Wind noise in hearing aids: I. Effect of wide dynamic range compression and modulation-based noise reduction

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Abstract

Objective: The objectives of this study were: (1) to examine the effect of wide dynamic range compression (WDRC) and modulation-based noise reduction (NR) algorithms on wind noise levels at the hearing aid output; and (2) to derive effective strategies for clinicians and engineers to reduce wind noise in hearing aids. Design: Three digital hearing aids were fitted to KEMAR. The noise output was recorded at flow velocities of 0, 4.5, 9.0, and 13.5 m/s in a wind tunnel as the KEMAR head was turned from 0° to 360°. Study sample: Flow noise levels were compared between the 1:1 linear and 3:1 WDRC conditions, and between NR-activated and NR-deactivated conditions when the hearing aid was programmed to the directional and omnidirectional modes. Results: The results showed that: (1) WDRC increased low-level noise and reduced high-level noise; and (2) different noise reduction algorithms provided different amounts of wind noise reduction in different microphone modes, frequency regions, flow velocities, and head angles. Conclusions: Wind noise can be reduced by decreasing the gain for low-level inputs, increasing the compression ratio for high-level inputs, and activating modulation-based noise reduction algorithms.

Key Words: Hearing aid; wind; wide dynamic range compression; modulation-based; noise reduction

Wind noise can be problematic for hearing device users, especially for those who enjoy outdoor activities. With more children being fitted with hearing aids or receiving cochlear implants and more adults using hearing devices at younger ages, wind noise can become a prominent problem because these users tend to lead more active lives. In addition, hearing aid technologies have taken a great leap in the past 15 years. Modern digital hearing aids are often implemented with many signal processing algorithms and the effects of these algorithms on wind noise are not reported in the literature. In this study, we examined the effects of two signal processing algorithms that are commonly implemented in digital hearing aids, i.e. wide dynamic range compression and modulation based noise reduction, on wind noise level at the hearing aid output.

Wind (a naturally occurring air moment) or flow (a man-made air moment) noise in hearing aids is generated when the air flow impinges on the microphone diaphragms. Previous studies reported that flow noise level recorded at the hearing aid output depended on the flow velocity, the microphone mode, the hearing aid styles, and the head angle to the direction of the flow (Fortune & Preves, 1994; Grenner et al, 2000; Beard & Nepomuceno, 2001; Thompson & Dillon, 2002; Chung et al, 2009, 2010). The flow noise level measured at the hearing aid output generally increases with the flow velocity before the noise level reaches the maximum output level of the hearing aid. After that, an increase in flow velocity does not result in a further increase in flow noise level at the hearing aid output (Chung et al, 2009, 2010).
In addition, directional microphones usually yield higher flow noise levels than omnidirectional microphones (see Chung et al, 2010 for a detailed explanation). The exception is that directional microphones can reduce background noise, and therefore may generate lower noise levels than omnidirectional microphones at angles with the lowest flow noise levels in the high frequency region (Beard & Nepomuceno, 2001; Chung et al, 2009, 2010).

When the flow noise level is measured at different head angles, behind-the-ear hearing aids yield the highest flow noise levels when the flow comes from the front or the back. Flow noise levels are the lowest when the flow comes from the sides (Chung et al, 2009). Custom hearing aids generate the highest noise levels when the flow comes from the front or the back. They also yield high noise levels when the head is turned to 190 – 250° for hearing aids worn in the right ear, or to 110 – 170° for hearing aids worn in the left ear (both on the wind side, Beard & Nepomuceno, 2001; Chung et al, 2010).

WDRC algorithms are widely implemented in hearing aids to allow users to hear soft and loud sounds without the need to adjust the volume control (Dillon, 1996; Kuk, 1996; Cox, 1999; Souza, 2002). These algorithms generally provide more amplification to low-level than high-level sounds so that low-level sounds are audible and high-level sounds are not uncomfortable to hearing aid users. The compression ratios of WDRC are typically kept at or below 3:1 because higher compression ratios along with fast time constants are reported to degrade speech intelligibility (Souza, 2002). Despite their popularity in hearing aids, the effects of WDRC algorithms on wind noise have never been reported.

