Active Control in Hearing Aids reducing low-frequency variations

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Abstract

One of the major challenges in fitting hearing aids are substantial variations in the real-ear responses between subjects as well as differences between reinsertions of the hearing aid within one subject. In mid and low frequencies, these variations mainly originate from untight fits leading to leakage. To some extent, they are also due to variations in the individual acoustic impedance (residual ear-canal volume and middle ear compliance) and uncertainties regarding vent geometry and transducer roll-off.

The traditional approach would be an in-situ measurement along with a static compensation. However, the acceptance of this solution by hearing aid professionals is limited because it is time-consuming. In addition, the insertion of a probe tube typically influences the leakage, which leads to potentially inaccurate measurements at low frequencies.

A novel solution is the use of active control with a microphone inside the ear-canal. A classical closed-loop feedback control approach not only reduces noise such as bone-conducted own-voice leading to the occlusion effect, but also decreases the spread in the real-ear responses.

The presented results show considerable reductions in variations in the real-ear responses of a typical configuration. The simulations are verified by means of measurements with normal-hearing subjects.

Introduction

The application of active closed-loop control in hearing aids to reduce the occlusion effect has long been suggested by both academic publications [1] and various patent applications. A servo-control approach according to Fig. 1 has the potential of attenuating any contribution to the sound pressure in the ear-canal sensed by the canal-microphone not originating from the hearing aid, e.g. bone-conducted own-voice. This approach inherently also reduces the low-frequency spread of the real-ear responses, which is equivalent to the spread in the plant P (electrical receiver input to electrical canal-microphone output) for low and mid frequencies (below some kHz).

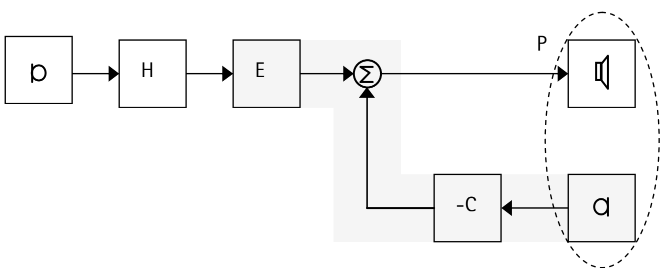

Fig. 1: Block diagram of an Active Control Hearing Aid.

The effective sound pressure reduction in the ear-canal equals the sensitivity S of the closed loop and – in agreement with classic control theory basics such as presented in [2] – is given by

$$S = \frac{1}{1+P\cdot C}. \tag{1}$$

P and C are the respective transfer functions of the plant and the compensator, where all variables are assumed to be complex-valued and frequency dependent. In order to ensure a flat transfer function for the forward path (processed output of block H to electrical receiver input) the pre-equalizer E is required to invert the sensitivity S. Thus

$$E = 1+\overline{P}\cdot C, \tag{2}$$

where $\overline{P}$ is the pre-defined average of the plant measurements. The controlled plant P_{ctrld} results to be

$$P_{ctrld} = P\cdot\frac{1+\overline{P}\cdot C}{1+P\cdot C}, \tag{3}$$

$$\lim_{|C|\to\infty} P_{ctrld} = \overline{P}. \tag{4}$$

According to (4), an infinite compensator gain would lead to a perfectly controlled plant.

Simulation Model

A simple mathematical model for the plant P assumes

- low frequency roll-off: M-th order high-pass characteristic ($f_{HP,m}$) due to transducer roll-offs (pressure equalization) and leakage/venting
- high frequency roll-off: second-order low-pass characteristic (f_{LP}, ξ) due to the mechanical resonance of the receiver

as well as a pure time delay τ representing the plant death time and a frequency-independent gain k representing transducer sensitivities.

$$P = \frac{k\cdot e^{-j\cdot 2\pi\cdot f\cdot\tau}}{\prod_{m=1}^{M}\left(1-j\cdot\frac{f_{HP,m}}{f}\right)\cdot\left(1+j\cdot 2\xi\cdot\frac{f}{f_{LP}}-\frac{f^2}{f_{LP}^2}\right)} \tag{5}$$

This model neglects high-frequency effects due to the transducer tubings and variations in the individual acoustic ear impedance, assuming the receiver load impedance to be purely capacitive. Fig. 2 (upper sub-plot) shows simulated low-frequency variations of the 2nd-order roll-off ($M = 2$), representing e.g. changes in leakage or venting.

