Effect of improving audibility on better-ear glimpsing using non-linear amplification

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(Received 2 May 2017; revised 28 November 2018; accepted 30 November 2018; published online 21 December 2018)

Better-ear glimpsing (BEG) utilizes interaural level differences (ILDs) to improve speech intelligibility in noise. This spatial benefit is reduced in most hearing-impaired (HI) listeners due to their increased hearing loss at high frequencies. Even though this benefit can be improved by providing increased amplification, the improvement is limited by loudness discomfort. An alternative solution therefore extends ILDs to low frequencies, which has been shown to provide a substantial benefit from BEG. In contrast to previous studies, which only applied linear stimulus manipulations, wide dynamic range compression was applied here to improve the audibility of soft sounds while ensuring loudness comfort for loud sounds. Performance in both speech intelligibility and BEG was measured in 13 HI listeners at three different masker levels and for different interaural stimulus manipulations. The results revealed that at low signal levels, performance substantially improved with increasing masker level, but this improvement was reduced by the compressive behaviour at higher levels. Moreover, artificially extending ILDs by applying infinite (broadband) ILDs provided an extra spatial benefit in speech reception thresholds of up to 5 dB on top of that already provided by natural ILDs and interaural time differences, which increased with increasing signal level.
 https://doi.org/10.1121/1.5083823

[ICB] Pages: 3465–3474

I. INTRODUCTION

Spatial release from masking (SRM) refers to the improved perception of a target signal when it is spatially separated from a masker signal versus when it is co-located. SRM helps in the understanding of speech in noise for listeners with normal hearing (NH) as well as hearing-impaired (HI) listeners. However, the amount of SRM is typically reduced in HI listeners for a number of reasons, including loss of audibility, reduced spectral and temporal resolution, impaired processing of the signal's temporal fine structure, and aging-related effects such as reduced selective attention, short-term memory, and executive function (e.g., Dubno *et al.*, 2002; Gallun *et al.*, 2013; Glyde *et al.*, 2013a). In general, SRM can be due to a release from energetic masking as well as informational masking, depending on the nature of the involved stimuli as well as the listener's task (e.g., Freyman *et al.*, 1999; Brungart *et al.*, 2001; Kidd *et al.*, 2007). Energetic masking occurs when a similar set of neurons is excited by the masker and the target stimuli due to their temporal and spectral overlap within the auditory periphery, whereas informational masking occurs due to maskers interfering with auditory stream segregation processes or selective attention (e.g., Pollack, 1975; Shinn-Cunningham, 2008; Kidd *et al.*, 2010; Culling and Stone, 2017).

The SRM obtained when a masker is placed only on one side of the head and the target signal arrives from the front (or the other side of the head) can be largely attributed to the head shadow effect resulting in a single ear with a consistently better signal-to-noise ratio (SNR). This situation becomes more complex when listeners are surrounded by more than one fluctuating masker, i.e., when maskers are placed not only on one side of the head but rather on both sides of the head. In such conditions, the ear with the better SNR is not consistent but rather keeps fluctuating between the ears. The SRM obtained in such a scenario can be largely attributed to a phenomenon known as better-ear glimpsing (BEG; Brungart and Iyer, 2012; Glyde *et al.*, 2013b), even though the underlying mechanisms are still unclear and may be far too sluggish for an actual glimpsing process (e.g., Cooke, 2006; Culling and Mansell, 2013; Stone and Canavan, 2016). BEG is best studied in symmetric masker conditions (i.e., with the target speech from the front and a single fluctuating masker from either side of the head: Brungart and Iyer, 2012; Glyde *et al.*, 2013b) and provides a release from energetic masking by relying on the listener's ability to take advantage from either ear that provides the better short-term SNR. This process relies on the occurrence of interaural level differences (ILDs; Glyde *et al.*, 2013b) which, in the real world, are always accompanied by interaural time differences (ITDs). However, it has been reported by Glyde *et al.* (2013c), that, in many spatial masker conditions, ILDs (due to BEG) contribute more to SRM than ITDs, although some studies have reported contrary results (Culling *et al.*, 2004; Kidd *et al.*, 2010). Independent of the relative contribution of ILDs to SRM, it is known that ILDs mainly exist at high frequencies, but, unfortunately, that is where most HI listeners show the strongest hearing loss (e.g., Dillon, 2012, pp. 286–335). Hence, the more severe

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the hearing loss, the lesser will be the audibility of the target signal. As a consequence, HI listeners cannot take full advantage of BEG, leading (or contributing) to their difficulties in understanding speech in noise.

The most commonly applied solution to compensate for the loss of audibility is to provide hearing aid amplification, which for most hearing losses requires increased amplification at high frequencies. For instance, Glyde *et al.* (2015) have shown that by providing extra (linear) amplification on top of what is recommended by common prescription rules such as National Acoustic Laboratories—Revised Profound (NAL-RP; Byrne *et al.*, 1990), the SRM provided by BEG can be significantly improved. However, providing amplification that provides audibility [or sensation level (SL)] that is similar to NH listeners, in particular at high frequencies, is challenging. On the one hand, this is due to the reduced auditory dynamic range leading to loudness discomfort issues and, on the other hand, due to technical limitations such as acoustic feedback. In this regard, and as an alternative to increasing amplification (and thereby improving audibility) at high frequencies, Rana and Buchholz (2016) followed the basic idea described, for instance, by Moore *et al.* (2016) and introduced ILDs at low and mid frequencies, where hearing loss is usually less pronounced. They found that both NH and HI listeners could utilize such artificially enhanced ILDs to provide a substantial amount of SRM due to BEG. However, the performance obtained in HI listeners was poorer than for NH listeners, which was most likely due to differences in audibility. Therefore, in a follow-up study, Rana and Buchholz (2018) carefully controlled the audibility of their speech signals across frequency in a cohort of HI listeners and then tested their speech intelligibility performance in noise with artificially maximized (broadband) ILDs at different SLs. The results revealed that HI listeners can utilize these artificial ILD cues in BEG to a similar extent as NH listeners, as long as both groups are tested at equal audibility (or sensation) level. The observed increase of SRM with increasing SL was in general agreement with other studies that utilized natural spatial cues (e.g., Best *et al.*, 2017; Glyde *et al.*, 2015; Jakien *et al.*, 2017).

