An Experimental Setup for the Estimation and Emulation of Bone and Air Conducted Components of One's Own Voice

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Introduction

Hearables have become ubiquitous in everyday life and include many more functionalities besides the classic hearing aid. While active signal processing highly depends on the device and application, the uniting feature of all hearables is that they are in-ear devices. As such, they obstruct the ear canal while worn, altering the user's auditory perception. Particularly affected by this is the perception of one's own voice. Under these circumstances, users often describe their voice as sounding "hollow" or "boomy" [1,2,3]. Among hearing aid users, this is known to be a significant source of dissatisfaction [3,4].

Von Békésy [5] identified two types of conduction relevant to the perception of one's own voice: Air conduction and bone conduction.

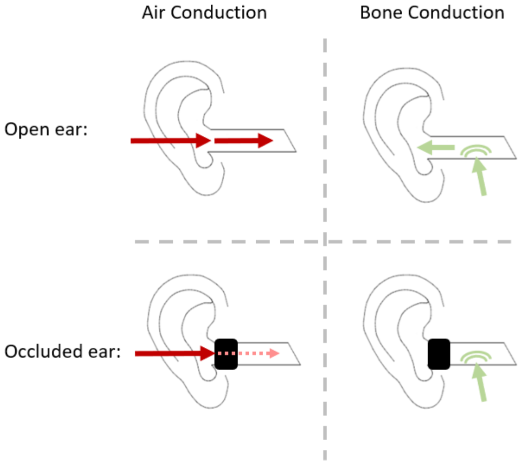

Figure 1: The effect of occlusion devices on air-conducted and bone-conducted sounds.

Air-conducted sounds propagate from the speaker's mouth through the air into their ear canals, including diffraction around the head and reflections from the speaking environment [6]. These sounds are affected by the *insertion loss*, as the occlusion device attenuates incoming airborne sound.

Bone-conducted components are transmitted by vibrations of the skull and surrounding tissues that couple into the auditory system at the outer, middle, and inner ear [7]. At the outer ear, vibrations of the ear canal walls act as a volume velocity source radiating sound into the ear canal. In the open ear case, low-frequency sounds transmitted by that mechanism dissipate outwards through the ear canal entrance. When an occlusion device is present, however, this outwards dissipation is (partly) removed, increasing the level at frequencies up to approximately 1kHz [2]. This is the *occlusion gain*.

Both effects are illustrated in Figure 1 and commonly aggregated into a single effect called the *occlusion effect*. However, investigating their perceptual influences on one's voice individually necessitates independent control over them. Yet, during vocalization, both effects always co-occur. Thus, there is a need for a setup that allows for separation and individual control of the insertion loss and occlusion gain in listening experiments.

In previous research, large ear muffs were successfully used to separate air-conducted and bone-conducted sounds [7,8]. Building on this approach, we aim to control the levels of insertion loss and occlusion gain independently by imposing real-time filtered voice into the ear muffs through internal loudspeakers. Effective separation requires attenuation of air-conducted sounds beyond a certain design threshold. As a result, the ear canal sound pressure is effectively only constituted by bone-conducted sounds to the outer ear. To prevent any coloration of these sounds compared to the open ear condition, it is essential to ensure that the ear muffs do not introduce any occlusion gain themselves. For this purpose, the literature states the ear muffs should have a large internal volume. Reported volumes range from 245cm^3 [8] to 6700cm^3 [9]. Careful consideration, however, showed that their inherent occlusion gain is only indirectly tied to the internal volume - that is, through the load impedance created by the ear muff. In this work, we present our approach to designing such an ear muff and preliminary experimental data on its properties.

Design Considerations

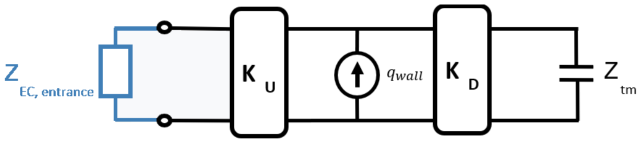

Figure 2: Lumped-element Model of the outer ear. Z_{tm} is the acoustic impedance of the tympanic membrane, $Z_{EC,entrance}$ the acoustic impedance of the ear canal entrance as seen from inside the ear canal, and q_{wall} the volume velocity source modeling the ear canal wall vibrations.

Figure 2 shows a simplified lumped-element model of the outer ear on which we based our design to minimize the occlusion gain. The tympanic membrane impedance is approximated according to Shaw and Stinson [10]. In its open state, the impedance of the ear canal entrance (open ear radiation impedance) resembles a mass at low frequencies. Occluding the ear or placing an ear muff over it alters this impedance. This impedance change is what causes the occlusion gain.

From the model, it becomes clear that the ear muff is not expected to introduce any occlusion gain if its load impedance can be matched to the open ear radiation impedance. To achieve this matching while keeping the dimensions as small as possible, we modeled the ear muff as a series of transmission lines with different cross-sectional areas representing the pinna, empty internal volume, and a section filled with acoustic foam. The model is depicted in Figure 3.

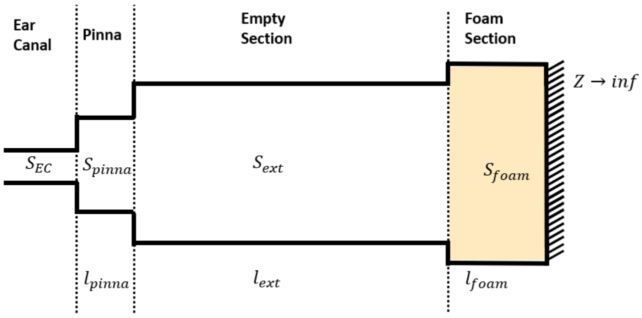

Figure 3: Simplified transmission line model of the ear muff.

