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Hearing impairment, hearing aids, and cues for self motion

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Abstract

When listeners turn their heads, the resulting change in binaural level and timing cues constitutes useable information about the location of signals in the world, particularly on front/back location. However, in order to make use of these dynamically changing cues, listeners must be able to compare the speed and direction in which a signal moves with the speed and direction of their head movements. Hearing impairment is frequently co-morbid with vestibular impairment, rendering access to self-motion cues less reliable, and it is also associated with an increase in the minimum audible movement angle, a measure of auditory motion processing. Hearing impairment is also typically associated with raised thresholds in the range of frequencies in which the filtering effects of the outer ear provide other cues useful for resolving front/back localization ambiguities. By moving sound sources as a function of the instantaneous position of the listener’s head, we created a front/back illusion, and demonstrated that listeners with hearing impairment rely more heavily on self-motion cues than on high frequency information, even though their ability to use binaural cues can be impaired. The use of hearing aids had heterogeneous effects in different listeners, although in no case did they return a listener to normal performance. To examine this phenomenon we made recordings of the output of hearing aids driven with a structured noise sequence whose level transitions were statistically controlled. We found that hearing aids affect spectral cues as well as interaural level and temporal envelope differences. Taken together with recordings made on a rotating KEMAR manikin, we demonstrate that hearing aids may interfere with a listener’s ability to process acoustical cues for self motion.

Keywords: Head movement, acoustic movement, sound localization, hearing aids, signal processing, compression
Hearing impairment, hearing aids, and cues for self motion

1 Introduction

We have previously demonstrated that moving a sound at twice the rate of any head rotations results in a front-to-back or back-to-front spatial illusion [1]. This works because head movements can be used to disambiguate front/back confusions. If a listener turns 10° to the right, then a front-located sound source will move 10° to the left of the listener, whereas a rear-located sound source will move 10° to the right. Thus by turning the head and noting the direction in which a sound moves, one can determine whether it is coming from the front or the back. Using motion tracking to smoothly pan a signal around a ring of loudspeakers as a function of head movement makes it possible to artificially maintain these geometric relationships for signals in the opposite hemifield. The resulting illusion [2], is powerful and largely inescapable provided the signals are low pass filtered. Once high frequency energy is added, however, the pinna cues to sound location directly conflict with those indicated by the dynamic movement cues. That is, the salient spectral cues to location [3] indicate a direction opposite to that indicated by how the signal appears to move relative to the head. Thus it is unsurprising that for normal hearing listeners, the salience of the illusion appears to decline as high frequency energy is added.

For hearing impaired listeners, on the other hand, while the illusion tends to be weaker overall, adding high frequency energy typically does not result in a change in its apparent strength [4]. The most parsimonious explanation for this is that many hearing impaired listeners do not have access to the high frequency pinna cues that cause the confusion, and rely instead solely on dynamic motion cues for front/back location. This reliance on movement makes the front/back illusion a reasonable test of the ability of listeners to use the relationship between their own head motion and the resulting movement of an acoustic source to establish the front back location of the signal [4]. To ideally integrate one's own motion into spatial auditory perception requires that three conditions be satisfied: 1) that the listener have sensitivity to and the ability to use high frequencies where spectral pinna cues are most informative to spatial location, 2) that the listener be able to accurately determine the way in which binaural cues change over time, and 3) that the listener have accurate information on their own movement. Hearing impairment is associated with degradation in all three of these conditions: typically a decrease in sensitivity to high frequency sounds [5], poorer processing of binaural cues [6, 7], and often co-morbid vestibular and balance disorders [8]. With these requirements in mind, we used the front/back illusion as an assay of a hearing aid user’s ability to bind together what is fundamentally a multi-sensory process.

Generally speaking, hearing aids are designed to increase audibility, with spatial fidelity being a lesser objective. A few studies show some benefit in spatial localization performance [9-11] particularly when using open fittings [12, 13] or completely-in-the-canal aids [14, 15]. Other studies, however, show minimal improvement or even a decline in localization performance [16-18]. Given the heterogeneity of the findings in the literature, perhaps the only unambiguous
conclusion that can be drawn is that hearing aids do not return a hearing impaired listener to normal performance in spatial hearing tasks [4, 14, 19].