One common problem that was raised by all the above studies is the inability of some HI listeners to perform at higher SLs due to loudness tolerance problems. Hearing aids use Wideband Dynamic Range Compression (WDRC: Kates, 2008, pp. 221–259) to ensure loudness comfort and, at the same time, provide increased audibility to soft sounds. However, even though WDRC provides increased audibility to soft sounds, it is unclear how far this increase in audibility can improve the effectiveness of BEG. Depending on how WDRC is implemented in a hearing aid, it can reduce temporal fluctuations as well as the spectral contrast of the incoming signals (e.g., Plomp, 1988). Implementations that avoid such distortions (i.e., by applying slow acting compression) may provide less amplification to soft sounds in particular (Dillon, 2012, pp. 170–197). Hence, it may well be that the detrimental effect of signal distortions in hearing aids counteract the benefit provided by the increase in audibility of soft sounds, resulting in a negligible or even negative effect on BEG. This potential problem may be further aggravated when two independently operating hearing aids are fitted to the left and right ear of a listener, which can result in distorted ILD cues (e.g., Wiggins and Seebeer, 2012; Buchholz, 2013). To the best of the authors knowledge, there are no studies that have systematically investigated the effect of increasing stimulus level (and thus audibility) on BEG using non-linear amplification. However, there are a few studies that have investigated the effect of audibility using non-linear amplification on speech recognition tasks. For instance, Davies-Venn *et al.* (2009) tested listeners with different degrees of hearing loss on a nonsense syllable recognition task who were fitted with hearing aids with fast-acting WDRC. They reported a significant improvement in performance when the stimulus level was increased from 50 to 65 dB sound pressure level (SPL), but the performance reduced when the input level was further increased from 65 to 80 dB SPL. Hence, it may be similarly expected that speech recognition in noise is improved by non-linear amplification at soft signal levels, but reduced at high signal levels.

Overall, this study progresses the work done by Rana and Buchholz (2016) as well as Rana and Buchholz (2018) in terms of understanding the role of audibility on BEG and the ways that BEG can be maximized in HI listeners to improve speech intelligibility in noise. The specific aims of the present study were to investigate: (i) the effect of non-linear amplification using WDRC on BEG and thus, to better understand the interaction between the provided increase in audibility and the inherent signal distortions, and (ii) the spatial benefit (or SRM) that can be achieved in addition to the one already provided by natural ILDs and ITDs by artificially enhancing ILDs, in particular at low frequencies, which was realized here by applying infinite ILDs. The results obtained from this study will help to better understand and to design non-linear amplification schemes that optimize the benefit provided by BEG in real-world environments.

II. METHOD

Speech intelligibility, as well as SRM, was measured for NH and HI listeners using stimuli with different combinations of spatial cues as well as using different amplification schemes.

A. Participants

Ten NH listeners (hearing thresholds <15 dB hearing level) aged between 25 and 41 yr (mean age of 33.5 yr) and 13 HI listeners aged between 68 and 79 yr (mean age of 74 yr) with a symmetric (<10 dB difference between ears from 250 Hz to 4 kHz), sensorineural, mild to moderate-severe hearing loss participated in this study. Hearing thresholds were tested prior to the experiment for all participants to either confirm normal hearing or to establish their degree of hearing loss. The mean and ±1 standard deviation of the four-frequency (500, 1000, 2000, 4000 Hz) average hearing loss of the HI subjects was 31 ± 8 dB. Individual and mean pure tone thresholds averaged across the left and right ear are shown in Fig. 1. All participants had Australian English as their first language and had no reported attention deficit disorder or intellectual disability. The complete testing was conducted in a sound-treated audiological test booth at the

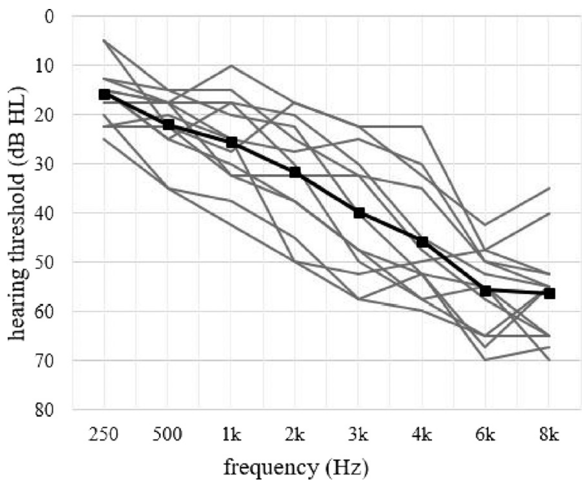

FIG. 1. Mean (black line) and individually averaged (gray line) pure tone audiograms averaged over the left and right ear of the 13 HI listeners.

NAL and took about 4 hours per subject, which was divided into two appointments of 2 hours each. Participants received a small gratuity for their participation. Ethical clearance was taken from the Australian Hearing Human Research Ethics Committee and the Macquarie University Human Research Ethics Committees.

