A method for quantifying the effects of non-linear hearing-aid signal-processing on interaural level difference cues in conditions with multiple sound sources

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Measuring the effects of non-linear hearing aid algorithms on Interaural Level Differences (ILD) in multi-source environments is not straightforward because the principle of superposition cannot be applied. In order to correctly drive the hearing aids' compressors, for example, the hearing aids must be presented with the intended mixture of signals. At the output side of the hearing aids, the signal mixtures must then be separated in order to be able to compute source-specific 'aided ILDs'. Such signal separation was realized by using methods previously developed by Hagerman and Olofsson [Acta Acustica **90**, 356-361 (2004)] that were extended to work in a free-field set-up with up to three spatially separated concurrent sound sources presented from fixed positions. In the following, we report on how compression in bilateral hearing aids may change ILD cues by a considerable magnitude. We further show how ILD effects depend on the compression parameters as well as the types and number of signals presented.

INTRODUCTION

It is well-known that non-linear hearing aid signal processing can distort ILDs that occur naturally in complex multi-source environments. But so far, the magnitude of these distortions has not been described in great detail with typical hearing aid settings.

In a paper by Singh *et al.* (2009) an experiment was described in which spatial hearing by older adults with normal and impaired hearing in a complex and uncertain acoustic scene was investigated. The experiment run by Singh *et al.* uses a set-up with three concurrent talkers presented from three separate loudspeakers located directly in front, 45 degrees to the right and 90 degrees to the right. In a follow-up study, investigating the perceptual benefit of linear and compressive hearing instruments, in such a set-up, a precise handle was needed on the experimental contrast. Therefore, the current work was initiated to produce a tool with which the experimental contrast could be quantified.

Previous research conducted at the Eriksholm research centre (Naylor and Johannesson, 2009) looking into the side-effects of compression provided a basis for further extension of the work by Hagerman and Olofsson (2004) to the present context.

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DEMIXING METHOD

To be able to estimate the ILD of a single sound source in a complex acoustic context after hearing aid processing, access to the source-specific output of both the left and right hearing aid is necessary. The method used for this is an extension of the method developed by Hagerman and Olofsson (2004). The principle of cancellation by means of phase inversion is illustrated in Eq. 1:

$$\left.\begin{aligned} X_{++} &= T + M_1 \\ X_{+-} &= T - M_1 \end{aligned}\right\} \Rightarrow T_{out} = \frac{1}{2}(Y_{++} + Y_{+-}) \qquad \text{(Eq. 1)}$$

The equation illustrates how a number of source signals, X, can be constructed with either positive or negative phases of the constituent parts. In this case X contains two source signals, a target, T, and a masker, M_1. If we are interested in retrieving T at the output of a hearing aid, two versions of the source signal X must be constructed, as shown in Eq. 1. These two signals are then played back via loudspeakers, and the corresponding output, Y, of each hearing aid is recorded. This allows reconstructing an estimate of the target signal, T, by linear combination of the recorded output signals. The masker, M_1, can be found in a similar way.

The method described above has also been extended to the case of three concurrent sound sources, but for the sake of simplicity and communication of the basic principle, only the two-source case is reported here.

Methodological investigations have shown that the signal-to-interference level ratio of the estimated output is at least 20 dB in the current context. This holds even for fast compression. With a signal-to-interference ratio of at least 20 dB, it is possible to compute reliable estimates of ILDs in the demixed signals in any of the 1/3-octave bands used for analysis.

ILD recordings

All recordings were made in an anechoic chamber. The physical set-up used for the recordings is shown in Fig. 1. The distance from the manikin to the loudspeakers was 1.4 m.

Recordings were controlled from a Matlab-based application, playing back .wav-based sounds from an Echo Gina 3G sound card, at a sampling frequency of 96 kHz, via active Genelec 8030A loudspeakers. The recordings were synchronized to the playback at a sample level. They were made using a B&K type 4128 HATS, connected to a B&K 5935 dual microphone power supply and pre-amplifier connected to the sound card.