Modulation-based noise reduction algorithms are implemented in nearly all digital hearing aids. These algorithms are reported to reduce noise interference, listening effort, and aversiveness of sounds as well as increase listening comfort and cognitive capability in background noise (Boysman & Dreschler, 2000; Bentler & Chiou, 2006; Chung et al, 2006; Mueller et al, 2006; Palmer et al, 2006; Sarampalis et al, 2009). Common features for these noise reduction algorithms are to use modulation rates to infer the presence or absence of speech and to use modulation depth to estimate the signal-to-noise ratio (SNR) in the incoming signal (Powers et al, 1999; Holube & Velde, 2000; Chung, 2004; Bentler & Chiou, 2006). During speech production, the vocal tract changes shape to produce different sounds. The opening and closing of the vocal tract generate a slow modulation in the speech envelope, typically at the rate of 2 to 10 Hz. Sounds with modulation rates outside of this range are assumed to be noise (e.g. car noise, jackhammer). Additionally, the algorithms estimate the SNR of the incoming signal by tracking the peaks (maxima) and valleys (minima) of the signal. A high SNR ratio is inferred if there is a large difference between the maxima and the minima (i.e. high modulation depth), and vice versa. If the modulation rate is within the speech range (i.e. speech present) and the modulation depth is high (i.e. high SNR) in a signal processing channel, the gain of the channel is minimally altered. If the modulation rate is outside of the speech range (i.e. speech absent) and/or modulation depth is low (i.e. low SNR), the gain of the channel is reduced (Powers et al, 1999; Chung, 2004).

Modulation-based noise reduction algorithms implemented in different hearing aid models, or by different hearing aid manufacturers, also can be implemented with: (1) different specifications (e.g. the number of signal processing channels); (2) additional sound classification algorithms that classify the incoming sounds into different categories (e.g. speech, music, background noise, wind noise); (3) different decision rules which determine the amount of noise reduction in each frequency bands when noise is detected; and (4) different time constants that determine how fast the gain changes are executed (Chung, 2004). These parameters may also change depending on whether the hearing aid is in the directional or omnidirectional mode.

As these noise reduction algorithms are usually implemented with additional sound identification analysers (e.g. temporal-intensity patterns within a frequency channel and across frequency channels), they do not depend solely on modulation characteristics to infer the presence of speech or wind noise. Although speech and wind noise may have a similar modulation spectrum (Kates & Arehart, 2005; Kates, 2008), it is possible that these algorithms can still separate speech and wind noise based on other acoustic characteristics, and reduce wind noise interference. Nevertheless, no published study has examined this hypothesis.

The purposes of this study were: (1) to examine the effect of wide dynamic range compression on wind noise levels at the hearing aid output; (2) to explore if modulation-based noise reduction algorithms could reduce wind noise levels when the hearing aid was programmed to directional and omnidirectional microphone modes; and (3) to determine if these algorithms can be integrated into wind noise reduction algorithms to lessen wind noise interference for hearing aid, cochlear implant, and hearing protector users.

Materials and Methods

Digital hearing aids were programmed in an anechoic chamber when worn on a Knowles electronic manikin for acoustic research (KEMAR, Burkhard & Sachs, 1975). The default hearing aid program was omnidirectional microphone (OMNI), linear amplification (LIN) with flat frequency response, the noise reduction algorithm deactivated (NRoff), and all other signal processing algorithms deactivated. The hearing aids were then programmed to the following experimental conditions via hearing aid fitting software:

1. WDRC (compression ratio = 3:1)
2. Modulation-based noise reduction algorithm activated (NRon)
3. Directional microphone (DIR)

The KEMAR head with the hearing aids was then transported to a quiet wind tunnel. The hearing aid outputs were recorded at different flow velocities when the hearing aids were set to different programs. The KEMAR head and hearing aids were later transported back to the anechoic chamber for calibration measurements. Throughout the entire process, the hearing aids were kept on in KEMAR’s pinna and any physical contact with the hearing aids was avoided. These caution were taken to ensure that the frequency responses and hearing aid characteristics were identical in the anechoic chamber and in the wind tunnel measurements.
Hearing aid characteristics

Two behind-the-ear (BTE1 and BTE2) and one in-the-ear (ITE1) digital hearing aids were tested in this study. BTE1 and ITE1 were made by the same manufacturer. They had 16 compression and noise reduction channels and fast compression time constants ($\leq 50$ ms). BTE1 was also used in Chung and colleagues (2009, BTE2) and ITE1 was used in Chung and colleagues (2010, ITE1). BTE2 was a digital hearing aid made by another manufacturer. It had 17 compression and noise reduction channels. ITE1 did not have a vent and both BTE1 and BTE2 were coupled to KEMAR’s ear canal using a skeleton acrylic earmold without a vent. When worn on KEMAR, the two omnidirectional microphones that formed the directional microphone in the hearing aids were aligned horizontally.

All the hearing aids were high-end digital hearing aids. They were chosen because they either do not have a wind noise reduction algorithm or the algorithm could be deactivated in the hearing aid fitting software. None of them had transient noise reduction algorithms. During testing, all other signal processing algorithms (e.g. automatic microphone switching algorithms, automatic program switching algorithms, feedback reduction algorithms) were disabled.