B. Stimuli

Speech intelligibility was assessed with a corpus of 80 lists of 16 BKB (Bamford-Kowal-Bench)-like target sentences (Bench *et al.*, 1979) produced by a native female speaker of Australian English. The root-mean-square (RMS) level of all sentences was normalized and the average spectrum of each sentence list was equalized to match the average spectrum of the entire corpus, which is shown in the left panel of Fig. 2. The masker was realized by two separate noise-vocoded single talker speech maskers, which were identical to the ones described by Rana and Buchholz (2016). In brief, the two different-voice speech discourses from Cameron and Dillon (2007) were noise vocoded using a short-term Fourier Transform with 20-ms long time windows and four critical-bands wide spectral smoothing. This process realized a noise vocoder with about 5 effective frequency channels (within a bandwidth of 8 kHz) and made the two maskers, when combined, largely unintelligible. Each masker was equalized to match the long-term spectrum of the target speech. The noise vocoding was applied to minimize the influence of informational masking (see Rana and Buchholz, 2016) and thereby to focus on energetic auditory processes such as BEG (Brungart and Iyer, 2012) or equalization-cancellation (Durlach, 1963). Hence, this may have also minimized the potential influence of localization or perceived spatial separation of the target and masker signals on SRM, which is more commonly associated with informational masking (e.g., Freyman *et al.*, 1999) and may play only a minor role in energetic masking.

All stimuli were spatialized using non-individualized Head-Related Transfer Functions (HRTFs) and stored as binaural wave-files with a sampling frequency of 44.1 kHz. The target signal was always presented from the front of the listener (0° direction) and the two masker signals were either co-located with the target or spatially separated at +90° and −90°. The stimuli were spatialized using the HRTFs described by Cameron and Dillon (2007), which were measured on a Bruel & Kjaer Head (Denmark) and Torso Simulator. The magnitude spectra of the HRTFs are shown in the right panel of Fig. 2 separately for the left and right ear. For the spatially separated HRTFs the ILDs are indicated by the shaded areas. The corresponding ITDs are not shown here, but for the 0° direction the ITD was less than 23 μs (i.e., less than 1 sample) and around −680 and +680 μs for the −90° and +90° direction, respectively.

To investigate the individual contributions of ILDs and ITDs on speech intelligibility as well as SRM, in one spatially-separated condition the ITDs were removed and only the ILDs were preserved. This ILD-only condition was realized by deriving minimum-phase versions of the original HRTFs for the +90° and −90° directions. Finally, a condition with *infinite ILDs* and no ITDs was created by using the HRTFs for the 0° direction, but removing (i.e., by multiplying with zero) either the left ear or the right ear of the HRTF. This generated a masker signal that contained one masker that was only presented to the left ear and one masker that was only presented to the right ear. Even though this process maximized ILDs across all frequencies, the main spatial benefit would occur at low and mid frequencies where natural ILDs were rather small (Fig. 2, right panel) and most hearing losses were less pronounced (see Sec. I).

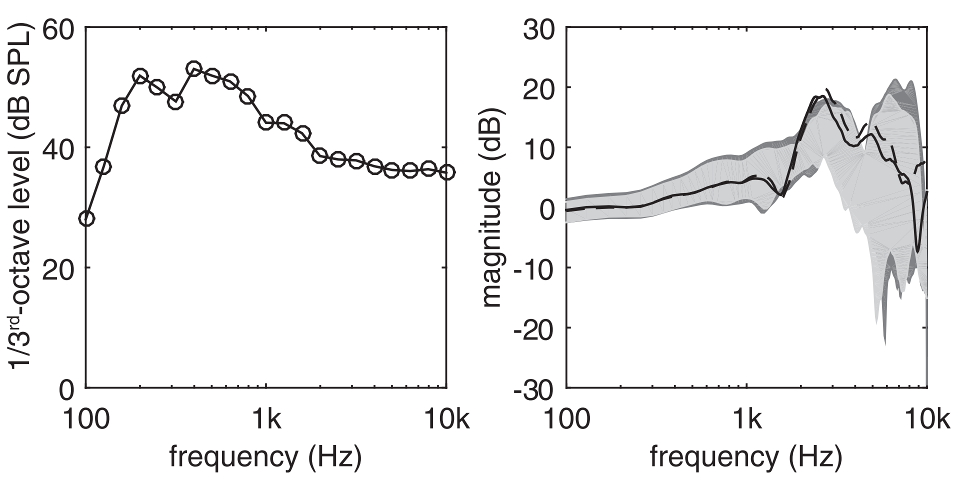

FIG. 2. The left panel shows the third-octave spectrum of the target speech in free-field (i.e., before spatialization) at 60 dB SPL, which was identical for the masker signals. The right panel shows the magnitude spectrum of the applied natural HRTFs for the front (0°) direction (left ear: solid line; right ear: dashed line) as well as for the +90° (light-gray shaded area) and −90° (dark gray-shaded area) direction, whereby the corresponding spectra at the left and right ear are indicated by the edges of the shaded areas. The shaded areas provide a direct representation of the provided ILDs.

In the co-located (i.e., 0°) condition the same HRTF as for the target speech was utilized, but a gain of −3 dB was applied to each ear to compensate for the level increase that resulted from adding two uncorrelated maskers. In summary, speech intelligibility was assessed either with the masker co-located with the target at 0° or with the maskers spatially separated at ±90° using (a) natural HRTFs (i.e., natural ITDs and ILDs), (b) natural HRTFs without ITDs (i.e., using ILDs only), and (c) artificially enhanced (infinite) ILDs. The binaural target and masker stimuli were presented to a real-time master hearing aid using a standard Windows computer connected to a RME™ Fireface UC USB sound card and running a purpose-build speech test software in MATLAB (see Sec. II C). The hardware of the master hearing aid consisted of a standard Windows computer connected to another RME™ Fireface UC USB sound card and received the binaural input from the test computer via balanced audio cables. The amplified output signal of the master hearing aid was presented to the test subjects through Sennheiser HD215 circumaural headphones, which were calibrated using a Bruel & Kjaer artificial ear.

