Auditory-model based assessment of the effects of hearing loss and hearing-aid compression on spectral and temporal resolution

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Most state-of-the-art hearing aids apply multi-channel dynamic-range compression (DRC). Such designs have the potential to emulate, at least to some degree, the processing that takes place in the healthy auditory system. One way to assess hearing-aid performance is to measure speech intelligibility. However, due to the complexity of speech and its robustness to spectral and temporal alterations, the effects of DRC on speech perception have been mixed and controversial. The goal of the present study was to obtain a clearer understanding of the interplay between hearing loss and DRC by means of auditory modeling. Inspired by the work of Edwards (2002), we studied the effects of DRC on a set of relatively basic outcome measures, such as forward masking functions (Glasberg and Moore, 1987) and spectral masking patterns (Moore *et al.*, 1998), obtained at several masker levels and frequencies. Outcomes were simulated using the auditory processing model of Jepsen *et al.* (2008) with the front end modified to include effects of hearing impairment and DRC. The results were compared to experimental data from normal-hearing and hearing-impaired listeners.

INTRODUCTION

Many studies have investigated whether amplification with multi-channel compression can be beneficial for speech intelligibility compared to linear amplification. While some studies have reported that multi-channel compression provides an advantage (Moore *et al.*, 1999; Souza and Bishop, 1999; Souza and Turner, 1999), others have reported no benefit or even a detrimental effect relative to linear gain (Franck *et al.*, 1999; Stone *et al.*, 1999). Thus, the benefit of multi-channel compression for improving speech intelligibility remains unclear.

Edwards (2002) suggested using a set of relatively simple outcome measures, based on narrowband signals, for the evaluation of hearing-aid signal processing. This also allows the use of computational models of the auditory system, which have recently been proven to be able to account for the effects of sensorineural hearing loss on

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auditory signal detection and discrimination (Jepsen and Dau, 2011; Panda *et al.*, 2014).

The purpose of the current study was to follow this approach in a systematic manner. Two psychoacoustic experiments were chosen, one to evaluate temporal resolution and the other to evaluate spectral resolution. Both normal-hearing (NH) and hearing-impaired (HI) subjects were tested to evaluate how hearing loss affects these outcome measures. In addition to behavioral experiments, the results were also simulated in a computational model of the auditory system that can account for detection and masking data from NH (Jepsen *et al.*, 2008) and HI (Jepsen and Dau, 2011) listeners. The modeling framework allowed the evaluation of multi-channel dynamic range compression without tedious retesting. To simulate an aided-impaired auditory system, a preprocessing stage was added to the model.

METHODS

Subjects

Three NH subjects, aged 24-29 and three HI subjects, aged 69-74, were tested. The hearing loss of all HI subjects was mild to moderate and sensorineural in nature.

Stimuli and procedure

The decay of forward masking (Glasberg *et al.*, 1987) and spectral masking patterns (Moore *et al.*, 1998) were measured in both subject groups in the unaided condition.

In the first experiment, the masker was a band of noise, centered either at 1 or 4 kHz with a bandwidth of 500 and 2000 Hz, respectively. Its sound pressure level (SPL) was fixed at 75 dB and its duration was 220 ms including a 10-ms rise and a 5-ms fall raised-cosine ramp. The probe was a short, 20-ms long pure tone gated for its entire duration (no steady state) at the masker central frequency. Different time intervals between the offset of the masker and the onset of the probe (zero voltage points) were tested. Positive offset values therefore relate to a purely forward masking condition, whereas the negative values mean that there was either a full or partial temporal overlap of the probe and the masker (simultaneous masking).

In the second experiment, the masker was an 80-Hz-wide band of noise centered either at 1 or 4 kHz. Its level was fixed at 75 dB SPL and its duration was 220 ms including 10-ms rise and fall raised-cosine ramps. The probes were pure tones at various frequencies, gated simultaneously with the masker.

