reduce the insertion gain to around −3 dB from approximately 4–6 kHz. The two participants with hearing losses greater than 80 dB HL in the right ear were fitted monaurally on the left ear, and the remaining 10 participants were fitted binaurally.

For most participants, the maximum output limits were set to 95 dB SPL (2-cc coupler) in each of the 32 channels. This was equivalent to a wideband level of 110 dB SPL, and minimized the possibility of output-limiting distortion during stimulus presentation. For two participants with hearing thresholds greater than 65 dB HL at some frequencies, the maximum output limits were raised to at least 100 dB SPL in most channels. The front omni-directional microphone was used and the volume control was disabled. No other processing features, such as noise reduction or feedback cancellation, were loaded into the test aids for this study (see Figure 1 for all features).

The aids were individually fitted to 10 dB insertion gain (± 1 dB) from 0.75–6 kHz with an Audioscan Verifit real-ear measurement system and pink noise at 60 dB SPL. The ULD processing and this insertion gain were used during fitting to satisfactorily reduce comb-filtering effects that could interfere with the fitting process, and was low enough to avoid feedback ringing and howling sounds without feedback management algorithms. The 2-cc gain values in the fitting software were set to the 0.75 and 6 kHz values for lower and higher frequencies, respectively.

After the insertion gain was adjusted to 10 dB from 0.75–6 kHz, the 2-cc coupler gain values in the fitting software were reduced by 22 dB at 0.25 kHz, 16 dB at 0.5 kHz, and 10 dB from 0.75 kHz and above. This made the amplitudes of the aided and unaided signals in the ear canal approximately equal from 0.75 to 6 kHz to give the worst-case acoustic comb-filtering effect. This fitting rationale did not account for differences in audibility among individual participants. Since this fitting made the real-ear aided response similar to the real-ear unaided response, the stimuli used in this study will be supra-threshold. The 2-cc coupler gain was rolled off below 0.75 kHz since little amplification was possible at those frequencies due to the leakage of sound from the open ear canal. Above 6 kHz, the aids were not calibrated so the gain rolled off with the natural response of the microphone and receiver.

Experimental setup

The experimenter used Abobe Audition 1.5 software to control the presentation of the music stimuli to the participants via a 24-bit, external sound card, which was connected to a low-distortion, 100 Watt power amplifier that drove a high-fidelity Dynaudio Acoustics BM5 near-field, studio monitor speaker. The speaker was located in a carpeted, double-walled sound booth that had a reverberation time of 200–140 ms from 0.25–8 kHz, respectively. Other equipment was outside the booth to minimize noise. The speaker had a wide, flat frequency response (0.05 to 21 kHz, ± 3 dB) and could handle the expected music levels (maximum output 103 dB SPL continuous and 125 dB SPL peak at 1 metre). The speaker was located 1 metre in front of the participants' seating position and head height (1 metre).

Music stimuli

A list of classical music samples were selected from recordings of orchestral and opera repertoire since these were among the styles preferred by 8 of the 12 hearing-impaired participants. Mono, 22.05-kHz bandwidth versions of these samples were created from the compact disc recordings for presentation via a single speaker. The aids were fitted to the normal-hearing pilot participants as described above, and the three aided processing conditions (ULD, ULD + 2ms, and LD) were blindly compared for each music sample. The two music samples where differences in processing were most clearly heard by the normal-hearing pilot participants were selected for the impaired-hearing participants. These critical music samples were 7–8 second excerpts from Borodin's String Quartet No. 2: Nocturne; and Pergolesi's Stabat Mater (female opera singer with string instruments).

Adjustment of presentation levels

The presentation level of each music sample was adjusted to a comfortable listening level for each impaired-hearing musician while wearing the aids in the muted condition. The samples were initially presented at a level that was anticipated to be soft based on the hearing loss. The presentation level was adjusted by the experimenter using an ascending approach until rated as 'moderately loud' (mf: mezzo forte) on an 8-point scale (see Table 2) comprised of standard musical loudness terms. Downward adjustments were made if this loudness was exceeded. When the desired loudness was achieved the sample's presentation level was fixed for all experiments. Figure 3 shows the long-term-average, one-third-octave, root-mean-square, presentation level (converted to dB HL) for each music sample averaged across participants relative to the mean audiogram. The corresponding wideband reference levels in dB SPL units were 72 and 73 dB SPL for the Borodin and Pergolesi samples, respectively.