The main signal processing associated with the stimulus playback and master hearing aid processing is illustrated in Fig. 3, with H_0 the free-field-to-ear-drum transfer function, H_0^{-1} the ear-drum-to-free-field transfer function (or the inverse of H_0), G_{HA} the main hearing aid processing (or gain), and H_{EQ} the transfer function of a filter that equalizes the frequency response of the applied headphones. The free-field-to-ear-drum transfer function H_0 approximated the absolute spectrum of the HRTF for the 0° direction shown in Fig. 2 (right panel) averaged across the left and right ear. This function was realized in the master hearing aid by applying appropriate gains to the individual frequency channels of the hearing aid filterbank. The ear-drum-to-free-field transfer function H_0^{-1} was realized by a 1024-samples long minimum-phase Finite Impulse Response (FIR) filter that was applied offline to the binaural signals on the test computer. The headphone equalization filter was realized by a 512-samples long minimum-phase FIR filter that was derived with a Bruel & Kjaer artificial ear and averaged across the left and right ear. Hearing aid amplification was set in sixteen 1/3-octave wide frequency channels either by applying linear amplification according to the NAL-RP prescription or by applying non-linear amplification according to the NAL-NL2 prescription (e.g., Dillon, 2012, pp. 313–314).

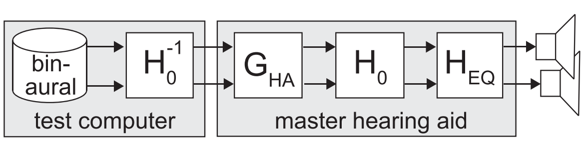

FIG. 3. Illustration of the signal processing relevant to the stimulus playback and hearing aid signal processing. Binaural stimuli were generated for different spatial test conditions, transformed to free-field equivalent levels using a filter with transfer function H_0^{-1}, and presented to a real-time master hearing aid using a test computer. The master hearing aid applied either linear or non-linear amplification as indicated by the gain G_{HA}, mapped the resulting signals back to ear-drum levels, and played them to the subjects using equalized headphones (via a filter with transfer function H_{EQ}). Further details are described in the text.

Amplification was derived individually for the subject's hearing loss averaged across the left and right ear.

The effect of different compression parameters (such as compression threshold, compression ratio, number of frequency channels, attack time, release time) on speech intelligibility has been discussed extensively throughout the relevant literature (e.g., Yund and Buckles, 1995; Boike and Souza, 2000; Souza., 2002; Moore *et al.*, 2004; Dillon, 2012, pp. 170–197). Here, fast-acting WDRC was applied to each frequency channel separately with attack and release times of 10 and 100 ms, respectively. The compression thresholds were set to the 1/3-octave levels of LTASS speech (Byrne *et al.*, 1994) with a broadband level of 52 dB SPL. For signal components below the compression threshold, amplification was linear (i.e., the compression ratio was set to 1). For more intense signal components, the average compression ratios for the HI group was between 1 and 1.5 at frequencies below 1 kHz and around 2 at frequencies above 1 kHz. The resulting gains, averaged across the HI subjects, are shown in the left panel of Fig. 4 for the applied speech signals at 50, 60, and 70 dB SPL. The resulting free-field levels are shown in the right panel. The applied (linear) gain according to NAL-RP is also shown as well as the corresponding speech spectrum before and after linear amplification with an input level of 60 dB SPL.

C. Procedure

Similar to Rana and Buchholz (2016), adaptive speech reception thresholds (SRTs) were measured for target sentences in the presence of different maskers using a MATLAB program installed on a personal computer. The participant's task was to repeat as many words as they heard in each target sentence while ignoring the distracting signals. At least 17, and up to 30, sentences were presented and the SNR was adjusted adaptively to achieve 50% correct word identification (SNR50) by keeping the masker level constant and varying the target level. Further details can be found in Keidser *et al.* (2013).

In order to investigate the effect of non-linear amplification on the utilization of spatial cues in understanding speech as well as in SRM, all HI listeners were tested at an overall masker level of 50, 60, and 70 dB SPL, as measured in the free-field before amplification was applied. As a reference condition, all HI listeners were also tested with linear amplification at a single masker level of 60 dB SPL. The NH subjects were tested with linear amplification of 0 dB and a combined masker level of 60 dB SPL. The masker levels were originally chosen after extensive pilot testing and allowed to investigate both the effect of linear versus non-linear amplification and the effect of compression as a function of masker level while ensuring loudness comfort. However, this resulted in most of the masker levels to be lower than, for instance, the 65 dB SPL assumed in the NAL-RP prescription formula (Dillon, 2012, pp. 313–314), and may have magnified the effect of hearing loss (i.e., reduced audibility) on overall performance, especially with the linear amplification.