In both experiments, the level of the probe at each frequency was varied adaptively using a 3AFC 1-up-2-down paradigm, until the masked threshold was reached. Absolute thresholds (in the absence of any masker) were also measured.

Additionally, for the HI subjects, temporal masking curves (TMCs; Nelson *et al.*, 2001) were measured in order to obtain basilar membrane input/output functions (BMIO). The probe was a 16-ms long pure tone gated for its entire duration (no steady state portion) at the frequency of interest (1 or 4 kHz) with a fixed level of 8 dB sensation level (SL), based on a prior measurement. For each probe, the masker was a 220 ms-long (gated with 8-ms rise and fall raised-cosine ramps) pure tone at the probe frequency (on-frequency condition) or one octave below (off-frequency condition). The masker-signal temporal gap and the masker level were varied adaptively in two dimensions using the Grid method (Fereczkowski *et al.*, 2015). Plotting the measured off-frequency versus the on-frequency TMC, a behavioural estimate of the BMIO at the probe frequency was obtained.

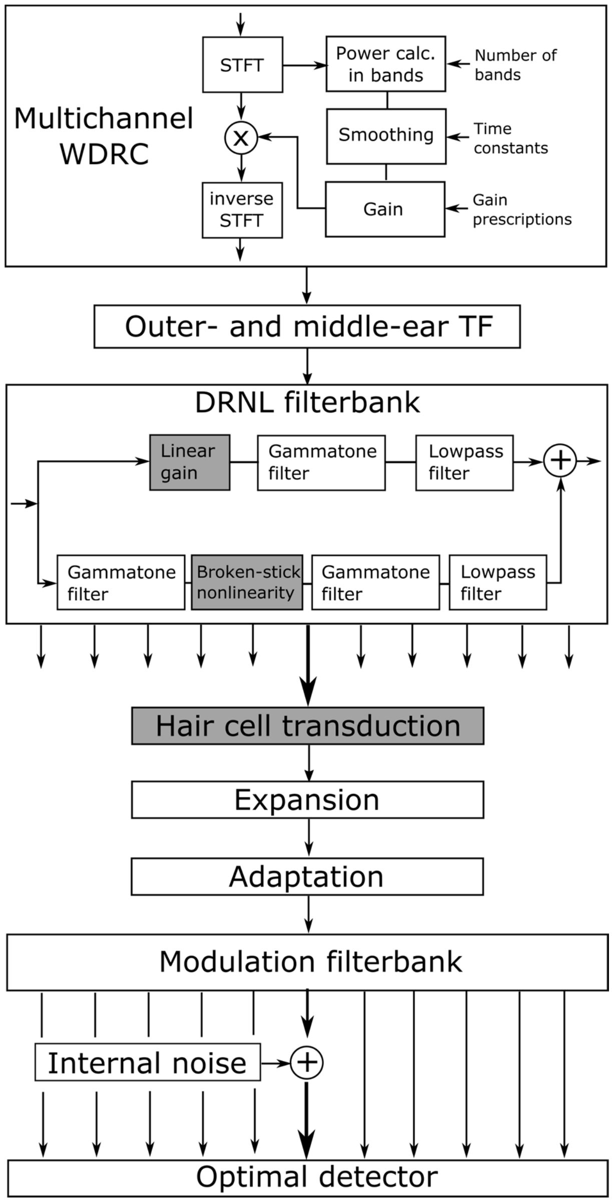

Fig. 1: Structure of the CASP model (Jepsen and Dau, 2011) with the preprocessing stage simulating a hearing-aid multi-channel wide dynamic range compression system.