The input-output function of many hearing aids have an expansion region to reduce the gain of sounds below 35–40 dB SPL, a WDRC region to provide variable gains for sounds between 40–90 dB SPL, and an output limiting region to reduce the hearing aid output to several dB below the maximum output level of the hearing aid fitting software. The expansion region, however, only affects the hearing aid output if the incoming sound fell below 38 dB SPL.

Table 1 lists all the experimental conditions for flow noise measurements. The omnidirectional mode of BTE1 (BTE1 OMNI) was tested when it was programmed to the linear (i.e. compression ratio = 1:1, LIN) or the compression (compression ratio = 3:1, WDRC) modes. Both microphone modes of ITE1 and BTE2 were tested with the noise reduction algorithm activated (NRon), and deactivated (NRoff). Two hearing aids from different hearing aid manufacturers were tested in order to examine the noise reduction effects of different implementations of modulation-based noise reduction algorithms.

Hearing aid programming

Sound field calibration was conducted prior to hearing aid programming. A pink noise with bandwidth of 1–8000 Hz was presented from a sound editing software (Adobe Audition 1.0), a sound card (Delta1010, M-Audio), and then a powered loudspeaker (Mackie HR824) in an anechoic chamber. The pink noise was picked up using a 1.27-cm microphone (ER-11 1/2” microphone) placed in the medial opening of a Zwischen coupler. The pre-amplifier of the microphone (ER-11 pre-amplifier) was set to have flat frequency response and –10 dB gain, which allowed undistorted output for sounds up to 140 dB SPL at the microphone input. The signal was then sent to a sound card (Sound Blaster Extigy) and a second computer.

The spectrum of the signal was analysed in one-third octave bands from 1 to 10000 Hz in real time using SpectraPlus (Pioneer Hill Software). The hearing aid gains for low-, medium- (if available), and high-level inputs were adjusted to be identical in each frequency channel using the manufacturers’ hearing aid fitting software (i.e. linear amplification). The gains were also adjusted across frequency channels to match the spectrum of the original pink noise to ensure that the frequency response of the recording path (i.e. from the hearing aid to the computer sound card) was as flat as possible.

All hearing aids were programmed in identical procedures. Figure 1 shows the frequency responses of the hearing aids in response to the 75 dB pink noise.

BTE1 OMNI was also programmed to 3:1 by increasing the gain of low-level sounds across the frequency region of their linear programs and reducing the gain of high-level sounds for the equal amount. For each hearing aid, the compression ratios were verified in a Fonix 7000 Hearing Aid Analyser (Frye Electronics) at 250, 500, 1000, 2000, and 4000 Hz first. The gain differences between low-, medium- (if available), and high-level gains in order to yield 3:1 compression were noted in each frequency channel. The frequency response was then verified when the hearing aid was worn on KEMAR. The gains were adjusted across the frequency channel to yield a relatively flat frequency response. The gain relationship within each frequency channel was maintained to keep the 3:1 compression ratio.

For the NRon mode, the modulation-based noise reduction algorithms of ITE1 and BTE2 were activated using the hearing aid fitting software.

Wind tunnel characteristics

The wind tunnel was an Eiffel type which measured 10.36 metres long and 2.3 metres high (Brown & Monegue, 1995). It had a testing section with dimensions of 52.7 cm wide $\times$ 60.3 cm high $\times$ 121.9 cm long. The wind tunnel was designed to generate quiet laminar flow in the testing section in the absence of objects/obstructions. It was

<table>
<thead>
<tr>
<th>No. of channels</th>
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<th>Ears</th>
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<th>MBNR</th>
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<tbody>
<tr>
<td>BTE1</td>
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<td>L OMNI</td>
<td>OMNI</td>
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<tr>
<td>ITE1</td>
<td>16</td>
<td>L OMNI</td>
<td>OMNI</td>
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<td>BTE2</td>
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<td>2 OMNI</td>
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Table 1. Hearing aid characteristics and testing conditions.
The signal was subsequently fed into the external sound card in the Zwislocki coupler and the ER-11 preamplifier amplified the process. The ER-11 microphone picked up the hearing aid output measurements was identical to that used in the hearing aid programming the wind tunnel testing section on the wooden base to serve as the section. A mark was also made equidistant to the lateral walls of installed in a circular opening on the wooden base of the testing wind tunnel test section with respect to length, width, and height. A circular wooden dial with angle markings in 10° increments was wind tunnel which varied for different microphone modes at different head angles.

Prior to flow noise measurements, the fan speeds required to obtain the flow velocities were calibrated using Pitot tube measurements and a linear regression equation. The fan speeds required to generate 4.5, 9.0, and 13.5 m/s were 10.3, 19.3, and 28.2 revolutions per second, respectively. More details of the wind tunnel and flow characteristics are reported in Brown & Mongeau (1995) and Chung and colleagues (2009).