Both groups were tested at each masker level in a co-located condition as well as in three spatially-separated

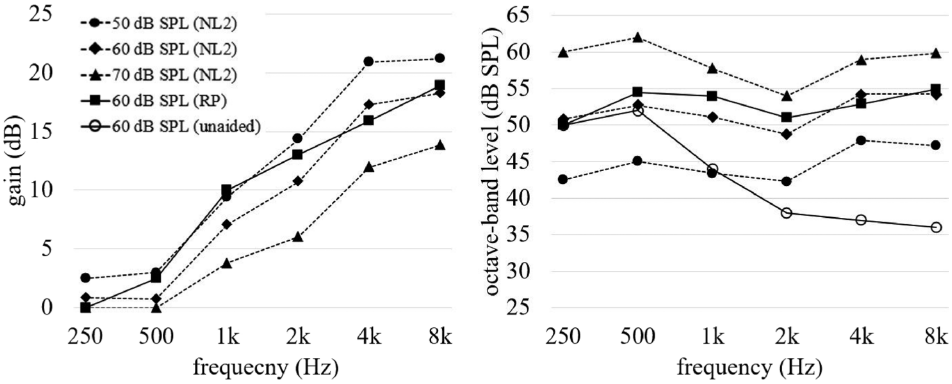

FIG. 4. The average gains prescribed by the (non-linear) NAL-NL2 formula for the tested speech levels of 50, 60, and 70 dB SPL are shown in the left panel and the corresponding speech spectra after amplification are shown in the right panel (dashed lines). The corresponding linear gains and the resulting speech spectrum are also shown (solid lines, squares) together with the unaided speech spectrum at a free-field level of 60 dB SPL (solid line, circles).

conditions which differed in their auralization method, using: (1) natural HRTFs (i.e., ILDs and ITDs), (2) natural ILDs only, and (3) artificially enhanced (infinite) ILDs. For each condition, the SRTs were averaged over two measurements, and the SRM was calculated by subtracting SRTs in the spatially separated conditions from the SRTs in the corresponding co-located condition. Before any speech testing started, loudness assessments were done in each test subject for the target and masker signals separately to ensure loudness comfort for all tested stimuli using an 8-point rating scale (1-very soft; 2-soft; 3-comfortable, but slightly soft; 4-comfortable; 5-comfortable, but slightly loud; 6-loud, but ok; 7-slightly uncomfortable; 8-uncomfortably loud). Any condition with an individual loudness judgment above point 6 was dropped in the speech test for that subject.

III. RESULTS

The data were analyzed using IBM SPSS version 22.

A. SRTs

The mean SRTs for the 13 HI subjects are shown in Fig. 5 (left panel, filled symbols) together with 95% confidence intervals for the four different spatial cue conditions with amplification as parameter (i.e., non-linear amplification at 50, 60, and 70 dB SPL masker level and linear amplification at 60 dB SPL masker level). For reference purposes the SRTs for the NH subjects are also shown (open symbols). Since the main interest of this study was to examine the effect of restoring audibility using non-linear amplification on the utilization of different combinations of spatial cues, a two-way repeated measure analysis of variance (ANOVA) was conducted with masker level and type of spatial cues as main factors. The results revealed a significant main effect of masker level [$F(2, 24) = 46.03$, $p < 0.01$) as well as type of spatial cues [$F(3, 36) = 282.36$, $p < 0.01$], but no significant interaction. As is shown in Fig. 5 (left panel), the SRTs improved with an increase in masker level across all conditions. However, the rate of improvement was much faster when the level was increased from 50 to 60 dB SPL than when increased from 60 to 70 dB SPL. The results of a paired t-test with Holm-Bonferroni correction (Holm, 1979) revealed a significant difference in SRTs between 50 and 60 dB SPL across all conditions ($p < 0.01$) but no significant difference between 60 and 70 dB SPL, except when artificially enhanced (infinite) ILDs were used. Further, based on an independent t-test, it was also found that even at the highest masker level of 70 dB SPL with non-linear amplification, the SRTs of the HI listeners were on average 4 dB higher (i.e., worse) than for the listeners with NH at 60 dB SPL masker level ($p < 0.01$).

Another main interest of this study was to measure the extra benefit that artificially enhanced ILDs can provide on top of natural spatial cues. Results of a paired t-test with Holm-Bonferroni correction showed that SRTs with artificially enhanced ILDs were significantly ($p < 0.01$) lower than the corresponding SRTs with natural ILDs alone as well as when natural ILDs were combined with natural ITDs for HI listeners as well as NH listeners. The contribution of ITDs to the SRTs on top of natural ILDs was also analyzed by subtracting SRTs with natural ILDs and ITDs from SRTs with natural ILDs alone. Based on a paired t-test with Holm-Bonferroni correction, a significant contribution of ITDs was found only for HI listeners for a masker level of 60 dB SPL,

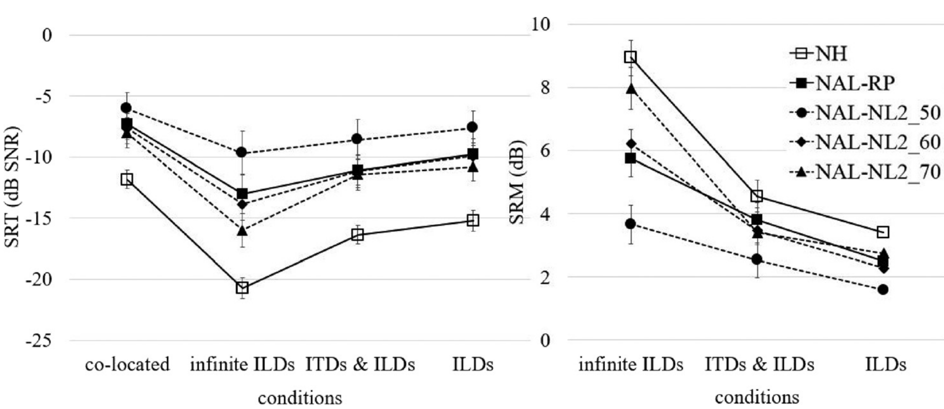

FIG. 5. Mean SRTs, mean SRM, and 95% confidence intervals across conditions.

for both linear and non-linear amplification, and for NH listeners.