Model of the auditory system

Simulations of the first two experiments were performed in the unaided condition for the NH and the HI listeners and in the aided condition for the HI only. The CASP model (Jepsen *et al.*, 2008) was used. The model structure is shown in Fig. 1. To simulate the effects of individual hearing loss, changes were made in the dual-resonance nonlinear (DRNL) filterbank (Lopez-Poveda and Meddis, 2001) and the inner hair cell (IHC) stage, based on the TMC data (Jepsen and Dau, 2011). First, the parameters of the DRNL broken-stick nonlinearity and the linear path gain at 1 and 4 kHz were adjusted to best fit the measured BMIO. Then a linear interpolation of the parameters across frequencies was performed. The IHC loss was estimated as the difference between the total loss (from the audiogram) and the OHC loss (inferred from the fitted DRNL input-output function).

Hearing-aid simulator

A multi-channel DRC hearing-aid simulator was developed for this study (Fig. 1). In the simulator, the input signal is broken down into time-frequency units using the short time Fourier transform (STFT). Then, for each time slice, the fast Fourier transform (FFT) bins are assigned to bands (channels). The spacing was quasi-logarithmic, based on the equivalent-rectangular-bandwidth scale. In this study, the number of channels was set to 19. In each frequency band, the power was calculated. Then the power was smoothed by a one-pole filter using the desired time constants. Attack and release times were set to 1 and 10 ms, respectively (RC time constants). Based on the smoothed power estimate, a gain matrix was calculated. The amount of gain in each channel depends on the prescription. Here, the NAL-NL1 targets were used as a starting point. The band-specific values were projected back onto the original FFT bins. The resulting gain matrix was then multiplied by the STFT representation of the input signal and STFT synthesis was performed to obtain the output time signal.

RESULTS

Figure 2 shows the masked thresholds for the decay of forward masking. Figure 3 shows the masked thresholds for the simultaneous spectral masking patterns.

In both figures, the left column shows results for the masker centered at 1 kHz while the right column indicates results for the 4-kHz case. The average data and model simulations for the NH listeners are shown in the top panel. The consecutive three panels show individual unaided data with both unaided and aided model simulations.

The absolute thresholds (in absence of the masker) are not shown in the figures. Due to hearing loss, for all of the HI subjects, these thresholds are elevated. Introducing hearing-aid amplification leads to a significant decrease in these thresholds.

