In-situ compression for vented fittings

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The present work contains a simulated analysis of the properties behind the mixing of processed and direct sound at the eardrum. Specifically, the achieved in-situ, input-level dependent eardrum gain of compressive hearing aids with vented fittings is investigated and compared to the prescribed gain dictated by the fitting rationale. The results show potentially huge discrepancies between the achieved gain and the intended gain at the eardrum, due to the mixing of the processed and the direct sound. The results generalize to open and leaky fittings.

INTRODUCTION

One of the cornerstones of hearing aid (HA) fitting is acoustic accuracy. Without control of the sound at the eardrum, there is a risk that the HA user is insufficiently or overly compensated for their hearing loss, and well researched and optimized hearing aid signal processing features may lose power. Deficient management of the acoustics may thus increase the variation in the HA user's performance and satisfaction.

A central challenge in fitting hearing aids is therefore to ensure that the gain at the individual eardrum matches the gain prescribed by the fitting rationale. This task is not trivial, since the sound pressure at the eardrum depends on inter-individual variations in the anatomy of the ear and particularly on the openness of the acoustic fitting to the ear.

Variations in the processed HA sound pressure at the eardrum may be corrected for by a real ear to coupler difference (RECD) measurement (Moodie *et al*., 1994). However, as Hoover *et al*. (2000) indicated, it is not sufficient only to consider the processed sound. The direct sound transmitted directly through an open, vented or loose earmold also plays a significant role.

The goal of this study was therefore to make a systematic analysis of the interaction of the processed and direct sound at the eardrum in a vented *in-situ* fitting.

GENERAL SUMMATION OF HARMONIC SIGNALS OF SAME FREQUENCY

When two sound sources add at a point in space, the resulting sound pressure at each frequency depends on the ratio between the amplitudes of the two sources and the difference in phase. The sum of two harmonic signals can be written as:

$$s = A_1 \cos(2\pi ft + \phi_1) + A_2 \cos(2\pi ft + \phi_2) \qquad \text{(Eq. 1)}$$

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where A denotes the amplitudes of the harmonic signal, f the frequency, t the time and φ the phase. Both A and φ may be frequency dependent. The sum of two signals may be expressed as:

$$s = \sqrt{A_1^2 + A_2^2 + 2A_1A_2\cos\Delta\phi} \cdot \cos(2\pi ft + \Theta) \qquad \text{(Eq. 2)}$$

$$\Theta = \arctan\left(\frac{A_2 \sin\Delta\phi}{A_1 + A_2\cos\Delta\phi}\right) \qquad \text{(Eq. 3)}$$

Here $\Delta\varphi = \varphi_2 - \varphi_1$. This rewriting of the equation shows that the resulting waveform is a phase shifted harmonic with the same frequency as the two components. The amplitude of the resulting sum depends on the amplitudes of the signals and the phase difference between the two. In order to uncover specific details about the properties of adding two harmonic sources, the following sections will study the implications of this equation in detail.

The general phenomenon is illustrated in Fig. 1, which illustrates the addition of two generic harmonic signals with differing magnitude and phase. For the purpose of illustration, A_1 = 0 dB and $A_2 \propto f$. $\Delta\varphi$ is either 0 (in phase), π (in antiphase) or $\propto f$ (varying proportional to frequency, equivalent to a delay).

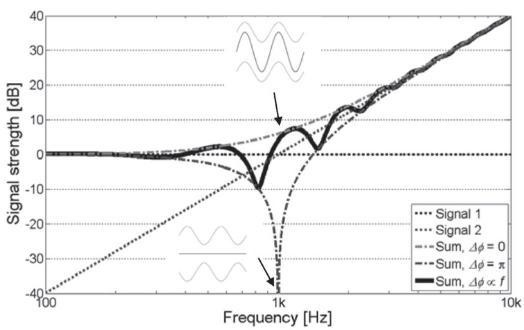

Fig. 1: Illustration of the phase disruption phenomenon. Inserts illustrate constructive (in-phase) and destructive (antiphase) interference.

