For hearing aid noise reduction, babble is not just babble

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Most modern hearing aids provide single-microphone noise reduction without specifying how they work. The current study investigates how noise reduction is applied to babble noise in current premium hearing aids. Coupler gain measurements were performed in an acoustic test chamber. The signals used were standardized test signals, as well as babble noises compiled with different numbers of speakers (2, 4, 6, 8, and 10 speakers). The output of the hearing aid was measured with the noise reduction off and the strongest setting available. The gain reduction was calculated as the difference between the two settings. The results showed that, for an unmodulated test signal, the noise reduction algorithms applied quite different amounts of gain reduction across frequency. For the babble noise, some of the algorithms reduced gain very little, even for the 10-person babble. Other algorithms applied a graduated response, i.e., most gain reduction for 10-person babble, and the least amount of noise reduction for 2-person babble. Along with previous studies, this study highlights the need to have a standardized benchmarking procedure to define not only how noise reduction works in hearing aids but also which listening situations in which the noise reduction is active.

INTRODUCTION

For many hearing aid users, listening to speech in noisy situations is an important goal. For this reason, most modern hearing aids have single-microphone noise reduction. The general aim of hearing aid noise reduction is to reduce background noise while preserving speech information and sound quality. This is usually done by detecting in which frequency regions noise is more intense than speech and reducing gain in these regions. While there is only limited evidence that noise reduction improves speech intelligibility in noise, there is evidence of other benefits including improved sound quality and listening comfort, reduced noise annoyance, as well as possible improvements in listening effort and cognitive load (see Brons *et al.*, 2013 for recent discussion).

Previous studies have found large differences in how noise reduction works in commercial hearing aids (Bentler and Chiou, 2006; Brons *et al.*, 2013; 2014; Hoetink *et al.*, 2009; Smeds *et al.*, 2010). Quantitatively, there are more than 10-dB differences in how gain reduction is applied in a given frequency region. These differences can be heard by normally-hearing and hearing-impaired listeners

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(Brons *et al.*, 2013; 2014). There are also differences in which signals activate the noise reduction (Hoetlink *et al.*, 2009) and the signal-to-noise ratio required to activate noise reduction (Smeds *et al.*, 2010; Brons *et al.*, 2013; 2014). This demonstrates that commercial implementations of noise reduction can behave differently in differerent listening situations.

This experiment is part of a larger study to investigate how different noise reduction algorithms in hearing aids are active in various listening situations. During preliminary measurements, it was observed that sometimes babble noise would result in a large gain reduction. For other measurements using the same hearing aids and a different babble signal, no gain reduction was applied by the same noise reduction setting. Given that listening situations with babble as background noise are highly relevant for hearing aid users, we decided to investigate this further. The purpose of this experiment is to investigate the effect of varying the number of babble speakers on how noise reduction is applied in current premium hearing aids.

METHOD

Five premium receiver-in-ear (RIE) hearing aids were programmed linearly to a mild, sloping hearing loss. Except for the noise reduction algorithm, all other advanced signal processing strategies were turned off. Recordings from the hearing aids were performed in an ear simulator in an anechoic test box. The amount of gain reduction applied by the noise reduction was calculated by comparing the output of the hearing aids with the noise reduction i) off, and ii) on with the strongest noise reduction setting available.

Hearing aids, programming, and verification

The hearing aids included were the latest premium hearing aids from five manufacturers, as of June 2015. They were all RIE (Receiver-In-the-Ear) form factor. The receiver was the lowest power level available for each aid. The hearing aids were programmed linearly with all other adaptive features off, including directional microphones and automatic program changes. If possible, expansion was switched off and the maximum power output was set to its maximum value. In the respective fitting softwares, occluded earpieces were selected. Each hearing aid was programmed with two listening programs: one with noise reduction off and the other with noise reduction on, with the strongest setting available.

Using the NAL-NL stand-alone software (v1.927), coupler targets were generated for the standardized N2 hearing loss (Bisgaard *et al.*, 2010), which is a mild, sloping hearing loss (Fig. 1). The targets were calculated using the NAL-NL2 rationale (Keidser *et al.*, 2011) specified for a 65-dB speech input. The audiological input variables used were: hearing thresholds measured using supra-aural headphones with default adult acoustic transforms. The user’s sex was unspecified. The fitting variables were: bilateral, behind-the-ear fitting with RIE tubing with an occluded earmold. The compressor was assumed to be 18-channel with an intermediate compressor speed.