The extent to which WDRC in non-linear amplification can negatively or positively affect the utilization of spatial cues when compared to linear amplification was also investigated. The results of a paired *t*-test revealed no significant differences ($p > 0.01$) between SRTs with linear and non-linear amplification across all conditions when the same masker level (i.e., 60 dB SPL) is considered. Finally, the test–retest variability was assessed by subtracting the SRTs of the second trial from the SRTs of the first trial. According to a paired *t*-test, the mean intra-subject difference across all conditions, for both NH and HI listeners, ranged from −0.7 to 1.32 dB, and was not significant ($p > 0.01$). These results are in line with Rana and Buchholz (2016).

B. SRM

SRM was calculated by subtracting the spatially separated SRTs from the co-located SRTs. The mean SRM with 95% confidence intervals is shown in Fig. 5 (right panel) for the HI (filled symbols) and NH (open symbols) listeners. Similar to the SRT analysis, a two-way repeated measure ANOVA was applied to the SRM for non-linear amplification with masker level and type of spatial cues as main effects. Results revealed a significant main effect of masker level [$F(2, 24) = 22.72$, $p < 0.01$) as well as type of spatial cues [$F(2, 24) = 148.9$, $p < 0.01$], but no significant interaction. A paired *t*-test with Holm-Bonferroni correction revealed a significant effect of masker level on SRM only for the masker with artificially enhanced ILDs. To investigate if the best performance of the HI listeners, i.e., the SRM at a masker level of 70 dB SPL, was similar to the performance of the listeners with NH, an independent *t*-test was conducted. Results showed no significant difference in SRM between the two groups. This observation is different to the SRTs, which, in HI listeners, were significantly higher (i.e., worse) than for NH listeners in all conditions. Hence, even though SRM may be restored in HI listeners by providing sufficient non-linear amplification, this is not the case for overall performance in speech intelligibility, at least for the amplification levels considered in the present study.

To measure the extra spatial advantage of using artificially enhanced ILDs in SRM, a paired *t*-test with Holm-Bonferroni correction was conducted. The SRM obtained with artificially enhanced ILDs was significantly ($p < 0.01$) larger than the SRM obtained with natural spatial cues, both with and without ITDs. The SRM data shown in Fig. 5 (right panel) suggests that the SRM obtained with natural ILDs and ITDs was on average about 1 dB larger than for the case that only ILDs are applied. However, the results of a paired *t*-test revealed that this contribution of ITDs was only significant for HI listeners at 60 dB SPL, for both linear and non-linear amplification, and for NH listeners. Finally, using a paired *t*-test to compare the SRM between linear and non-linear amplification at a masker level of 60 dB SPL found no significant difference for all conditions. This is in agreement with the SRT data and indicates a negligible impact of WDRC on SRM across all spatial cues.

IV. DISCUSSION

The effect of increased SL on the ability of HI listeners to utilize artificially enhanced ILDs in BEG has been investigated by Rana and Buchholz (2018) using linear amplification. The present study builds upon their methods and results but for the case that non-linear amplification is additionally applied. Similar to Rana and Buchholz (2018), infinite ILDs were used to artificially enhance ILDs. Even though this approach maximises ILDs across all frequencies, the main benefit that is provided to HI listeners occurs at low (and potentially mid) frequencies, where natural ILDs are rather small and hearing loss is less pronounced.

A. Effect of audibility

The role of reduced audibility on speech intelligibility as well as SRM has been investigated in a number of studies. For example, Rana and Buchholz (2018) controlled audibility in HI (and NH) listeners carefully across frequency and then measured SRTs in a noise-vocoded two-talker masker at different masker SLs. For each increase in SL of 10 dB they found an average improvement of 2 dB in co-located SRTs, 4 dB in spatially-separated SRTs, and a corresponding 2 dB in SRM. Glyde *et al.* (2015) provided 50% extra amplification on top of the frequency-dependent amplification prescribed by the NAL-RP formula, and HI listeners showed an improvement in co-located SRTs by about 1 dB, in spatially separated SRTs by about 4 dB, and hence in SRM by about 3 dB. Similarly, Jakien *et al.* (2017) investigated speech intelligibility at 19.5 and 39.5 dB SL relative to their individual SRTs in quiet, and depending on their stimulus conditions, they found different degrees of improvements in SRTs and SRM with increasing SL of up to 4 dB.

In all of these studies the benefit provided by increased SL was most likely due to improved access (due to increased audibility) to the softer components of the target speech as well as to the available spatial cues (e.g., Glyde *et al.*, 2015). Thereby, the utilization of linear amplification may have played a crucial role as it does not only preserve the temporal and spectral behaviour of the signals at the listeners' ears, but also preserves the available ILDs (Robinson and Gatehouse, 1996; Byrne and Noble, 1998). However, to avoid any loudness discomfort associated with the linear amplification and the smaller dynamic range for HI listeners, in this study different levels of audibility were provided using non-linear amplification. HI listeners were tested at three different masker levels (i.e., at 50, 60, and 70 dB SPL) using a master hearing aid with multi-channel, fast-acting WDRC that was fitted to the individual test subjects according to the NAL-NL2 prescription formula (see Sec. II B). It was found that when the masker level was increased from 50 to 60 dB SPL, the co-located SRTs improved by about 1.6 dB, whereas the spatially separated SRTs improved by about 3.5–5.5 dB, with the larger increase observed with the artificially enhanced ILDs. Hence, the SRM was increased by about 2–4 dB. When the masker level was further increased from 60 to 70 dB SPL, the co-located SRTs improved by only 0.4 dB whereas the spatially separated SRTs improved by only 1–3 dB, again with the larger increase observed with the artificially enhanced ILDs. The improvement in SRTs was found to be significant for all conditions when the masker level was increased from 50 to 60 dB SPL, but when the masker level was increased from 60 to 70 dB SPL it was only significant for the spatially-separated condition with artificially enhanced ILDs. The size of the improvement in co-located and spatially-separated SRTs (and thus in SRM) with an increase in masker level at soft levels was very similar to the improvement found with linear amplification as described by Rana and Buchholz (2018) as well as in Glyde *et al.* (2015). The observed reduction of this improvement at higher signal levels is in agreement to other studies that measured the effect of non-linear amplification on overall speech intelligibility (e.g., Davies-Venn *et al.*, 2009) and reflects the increased effect of compression at higher signal levels as further described below.