The upper dash-dotted gray line in Fig. 1 shows the resulting amplitude in the case where the two signals have a fixed in-phase relationship ($\Delta\varphi$ = 0), whereas the lower dash-dotted line represents the antiphase situation. Under the condition that the signals are in antiphase and have the exact same amplitude, the total signal cancels out and becomes infinitely small. This is known as destructive interference or *phase cancellation*. On the other hand, if the two signals are in phase, the amplitudes add constructively. This yields a 6 dB higher result at that frequency since the two input signals also have the same amplitude. This can be seen in the upper dash-dotted gray line at 1 kHz. The two dash-dotted curves define the interval in which the sum of two harmonic signals may lie, depending on the phase relation.

Digital hearing aids have an inherent processing delay, which gives a frequency-dependent phase. The thick black line in Fig. 1 therefore exemplifies how the total sound pressure might look like, if the relative phase depends linearly on frequency. At frequencies where the relative phase is almost π and the relative amplitude is not quite 0 dB, the signals do not cancel out as such. Therefore this phenomenon may be called *phase disruption* (as opposed to phase cancellation).

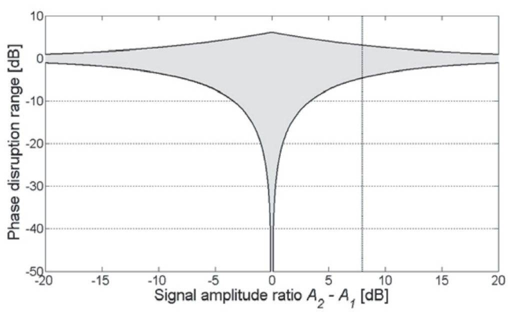

Fig. 2: The phase disruption range in which the sum of two harmonic signals may lie is shown by the gray area. In this plot, 0 dB corresponds to the maximum amplitude of the two signals.

Figure 2 shows the general amplitude range in which the sum of the two harmonic sources lies. The figure reveals that if one signal is, e.g. 8 dB more powerful than the other signal (as indicated at the vertical line), the sum of the two may lie between 4.4 dB less and 2.9 dB more than the most powerful signal. The resulting signal strength in this case is determined by the phase relation between the two sources.

IN-SITU ACOUSTICS IN HEARING AIDS

This section describes the relevance of the general analysis to HA acoustics. In order to address this question, the in-situ hearing aid, tubing, ear canal and eardrum (as illustrated in Fig. 3) has been simulated by use of a transmission line model approach, by which any transfer function in a given acoustic plane wave system, such as the in-situ hearing aid, can be accurately calculated (Egolf, 1980).

In HA acoustics, the two main sound sources influencing the sound pressure at the eardrum are the *processed* and the *direct* sound. The processed sound is played back by the HA receiver. The insertion gain of the HA by itself is denoted the real-ear hearing aid gain (REHAG) during this study. The REHAG is the target gain as defined by the fitting rationale. The direct sound is external sound transmitted directly through any intentional (vent) or unintentional (leakage) sound path through or around an earplug. The spectral shaping of external sounds through this acoustic pathway is characterized by the real-ear occluded gain (REOG). Both the REHAG and the REOG are insertion gains, i.e. the measured eardrum sound pressure relative to that of the unoccluded ear. The total gain, i.e. the absolute sum of the REHAG and the REOG, is characterized by the in-situ real-ear insertion gain (REIG), which is frequently measured in the clinic to validate the accuracy of the fitting.

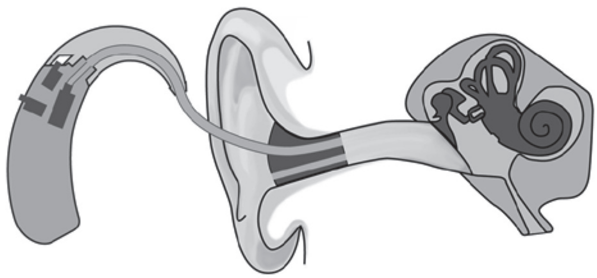

Fig. 3: Illustration of a ventilated in-situ HA system. The vent is connecting the residual volume to the outside.

Both the processed and the direct sound are greatly influenced by a vent. In essence, the vented in-situ HA forms a Helmholtz resonator. The Helmholtz resonator may be excited by either the HA receiver or by external sound, and is characterized by the Helmholtz frequency at which the system resonates.