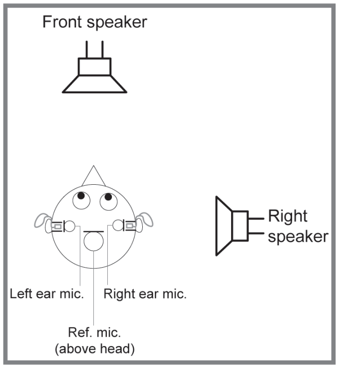

Fig. 1: Set-up used for measuring ILDs on a B&K HATS.

Hearing aids

The hearing aids used for the investigation were of the Receiver In The Ear (RITE) type. This type is characterized by using a microphone location above the ear as in a Behind The Ear (BTE) device. The receivers were placed in the ear canal of the HATS using open domes, which results in direct sound dominating the ear-input signals up to about 1 kHz. Above this frequency, the amplified sound dominates.

The gain in the hearing aids was prescribed for a moderate sloping hearing loss. The average hearing loss for the octave frequencies from 500 Hz to 4 kHz was 40 dB HL, hearing loss of 45 and 65 dB HL at 2 and 8 kHz. The prescription used was NAL-NL1, for an input level of

65 dBA SPL. Around this input level either linear amplification or 2:1 compression was applied for inputs ranging from 50 dB SPL to 80 dB SPL. For inputs below 50 dB SPL the gain was linear until it rolled off at very low levels due to expansion. Above 80 dB SPL the gain was linear until the output limiter was activated.

To avoid confounding effects of adaptive signal processing and binaural hearing aid functionality all functionality except the one described above and anti-feedback suppression was turned off.

Post processing of recorded signals

Given either a bilateral recording of a single source signal or a bilateral set of demixed signals, the ILDs were computed in the following way. Each signal was filtered into 1/3-octave bands with center frequencies ranging from 160 Hz to 6.4 kHz, and the L_{eq} was computed in non-overlapping time frames of 10 ms. Based on the levels at both the left and the right ear, the ILDs could now be computed as the difference in level for each 10 ms time frame. Also, a long term ILD was computed by first computing the long term level in all 1/3-otave bands at each ear and then computing the ILD as the difference between these ear levels.

In the following, the term “aided ILD” will be used to denote the source-specific ILD computed from the output of the left and right hearing aids on the manikin used for the recordings.

RESULTS

The aided ILD for a single pink noise sound source positioned directly to the right of the manikin is shown in Fig. 2.

From Fig. 2 it is seen that the aided ILD from the linear hearing aids is larger than the one from the compressive hearing aids. The difference in high-frequency average ILD amounts to about 3.5 dB or about 20% of the magnitude of the ILD from the linear hearing aids. In an acoustic sense, this single source aided ILD can be considered a phantom cue for spatial hearing.

The reason for the reduction in the aided ILD from the compressive hearing aids is due to the interaction between the head shadow and compression. Since the level at the right ear may be up to 20 dB larger than that of the left ear for high frequencies, the gain in the left hearing aid will be higher than that of the right hearing aid, which will cause the level differences between the outputs of the hearing aids to be smaller than those observable at the inputs.

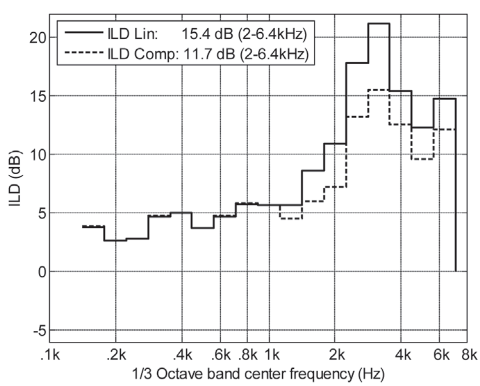

Fig. 2: ILDs measured on a B&K HATS as a function of 1/3 octave band centre frequency. The ILDs are obtained for a single pink noise source located directly to the right of the manikin. Shown by the full and dashed lines are the aided ILDs, obtained with linear and compressive RITE hearing aids on both ears, respectively. In the legend, next to the label, the average aided ILD of the 1/3 octave bands with centre frequencies from 2 to 6.4 kHz is also given.

The amount of aided ILD reduction observable is also affected by the compression knee-points in the hearing aids. Given that knee-points of 50 dB SPL were used in this study, only levels exceeding the knee-point were subject to compression, and thus the effect observed (ca 3.5 dB of ILD reduction) is smaller than indicated by the compression ratio.