Typical hearing aids have their microphones situated behind-the-ear, outside the pinna and away from the ear canal, so while they increase the audibility of high frequency cues, the cues themselves may not be relevant and informative for spatial location. Another potential issue for sound localization is that most hearing aids use dynamic range compression. Compression is used to allow the level of quiet signals to be increased to assist with audibility, but reduce or not further amplify signals with levels high enough to cause discomfort. The vast majority of currently available hearing aids have compressors that are driven only by the signals arriving at their own microphone(s). For a binaural fitting, this means that the left and right independently acting compressors may misrepresent binaural cues [20]. For example, a high level signal on the left side of the head might engage the compressor in the left hearing aid but (because of the head shadow) may not be of sufficient level to engage the compressor in the right aid. This would lead to a reduction in the ILD of the received signal, and therefore would suggest a source location closer to the midline than its true position: a distortion of auditory space [21]. With these static spatial distortions in mind, it is reasonable to suppose that the spatial distortion of moving signals would be at least as impacted, if not more so.

Given observed changes in localization and orienting performance while wearing hearing aids [22] and the minimal data available on the acoustic artefacts introduced by head movements and hearing aid processing, we conducted an experiment on an acoustic manikin (KEMAR) in a hemi-anechoic chamber, comparing the acoustic output of a pair of hearing aids to the output of a pair of in-ear reference microphones. Recordings were made of speech-shaped noise, varying in level according to a statistically controlled sequence, and presented from a stationary loudspeaker. We also made recordings while the manikin was oscillated between ± 90° and used real-time motion tracking to trigger changes in level at specific angles. We analysed these recordings and measured hearing-aid-induced changes in ILD and envelope ITD as a function of level and source angle, as well as those changes introduced by movement of the manikin.

2 Methods

2.1 Signal Capture

A KEMAR manikin was fitted with a pair of hearing aids (representative of those fitted by the NHS in the United Kingdom) with open fit tulip domes, set to omnidirectional microphone mode. The fitting was based on the 5dB rounded left-ear values of the average audiogram of a hearing impaired listener [5] and the hearing aids were fitted by an NHS audiologist according to standard practice. The resulting gain structure and compression settings were typical for a hearing aid fitted to an average hearing impaired user (See Table 1 below).
The outputs of these hearing aids were recorded with B&K 4189 microphones fitted in the manikin’s ear canals using Zwislocki couplers and attached via B&K UA-0122 flexible right-angle adaptors to B&K 2671 pre-amps and then a B&K 2672 amplifier. The signals arriving at the ear canals directly were recorded using a pair of in-ear microphones (The Sound Professionals SPTFB-2). All four signals were captured using a Ferrofish A-16 AD/DA and an RME MADI PCIe interface. All signals were played and recorded via Matlab using the open source dynamic link library “playrec” (www.playrec.co.uk) through a calibrated acoustic presentation system [4].

### 2.2 Signal Generation

For part one, we generated a continuous speech-shaped noise signal that was changed in level every 250ms. The levels could be any one of 9 levels from 50 dB to 90 dB in 5 dB steps (in accordance with ANSI S3.22-2003), the presentation order of which was determined by a 9^3 deBruijn sequence [23]. The statistical properties of the sequence ensured that each pairwise level transition occurred exactly nine times for each signal. This allowed us to analyse individual transitions embedded in the ongoing signal. Level changes were smoothed with a 10ms raised cosine crossfade. The signals was played at angles of 0°, 30°, 45°, 60°, 90°, 120° and 180° (with positive angles indicating the right side of the manikin). For part two, we also used speech-shaped noise, but the signal was triggered to change in level when the manikin reached a predetermined trigger angle. The signal was double to triple buffered in 15 msec chunks, and the level was toggled between 50 and 70 dB as the manikin was swept past a trigger angle of either 0°, -45°, or +45° (in three separate conditions). Overall system latency plus buffer duration meant that the latency from trigger angle to a change in level varied between 31 and 46 ms. This time smearing (although not common) required analysis of ILDs to be restricted to bins of 15 ms, unlike for part one where these comparisons could be made on a sample-by-sample basis.

### 2.3 Motion Tracking

Motion tracking was performed using a 6-camera infrared Vicon system (MX 3+ with Ultranet) and a head-mounted crown of retroreflective markers. The position of the manikin’s head was sampled at a rate of 67 Hz to within an error of less than 0.25° via a dynamic link library in Matlab. For the recordings made during motion, the experimenter sat behind the manikin which was strapped to a rotating chair and turned the manikin in a practiced and controlled fashion closely approximating a sigmoidal angular trajectory between -90° and +90°.

