Assessing the effects of hearing-aid compression on auditory spectral and temporal resolution using an auditory modeling framework

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Abstract: Sensorineural hearing loss results both in a reduced sensitivity to sound, as well as suprathreshold deficits, such as loudness recruitment and degraded spectral and temporal resolution. To compensate for loudness recruitment, most hearing aids apply level-dependent amplification, such as multi-channel wide dynamic-range compression. However, the most appropriate choice of parameters, such as the time constants and the number of channels, has been controversial. Speech intelligibility has been often considered as an outcome measure and it has been difficult to delineate the effects of hearing-aid signal processing on the representation of signals, due to the complex spectro-temporal structure of speech. In the current study, hearing-aid compensation strategies were evaluated using synthetic stimuli in psychoacoustic experiments with hearing-impaired and normal-hearing listeners. A computational model of the auditory signal processing was used to assess the effects of linear amplification and multi-channel fast-acting compression on spectral and temporal masking. Improvements in the decay of forward masking were predicted with both types of amplification due to the increased audibility. On the other hand, spectral masking was reduced with compression, but not linear amplification, due to the increased signal-to-noise ratio across frequency. The results provide insights into the effects of hearing-aid amplification on basic auditory processing.

Keywords: Psychoacoustics, Hearing aids, Hearing loss

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1. INTRODUCTION

Loss of sensitivity and loudness recruitment are among the primary consequences of a sensorineural hearing loss (SNHL). Reduced sensitivity can be compensated for in a hearing aid by applying a linear (level-independent) gain. However, to avoid an excessive amplification of intense sounds, level-dependent amplification schemes are commonly employed, such as wide dynamic range compression (WDRC). Such systems usually operate in several independent frequency channels [1]. This allows for a reduction of across-frequency interactions [2] and may provide a better fit to an individual frequency-dependent hearing loss [1,3]. However, using many compression channels also results in a reduced spectral contrast and, in turn, may negatively affect speech recognition performance [4,5]. The speed of gain reduction and recovery in a compressor are governed, respectively, by its attack and release time constants. Attack times ranging from 1 to 10 ms have been reported [6,7]. Such short attack times allow the hearing aid to quickly react to sudden increases in the signal level. The appropriate values for the release time are less well-defined. A compressor is usually considered as fast-acting if the release time is shorter than 200 ms, and as slow-acting otherwise [1]. A short release time allows the gain to more closely follow rapid fluctuations of the input signal. As a result, more gain is applied to short low-intensity segments, which can improve audibility of speech components occurring in the dips of the background noise. On the other hand, the same mechanism leads to a smoothing of the temporal envelope of the signals, which can affect speech recognition in hearing-impaired (HI) listeners [7,8].

Studies on the effects of hearing-aid compression (vs. linear amplification) on the perception of speech have provided mixed results. Some studies have shown that compression improves speech recognition performance to a greater extent than linear amplification [9–12]. Other studies have shown detrimental effects of compression on, for example, vowel identification [5] and speech-in-noise
recognition performance in HI listeners [13]. These divergent sets of results may partly be attributed to the complex spectro-temporal structure of speech and variety of cues involved in its recognition. The cues that are used for speech recognition may be affected by the degree of hearing loss [e.g. 8], and depend on stimulus parameters such as the signal level, the type of background interference, the signal-to-noise ratio (SNR) etc.

Alternatively, it has been proposed that the effectiveness of hearing-aid signal processing could be evaluated using synthetic, non-speech stimuli in a series of basic psychoacoustic tasks related to spectral and temporal processing in the auditory system [14]. Such an approach could make the interplay between a given hearing loss and a specific hearing-aid processing more transparent. Fewer auditory cues would be involved than in the case of speech. A disadvantage of this approach is that the outcome measures involved may not be directly related to the challenges experienced in real-world communication situations.