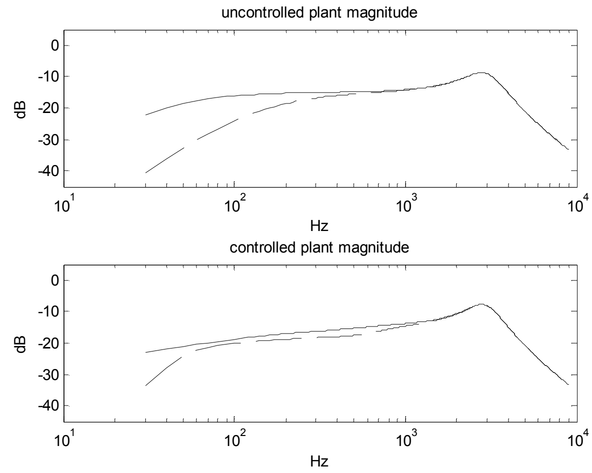

Fig. 2: Simulated spread in uncontrolled plant P (upper sub-plot) and in controlled plant P_{ctrld} (lower sub-plot).

According to (4), it would be desirable to maximize the compensator gain. However, gain and bandwidth are limited by stability constraints and can, to some extent, be traded for each other. The classical control theory toolbox provides phase-lag elements (also referred to as low-shelf filters) as compensators.

$$C = k_C \cdot \frac{f_z}{f_p} \cdot \frac{1 + j \cdot 2\xi_z \cdot \frac{f}{f_z} - \frac{f^2}{f_z^2}}{1 + j \cdot 2\xi_p \cdot \frac{f}{f_p} - \frac{f^2}{f_p^2}} \tag{6}$$

In (6), f_p and f_z are the cut-off frequencies, while ξ_z and ξ_p represent the respective damping factors.

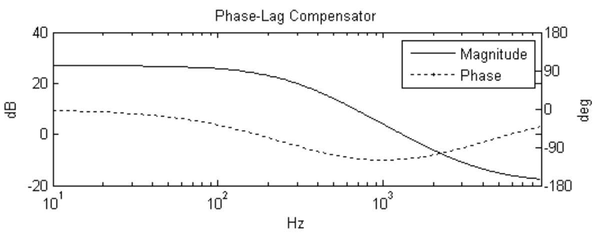

Fig. 3: Typical phase-lag compensator filter C ($f_z > f_p$).

Applying this simple 2nd-order compensator to the simulated plants yields the controlled plants in Fig. 2 (lower sub-plot). A considerable reduction in low-frequency variations is achieved. The slight increase in the plant variation at higher frequencies is associated with the overshoot in the sensitivity function S and can be reduced (or rather 'smeared' over a larger frequency region) at the expense of higher filter orders for the compensator C.

Measurements

The effect of the same compensator on measured plants in N=20 normal-hearing subjects is shown in Fig. 4. Here, the low-frequency spread is mainly due to variations in residual ear-canal volume and middle ear compliance, along with uncertainties regarding transducer sensitivities.

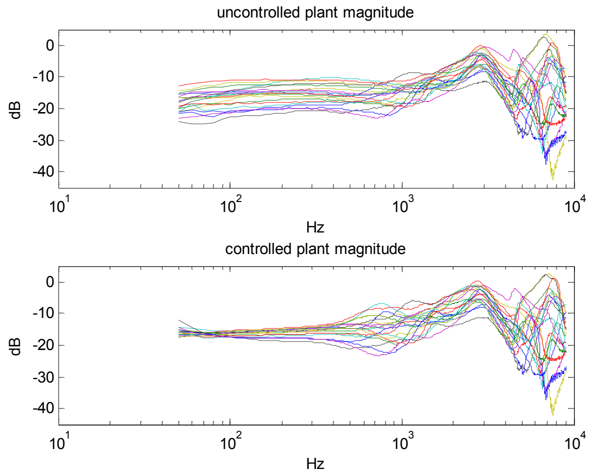

Fig. 4: Measured spread in uncontrolled plant (upper sub-plot) and in controlled plant (lower sub-plot), N=20.

Table 1 shows the absolute spreads in the uncontrolled and controlled plant.

Table 1: Variations in original and controlled plants

Variations	Frequency		
	100 Hz	500 Hz	1 kHz
original plant	12dB	10dB	11dB
controlled plant	2dB	10dB	12dB
Δ	-10dB	0dB	+1dB

Conclusions

A classical closed-loop feedback control approach does not only reduce undesired noise such as bone-conducted own-voice leading to the occlusion effect, but also substantially decreases the spread in the real-ear responses. The presented measurement results using a simple time-invariant 2nd-order phase-lag compensator show considerable reductions of up to 10dB at 100Hz in variations in the real-ear responses of a typical active control configuration.

Literature

[1] Mejia, J., Dillon, H., Fisher, M.: Active cancellation of occlusion: An electronic vent for hearing aids and hearing protectors. J. Acoust. Soc. Am. 124 (1), July 2008

[2] Elliott, S.: Signal Processing for Active Control. Academic Press, 2001