Experiment 1: Aided processing conditions

The participants were not trained to listen for delay effects nor screened for their ability to hear differences between processing conditions. Blind paired comparisons of all aided processing conditions were made (ULD vs. ULD + 2ms, ULD + 2ms vs. LD, ULD vs. LD). The order of processing condition pairs and the first condition presented for each pair were counter-balanced across participants for each music stimulus. All comparisons were made with one music sample and then with the other sample, and the order of music samples was counter-balanced across participants. The participants indicated their preferred processing condition, and rated the strength of each preference as "Slightly prefer", "Prefer", or "Strongly prefer". A comparison could be repeated as many times as was required for a choice to be made. If there was no preference, a forced choice was made by the participant and the strength was recorded as "Slightly prefer". Paired comparisons allowed many conditions to be tested during a single 2-hour session, which included a rest break between Experiments 1 and 2.

Table 2. Eight-point scale of music loudness terms used to individually adjust the presentation levels of the music stimuli to *mf: mezzo forte* (moderately loud).

Symbol	*Meaning*	*English translation*
fff	Fortissimo possible	As loud as possible
ff	Fortissimo	Very loud
f	Forte	Loud
mf	Mezzo forte	Moderately loud
mp	Mezzo piano	Moderately soft
p	Piano	Soft
pp	Pianissimo	Very soft
ppp	Pianissimo possible	As soft as possible

Experiment 2: Unaided versus aided conditions

Blind paired comparisons of unaided versus each aided condition were made (Mute vs. ULD, Mute vs. ULD + 2ms, Mute vs. LD). The procedures and counter-balancing of processing conditions and music samples across participants were as for Experiment 1.

Results

Nonparametric statistics were used due to the ordinal nature of the data. Two-tailed sign tests and Wilcoxon signed rank tests were performed with the null hypothesis that the median was not equal to zero. For the Wilcoxon signed rank tests, the values ± 1, ± 2, and ± 3 were respectively assigned to the "Slightly prefer", "Prefer", and "Strongly prefer" responses, with the sign indicating the preferred condition. The sign tests only used the sign of these values.

Experiment 1: Aided processing conditions

Figure 4 shows preferences among each pair of aided processing conditions with the Borodin or Pergolesi music samples. At a nominal hearing-aid delay of 3.4 ms, there was a significant preference for the minimum-phase design (ULD + 2ms) with the Borodin sample (sign test, $p < 0.04$), and the linear-phase design (LD) with the Pergolesi sample (sign test, $p < 0.04$). However, these preferences were not significant with a Wilcoxon signed rank test that takes into account the strength of the preferences (ULD + 2ms with Borodin: $W = 16.0$, $p = 0.078$; LD with Pergolesi: $W = 63.5$, $p = 0.060$). Preferences were not significant for the other aided processing condition comparisons.

Experiment 2: Unaided versus aided conditions

Figure 5 shows the preferences between the muted and aided processing conditions with the Borodin or Pergolesi music samples. Preferences were not significant for all comparisons.

Discussion

The current study investigated whether preferences existed between two of the lower delays of current hearing aids with hearing-impaired participants wearing real, open-canal, devices set to acoustically maximize comb-filtering effects. Furthermore, musicians were recruited and listened to important stimuli (music) that were also critical for comb-filtering effects. Thus, the current study used what were probably worst-case acoustic conditions for comb-filtering effects in sensitive test conditions. This was motivated by the following: (1) The only previous study to investigate such low delays with ear-level devices used a higher insertion gain, and non-musicians found these delays not at all disturbing with a single music stimulus (Groth & Søndergaard, 2004); (2) Simulations showed that such delays would be more disturbing with less gain (Stone et al, 2008); (3) Stimuli that have energy across a broad range of frequencies can be critical

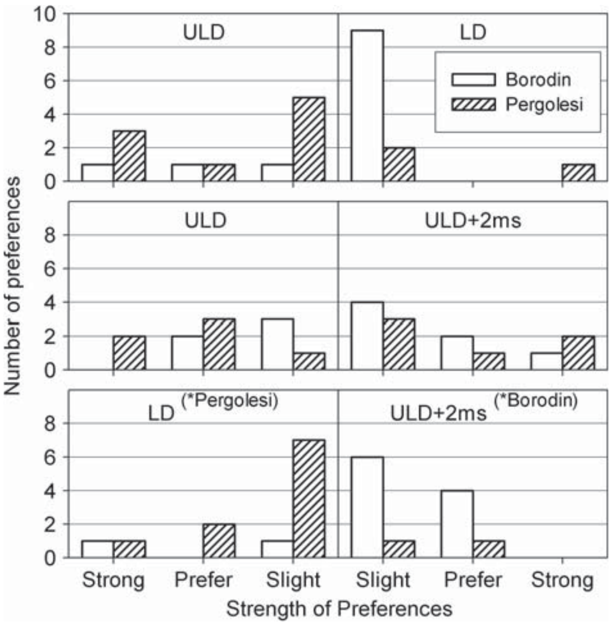

Figure 4. Preferences among aided processing conditions with the Borodin and Pergolesi music stimuli.