The processed sound is affected by the so-called *vent effect*, which is defined as the processed sound pressure at the eardrum for a vented earmold relative to that of a fully occluded earmold. The vent effect (VE) is depicted in Fig. 4 for different vent diameters, neglecting any leakage for simplicity. The figure shows that low frequencies of the processed sound are attenuated at the eardrum, whereas high frequencies are unaltered. The vented REHAG is the sum in dB of the prescribed gain and the VE.

Figure 4 also shows the REOG for the same vent diameters. High frequencies of external sounds are attenuated whereas low frequencies are passed through. Furthermore, the Helmholtz resonance provides a small boost of sound at the same frequency in both the VE and the REOG. The amplitude of the Helmholtz resonance depends on acoustic losses in the in-situ HA system. Typically, the amplitude of the resonance ranges between 4-6 dB for in-situ vented earmolds, but may go as high as 10-12 dB in couplers and even disappear completely for leaky, closed in-situ earmolds. See also Dillon (2001) for more information.

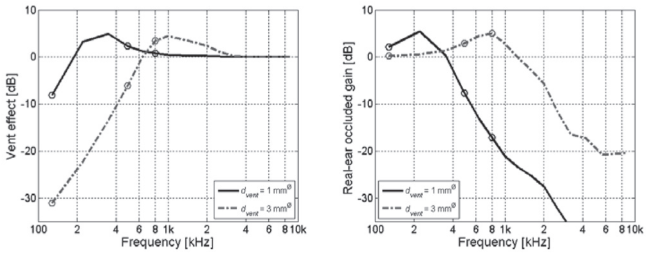

Fig. 4: The vent effect (left panel) and the REOG (right panel) as function of frequency exemplified for two vents with different diameters and equal lengths. Values indicated in the plots with circles at f = 125 Hz, 500 Hz and 800 Hz are used in Fig 6.

Figure 5 shows a simulated example of the mixing of the REOG and the vented REHAG at the eardrum. The HA is fitted with a flat 30 dB hearing loss, and the plug has a vent of 3 mmØ. The simulated REIG, shown by the solid line, is calculated at a certain input level, assuming that the relative phase of the two signals is proportional to frequency. The shape of the solid line is well known to clinicians, and is a signature of a vented or open fitting. The REIG clearly shows peaks and valleys in the frequency region where the two sources are similar in amplitude. This phase disruption phenomenon has the characteristics of a comb filter in the frequency range where the amplitude difference is small. Perceptually, this so-called coloration effect alters the timbre of the sound, and may be perceived as a degradation of sound quality in comparison to a closed fitting.

As an example of the nature of the phase disruption phenomenon, it may be observed in Fig. 5 that a 10 dB **increase** of REHAG at 300 Hz would ironically **decrease** the in-situ gain at that frequency, since the REOG and the REHAG are in antiphase at this particular frequency. Furthermore, Fig. 5 shows that in the case where one source is much stronger than the other, the phase disruption phenomenon is negligible. In that case, the stronger source dominates the resulting sound pressure.

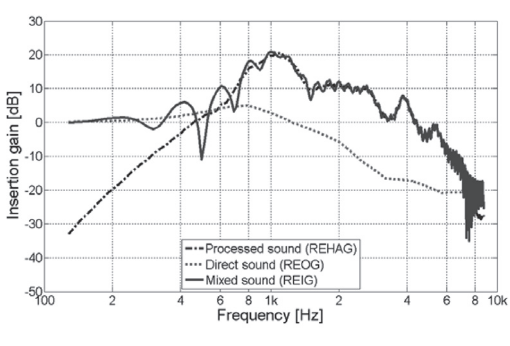

Fig. 5: Simulated example of the mixing of the REHAG and the REOG at the eardrum for a flat 30-dB-HL and a 3-mmØ cylindrical vent.

IN-SITU COMPRESSION CURVES

For compressive hearing aids, the phase disruption phenomenon becomes particularly interesting since the relative amplitude of the processed and direct sound varies with input level.