B. Effect of compression

In Sec. IV A it was mentioned that, when the masker level was increased from 50 to 60 dB SPL, a similar large increase in both SRTs and SRM was achieved as previously observed with linear amplification. When the masker level was further increased from 60 to 70 dB SPL, this increase was substantially reduced in the co-located as well as all the spatially separated conditions except for the case of artificially enhanced ILDs. This behaviour can be explained by considering the details of the applied WDRC implementation described in Sec. II B. Since the compression thresholds were set to the hearing aid frequency band levels of LTASS speech with a broadband level of 52 dB SPL, the 50 dB SPL masker was largely processed linearly, which was also the case for some part of the masker at 60 dB SPL. Given that the measured SRTs were at highly negative SNRs (i.e., around −6 to −16 dB), most segments of the corresponding target speech were also in the linear range of the WDRC. Hence, it is not surprising that, due to the more-or-less linear behaviour of the WDRC, a similar increase in performance can be observed with increasing masker level as previously reported with linear amplification. However, this was not the case for the highest masker level of 70 dB SPL, for which most of the masker components and a substantial part of the target speech would have been in the compressive region of the WDRC. As a consequence, the benefit achieved by increasing the masker level from 60 to 70 dB SPL was reduced. In the same vein, it can be explained why the condition with artificially enhanced ILDs was less affected by WDRC. In this condition, a substantial part of the SRM was provided by BEG at low and mid frequencies. At these frequencies the compression ratios were less than 1.5 (see Sec. II B) and thus the WDRC acted rather linearly over the entire input level range. Finally, the similar performance observed for the non-linear and linear amplification at a masker level of 60 dB SPL can be explained by the similarity in hearing aid gains that was provided by NAL-NL2 and NAL-RP at this level (Fig. 5, left panel) together with the above observation that the target signal, as well as a substantial part of the (fluctuating) masker signal, was largely in the linear range of the WDRC.

The observation that there was not much compression prescribed by NAL-NL2 at low (and mid) signal frequencies, at least for the rather typical hearing losses considered in this study (i.e., with a rather mild low-frequency and sloping hearing loss), supports the idea that enhancing ILDs at low-frequencies may be an interesting solution for improving BEG with hearing aids and thereby improving speech intelligibility in noise. However, the effective compression implemented by the WDRC schemes in more recent hearing aids is rather low for the applied stimuli (i.e., at speech modulation rates of around 4–10 Hz) due to the application of slow (or dual) time constants. Even though those devices may show less effective compression on speech intelligibility and SRM at high signal levels, they also may not provide the same amount of gain (and thus audibility) that was applied here. Hence, the advantage of reduced distortions by the more recent WDRC schemes may be offset by a less effective compensation of audibility and thus, most likely, by a reduced spatial benefit. This should be further investigated in the future.

C. Extra benefit provided by infinite ILDs

The results of this study showed that artificially enhancing ILDs by using infinite ILDs provides an extra benefit in SRTs by about 3 dB when linear amplification is used at a masker level of 60 dB SPL. When non-linear amplification is used, SRTs with artificially enhanced ILDs are better by about 2 dB at 50 dB SPL, 4 dB at 60 dB SPL, and by about 5 dB at 70 dB SPL. Interestingly, the additional benefit increased with increasing masker level using non-linear amplification, which can be explained by the characteristics of the applied WDRC that is described in Sec. II B. Since natural ILDs only exist at high frequencies, they were strongly affected by the applied compression. As a consequence, the rate of improvement in SRTs with an increase in masker level was rather small (see Fig. 5). This was different for the case of artificially enhanced ILDs, which most likely provided most of their spatial benefit at low and mid frequencies, at which the applied WDRC acted rather linearly. Hence, it can be concluded that artificially enhancing ILDs at low frequencies would be helpful in improving speech intelligibility in noise and can provide an extra benefit on top of what natural ILDs provide.

However, the noted extra benefit might be questionable when natural ILDs are combined with natural ITDs, which may provide a substantial spatial advantage via masker cancellation at frequencies below about 1.5 kHz (e.g., Durlach, 1963). To investigate the additional spatial benefit provided by natural ITDs and the one provided by artificially enhanced ILDs, speech intelligibility was also measured in a condition with natural ILDs and ITDs. Providing ITDs in addition to ILDs improved SRTs by about 1 dB across all the different amplification methods for both NH and HI listeners. Consequently, the extra benefit provided by artificially enhanced ILDs on top of natural ITDs in combination with natural ILDs reduced to about 2 dB when linear amplification was used. With non-linear amplification the extra benefit reduced to 1 dB at 50 dB SPL, 3 dB at 60 dB SPL and about 4 dB at 70 dB SPL.

Even though artificially enhancing ILDs as well as adding ITDs provided a significant spatial benefit on their own, the extra benefit in speech intelligibility when artificially enhanced ILDs are combined with natural ITDs is unclear and may change depending on the individual weighting of binaural cues by the listeners (e.g., Clayton *et al.*, 2017), the potential *binaural interference* by spatial information in neighbouring or remote frequencies (e.g., Epp *et al.*, 2017), and the considered stimuli as well as spatial configuration. Moreover, it should be noted here that the rather small effect of the ITDs on speech intelligibility of about 1 dB may be partially explained by the noise-vocoding process of the masker signal, which may have degraded important envelope ITD cues (e.g., Stone and Moore, 2008).