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Table 1: Band-by-band HA Compression Settings

<table>
<thead>
<tr>
<th>Frequency [Hz]</th>
<th>250</th>
<th>500</th>
<th>750</th>
<th>1K</th>
<th>1.5K</th>
<th>2K</th>
<th>3K</th>
<th>4K</th>
<th>8K</th>
</tr>
</thead>
<tbody>
<tr>
<td>Attack Time (ms)</td>
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<td>14</td>
<td>16</td>
<td>14</td>
<td>16</td>
<td>13</td>
<td>13</td>
<td>10</td>
<td>11</td>
</tr>
<tr>
<td>Release Time (ms)</td>
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<td>22</td>
<td>24</td>
<td>46</td>
<td>56</td>
<td>51</td>
<td>55</td>
<td>50</td>
<td>26</td>
</tr>
<tr>
<td>Compression Ratio</td>
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<td>1.5</td>
<td>1.5</td>
<td>1.9</td>
<td>2.0</td>
<td>1.8</td>
<td>2.0</td>
<td>1.6</td>
<td>1.2</td>
</tr>
</tbody>
</table>
3 Results

3.1 Static Manikin Recordings

Figure 1 shows the reference microphone and hearing aid outputs (left in black, right in orange) during an isolated transition in signal level (left two panels), in this case, the mean recorded output for a signal at 45° transitioning from 50 to 90 dB. The reference microphone followed this change from 50 to 90 dB faithfully (left), while the HA amplified the level of the lower signal (right). The right two panels show the ILD computed from these recordings. As can be seen, the ILDs for the reference microphone do not change as the signal level changes. The hearing aids, however, initially exaggerate the ILD at 50 dB which then drops after the transition to 90 dB. There is a large momentary spike in ILD seen in the HA output at the transition point. This can be thought of as a distortion in the envelope ITD. Generally, because the magnitude of static ILD distortions changes as a function of signal level change and angle, we quantified the distortions seen at all possible level transitions (mean ILD measured between the white arrows in the right hand panel).

The change in hearing aid output ILDs over the transitions of the 9 possible levels at 3 signal angles are displayed in Figure 2 as heatmaps. The leftmost panel shows the ILD changes at 0° presentation angle, the center panel at 45°, and the right panel at 90°. The relatively homogeneous field seen in the 0° panel is consistent with the fact that these signals arrive at roughly equal level to the two ears, meaning the compressors would act more or less in parallel with each other. Increasing differences are found at 45° and 90°, where the signals more strongly engage the right-side compressor. The largest differences occur between transitions across the 55 to 60 dB border.
3.2 Rotating Manikin Recordings

Figure 3 shows the changes in ILD for a signal presented at 0° over the course of a left-to-right movement of the manikin from -90° to +90°. Reference microphone outputs are in blue, hearing aids in orange. The ILDs recorded from the reference microphones changed in sigmoidal manner as KEMAR rotated from left to right for both 50 dB (left) and 70 dB signals (right). The hearing aid ILDs did not change in a similarly monotonic way. There is an artefact late in the trajectory as the hearing aid ILDs change in the opposite direction from those recorded from the reference microphone (see the ΔILD reversal from black to blue arrows) for both 50 and 70 dB signals.

For the next set of trials, a signal at the 0° loudspeaker was incremented from 50 to 70 dB when the manikin passed a trigger angle (Figure 4). The left panel shows the ILD when the trigger point was at 0° (so that at the level change point, the signal arrived at roughly equal level to both hearing aids). No immediate effect of the level change was observed, but one can still observe an artefact similar to that seen in the constant level trials (see the ΔILD reversal from black to blue arrows). In the second panel, the trigger was -45°. In this case the signal was +45° off the acoustic midline of the manikin when it changed. In this case there appears a small change in ILD immediately after the level increment (white to blue patch). In the third panel, the trigger was +45°. In this case, a transient compression-induced ILD artefact may be seen. This spike may also be seen in the output of the reference microphones, indicating cross-talk from the hearing aid output (see Section 4.3.2).

Figure 5 shows ILDs in response to a decrement in signal level at the 0° loudspeaker as the manikin was rotated right to left. A similar departure from linearity is observed here, consisting of an apparent counter-rotation of the ILD-implied signal location between the black and blue arrows.
For a trigger angle at -45°, we see a similar pattern initially, but in this case, a transient compression-induced ILD artefact may be seen. A similar ILD distortion may be seen in the right panel as with a trigger point of +45°. In addition to distortions over a longer time course, these transient changes in ILD may also negatively impact the spatial representation of a signal changing in level as the head is moving.

![Figure 5: ILD during head movement: level decrement](image)

4 Discussion

4.1 Overview

We measured the fidelity with which current hearing aids maintain spectral and dynamic movement cues available to the unaided listener. Unaided, the ILDs recorded at the ear canals of KEMAR faithfully followed those expected by modelling, with statically presented signals maintaining the same ILD over all possible changes in level. When KEMAR was rotated, the ILDs of the unaided signal changed in a similarly expected sinusoidal fashion. When recording from hearing aids we found that ILD and envelope ITD were different, potentially due to attack and release of compression. The differences were large enough to have perceptual consequences, and could be evoked by signals whose dynamic range was similar to that of natural speech signals. These distortions changed as a function of both signal level and signal-to-listener angle. The idiosyncratic nature of such acoustic artefacts makes them unpredictable and likely unlearnable. That is, a listener may not be able to adjust or compensate for the distortions in spatial cues because that would require a priori knowledge about the location and level of the signal, information that may not be available to the listener.