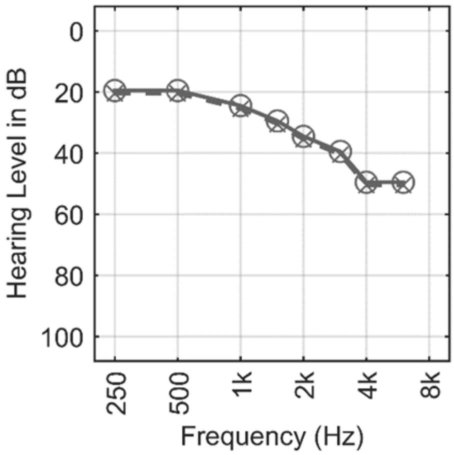

Fig. 1: The standardized N2 audiogram used to generate coupler targets for the hearing aids (Bisgaard *et al.*, 2010).

Fitting verification was performed in a 2-cc coupler in an Aurical HIT box with OTOsuite software (v. 4.75.00). For each of the hearing aids, coupler gain was matched to the NAL-NL2 targets within ±3 dB between 500 to 4000 Hz using the International Speech Test Signal (ISTS, Holube *et al.*, 2010). To check linearity, the ISTS was presented between 50 to 75 dB SPL at 5-dB intervals. In addition, it was verified that the coupler gain for the two hearing aid programs (noise reduction on and off) for each hearing aid were equal using the ISTS at 65 dB SPL.

Equipment

Recordings from the hearing aids were performed in an anechoic test box (Brüel and Kjær type 4232) using a PC with a Fireface UFX sound card. The recording software was Adobe Audition 3.0 (build 7283). The hearing aids were coupled to an IEC 60318-4 ear simulator (Brüel and Kjær type 4157) using a type DS 0540 earmold holder and sealed with adhesive gum. The recordings were performed with no earmoulds to optimise the seal. The microphones and amplifiers were Brüel and Kjær type 4192 measurement and reference microphone, type 2669 pre-amplifiers, and type 2692-C Nexus charge amplifier for very high input.

Signals

The three different types of signal are listed below. All signals were 60-seconds long and presented at 67 dB SPL. The frequency and modulation spectrum of the signals are plotted in Fig. 2 below.

1. The ISTS (Holube *et al.*, 2010) was included.
2. Wave files with babble noises consisting of varying amounts of speakers (2, 4, 6, 8, and 10-speakers) were created based on the ISTS. To do this, the pauses in the ISTS utterances were found. Then nine new sound tracks were

created, each with a start point at a randomly assigned pause and then looped back to the start. Then the new tracks were superimposed on the original ISTS to create wave files with the required number of babble speakers. Finally, the overall level of the wave files was adjusted to match the RMS level of the original ISTS wave file re: max.

3. An unmodulated speech-shaped noise was created by spectrally shaping the ANSI speech noise (ANSI S3.42, 1992) to match a real female speech signal (Cox *et al.*, 1987), which resembles the ISTS. The shaping was performed in 1/6-octave bands using an FIR filter with 2048 taps at sample rate 44.1 kHz.