Flow noise measurements
The KEMAR head and its ears were positioned in the center of the wind tunnel test section with respect to length, width, and height. A circular wooden dial with angle markings in 10° increments was installed in a circular opening on the wooden base of the testing section. A mark was also made equidistant to the lateral walls of the wind tunnel testing section on the wooden base to serve as the 0° reference. The recording equipment used in the flow noise measurements was identical to that used in the hearing aid programming process. The ER-11 microphone picked up the hearing aid output in the Zwischenlocki coupler and the ER-11 preamplifier amplified the signal. The signal was subsequently fed into the external sound card and recorded in a computer using an audio recording and editing software (Audition).

For each hearing aid, the location with the maximum flow noise levels was identified by slowly turning the KEMAR head from 0° to 360° when the hearing aid was programmed to the DIR mode and the flow velocity was set at 13.5 m/s (i.e. the maximum velocity). The recording level of the sound card was adjusted so that the maximum flow noise level was 10 dB below the limit of the dynamic range of the sound card. This process ensured that the hearing aid output level would not exceed the limits of the recording equipment and the upper dynamic range of the sound card was utilized for better amplitude resolution.

During flow noise measurement, the KEMAR head was turned relative to the direction of the flow every 10° from 0° to 360°. Thirty-second recordings were made at flow velocities of 0, 4.5, 9.0, and 13.5 m/s when the hearing aid was programmed to different conditions. The volume control and the recording levels of the sound card were noted.

ITE1 OMNI at 0 and 9.0 m/s and DIR at 4.5 and 13.5 m/s were measured again 14 days later to check the repeatability of flow noise measurements.

Calibration
The KEMAR head with the hearing aid was transported back to the anechoic chamber after flow noise measurements. The hearing aid output in response to a 75 dB SPL pink noise was recorded using identical equipment and volume control settings as in the wind tunnel. This calibration noise served as a reference for 75 dB SPL input level and the flow noise levels in the recordings were calculated using the MATLAB program. As output equaled input plus gain (Output = Input + Gain) for any audio system, this calibration procedure normalized the insertion gain of the hearing aid and the overall gain of the recording path to 0 dB (Chung et al, 2009, 2010).

Results
Overall noise levels and one-third octave band noise levels from 100 to 8000 Hz were analysed using MATLAB programs. In the following discussions, overall levels and/or one-third octave band levels at 125, 500, and 2000 Hz are plotted in Cartesian plots, instead of polar plots, in order to allow easy level comparisons.

WDRC
The overall noise levels of LIN and WDRC recorded at the four flow velocities are plotted in Figure 2. The plots of BTE1 LIN vs. WDRC showed that WDRC reduced the dynamic range of the flow noise. Low-level flow noise was amplified and high-level flow noise was reduced at all head angles. These results are consistent with the general actions of WDRCs reported in the literature (Dillon, 1996; Kuk, 1996; Souza, 2002).

For any compression system, the effective compression ratios of hearing aids for real-world stimuli decrease with increase in temporal fluctuations exhibited in the incoming sounds and decrease with increase in the compression time constants (Stone & Moore, 1992; Armstrong, 1993; Souza, 2002). As compression ratio is defined as change in input divided by change in output, the effective compression ratio for the 3:1 WDRC condition (CR\text{WDRC}) in this study can be calculated using the differences in output levels recorded in the 4.5 and 13.5 m/s conditions:
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where

$\text{CR}_{\text{LIN}} = 1:1$ as verified in Fonix Hearing Aid Analyser; and

$\text{Change in Input}_{\text{LIN}} = \text{Change in Input}_{\text{WDRC}}$, because the changes were induced by the change in flow velocity at the microphone input;

Therefore,

$$\text{CR}_{\text{WDRC}} = \frac{\text{Change in Output}_{\text{LIN}}}{\text{Change in Output}_{\text{WDRC}}} \quad (2)$$

The effective compression ratios for BTE1 calculated using formula (2) and the differences in outputs in the 4.5 and 13.5 m/s conditions ranged from 1.7:1 to 3.0:1 with an average of 2.2:1.

The average effective compression ratio was lower but relatively close to the compression ratio measured using pure-tone signals in the hearing-aid analyser. This finding is consistent with the previous notion that the effective compression ratio for sounds with temporal fluctuations is always lower than the compression ratio measured using steady-state stimuli (Stone & Moore, 1992; Souza, 2002). This is because the compressor does not have sufficient time to fully reduce the gain before it needs to adopt a lower gain reduction (when the flow noise level decreased) or release the compression (when the flow noise level dropped below the compression threshold). The fast compression time constants in BTE1, on the other hand, have prevented the effectiveness compression ratio from dropping close to 1:1 (linear amplification) as in systems with very long time constants.