With this model, it was possible to analytically calculate the load impedance the ear muff would impose and compare it to the open ear radiation impedance. The results are presented in Figure 4.

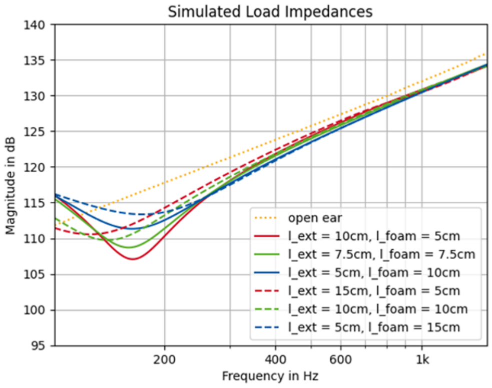

Figure 4: Simulated load impedances.

Based on these simulations, we 3D printed three prototypes, PT1, PT2, and PT3. They were designed as an extension to fit between parts of a commercial pair of ear muffs. All featured a length of 10cm. PT1 had an internal surface area of $53 cm^2$, while the other two were designed as two concentric shells, resulting in an internal area of $27 cm^2$. Furthermore, PT3 was printed with a thicker outer shell. Their actual load impedances were measured using the impedance tube approach proposed by Vorländer [11]. Figure 5 shows the results.

To obtain the achieved separation between air conduction and bone conduction in dB, we also measured the insertion loss created by all prototypes. This was done using a GRAS 45CA test fixture for sound incidents from the front. The design target was to achieve at least 20dB attenuation throughout the frequency range. Figures 6 and 7 show the resulting occlusion gain estimated using the model in Figure 2 and the measured insertion loss for all prototypes.

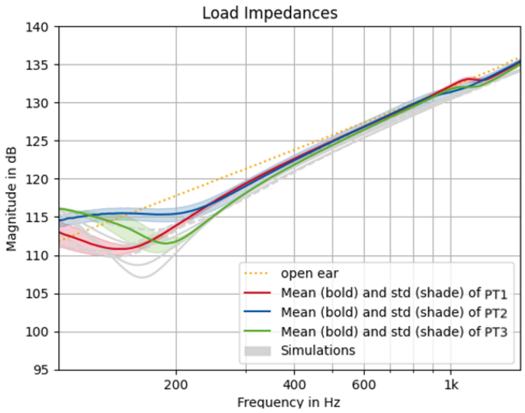

Figure 5: Measured load impedances.

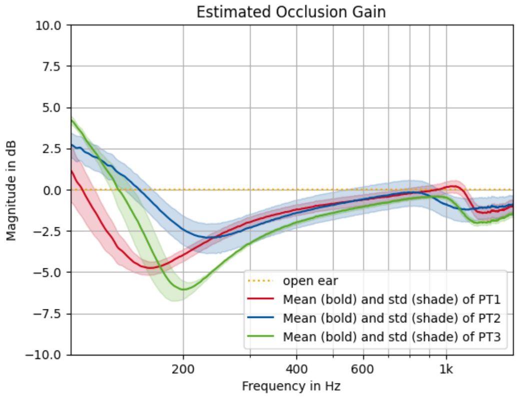

Figure 6: Estimated occlusion gain for all prototypes

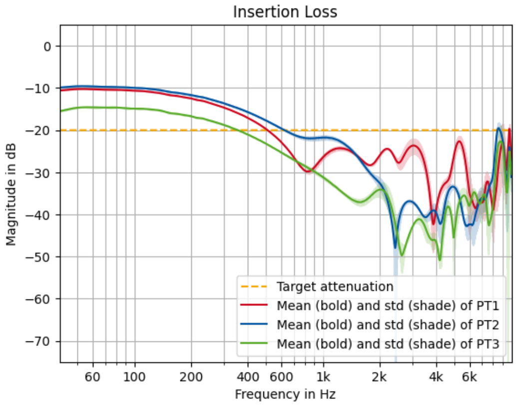

Figure 7: Measured insertion loss for all prototypes

Analysis

Figure 5 shows the measurements to match the simulations well. In both the simulation and the prototypes, however, a resonance behavior can be observed between around 150Hz and 200Hz. This is due to the air in the residual ear canal acting as a mass while the air inside the ear muff behaves like a compliance. This resonance causes the ear muffs' impedances to differ more severely from the open ear radiation impedance, causing a distinct negative occlusion gain in that frequency range. Below roughly 150Hz, the occlusion gains are positive with PT1 featuring the lowest positive occlusion gain at around 1dB. This confirms prior research stating a large internal volume to be necessary, as PT1 had a larger empty internal volume than PT2 and PT3.

The insertion loss measurements indicated that PT3 outperformed the other two prototypes across the entire frequency range. This is due to PT3 being printed with a thicker shell, thereby reducing sound transmissions through the walls. However, the target of at least 20 dB of broadband attenuation has not yet been achieved, as even PT3 only reached approximately 15 dB below 350 Hz. On this front, machining the prototype from solid material and creating tighter seals with the other ear muff components are anticipated to yield further improvements.

We calculated the achieved separation as the difference between Figure 6 and Figure 7 over frequency. With the most promising prototype PT3, separation between air and bone conduction was at least 20dB for frequencies above 400Hz. Below 400Hz, however, the lower insertion loss and distinct negative occlusion gain resulted in the separation reaching as low as 5dB. It is thus necessary to further reduce the occlusion gain by better matching the open ear radiation impedance, as well as increasing the insertion loss.

Acknowledgments

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