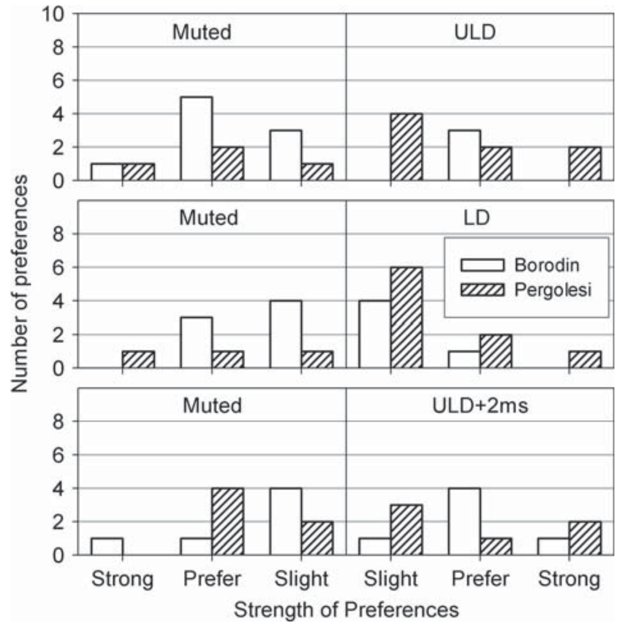

Figure 5. Preferences for a muted or delayed hearing-aid signal with the Borodin and Pergolesi music stimuli.

in that they can be highly sensitive to comb-filtering effects, and running speech is not a critical stimulus in this sense (Bramsløw, 2010); (4) Apart from nominal hearing-aid delay differences, more subtle differences in the variation of delay with frequency (phase-frequency response) were investigated; (5) The fitting allowed the evaluation of entirely removing the delayed aided signal and associated comb-filtering effects at low delays; (6) If differences between such low delays and phase-frequency responses were not important in such sensitive test conditions, then they may not be important in more typical conditions.

Figure 3 shows that averaged across participants, the one-third-octave, root-mean-square music presentation levels were above hearing threshold levels up to the 4-kHz band, while the music peaks may have been audible in the 5- and 6.3-kHz bands. Therefore, if delay effects were perceived, they were probably predominantly perceivable from 0.75–4 kHz, with the rolled-off gain reducing comb-filtering spectral ripple heights at lower frequencies, and increasing hearing threshold levels decreasing the audibility of delay effects at higher frequencies. More high-frequency gain to compensate for the sloping hearing loss would have increased the bandwidth over which ripples were above hearing threshold levels. However, this would have reduced the ripple height from the worst case, and ripple perception may have been reduced by the likely flattening of the auditory filter slopes with the increased sound pressure levels (Moore & Glasberg, 1997).

It is not clear whether differences between processing conditions were audible, since it is possible that some or all of the “Slightly prefer” responses were forced choices, where either differences were heard but with no preference, or differences were not heard at all. Preferences were not significant between the 1.4- or 3.4-ms nominal hearing-aid delays, or between any delayed signal and no delayed signal, in worst-case acoustic conditions and with critical music stimuli for the comb-filtering effect. This agrees with the findings of Groth & Søndergaard (2004) for delays of 2 and 4 ms with their music stimulus, although this was with non-musicians and a less worst-case fitting that probably only maximized comb-filtering ripples around 1 kHz. With a similar fitting to that of Groth & Søndergaard (2004), Bramsløw (2010) found that removing the largest ripples around 1 kHz was preferred with critical stimuli for delay effects, although delays below 5 ms were not evaluated. While the current study used a similar forced choice, paired comparison methodology to Bramsløw (2010) but with lower delays, comparisons with Bramsløw (2010) are limited by differences in gain, stimuli, and participants (e.g. musicians). The muted condition was thought of as being close to ideal in terms of sound reproduction, since the ear canal was open and there was no delayed hearing-aid signal. However, the canal was not fully open due to the presence of the tubing that tended to reduce the insertion gain to around -3 dB from approximately 4–6 kHz, and the addition of the delayed signal tended to remove this negative insertion gain. It is not clear whether this affected preferences, although at some of these frequencies the long-term average level of the music stimuli was below mean hearing thresholds (see Figure 3), and high-frequency spectral ripples may be difficult to resolve with an impaired auditory system.