**Noise reduction**

**Noise patterns**

In general, Figures 3 and 4 show that NRon generally yielded lower noise levels for both OMNI and DIR modes in all plots. The level differences between the NRoff and NRon conditions, i.e. NRoff minus NRon (NRoff − NRon), are plotted in Figure 5 in order to examine the differences more closely. Positive values in Figure 5 indicate the amount of noise reduction in dB provided by the noise reduction algorithm. Negative values indicate that the noise level was higher when the noise reduction algorithm was activated. Dotted lines are added at the 0 dB tick marks for easy level comparisons.

Figure 5 shows that the noise reduction algorithm was consistently more effective in reducing noise level in the absence of a flow (i.e. 0 m/s) than in the presence of a flow. In addition, the algorithm was generally more effective in OMNI than DIR because OMNI often had higher positive values (higher amount of noise reduction) than DIR, except at 125 Hz. Further, most of the NRoff minus NRon difference values were positive for OMNI in all plots, whereas many level differences for DIR had negative values at 125 and 500 Hz (i.e. 110° to 360° at 500 Hz), especially at higher flow velocities. At 13.5 m/s, for example, NRoff minus NRon yielded negative values in many head angles, indicating that the noise level was higher at 125 and 500 Hz when the noise reduction algorithm was activated. Nevertheless, these negative differences were rarely observed in overall levels or in 2000 Hz one-third octave band levels, suggesting that the noise reduction algorithm was able to reduce noise at higher frequencies and did not harm the overall performance of the hearing aid even at high flow velocities.

Figures 6 and 7 showed the noise levels measured at the BTE2 output, and Figure 8 shows corresponding level differences for NRoff minus NRon (NRoff − NRon). The BTE2 results were consistent with those of ITE1 that noise levels were generally lower in the NRon than the NRoff condition (Figures 6 and 7). Figure 8, however, showed that BTE2 and ITE1 had different NRoff − NRon patterns. The NRoff − NRon values of BTE2 at 0 m/s flow velocity were not necessarily the highest among all flow velocities. For example, values obtained at flow velocity of 13.5 m/s were consistently higher than those at 0 m/s at 125 Hz one-third octave band. In addition, the amount of noise reduction provided in the OMNI and DIR modes were generally similar for BTE2, except around 90° where the algorithm provided lower amount of noise reduction in the DIR than the OMNI mode at 500 and 2000 Hz. Further, activating the noise reduction algorithm generally did not increase the noise levels for BTE2. Figure 8 only shows occasional negative NRoff − NRon

![Figure 2](#)

**Figure 2.** Overall levels of hearing aid output when BTE1 OMNI was programmed to linear amplification (LIN), or wide dynamic range compression (WDRC) with 3:1 compression. Hearing aid gain was normalized to 0 dB.
values, which are likely due to angle-to-angle fluctuations in flow noise levels in different settings instead of systematic deviations.

**Statistical Analyses**

Two (hearing aid) × 2 (microphone mode) × 4 (flow velocity) repeated measure ANOVAs, with repeated measures on hearing aid output levels, were conducted to examine if the noise reduction algorithms provided statistically different amounts of noise reduction at the overall and 500-Hz one-third octave band levels. The overall levels were chosen because they reflected the overall performance of the noise reduction algorithms. The 500-Hz levels were chosen to examine the effect of noise reduction algorithms on flow noise in the low frequency region. Although wind noise tends to have high energy concentration at very low frequencies (e.g. around 125 Hz), hearing aid frequency responses typically have a low-frequency roll off at 125 Hz, partly because such low frequencies do not carry important information for speech understanding. The level differences of \( NR_{off}/NR_{on} \) were, therefore, not analysed at 125 Hz one-third octave band. The average amounts of noise reduction provided in each flow velocity condition are tabulated in Table 2.

The ANOVA results showed significant main factors of microphone mode (\( p < 0.05 \)), flow velocity (\( p < 0.05 \)), and significant interactions of hearing aid * microphone mode (\( p < 0.05 \)), hearing aid * flow velocity (\( p < 0.05 \)), microphone mode * flow velocity (\( p < 0.05 \), Table 3). Significant interactions were also found for the interaction

**Figure 3.** Overall and one-third band noise levels measured at the hearing aid output when the omnidirectional (OMNI) microphone mode of ITE1 was programmed to linear amplification (LIN), with (NRon) or without (NRoff) activating the noise reduction algorithm. Hearing aid gain was normalized to 0 dB.
of hearing aid*microphone mode*flow velocity (p < 0.05) at 500 Hz, but the main effect of hearing aid did not reach statistical significance in any analysis (p > 0.05).