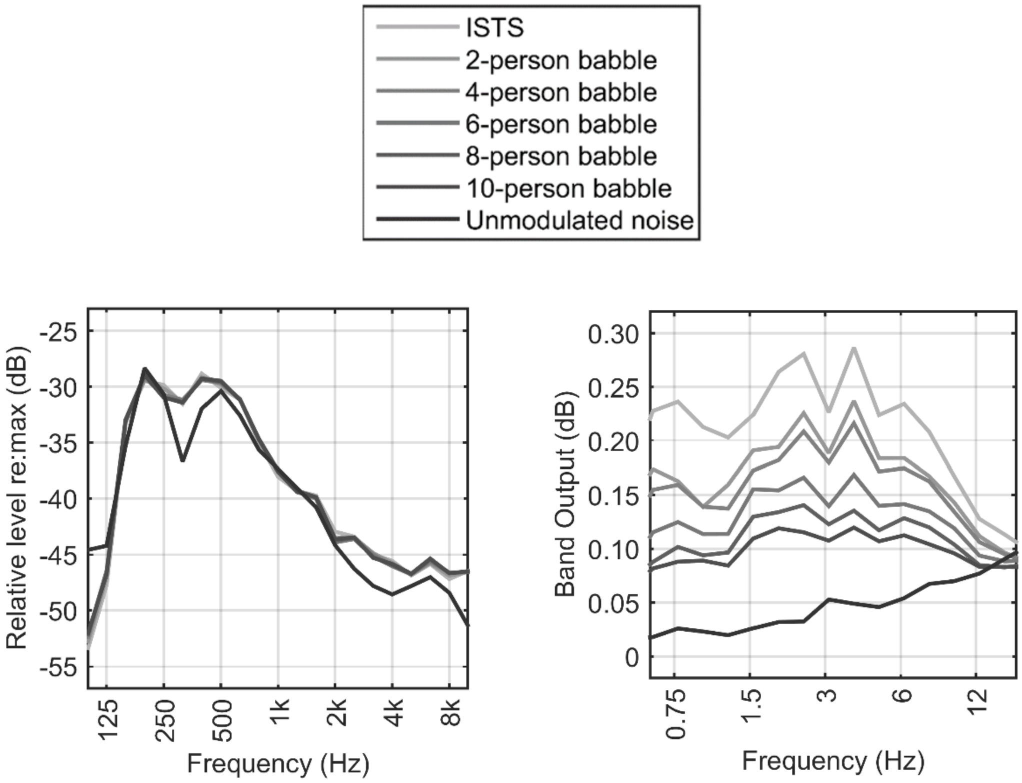

Fig. 2: The long-term average frequency (left panel) and modulation spectrum (right panel) of the signals.

Procedures

For each combination of signal, hearing aid, and program (noise reduction on and off), recordings of the hearing aid output were performed in the ear simulator. For each combination, the output was calculated in 1/3-octave bands between 250 to 8000 Hz using a 125-ms analysis window. Gain reduction was calculated as the difference in the output for the two programs, after a 35-second pre-conditioning time.

RESULTS

Recordings were made for each of the five hearing aids with noise reduction i) off, and ii) on, with the strongest setting of noise reduction. The difference in output for the noise reduction switched on and off were used to calculate the long-term average gain reduction due to the noise reduction (Fig. 3). The results presented in this manuscript were carefully cross-verified using similar signals on different software and hardware (the noise reduction measurement functionality of the OTOsuite software with the Aurical HIT box).

As expected, none of the noise reduction algorithms applied gain reduction to the ISTS signal. All the noise reduction systems applied the most gain reduction to the unmodulated noise. The maximum amount of gain reduction in a given frequency area varied from 7 dB to more than 15 dB.

Two of the hearing aids (A and D) applied a graduated response and applied most gain reduction to 10-person babble and least noise reduction to 2-person babble. The remaining three hearing aids seemed to apply the same graduated response, but only became active when the babble consisted of 8 or 10 speakers. (B, C, and E).

DISCUSSION

For the standardized test signals, none of the hearing aids reduced gain for the ISTS, suggesting that all noise reduction systems could appropriately detect speech, at least in quiet. All hearing aids applied the most gain reduction for the unmodulated noise. There were large differences in how much gain reduction was applied across frequency, with some only applying 7 dB in certain frequencies and others applying more than 15 dB gain reductions across the whole frequency spectrum. The range of differences is consistent with previous studies (Brons *et al.,* 2013; 2014; Hoetlink *et al.,* 2009; Smeds *et al.,* 2010).

For the babble noise signals, some of the hearing aids applied very little gain reduction (< 5 dB), even for a 10-person babble. Two of the hearing aids (A and D) applied a graduated approach and began to reduce gain for the 2-person babble and gradually increase the amount of noise reduction as the number of speakers increased. The other three hearing aids (B, C, and D) did not become active until there were 8 or 10 persons present in the babble.

The matter of how much gain reduction is appropriate has not been established. In addition, potential confounders include the level of the signal across frequency, and the hearing aid user's hearing thresholds, potential cognitive factors, as well as how the algorithm is implemented (Arehart *et al.*, 2015). Generally speaking if a noise reduction algorithm applies too much gain reduction then it may reduce audibility for speech. If too little gain reduction is applied, then the effect of noise reduction will not be audible to the user, and some of the positive benefits of noise reduction, such as improved listening comfort and reduced annoyance, will not be as optimal as they could be. There is little published evidence of how much gain reduction is appropriate, but Brons *et al.* (2013; 2014) have demonstrated that the differences in how noise reduction is applied can impact speech intelligibility, noise annoyance and listener preference.