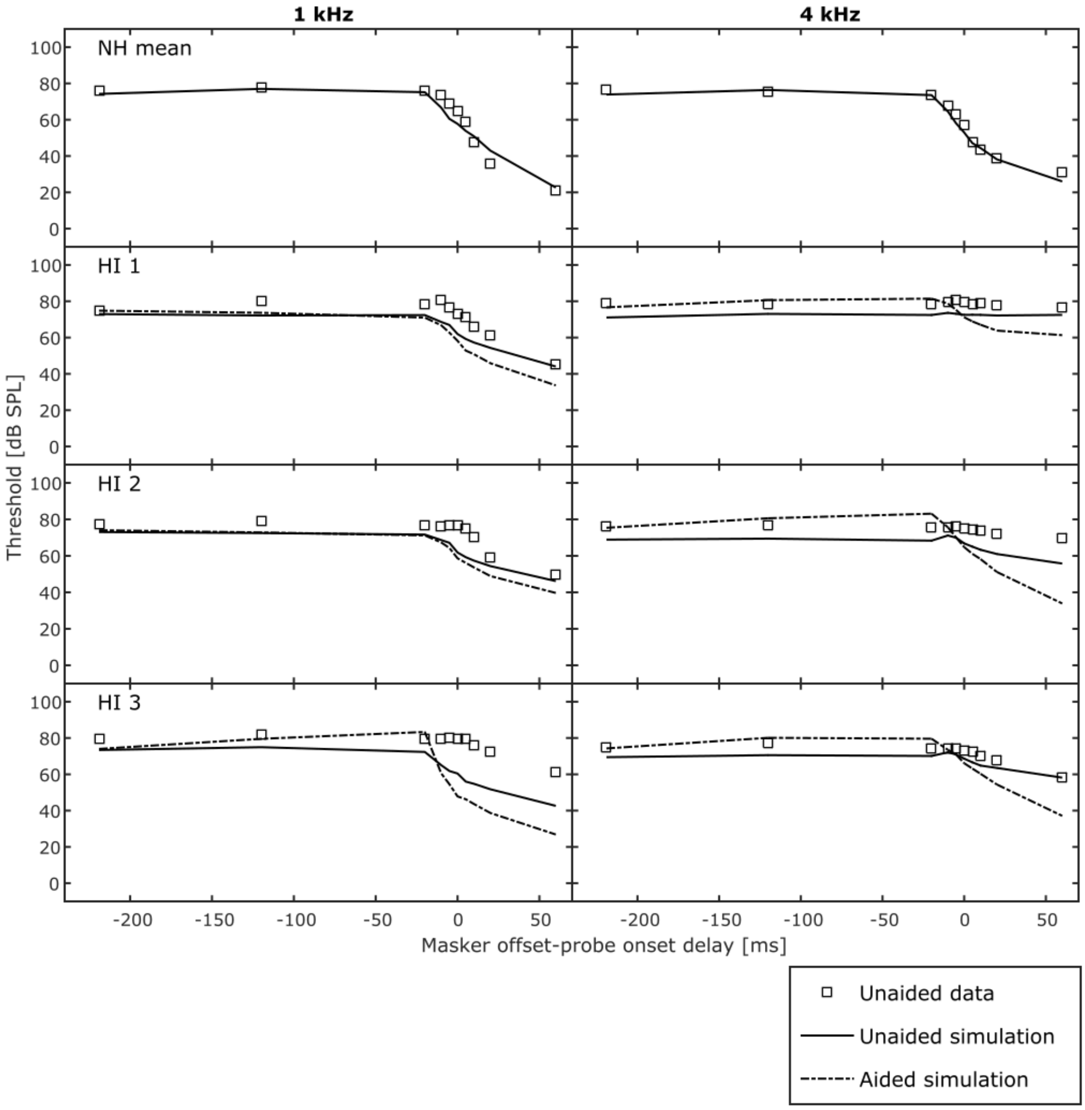

Fig. 2: Masked thresholds in the decay of forward masking experiments. Left and right panels show results for probe frequency of 1 and 4 kHz respectively. The top panel shows average unaided data and model simulations for the NH subjects. The consecutive panels show the unaided data with model simulations and aided simulations individually for each of the HI subjects.