Post hoc Fisher’s PLSD tests conducted on the ITE1 data indicated that the modulation-based noise reduction algorithm provided an average of 2 dB and 2.4 dB more noise reduction in the OMNI than the DIR mode for overall and 500-Hz levels, respectively (p < 0.05, see Table 2). In addition, the amount of noise reduction decreased with the increase in flow velocity. The average amount of noise reduction provided at the overall and 500 Hz levels were significantly higher for 4.5 than 13.5 m/s (p < 0.0083, p adjusted for six tests); for 0 m/s than 13.5 m/s (p < 0.00083); for 4.5 than 9.0 m/s (p < 0.00083); for 0 than 9.0 m/s (p < 0.0083); and for 0 than 4.5 m/s (p < 0.0083). Only the level differences between 13.5 and 9.0 m/s were not statistically significant.

For BTE2, post hoc Fisher’s PLSD tests indicated that the modulation-based noise reduction algorithm provided an average of 0.7 dB and 0.9 dB more noise reduction in the OMNI than the DIR mode for overall and 500 Hz levels, respectively (p < 0.05). As the average differences between OMNI and DIR were within 1 dB (Table 2), BTE2 essentially provided the same amount of noise reduction in the two microphone modes.

Regarding the amounts of noise reduction at different flow velocities, BTE2 provided higher amounts of noise reduction at 500 Hz for 0 than 13.5 m/s (p < 0.0083, p adjusted for six tests), at the overall level for 13.5 than 9.0 m/s (p < 0.0083) and for 13.5 than 0 m/s (p < 0.0083), and at both the overall level and 500 Hz for 0 than 9.0 m/s (p < 0.0083) and for 0 than 4.5 m/s (p < 0.0083). The average differences between 9.0 and 4.5 m/s were not significant in either
the overall level or the 500 Hz one-third octave band level analyses ($p > 0.0083$).

These results of ITE1 and BTE2 suggested that modulation-based noise reduction algorithms implemented in different hearing aids provide different amount of wind noise reduction: (1) at different flow velocities; (2) at different head angles relative to the direction of the flow; and (3) at different microphone modes. The average amount of overall noise reduction ranged from 1.7 to 6.3 dB in the presence of a flow, and up to 8.3 in the absence of a flow. These findings indicated that modulation-based noise reduction algorithms can potentially help increase the comfort of hearing-aid use in wind and reduce the annoyance of wind noise for hearing-aid users.

**Repeatability**

The flow noise measurement on BTE1 was highly repeatable (Figure 9). The average differences on the overall levels between Run 1 and Run 2 were within 0.4 dB for OMNI at 4.5 and 13.5 m/s, and within 1.3 dB for DIR at 0 and 9.0 m/s. The greatest difference was 12.5 dB which occurred at 30° at 13.5 m/s for DIR. This occurred in a region in which the flow noise levels fluctuated considerably from angle to angle (i.e. the BTE1 DIR flow noise levels changed from 109.6 to 78.5 dB between 0° and 50°). Such fluctuations were likely due to turbulence created by the flow passing the pinna and creating fluctuating pressure on the microphone. The test-retest measurements, therefore, varied considerably. Similar findings were also reported by Chung and colleagues (2009).

**Discussion**

**Effects of WDRC**

WDRC increased low level noise but decreased high level flow noise (Figure 2), indicating that flow noise at the hearing aid output was
Effect of signal processing on wind noise

affected by the amount of gain applied in the signal processing path. These results were consistent with the general actions of WDRC (Dillon, 1996; Kuk, 1996; Souza, 2002). The implication is that clinicians and hearing-aid engineers can manipulate various hearing aid settings to reduce wind noise levels.

One of the strategies is to reduce the gain for low-level inputs or reduce the compression ratios for low- to mid-level inputs when low-level wind noise is detected. As the results of this study were obtained with a hearing aid with fast compression time constants and slower time constants have been reported to reduce the effective compression ratio of the hearing aid (Armstrong, 1993; Souza, 2001), another strategy is to manipulate the time constants to increase the effective compression ratio for high level sounds.

Additionally, most digital hearing aids are either implemented with a high-level compressor or automatic gain control algorithms to prevent very high-level sounds from exceeding the dynamic range of the digital chip or the maximum output level of the receiver. Wind noise, therefore, can be reduced by: (1) increasing the compression ratio of the high-level compressor (i.e. reducing the gain as input increased); or (2) by decreasing the compression threshold of the high-level compressor/automatic gain control algorithm (i.e. starting to reduce the gain at a lower input level). These strategies can potentially help reduce the noise levels generated at the hearing aid output by strong wind.