4.2 Perceptual Consequences

The change in ILD seen as a function of manikin angle was not monotonic over the course of a head movement. Indeed the signal movement suggested by the ΔILD over certain angles was sometimes opposite to that of the unamplified signal (see between black and blue arrows in Figures 3, 4, and 5). For example, the ILD at a trajectory point of 70° (blue arrow) matches that of the ILD at 10° (black arrow), which would indicate a counter rotation of about 60°. The perceptual consequence of this reversal in the direction of ILD change would be a signal that during this time period would appear to be rotating with and faster than the head. Such a movement has the potential to produce front/back confusions, as has been previously
demonstrated [1, 2, 4, 24]. Because, however, the change in ITD during this period of movement should, in principle, be unaffected by hearing aid processing, what would result is a scenario in which ΔILD indicated a location to the back, ΔITD indicated a location to the front, and spectral cues were largely uninformative to front/back location. The resulting ambiguity about front/back location could cause the decrease in the apparent illusion salience frequently observed in users of hearing aids [4], as well as confusion about the location of static front or back signals.

4.3 Caveats

4.3.1 ILD versus ITD

We do not currently know whether the front/back illusion is driven more strongly by changes in ILD or ITD. The illusion is strongest when signals are low-pass filtered below 500 Hz, suggesting a primary role of ITD in the process, but to date no studies have determined the perceptual weighting of ITD vs ILD in the front/back illusion. Given that in some instances during a head movement ΔILD and ΔITD can indicate signal locations in the opposite hemifield from one another, one possibility is the percept of multiple auditory images. Pöntynen et al [24] describe such a multiple image percept in their work on the front/back illusion. Moving head tracked full band signals that contained high frequency pinna cues resulted in listeners experiencing two signal locations, one at the front indicated by one cue and one at the back indicated by the other.

4.3.2 Signal Cross Talk

It should be noted that our method of simultaneous recording of hearing aid output and acoustic signals arriving at the ear likely produced cross-talk. The hearing aids were fitted with open tulip-domes, which would allow some of the amplified signal to reach the in-ear microphones fitted to KEMAR. This technique was employed to ensure that the motion of the manikin was identical for the acoustic signal and amplified signal, permitting us to directly compare the two sets of recordings, but this did prevent us from capturing the unadulterated air-borne acoustic signal. Without the use of a robotic arm or programmable turntable, however – neither of which tend to be silent – a truly uncontaminated and yet movement-matched signal comparison would not be possible. Because the amplification crosstalk results in a mix between the undistorted acoustic input and the amplified input, however, we argue that at worst we are underestimating the true ILD distortions produced by hearing aids on a moving listener.

4.3.3 Band-by-band Analysis

We have not analysed our data for changes in ILD within frequency bands. Typical hearing aids use multi-band compression, applying changes in level only to a portion of the spectrum. We used speech-shaped noise to drive our hearing aids, and such a signal is not as strongly amplitude modulated within frequency bands as speech is. As such we may be underestimating the changes in level that occur within particular frequency bands, but further analysis will follow.

4.4 Linked vs unlinked compression

While this study may be the first to describe compression-induced ILD distortions in moving listeners, it is certainly not the first to examine ILD distortion. ILD cues for statically presented
have previously been shown to be altered by the use of binaurally unlinked compression, leading to a corresponding change in ILD discrimination performance [25]. The alternative is the use of bilaterally linked compression, in which two aids communicate their compression values in quasi realtime and the maximal compression value is used for both. This has the potential to maintain ILD cues, while still employing dynamic range compression for level adjustment. Such a technique has been shown to improve speech perception in noisy, spatially complex environments, at least for normal hearing listeners [26].

4.5 Self-motion Awareness

As a final note, it should be acknowledged that in order to ideally utilize self motion for spatial perception, one must not only have robust information on dynamic spatial acoustic cues undistorted by amplification, one must also have robust information on one’s own movement. Thus the function of the vestibular, proprioceptive, and motor-planning systems also play an integral role in self-motion processing, changes in any of which may interact with signal direction, hearing aid signal processing, and microphone directionality.

4.6 Conclusions

We have documented distortions in the ILDs of an ongoing amplitude modulated signal that are introduced by the use of dynamic range compression in hearing aids. These distortions depend on both the angle of the signal relative to the head and changes in level of the signal, rendering them unpredictable without prior knowledge of the signal and its location. The unpredictable nature of the ILD distortions makes it unlikely that a listener could learn and/or adjust to them.

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5 References