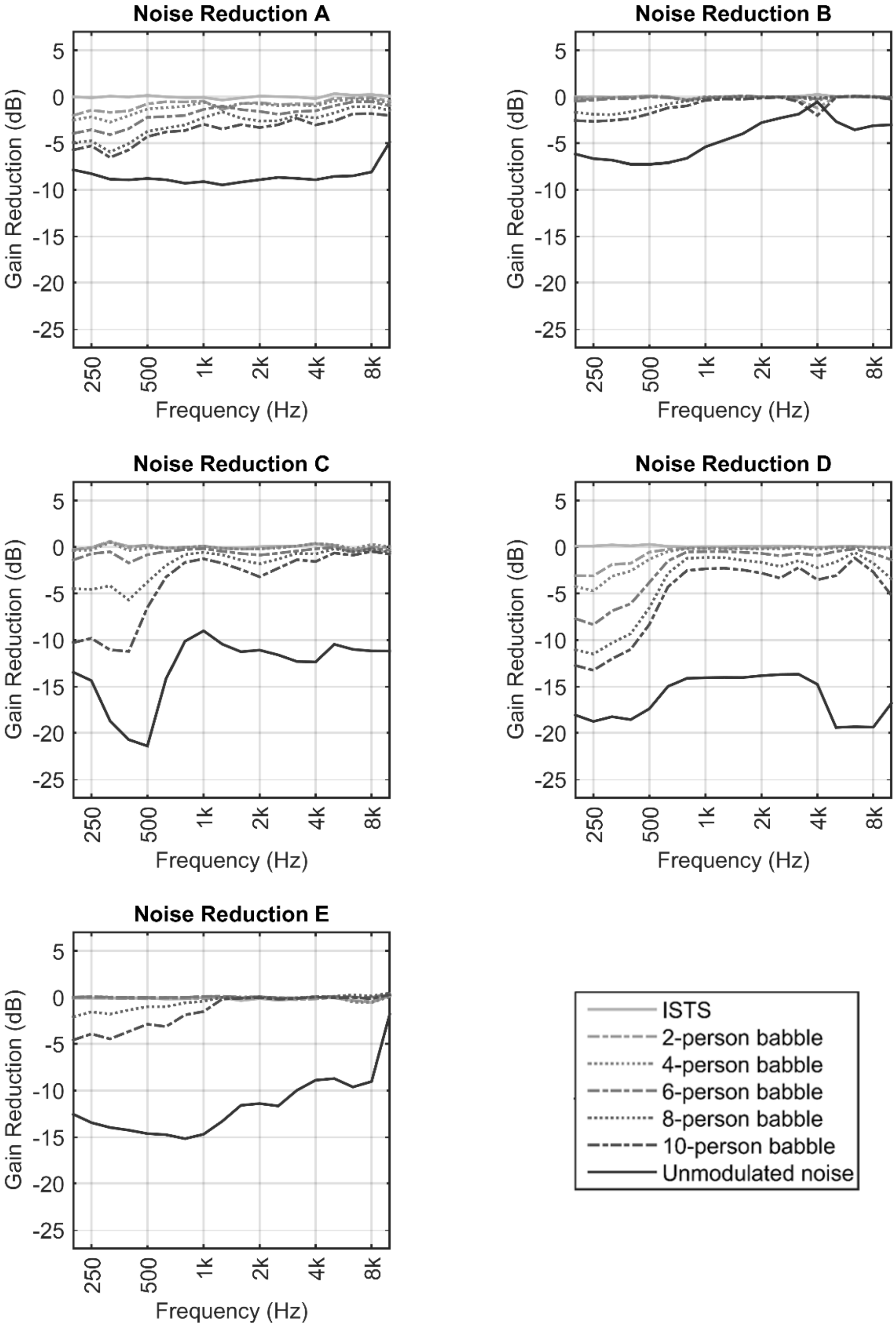

Fig. 3: Long-term average gain reduction in one-third octave bands for each of the seven test signals used. Every line represents the difference between noise reduction on and off averaged over 25 seconds after a 35-second pre-conditioning time.

This study demonstrates that there are differences among hearing aids in terms of which listening situations that the noise reduction activates. The range of listening situations explored by this experiment was quite broad, varying from a small group (two speakers) to a fairly large group (ten people speaking at the same time). There were considerable differences in how active the noise reduction systems were in these situations. Listening in groups is an important need for hearing aid users (Kochkin, 2010), yet a noise reduction algorithm may not be active in the situations that the user or audiologist expect it to be.

Other authors have raised the need for standardized measurements to describe how these noise reduction systems work (Bentler and Chiou, 2006; Brons *et al.*, 2013; 2014; Hoetink *et al.*, 2009; Smeds *et al.*, 2010). We suggest that such a test battery should include a range of realistic, yet well-described test signals. This would help the audiologists discern when the noise reduction systems are active in order to help them select an appropriate hearing aid for the listening needs of the user, and possibly to help fine-tune the hearing aid. If babble noise is included in a test battery, it is important to specify how it is compiled.

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