An important precaution for implementing these strategies is not to compromise speech intelligibility. Clinicians can set a ‘wind’

Figure 6. Overall and one-third band noise levels measured at the hearing aid output when the omnidirectional (OMNI) microphone mode of BTE2 was programmed to linear amplification (LIN), with (NRon) or without (NROff) activating the noise reduction algorithm.
program and instruct hearing-aid users to use the program only in situations requiring minimum communication (e.g. during jogging and fishing). Engineers may take advantage of the sound scene analysis algorithms implemented in most digital hearing aids to utilize these strategies only when speech is not detected in the incoming signal and to release the settings somewhat when speech is detected.

Effects of noise reduction algorithm
The NRon – NRoff differences shown in Figures 5 and 8 indicated that modulation-based noise reduction algorithms implemented in both digital hearing aids generally reduced flow noise levels at the hearing aid outputs. The exception was that DIR mode of ITE1 exhibited slightly higher noise levels at 125 and 500 Hz when the noise reduction algorithm was activated than when deactivated (i.e. the negative values on Figure 5). These negative values, however, are not observed in the overall levels of ITE1 DIR, indicating that the algorithm provided more noise reduction at high frequency regions that have off-set the increase in noise level at low frequency regions. The activation of the noise reduction algorithm, therefore, did not increase the overall noise level at the hearing aid output.

Data collected in the absence of a flow (i.e. 0 m/s) reflected the amounts of noise reduction provided in relatively steady background noise in the far field of the hearing-aid microphones (i.e. > 20 cm, Thompson, 2002). The results showed that the amount of noise reduction is generally higher in the absence of the flow than in the

Figure 7. Overall and one-third band noise levels measured at the hearing aid output when the directional (DIR) microphone mode of BTE2 was programmed to linear amplification (LIN), with (NRon) or without (NRoff) activating the noise reduction algorithm.
presence of the flow (Figures 3 to 8), indicating that both noise reduction algorithms were more effective in identifying and reducing ambient/background noise with relatively low temporal fluctuations than flow noise. The amount of noise reduction provided by the algorithms in the absence of wind, therefore, cannot be used to predict the amount of noise reduction in the presence of wind.

Data collected in the presence of a flow reflected the amount of noise reduction provided in temporally fluctuating flow noise that were generated in the near field (within several centimetres) of the microphones. ITE1 and BTE2 provided a different amount of noise reduction depending on the microphone mode, frequency region, head angle, and flow velocity (see Figures 3 to 8). Although the overall amounts of noise reduction provided by the two hearing aids were not statistically different, Figures 5 and 8 show that they can be quite different at different frequencies, even at the same head angle (e.g. around 90° for ITE1 and BTE2 in different frequencies).

In addition, noise reduction algorithms implemented in both ITE1 and BTE2 provided higher amounts of noise reduction in the OMNI than the DIR mode. This could, again, be due to the differences in the temporal fluctuations of the flow noise. Previous reports showed that wind noise can have similar modulation characteristics as speech (Kates, 2008) and DIR generally had higher temporal fluctuations than OMNI (Chung et al, 2010). It is possible that the detection and analysis unit of the noise reduction algorithms had more difficulty detecting the existence of flow noise in DIR than OMNI. It is also likely that

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**Figure 8.** The amount of noise reduction, NRoff minus NRon (NRoff – NRon) provided by BTE2 in the OMNI and DIR modes.
The decision rules of the algorithms are written in such a way that less gain reduction was applied to sounds with high temporal fluctuations in order to lower the chance of reducing speech signals. Despite the above limitations, both noise reduction algorithms were able to utilize other differences in acoustic characteristics between speech and wind noise. They applied some gain reduction to lower noise levels at the hearing aid output or at least did not increase the overall level. The clinical implication is that at least two modulation-based noise reduction algorithms can be incorporated into wind noise reduction strategies to help reduce wind noise in many head angles and do no harm in others. Engineers, however, should examine the effectiveness of individual algorithms in the presence of a flow before automatically incorporating these noise reduction algorithms as a part of wind noise reduction strategies. This is because some algorithms might increase instead of reduce flow noise in certain frequencies (e.g. ITE1 DIR at 125 Hz in Figure 4).