Only a few studies have considered this approach. Strelczyk et al. [15] investigated the effects of compression channel bandwidth on differential loudness growth and loudness summation, in an attempt to restore normal loudness perception in listeners with sensorineural hearing loss. A fairly good restoration of loudness perception was possible using a system with level-dependent channel bandwidth. However, the results of a follow-up study [16] indicated that the channel bandwidth only had a minimal effect on vowel identification. Brennan et al. [17] found that hearing-aid WDRC, compared to linear amplification, reduced modulation detection thresholds whereas gap detection thresholds did not depend on the processing scheme. The same authors studied the effect of compression with different configurations of time constants on the recovery from forward masking in HI listeners [6]. The improvement in forward-masking thresholds was found to be largest using very short time constants (1 ms attack, 5 ms release) where the gain for the masker was reduced and quickly recovered to higher values for the following low-level probe, increasing its level and improving the output SNR. However, in that study, even the slowest compressor (with an attack time constant of 150 ms and a release time constant of 1500 ms) reduced the amount of forward-masking, compared to the unaided condition, since the compressor still provided a substantial amount of gain. The benefit of slow-acting compression observed by the authors may thus have resulted from the increased overall gain, which could also be achieved by linear amplification. Unfortunately, a linear amplification condition was not included as a reference condition in their study.

Another advantage of considering basic psychoacoustic measures is that auditory modeling frameworks have been developed in the past and evaluated in such conditions. Such modeling work has been quite successful in predicting spectral and temporal masking effects in the normal [e.g. 18] and impaired auditory system [e.g. 19,20] whereas focus has been put on aided hearing. Only one study [21] simulated “hearing profiles” of listeners with normal and impaired hearing and investigated the effects of hearing-aid signal processing on measures of spectral and temporal acuity using a computational auditory model. A physiologically-inspired hearing-aid compression scheme was shown to significantly improve the profiles. However, no attempt has been made to verify those predictions in aided hearing-impaired listeners, nor to evaluate the effects of such processing on speech recognition performance or other ecologically relevant outcome measures.

In the present study, an attempt was made to measure and predict using a computational modeling framework the unaided performance of (average) normal-hearing (NH) and the unaided and aided individual hearing-impaired listeners in tasks related to spectral and temporal resolution. For the model predictions, the computational auditory signal processing and perception (CASP) framework [18] was used. The aided performance of the individual HI listeners was predicted using a simulated hearing-aid processor embedded in the same modeling framework. The predictions were verified in corresponding listening experiments using the same hearing-aid processing.

2. METHODS

2.1. Experimental Paradigms and Measurements

2.1.1. Listeners

Three NH and ten HI listeners took part in the current study. The NH group consisted of younger listeners, aged between 24 and 29 years, with hearing level (HL) below 20 dB in the 250–8,000 Hz range. Nine of the HI listeners were older (aged 66–76 years), 7 males and 2 females, and one of them was a younger male (aged 27 years). Their hearing loss was sensorineural, based on an air-bone gap of 10 dB or less. The degree of hearing loss varied from mild to moderately severe. The average HL was 18, 24, 29, 35, 56, and 59 at 250, 500, 1,000, 2,000, 4,000 and 8,000 Hz, respectively.

2.1.2. Stimuli

In the forward masking experiment [18,22], a broadband noise masker was followed by a tonal target. Masked thresholds were measured for a set of time delays (gaps) between the masker offset and the target onset (zero voltage points). The target was a 20-ms long sinusoidal burst, gated with a raised-cosine window over its entire duration. Two target frequencies (1 and 4 kHz) were considered. The noise masker was 220-ms long with a 10-ms long rising ramp and a 5-ms long falling ramp and cut-off frequencies of 100 and 8,000 Hz. Its overall sound
pressure level (SPL) was set to 85 dB in the unaided condition and 75 dB in the aided conditions, to avoid loudness discomfort due to the hearing-aid gain.

In the spectral masking experiment, threshold-levels of tonal probes at different frequencies were measured in a simultaneous masking paradigm [23]. The masker was an 80-Hz-wide band of noise, arithmetically centered at either 1 or 4 kHz, depending on the experimental condition. Its overall SPL was fixed at 75 dB. The probe frequencies tested ranged from 250 to 4.000 Hz for the 1-kHz-centered masker and from 1,000 to 8,000 Hz for the 4-kHz-centered masker. The probe level was varied adaptively. The probe and the masker were both 220-ms long and were gated simultaneously with 10-ms long raised-cosine rise/fall ramps.