The spectral shapes of the aided ILDs are such that both show a peak at about 3.5 kHz where the difference in magnitude is about 5 dB. The peak frequency of 3.5 kHz is somewhat lower than that for the open ear, which is expected from prior work on microphone location effects in hearing aids.

When a second sound source is activated directly in front of the manikin, changes relative to the single source case are seen for the aided ILDs from the sound source positioned directly to the right as seen in Fig. 3. These are within 1 dB of the corresponding values for a single sound source, but still, the ILDs observed at the output of a pair of compression hearing aids are reduced by about 3 dB compared to the single source linear case for the same type of signal. The reason for the observed differences is the smaller level difference between the level at the left and right ear of the manikin caused by the activation of the frontal sound source. Given a smaller level difference between the ears it is expected that compression will to a smaller degree distort the ILDs.

The ILDs for the frontal source and the linear hearing aids are seen in the top part of the figure and are as expected quite close to zero. The corresponding aided ILDs for the compression hearing aids, seen in the bottom part of Fig. 3, are notably different from zero, for frequencies around 2 kHz.. This is due to the asymmetry of the setup and the resulting changes in gain by the compressors. Thus, when several sound sources are active one may also observe a phantom ILD phenomenon from the outputs of bilaterally fitted, independently acting compression hearing aids. Also here, an average change in ILD of about 3 dB from the single source linear case (0 dB by definition) to two sources and compression is observed.

All results reported above have been for pink noise sound sources. These differ from most environmental sounds by having a constant level as a function of time. In the following, speech sources have been chosen to show what the effects of hearing aids are when the signal is more dynamic. As a first result, Fig. 4 shows the aided ILDs from the compression hearing aids being subjected to a single speech and pink noise sound source to the right of the manikin.

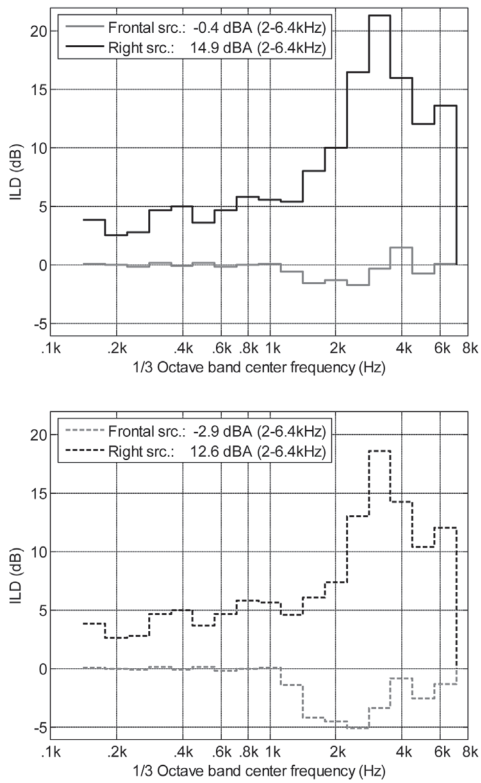

Fig. 3: ILDs measured on a B&K HATS as a function of third-octave band centre frequency. The ILDs in both panels were obtained for two concurrent pink noise sources, one located directly in front and the other one to the right of the manikin. Shown in grey and black are the aided ILDs for the frontal and right source, respectively. The top panel shows ILDs for the linear hearing aids and the bottom panel shows ILDs for the hearing aids with 2:1 compression.

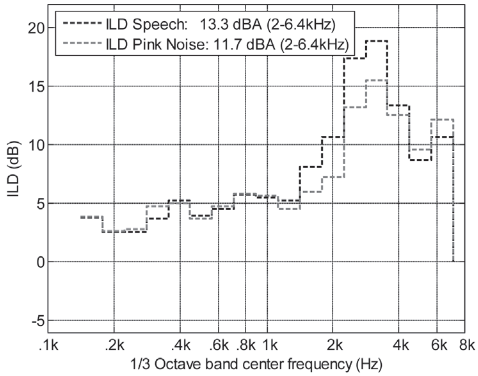

Fig. 4: ILDs measured on a B&K HATS as a function of third-octave band centre frequency. Both ILDs are obtained for a single sound source located directly to the right of the manikin. Shown by the black and grey dashed lines are the aided ILDs, obtained by recording either speech or pink noise with HATS equipped with compressive RITE hearing aids on both ears.