The prescribed gain could be calculated using any generic fitting rationale. In this study, the prescribed gain was calculated by use of a constructed but realistic fitting rationale under the standard assumption that the earmold is fully occluding, and that the REHAG is not negative. The prescribed gain was subsequently modified by the simulated vent effect (resulting in the vented REHAG) and added to the REOG, both provided in Fig. 4. The upper and lower limits of the phase disruption range are obtained by adding the vented REHAG and the REOG in both in-phase and in antiphase. The achieved REIG lies somewhere in this range depending on the precise phase relationship between the REHAG and the REOG.

Figure 6 shows a significant input-level dependent discrepancy between the prescribed gain (thick line) and the actually achieved gain at the eardrum (somewhere within the gray achieved gain range). For condition 1, where the hearing loss is mild (30 dB HL flat) and the vent large (3 mm^{Ø}), the discrepancy is significant up to at least 800 Hz. Even for condition 2, where the hearing loss is moderate (60 dB HL flat), and the vent is small (1 mm^{Ø}), the achieved gain is significantly influenced by the direct sound at low frequencies. The fitting target is only met accurately in condition 2 for 800 Hz, where the achieved in-situ gain range is narrow and lies on top of the prescribed gain in the entire input level range. At other conditions and frequencies, the achieved gain is either too high due to the Helmholtz resonance, too low due to the vent effect or too uncertain due to the phase disruption phenomenon.

White input-gain curves for specific phase differences are superposed on the gray achieved gain ranges in Fig. 6. These curves exemplify how the actual gain may look like as a function of input-level at the particular frequencies. They show that there is a risk of local minima in the achieved gain at certain input levels and frequencies, e.g. at 500 Hz for condition 1.

One of the main findings in Fig. 6 is in fact that it is not necessarily advantageous to compensate the REHAG to equal the prescribed gain for vented fittings, which is the main purpose with the real ear to coupler difference (RECD). This may be observed for condition 1 at 125 Hz, where the prescribed target of 0 dB gain is met at input levels higher than some 40 dB due to the REOG dominance. At input levels higher than 40 dB, the phase disruption range would actually increase drastically in the imagined case the REHAG is compensated to meet the prescribed gain. This increase in uncertainty in eardrum SPL arise from the fact that the compensated REHAG equals the REOG in this input range, thus increasing the risk of phase disruption. These findings are consistent with the conclusions by Hoover *et al*. (2000).

Figure 6 also shows that under some conditions, e.g. condition 1 at 800 Hz, a higher gain than prescribed is achieved at the eardrum. This is due to the Helmholtz resonance that gives a boost to the processed sound. Application of an RECD measurement would actually remove this Helmholtz resonance boost, but the vented REHAG would still equal the REOG at some level, so the phase disruption problem is not solved by the RECD alone.

As Fig. 6 shows, the instantaneous gain of a compressive HA is dependent on the time-varying input level and the non-linear input-gain curve at each frequency. This means that the frequency range and strength of coloration will vary with input level. The sound quality of a vented HA may thus be substandard for certain input levels, whereas it may be excellent for other input levels, depending on the gain of the HA and the ventilation of the plug. Therefore, it is important to take the in-situ acoustics into account during the design phases of experimental evaluation setups of HA features, where phase disruption and direct sound dominance may very well explain some of the variance from subject to subject experienced in the evaluation.

DISCUSSION AND CONCLUSION

The resulting acoustic effects of a vent may be subdivided into three cases, depending on the specific combination of vent size, hearing loss, frequency and input level:

1) The HA has full control over the sound, meaning that the processed sound is dominant and the direct sound is negligible in comparison. In this case, the user is provided with sufficient gain, and the perceived sound quality and the effect of HA features like noise reduction algorithms are limited only by the quality of the implementation. This case is largely predominant for severe hearing losses and at high frequencies.

2) The HA has no control over the sound. This is caused by the fact that the processed sound is too soft in comparison to the direct sound, either due to the vent-related attenuation of the hearing aid sound, a low gain prescription or a combination of the two. In this case, HA compression and other signal processing have no effect, since the vent conducted sound dominates. This is commonly utilized to ensure the intended 0 dB gain at low frequencies for open fittings, which are primarily used for hearing impaired with normal low frequency hearing.