2.1.3. Experimental setup and procedure

The experiments were conducted in a double-walled soundproof booth. The stimuli were played monaurally over Sennheiser HDa200 headphones, whose response was equalized digitally. The presentation level was calibrated using a 6-cc coupler. All signals were generated digitally on a PC running Matlab and delivered through an RME Fireface UCX soundcard with a 16-bit resolution that roughly mimics the resolution of a normal auditory system and translates to 24 rectangular carriers in the unaided and aided psychoacoustic experiments. The cochlear stage of the model was simulated by a dual resonance non-linear (DRNL) filterbank [30], which is capable of reproducing some of the cochlear nonlinearities. The parameters of the filterbank can be modified to account for individual hearing loss [19]. For each HI listener, the parameters of the DRNL filterbank and the inner hair cell (IHC) stage were modified based on the audiogram and an estimate of the basilar-membrane input/output (BM I/O) function [19,30]. The behavioral estimates of the BM I/O were obtained using the temporal masking curve (TMC) paradigm [31], using a time-efficient threshold-tracking method [32]. The simulations of the experiments were performed using the same framework [24] as in the actual experiments and the decision stage in the model employed the same correlation-based “matched filter” as in the original CASP model [18].
3. RESULTS

3.1. Forward Masking

The results of the unaided forward masking experiment and the model predictions are shown in Fig. 1. The masker SPL used for the unaided experiment was 85 dB. The masked thresholds are shown as a function of the masker-probe temporal delay (gap). The results obtained using the 1 kHz and the 4 kHz probes are shown on the left and right, respectively. The top panels represent the mean NH results and the remaining panels show four examples (out of ten) of the individual HI results. Each listener’s threshold in quiet (for the brief tonal probe) are indicated with a black star. The black squares represent experimental data. The black line represents model predictions. The RMS error of the model prediction is provided in each panel.

Fig. 1  Unaided forward masking functions obtained using the 85 dB masker. Masked thresholds are plotted as a function of the temporal delay between the masker offset and the probe onset. Negative values on the abscissa indicate full or partial temporal overlap of the probe and the masker (simultaneous masking). The results obtained using the 1 kHz and the 4 kHz probes are shown on the left and right, respectively. The top panels represent the mean NH results and the remaining panels show four examples (out of ten) of the individual HI results. Each listener’s threshold in quiet (for the brief tonal probe) are indicated with a black star. The black squares represent experimental data. The black line represents model predictions. The RMS error of the model prediction is provided in each panel.

Fig. 2  Unaided and aided forward masking functions obtained using the 75 dB masker. Layout as in Fig. 1. Unaided NH and HI model predictions are plotted with the black line. The blue line indicates predictions with linear amplification, the red line – with compression. The red squares indicate experimental data with compression.

3. RESULTS

3.1. Forward Masking

The results of the unaided forward masking experiment and the model predictions are shown in Fig. 1. The masker SPL used for the unaided experiment was 85 dB. The masked thresholds are shown as a function of the masker-probe temporal delay (gap). The forward masking curves of the NH listeners spanned over a relatively large dynamic range. The average value of the SPL in the fully-simultaneous masking condition (at the gap of −20 ms) exceeded 75 dB. The masked threshold decreased with increasing the masker-probe temporal delay. At the highest tested delay of 60 ms, the average threshold SPL was around 20–25 dB, a value close to the average threshold in quiet. The rate of the forward-masking decay was similar at both frequencies. The masked thresholds predicted with the model were in reasonably good agreement with the experimental data. The root mean square (RMS) error of the predictions was equal to 5.1 and 10.3 dB, at 1 and 4 kHz, respectively.

The forward-masked thresholds (at delays higher than 0 ms) and the threshold in quiet were typically higher in the HI than in the NH listeners. As for the NH listeners, at 60 ms, the HI listeners’ forward-masked thresholds decayed to values close to their individual thresholds in quiet. For all HI listeners, the thresholds in quiet were higher at 4 kHz than at 1 kHz, which effectively limited the dynamic range and the apparent rate of decay.