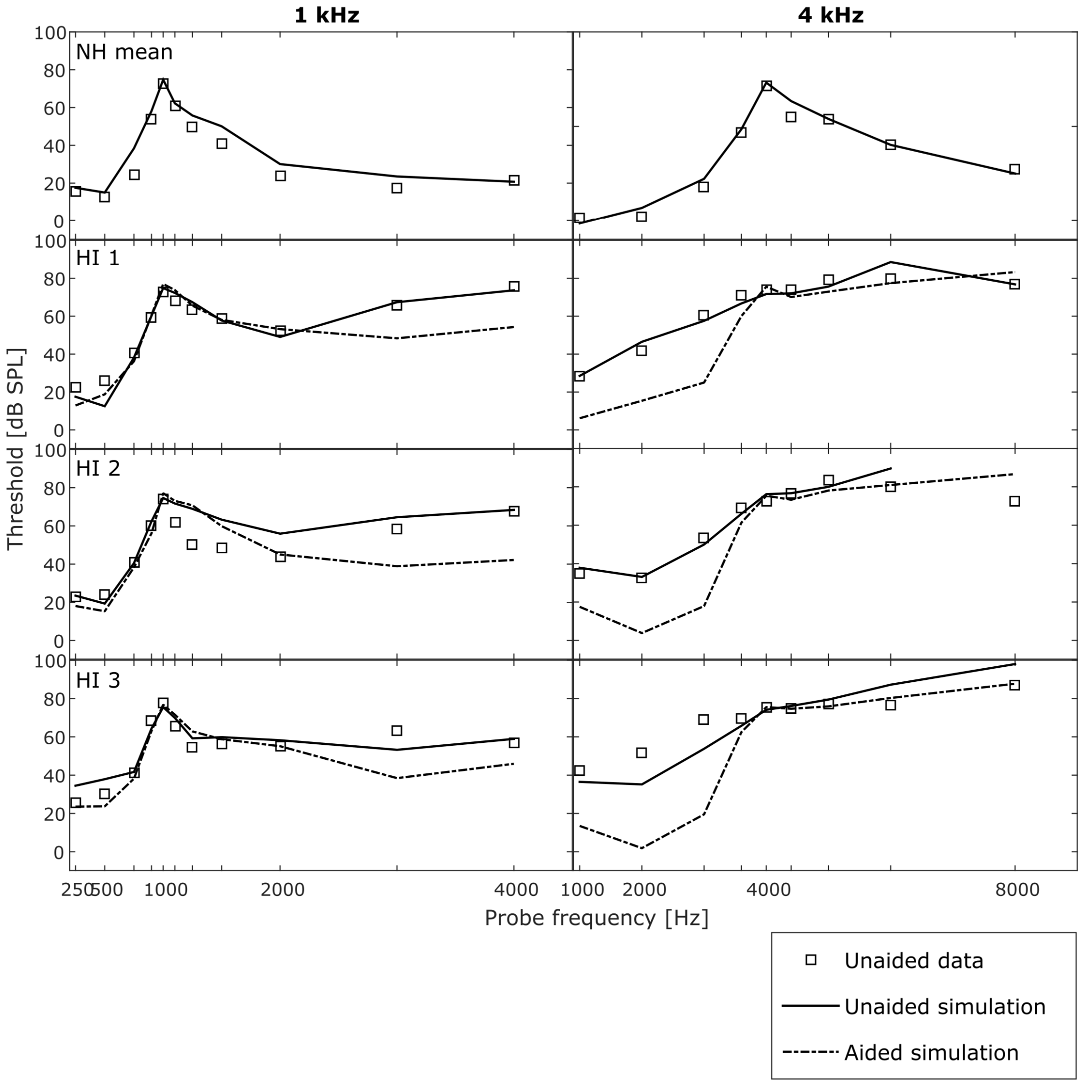

Fig. 3: Masked thresholds in the simultaneous spectral masking experiments. Left and right panels show results for probe frequency of 1 and 4 kHz respectively. The top panel shows average unaided data and model simulations for the normal hearing listeners. The consecutive panels show the unaided data with model simulations and aided simulations individually for each of the hearing-impaired listeners.

DISCUSSION

Decay of forward masking

Relative to the NH data, the HI listeners show elevated absolute thresholds (for detection of the short probe in silence) and a slower rate of recovery from forward masking. There is no elevation of the masked threshold in the full-overlap region (−219 to −20 ms). At 1 kHz, the audibility of the probe does not seem to be the only factor limiting the decay. However, at 4 kHz for HI1 and HI2, the curves are approximately at the level of the absolute threshold (not shown) regardless of the masker-probe separation. Simulated and measured decay curves have a similar dynamic range (except for HI3 at 1 kHz) but the model generally performs better, which is also the case for several subjects in Jepsen and Dau (2011). Introducing compression results in faster rates of recovery.

Spectral masking patterns

At 1 kHz, the lower skirt of the masking pattern does not differ significantly from the normal shape. However, the upper skirt is elevated. Only a part of this can be attributed to increased absolute thresholds. The rest indicates an increase in the upward spread of masking, related to broadening of auditory filters. In most cases, the model predicts the unaided masked threshold data. The thresholds are overestimated for probes above the masker frequency for subject HI2 at 1 kHz and subject HI3 at 4 kHz and underestimated below the masker frequency for HI3 at 4 kHz. Introducing multi-channel compression with the 1-kHz-centered masker leads to an improved performance only for the probe frequencies above 2 kHz. With the 4-kHz masker, the most significant improvement (decrease) in masked thresholds occurs for probe frequencies below the masker central frequency. There is also a slight reduction in the masking effect at probe frequencies above 4 kHz. In all cases, the absolute thresholds are significantly lower. Therefore, if the amount of masking (the difference between the masked and absolute thresholds) was plotted, the multi-channel compression would appear to restore the masking patterns to a more normal shape. However, this restoration would be due to the increased audibility and could be achieved using only linear gain.