3) The mixing of the processed and direct sounds results in a comb-filtering effect, which introduces a largely unpredictable, level-dependent coloration of the sound. This phase disruption effect is almost inevitable in some shape or form in realistic fittings. It has the effect that the sound quality may depend on signal level, and the effect of signal processing features such as directional microphones or noise reduction systems are potentially diminished.

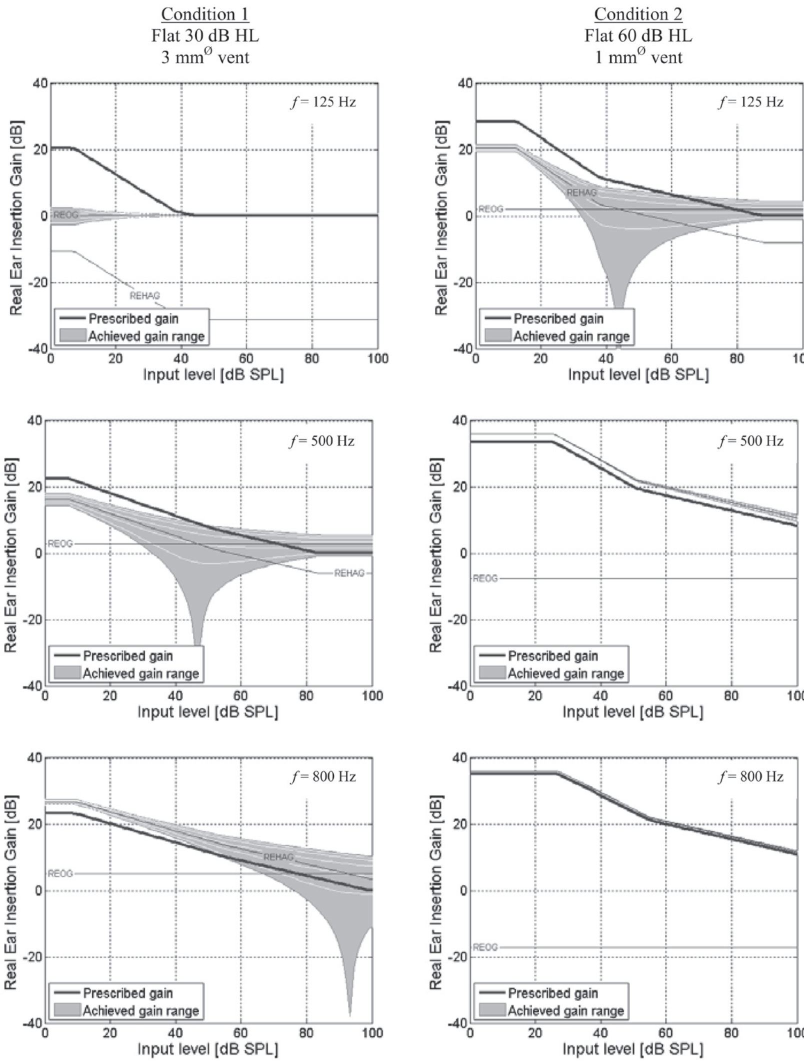

Fig. 6: The potential in-situ REIG range at three different frequencies with mild hearing loss of 30 dB HL and a large vent of 3 mm^Ø (left panels) and moderate hearing loss of 60 dB HL with a small vent of 1 mm^Ø (right panels). The prescribed gain is marked by a thick line and the REIG at the eardrum lies somewhere within the gray achieved gain range, and the specific shape depends on the relative phase at each frequency. The achieved gain range is superimposed with white lines indicating the REIG for specific phase differences of 0 to π in steps of π/6.

The conclusions of this study are general for vented or open fittings, and may even be extrapolated to closed earmolds, where slit leakage is very hard to avoid. Even though attention has been given to specific hearing losses with specific vent sizes and for a specific fitting rationale, the underlying mechanisms are the same for any non-tight earmold. This is both relevant with regards to sound quality during hearing aid use, but also important to consider in various research and development setups particularly with regards to HA feature evaluation.

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