Figure 2 shows the experimental data and the model predictions obtained with the masker at an SPL of 75 dB.
The trends present in the unaided model predictions were similar at 75 and 85 dB. However, the predicted simultaneous-masking thresholds were lower at the lower masker level, which further narrowed the dynamic range of the HI. At 4 kHz, the predicted curves were nearly flat. Introducing hearing-aid amplification led to a decrease in the predicted forward-masking thresholds. The masked thresholds obtained with compression were lower or equal to those obtained with linear amplification. The thresholds measured in the HI listeners with compression were, in general, lower than the corresponding unaided predictions, with several exceptions.

To further characterize the effects of hearing loss and hearing-aid amplification on forward masking, the decay of forward masking (DoFM) was quantified as the difference between the thresholds at −20 and 20 ms. The delay of 20 rather than 60 ms was chosen due to: 1) the proximity of the 60-ms forward-masked thresholds to the thresholds in quiet and 2) the change in the rate of the decay at about 20-ms observed in the NH and some of the HI data. The DoFM values for each probe frequency and amplification condition were calculated and are summarized as boxplots in Fig. 3. The statistical analysis\(^1\) indicated no significant differences between the unaided measured and predicted DoFM for the NH listeners at both frequencies and for the HI listeners at 4 kHz. At 1 kHz, the model underestimated the median HI DoFM by 9 dB \((p < 0.001)\). Across all conditions, the amount of DoFM was significantly greater in the NH than in HI listeners. The difference between NH and HI data at 1 kHz was about 10 dB \((p < 0.001)\), while the predicted difference was about 21 dB \((p < 0.001)\). At 4 kHz the measured and predicted differences were about 31 and 29 dB, respectively \((p < 0.001)\). The aided and unaided results obtained with the 75-dB masker are summarized in terms of the DoFM in Fig. 3 (bottom panel)\(^2\).

Introducing hearing-aid amplification resulted in a statistically significant increase in the predicted DoFM: by about 7 dB for linear \((p < 0.05)\) and 9 dB for compressive amplification \((p < 0.001)\) at 1 kHz, and by 6 dB for linear and 8 dB for compressive amplification (both \(p < 0.001)\) at 4 kHz. There was, however, no statistically significant difference between the mean predictions obtained with linear amplification and compression. At 1 kHz, the model underestimated the aided DoFM by 4 dB \((p < 0.001)\). At 4 kHz there were no statistically significant differences between the compression-aided DoFM predictions and data. Moreover, the aided DoFM measured in HI listeners was significantly smaller than the predicted NH DoFM (by 10 and 23 dB at 1 and 4 kHz, respectively, both \(p < 0.001)\).

### 3.2. Spectral Masking

The unaided NH and HI spectral masking patterns (SMPs) are shown in Fig. 4. The masked thresholds are shown as a function of the probe frequency. The layout is similar to the one used in Figs. 1 and 2. The NH SMPs were asymmetric with steeper low-frequency skirts and a distinct peak at the probe frequency corresponding to the center frequency of the masker. The NH model predictions

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\(1\)Bootstrap-based confidence intervals (CIs) for the difference in medians were used for comparisons between the experimental data and the corresponding model predictions (e.g. the NH data vs. the NH predictions) and across conditions (namely the NH data vs. the HI data, and HI predictions vs. the HI predictions). The compared groups were sampled with replacement and the difference in medians of the samples was computed 10,000 times. This way, an empirical distribution was computed, whose 2.5th and 97.5th percentiles were used as the upper and lower bounds of the CIs. The significance levels were Bonferroni-corrected for the number of comparisons and are indicated in the plot above the brackets. The whiskers shown extend to \(+/- 1.5\) times the interquartile range.