CONCLUSIONS

Sensorineural hearing loss results in a decreased audibility of pure tones, slower rate of decay of forward masking and flattened spectral masking patterns, consistent with earlier studies. Multi-channel compression appears capable of restoring, to some extent the performance of HI subjects in the above-mentioned tasks back to normal. More research is needed to disentangle the effects of linear (increased audibility) and level-dependent gain.

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REFERENCES

Edwards, B. (**2002**). "Signal processing, hearing aid design, and the psychoacoustic Turing test," Proc. International Conference on Acoustics, Speech, and Signal Processing, 2002, Orlando, Florida.

Fereczkowski, M., Kowalewski, B., Dau, T., and MacDonald, E.N. (**2015**). "Time-efficient multidimensional threshold tracking method," J. Acoust. Soc. Am., **137**, 2228-2228.

Franck B.A., van Kreveld-Bos C.S., Dreschler, W.A., and Verschuure, H. (**1999**). "Evaluation of spectral enhancement in hearing aids, combined with phonemic compression," J. Acoust. Soc. Am., **106**, 1452-1464.

Glasberg, B.R., Moore, B.C.J., and Bacon, S. (**1987**). "Gap detection and masking in hearing-impaired and normal-hearing subjects," J. Acoust. Soc. Am., **81**, 1546-1556.

Jepsen, M.L., Ewert, S., and Dau. T. (**2008**). "A computational model of human auditory signal processing and perception," J. Acoust. Soc. Am., **124**, 422-438.

Jepsen, M.L. and Dau, T. (**2011**). "Characterizing auditory processing and perception in individual listeners with sensorineural hearing loss," J. Acoust. Soc. Am., **129**, 262-281.

Lopez-Poveda, E.A. and Meddis, R. (**2001**). "A human nonlinear cochlear filterbank," J. Acoust. Soc. Am., **110**, 3107-3118.

Moore, B.C.J., Alcántara, J. I., and Dau. T. (**1998**). "Masking patterns for sinusoidal and narrow-band noise maskers," J. Acoust. Soc. Am., **104**, 1023-1038.

Moore, B.C.J., Peters, R.W., and Stone, M.A. (**1999**). "Benefits of linear amplify-cation and multichannel compression for speech comprehension in backgrounds with spectral and temporal dips," J. Acoust. Soc. Am., **105**, 400-411.

Nelson, D.A., Schroder, A.C., and Wojtczak, M. (**2001**). "A new procedure for measuring peripheral compression in normal-hearing and hearing-impaired listeners," J. Acoust. Soc. Am., **110**, 2045-2064.

Panda, M.R., Lecluyse, W., Tan, C.M., Jürgens, T., and Meddis, R. (**2014**). "Hearing dummies: Individualized computer models of hearing impairment," Int. J. Audiol., **53**, 699-709.

Souza, P.E. and Bishop, R.D. (**1999**). "Improving speech audibility with wide dynamic range compression in listeners with severe sensorineural loss," Ear Hearing, **20**, 461-470.

Souza, P.E. and Turner, C.W. (**1999**). "Quantifying the contribution of audibility to recognition of compression-amplified speech," Ear Hearing, **20**, 12-20.

Stone, M.A., Moore, B.C.J., Alcántara, J.I., and Glasberg, B.R. (**1999**). "Comparison of different forms of compression using wearable digital hearing aids," J. Acoust. Soc. Am., **106**, 3603-3619.