In comparison, at a nominal hearing-aid delay of 3.4 ms there were significant preferences (sign test) for either the minimum- or linear-phase design that depended on the music sample, although due to the small magnitudes of the median preference strengths they were not significant with a Wilcoxon signed rank test. There was a high number of “Slightly prefer” responses for this pair of processing conditions, and it is possible that many were forced choices where participants could not hear a difference between conditions. If the “Slightly prefer” responses were not forced, the data suggests that differences in phase-frequency response and music stimulus may be as important as differences in nominal hearing-aid delay with open-canal hearing aids. The phase difference resulted in a 0.3-ms difference in the variation of delay across frequency, which was 1.9 and 1.6 ms for the LD and ULD + 2ms conditions, respectively (see Figure 2). While this may be a small delay difference, when converted back to phase the difference between LD and ULD + 2ms was 9°, 45°, 90°, and 135° at 0.8, 2, 4, and 6 kHz, respectively. This may have noticeably shifted the locations of the comb-filter peaks and valleys at lower frequencies where the spectral ripples are resolved by the auditory system.

Figure 2 showed that the total hearing-aid delay (i.e. from acoustic microphone input to acoustic receiver output) substantially differed from the digital FIR filter delay with the minimum- and linear-phase designs. Other processing distributed through the digital signal processor acted as a 5-sample (0.3125 ms) delay line (not shown for clarity) in addition to the FIR filter delay, so the remaining delay ($>$1 ms) and phase contributions were due to other electronic circuits external to the digital signal processor, such as the microphone, pre-amplifier, analogue-to-digital converter, digital-to-analogue converter, and receiver. Electret microphone and receiver circuits are predominantly capacitive and inductive, respectively, which have opposing but not necessarily equal phase-frequency responses. While the hearing-aid delay at 100 Hz was substantially greater than the nominal hearing-aid delays of 1.4 and 3.4 ms, it probably had little perceptual effect due to the roll-off in the 2-cc coupler gain below 0.75 kHz combined with the leakage of the aided signal out of the open ear canal.

Other factors that may have affected preferences include room reverberation, the music samples, and the recruitment of musicians. Room reverberation was not investigated, although it does not significantly affect the disturbance of delays of 7 ms or less with occluded ear canals and own-voice stimuli (Stone & Moore, 2002). The use of a wider selection of critical and non-critical music stimuli for delay effects in highly and less sensitive test conditions may better reveal trends, and would be interesting topics for future research. This could include conditions where the audibility of the test stimuli is maintained proportionally among the individual hearing losses. Future research could also investigate whether small phase-frequency response differences are audible at higher insertion gains where the delayed signal dominates and the comb-filtering effect is negligible. The FIR filter has a sharper temporal response to impulsive sounds with the minimum-phase design (peak output after 0–0.0625 ms), which is similar to an analogue system, while the response of the linear-phase design to impulsive sounds builds up over 2 ms. Whether these temporal differences are audible and the minimum-phase design preferred by people who still prefer the sound of analogue aids, would also be interesting topics for future research.

Conclusions

Hearing-impaired musicians compared nominal hearing-aid delays of 1.4 and 3.4 ms, minimum- and linear-phase FIR filter designs, and such processed signals with no hearing-aid signal with open-canal, BTE hearing aids fitted to acoustically maximize comb-filtering effects. Two music stimuli that were highly sensitive to the occurrence delay effects were presented at moderately loud levels. Preference strength was rated on a three-point scale, and forced choices were made at the lowest preference strength when there was no preference or no difference was heard. With a delay of 3.4 ms, preferences were significant for the minimum-phase design with the Borodin music sample, and the linear-phase design with the Pergolesi music sample according to a sign test, but not a Wilcoxon signed rank test that takes the low preference strength into account. Delays of either 1.4 or 3.4 ms did not significantly alter user preference relative to each other or the condition with no delayed aided signal, which indicates that the introduction of signals delayed by up to 3.4 ms either was not detected or did not have consistently positive or negative effects.

Acknowledgements

We wish to thank the many musicians who assisted in this study and Prof. Peter Blamey (The Bionics Institute) for his advice regarding the experimental design. This study was conducted under Project 08/831H approved by the Royal Victorian Eye and Ear Hospital Human Research and Ethics Committee. Parts of this paper were presented in a poster at The International Hearing Aid Research Conference, Lake Tahoe, California, USA, 11–15 August 2010.

Declaration of interest: The authors report no declarations of interest.

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