\(2\)The reported statistical significance was established using the same method as for the upper panel. All of the HI conditions were compared to each other. However, only the median compression-aided HI data were compared to the median NH predictions, to limit the number of comparisons.
were in reasonably good agreement with the data. The model overestimated the masked thresholds by up to 18 dB (for the 1500-Hz probe with the 1-kHz masker) with the RMS error of 9.8 and 7.8 dB at 1 and 4 kHz, respectively. At 1 kHz the HI SMPs retained the steep low-frequency skirt and on-frequency peak that is typical for NH patterns. However, excessive upward spread of masking (USoM) was evident from the increased thresholds at the high frequencies. The model predictions reflected the individual patterns relatively well. Except for HI3, the RMS error was lower than for the NH predictions. The model typically overestimated the individual HI thresholds for probe frequencies between 1,100 and 3,000 Hz. This effect was present in the NH predictions as well. At 4 kHz, both the low- and high-frequency skirts of the HI SMPs became elevated. For some listeners (e.g. HI3) the masked thresholds increased monotonically with the probe frequency up to 4 kHz, above which the pattern became flat. This appears to be mostly due to the high absolute thresholds in quiet (indicated with the dotted line). As was the case at 1 kHz, the patterns predicted with the 4 kHz masker were in good agreement with the experimental data. The NH and HI predictions showed a similar RMS error, except for HI3, in which case the model predictions tended to greatly overestimate the high-frequency skirts of the HI patterns.

Figure 5 shows four examples of the aided HI predictions and the aided HI data obtained with compression. The unaided model predictions (black line) are included for comparison. The predictions obtained with linear amplification and compression are shown with the blue and red line, respectively. The data obtained with compression are shown with red squares.
SRM in the unaided HI listeners was equal to 27 and −1 dB (negative SRM) at 1 and 4 kHz, respectively. In both cases, the difference between the NH and the unaided HI was statistically significant ($p < 0.001$).

Hearing-aid compression increased the median SRM in the HI listeners at 1 kHz to 42 dB and to 14 dB at 4 kHz. The difference between the median SRM measured at 1 and 4 kHz in the NH and in the aided HI was statistically significant ($p < 0.001$). The difference between the aided and unaided SRM based on the HI listeners’ experimental data was statistically significant at both 1 and 4 kHz (with both $p < 0.001$). The SRM based on the model predictions was shown in the bottom panel of Fig. 6. The information about the statistical significance of the differences between each model prediction and the corresponding experimental condition is indicated below the data points. There was no statistically significant difference between the SRM estimated from the NH model predictions and from the experimental data at 4 kHz. At 1 kHz, the model underestimated the NH SRM by 6 dB. The median SRM of the HI listeners was also underestimated by the model predictions by about 6 to 19 dB, depending on the condition (with all $p < 0.05$). Moreover, there were no statistically significant differences in the SRM between the unaided and the linear conditions. In the model predictions using the 1-kHz masker, compression provided significantly higher SRM than linear amplification (36 vs. 18 dB, $p < 0.001$) and there was no statistically significant difference between the compression-aided HI and the NH. Based on the model predictions at 4 kHz, there was a statistically significant difference in the median SRM between the NH and the compression-aided HI SRM ($p < 0.001$), but no statistically significant differences between the compression, linear and unaided conditions.

4. DISCUSSION

In the present study, the effects of individual hearing loss and hearing-aid amplification on measures of spectral and temporal resolution were investigated. Two psychoacoustic outcome measures were considered: forward masking and spectral masking patterns. Psychoacoustic experiments were carried out in NH, as well as unaided HI listeners. Hearing-aid compression and linear (level-independent) amplification were then evaluated in the framework of an auditory-processing model, in terms of potential improvements in masked thresholds (where the NH served as a baseline). Furthermore, the model predictions obtained with hearing-aid compression were validated in an aided experiment in the same group of HI listeners.

The NH listeners displayed similar rates of decay of forward masking as reported previously [e.g. 18,22,23]. As expected, the forward-masking thresholds of the HI listeners were elevated and the amount of DoFM was
reduced. Model predictions indicated that the introduction of hearing-aid amplification would lead to an increase in the HI DoFM, with no difference between the compression and linear conditions. The aided DoFM was still significantly smaller than normal.