In the present setup, ITDs cannot be added to the artificially enhanced ILDs because they were realized by "infinite" ILDs (i.e., each masker is only applied to one ear). Since this is an artificial best-case scenario, future studies should also consider more realistic methods for introducing or maximizing ILDs at low and mid frequencies that can be implemented in hearing devices. This may be done by applying directional processing with multiple microphones (e.g., Kates, 2008, pp.75–109), using techniques like frequency transposition (e.g., Robinson *et al.*, 2007), or using ITD information to generate ILDs (Moore *et al.*, 2016). Independent of the realization method, providing ILDs at low frequencies requires sufficient amplification at low frequencies. With current hearing aid technologies, this can only be achieved by fitting occluding ear moulds, which is not comfortable for many listeners with hearing loss. However, new technologies may help with overcoming this problem, including active occlusion reduction (Mejia *et al.*, 2008) or novel receiver technologies with extended (low-frequency) bandwidth as, for instance, provided by Earlens™. Further, it would also be interesting to investigate the obtained spatial benefit using non-vocoded speech masker, although the thereby introduced informational masking and its influence on SRTs in the spatially separated condition may be negligible (e.g., Rana and Buchholz, 2016). Finally, it should be mentioned that the present realization of the artificially enhanced ILDs did also provide infinite ILDs at high frequencies where natural ILDs are present. Hence, this method may have also provided an additional spatial benefit at high frequencies. However, this benefit would have been rather small, because audibility would have already limited access to the rather large natural ILDs (see Fig. 2, left panel), at least in the HI listeners.

D. NH vs HI listeners

The improvement of SRM and SRT with an increase in audibility can help reduce the performance gap between NH and HI listeners, though this depends on the way audibility is controlled. Glyde *et al.* (2015) reported an average difference in SRM of about 2.5 dB between NH and HI listeners when performance was compared at equal audibility levels (relative to pure tone thresholds). In contrast, Rana and Buchholz (2018) equalized audibility carefully in a group of NH and HI listeners across frequency by measuring speech detection thresholds, and found that the difference in both SRTs and SRM between NH and HI listeners reduced to less than 0.5 dB. This suggests that, at least for their applied stimuli and methods, carefully controlling audibility across frequency can largely remove the performance gap between NH and HI listeners. In the present study, SRTs for the HI listeners at 70 dB SPL masker level were about 4 dB higher than for NH listeners. This difference between groups decreased to a non-significant difference of less than 1 dB in SRM suggesting that providing sufficient non-linear amplification can improve speech intelligibility performance, in particular in spatially-separated conditions, and can also improve SRM.

However, the spatially-separated conditions also exhibited the lowest SRTs and were therefore more affected by reduced (target) audibility. The gap of 4 dB in SRTs between the HI and NH group may have been further reduced if audibility would have been more carefully controlled across frequency using a procedure similar to Rana and Buchholz (2018). However, since prescriptive formulas such as NAL-NL2 have not been designed with the sole aim of restoring audibility, it is difficult to use a standard approach for investigating the effect of non-linear amplification for controlling audibility in the same systematic way as described by Rana and Buchholz (2018). However, future studies should further investigate how WDRC could be improved in hearing aids to optimize the benefit provided by BEG in noisy conditions. Finally, in addition to reduced audibility, other factors may have contributed to the differences in SRTs between the NH and HI group, including age-related factors, such as reduced cognitive abilities, or basic auditory function-related factors, such as reduced temporal and spectral resolution or reduced sensitivity to the signal's temporal fine structure (e.g., Divenyi and Haupt, 1997; Divenyi *et al.*, 2005; Murphy *et al.*, 2006; Singh *et al.*, 2008; Gallun *et al.*, 2013; Glyde *et al.*, 2015).

V. CONCLUSION

The present study confirmed that non-linear amplification using WDRC improves speech intelligibility in noise at low signal levels by increasing the audibility of the incoming signals, but this improvement is counteracted by the compressive behaviour at high signal levels. Similarly, the SRM provided by natural ILD and ITD cues is significantly improved at low but not at high signal levels. Moreover, artificially enhancing ILDs by applying infinite ILDs provided a substantial increase in SRM, due to BEG, on top of the SRM already provided by natural ILDs and ITDs. This extra benefit increased with increasing signal level, which was mainly due to the fact that most of the SRM was achieved at low and mid frequencies (up to around 1 kHz) where the applied WDRC scheme behaved rather linearly (due to the rather mild hearing losses of the HI subjects at those frequencies). These results confirm that extending ILDs to low frequencies may be an interesting solution for hearing aids to improve speech intelligibility in noise. However, the realization of low-frequency ILDs in hearing aids will come with certain challenges. For instance, obtaining optimal low frequency amplification might require occluding ear moulds, which may not be accepted by many listeners with mild or moderate hearing loss, or may introduce noticeable signal distortions or processing delays. Therefore, future studies should further investigate methods that optimize ILDs as well as non-linear amplification as a function of frequency such that hearing aids can provide the best benefit to HI listeners when communicating in noisy conditions. The best solution may well depend on the cognitive or other auditory abilities of the listener than described by the audiogram as well as the complexity of the experienced environment.

ACKNOWLEDGMENTS

The authors acknowledge the financial support of Macquarie University through an international research excellence (iMQRES) scholarship and the Australian National Health and Medical Research Council (NHMRC) via Grant No. APP1056332, and thank James Galloway for his technical support.

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