From Fig. 4 it can be seen that the aided ILD from the compression hearing aids for speech is about 1.5 dB larger on average and 3 dB larger at the peak frequency than the corresponding aided ILD for pink noise. This is due to interactions between signal dynamics and compression time constants. Thus, the speech signal yields less effective compression and consequently also less reduction of the aided ILD after compression.

The case of two concurrent speech sources directly in front and to the right for linear and compression hearing aids is shown in Fig. 5. It is seen that when linear hearing aids are used, the aided ILDs are as expected very close to the corresponding ones found for pink noise. When compression hearing aids are used, the aided ILD for the right source is larger than the corresponding one found for two pink noise sources, as shown in the right part of Fig. 3. Again, this is due to the smaller amount of compression due to the differences in signal dynamics. Therefore, also the phantom ILD phenomenon observed for the frontal source in the right part of Fig. 3 is reduced.

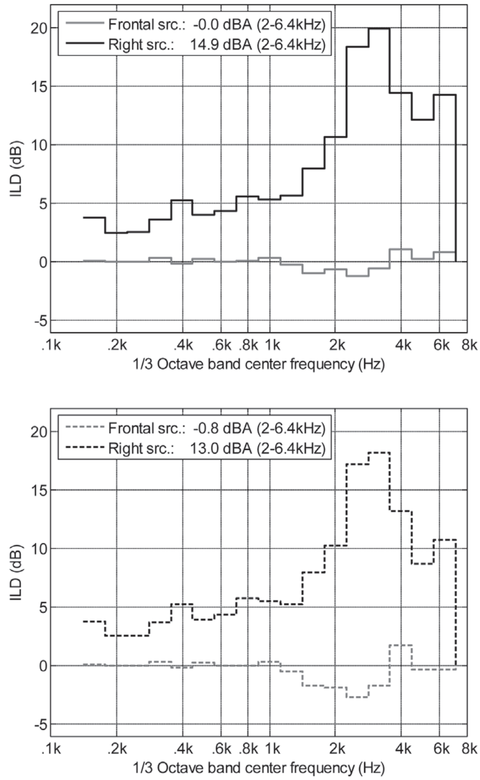

Fig. 5: ILDs measured on a B&K HATS as a function of third-octave band centre frequency. ILDs in both panels are obtained for two concurrent speech sources, one located directly in front and the other one to the right of the manikin. Shown in grey and black is the aided ILDs, of the frontal and right source. The left panel shows ILDs for hearing aids with 2:1 compression and the right panel shows ILDs for linear hearing aids.

SUMMARY

The work presented here has shown that aided ILDs are reduced by compression and the magnitude of this reduction has been reported for typical hearing aid settings, for one and two concurrent sound sources, as well as for pink noise and speech signals.

The knee points set for the WDRC hearing aids used in the present work, however, limit the amount of ILD reduction observed as a consequence of compression. However, WDRC implementations with lower knee points will lead to a larger degree of ILD distortion. Further, the amount of effective compression observable in the devices also affects the aided ILD magnitude. Thus, steady state signals, such as pink noise, will give rise to more compression and thus more ILD reduction than more dynamic signals such as speech.

When more than one sound source is active and the acoustic scene is asymmetrical, phantom aided ILD cues may be observed for 2:1 compression and steady state signals as used here. Extending the results of this work to fast compressors with large compression ratios of 3:1 or more, one should also observe phantom aided ILDs for speech sources.

The perceptual consequences of the distortions of the ILD produced by hearing aids are to be quantified in a follow-up experiment using the methods described by Singh *et al.* (2009).

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REFERENCES

Hagerman, B., and Olofsson, A. (**2004**). "A method to measure the effect of noise reduction algorithms using simultaneous speech and noise," Acta Acustica **90**, 356-361.

Naylor, G., and Johannesson, R. B. (**2009**). "Long-Term Signal-to-Noise Ratio at the Input and Output of Amplitude-Compression Systems," J. Am. Acad. Audiol. **20**, 161-171.

Singh, G., Pichora-Fuller, M. K., Behrens, B., and Neher, T. (**2009**). "The effect of hearing loss on auditory spatial attention", ISAAR Proceedings, Elsinore, Denmark.