The predicted increase in the DoFM of the aided HI is most likely due to the overall gain introduced by both amplification schemes, which increased the audibility of the stimulus. This is in line with the findings of Brennan [6], who found an improvement in forward-masking thresholds even with slow-acting compression. The advantage of fast-acting compression with a very short release time, such as the one used in the present study, is that the gain changes quickly enough to improve the output SNR. The linear processing amplifies the target and the masker by similar amount and therefore does not provide improvements in the SNR. In case of slow-acting compression, the release time constant is too large to provide different gain to the masker and the target, hence it effectively becomes similar to linear amplification.

The predicted lack of difference in the DoFM between fast-acting compression and linear amplification could be attributed to two factors. First, the CRs prescribed by the NAL-NL1 rule used in the present study are relatively low, especially in the high-frequency range [see, e.g. Fig. 2 in 33]. Low CRs pose a limitation to the amount of gain reduction applied to the masker, and therefore diminish the potential improvement of the output SNR. If a different rationale (with higher compression ratios) was considered, the benefit of compression should become apparent. Second, despite the very short nominal release time (10 ms), the effective release times were 6 to 7 times longer, as the power smoothing was done in the linear domain [26]. Another study [34] investigated the effects of compression speed on the recognition of consonants in a non-simultaneous masking paradigm. The NAL-NL2 prescription [33] was considered and thus the CRs were higher than in the present study. Compression with a short release time constant (10 ms) was shown to provide higher target output level and better SNR, leading to an improved recognition by HI listeners.

As expected, the NH SMPs were characterized by sharp peaks and relatively steep slopes. The median amount of the SRM (defined as the difference between the on-frequency threshold and the threshold for probe one octave above the central frequency of the masker) was about 40 dB. On the other hand, the SMPs of the HI listeners were much broader, and sometimes entirely flat, indicative of poor frequency selectivity. The values of the SRM were greatly reduced in the HI listeners, and even negative values were observed at 4 kHz. This can be attributed to the elevation of the listeners’ absolute thresholds, which was the greatest at high frequencies. Moreover, the model overestimated the upper skirts of the patterns in both the unaided and aided conditions. This is probably due to the limitations of the method of fitting of the individual model parameters. The method seems to underestimate the residual compression ratios at high frequencies [19] which leads to an excessive broadening of the DRNL filters in the high-frequency region. Linear amplification did not lead to any improvement in the predicted SRM. Only the addition of multi-channel compression improved the SRM to normal levels, and only at 1 kHz. At 4 kHz, the aided and unaided SRM were still similar, regardless of the compensation strategy. The benefit of compression at 1 kHz can be explained by the fact that, in a multi-channel system, the gain applied to the tonal target with relatively low input level is not affected by the high-intensity masker localized in another frequency range. Higher gain is applied to the target than to the masker, improving the SNR across frequency and facilitating the stimulus detection. This effect was probably diminished at 4 kHz by the low CRs prescribed by NAL-NL1 in the high-frequency range, which caused the lack of difference between linear and compressive amplification. The above-mentioned improvement in the output SNR is the only reason for the apparent improvement of frequency selectivity in HI listeners observed in the present study. It also requires a relatively large number of channels with bandwidth roughly equal to one ERB. Nevertheless, fast-acting multi-channel compression does not tackle the underlying cause of the decreased SRM, namely the broadening of the auditory filters in an impaired auditory system. In fact, such processing can be detrimental to intelligibility if it reduces the natural spectral peak-to-valley contrast, which can be used as a cue for speech recognition [35]. It remains unclear whether this potential reduction of spectral contrast outweighs the observed benefits of reduced spectral masking in more realistic conditions.

5. CONCLUSIONS

The auditory-processing model was shown to account for the effects of hearing loss and hearing-aid amplification on spectral and temporal processing. Linear amplification and fast-acting multi-channel compression were found to improve the measures of temporal masking in HI, due to increased stimulus audibility. However, contrary to the expectations, no additional benefit of compression was observed for the decay of forward masking.

On the other hand, significant improvements in terms of spectral masking were found with compression, but not with linear amplification. This is probably due to the increased SNR across frequency provided by the multi-channel compression.

The results of the present study provide insights into the interaction of hearing-aid amplification with basic
auditory processing. Future research will focus on the potential implications of the observed improvements on auditory perception in more realistic scenarios.

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