Further, as the amount of noise reduction tends to change depending on the head angle, engineers can design hearing aids

<p>| Table 2. | The average amount of noise reduction provided by ITE1 and BTE2 in the directional (DIR) and omnidirectional (OMNI) microphone modes at four flow velocities. |</p>
<table>
<thead>
<tr>
<th>Flow velocity</th>
<th>ITE1</th>
<th>BTE2</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>DIR</td>
<td>OMNI</td>
</tr>
<tr>
<td>13.5 m/s</td>
<td>1.7</td>
<td>5.0</td>
</tr>
<tr>
<td>9.0 m/s</td>
<td>2.0</td>
<td>5.1</td>
</tr>
<tr>
<td>4.5 m/s</td>
<td>3.7</td>
<td>6.0</td>
</tr>
<tr>
<td>0 m/s</td>
<td>8.3</td>
<td>7.7</td>
</tr>
<tr>
<td>500 Hz</td>
<td></td>
<td></td>
</tr>
<tr>
<td>13.5 m/s</td>
<td>0.3</td>
<td>4.8</td>
</tr>
<tr>
<td>9.0 m/s</td>
<td>1.2</td>
<td>5.4</td>
</tr>
<tr>
<td>4.5 m/s</td>
<td>3.6</td>
<td>5.5</td>
</tr>
<tr>
<td>0 m/s</td>
<td>10.8</td>
<td>9.9</td>
</tr>
</tbody>
</table>

<p>| Table 3. | Analysis of variance (ANOVA) table for the overall amount of noise reduction and the amount of noise reduction at 500 Hz one-third octave band. |</p>
<table>
<thead>
<tr>
<th>df</th>
<th>Sum of squares</th>
<th>Mean square</th>
<th>F-value</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Overall</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hearing aids</td>
<td>1</td>
<td>3.113</td>
<td>3.113</td>
<td>0.19</td>
</tr>
<tr>
<td>Microphone mode</td>
<td>1</td>
<td>253.05</td>
<td>253.05</td>
<td>108.41</td>
</tr>
<tr>
<td>Flow velocity</td>
<td>3</td>
<td>864.92</td>
<td>288.31</td>
<td>108.41</td>
</tr>
<tr>
<td>Hearing aid*Microphone mode</td>
<td>1</td>
<td>61.30</td>
<td>61.30</td>
<td>26.04</td>
</tr>
<tr>
<td>Hearing aid*Flow velocity</td>
<td>3</td>
<td>318.51</td>
<td>106.17</td>
<td>32.95</td>
</tr>
<tr>
<td>Microphone Mode*Flow velocity</td>
<td>3</td>
<td>273.94</td>
<td>91.31</td>
<td>69.43</td>
</tr>
<tr>
<td>Hearing aid<em>Microphone mode</em>Flow velocity</td>
<td>3</td>
<td>12.37</td>
<td>4.12</td>
<td>3.02</td>
</tr>
<tr>
<td>500 Hz</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hearing aids</td>
<td>1</td>
<td>0.02</td>
<td>0.02</td>
<td>0.00</td>
</tr>
<tr>
<td>Microphone mode</td>
<td>1</td>
<td>405.03</td>
<td>405.03</td>
<td>93.23</td>
</tr>
<tr>
<td>Flow velocity</td>
<td>3</td>
<td>2147.30</td>
<td>715.77</td>
<td>160.69</td>
</tr>
<tr>
<td>Hearing aid*Microphone mode</td>
<td>1</td>
<td>79.37</td>
<td>79.37</td>
<td>21.19</td>
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<tr>
<td>Hearing aid*Flow velocity</td>
<td>3</td>
<td>770.83</td>
<td>256.94</td>
<td>37.14</td>
</tr>
<tr>
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<td>323.44</td>
<td>107.81</td>
<td>39.50</td>
</tr>
<tr>
<td>Hearing aid<em>Microphone mode</em>Flow velocity</td>
<td>3</td>
<td>69.57</td>
<td>23.19</td>
<td>10.76</td>
</tr>
</tbody>
</table>

Figure 9. The differences between the flow noise levels measured on Day 1 and Day 14 for BTE1. Dotted lines indicate the head angles with high angle-to-angle level variations.
to constantly monitor and estimate the hearing aid outputs with and without activating the modulation-based noise reduction algorithm in wind. The hearing aids would then automatically activate the noise reduction algorithm if lower outputs are predicted in the NRon condition. This approach can also be coupled with the automatic directional and omnidirectional microphone switching algorithm to search for settings that would generate the lowest output in wind.

All measurements in this study were based on the assumptions that hearing aids had flat frequency responses and that the frequency responses of DIR and OMNI were matched. In practical use, frequency responses of hearing aids are shaped to compensate for users’ hearing loss. Additionally, some hearing aids would provide partial or no low-frequency equalization for DIR compared to OMNI, instead of full equalization used in this study. In such cases, relative flow noise levels between frequency channels and between microphone modes would be altered by the amounts of gain differences. Relative flow noise levels within a frequency channel across different head angles, however, would still be similar to the relative levels reported in this study.

Future studies are needed to examine how to better identify the presence of wind at the hearing aid input so that noise reduction algorithms can provide more consistent noise reduction in different microphone modes and across frequency regions. Future studies are also needed to explore other creative ways to reduce wind noise interference, increase listening comfort, and enhance speech intelligibility in wind.

Acknowledgements

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References