Dmitrii Burovikhin*, Michael Lauxmann, Ernst Dalhoff

Finite Element Modelling of a Hearing Contact Lens Coupled to the Middle Ear

Abstract: Hearing contact lens (HCL) is a new type of hearing aid devices. One of its main components is a piezo-electric actuator (PEA). In order to evaluate and maximize the HCL's performance, a model of the HCL coupled to the middle ear was developed using finite element (FE) approach. To validate the model, vibrational measurements on the HCL and temporal bones were performed using a Laser-Doppler-Vibrometer (LDV). The model was validated step by step starting with HCL only. Then a silicone cap was fitted onto the HCL to provide an interface between the HCL and the tympanic membrane. The HCL was placed on the tympanic membrane and additional measurements were performed to validate the coupled model. The model was used to evaluate the sensitivity of geometrical and material parameters with respect to performance measures of the HCL. Moreover, deeper insight was gained into the feedback behavior, which causes whistling sounds, and the contact between the HCL and tympanic membrane.

Keywords: piezo-electric actuator, hearing-aid device, feedback, hearing gain, finite element modelling.

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1 Introduction

In the 21st century thirty percent of European population is expected to have hearing difficulties and will need the help of hearing aid devices. However, there is a high level of dissatisfaction among the users of conventional hearing aids with acoustic excitation. According to the surveys among some of the main complaints are the feedback problems ('whistling in the ear'), the occlusion defect ('my voice sounds strange'), and poor sound quality due to distortion and exaggerated resonances [1].

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With the HCL from Vibrosonic GmbH there is a potential to make a quantum leap in the development of the hearing aid devices. The HCL is manufactured using microsystem technology (MEMS) and is so small, that it can be placed directly on the tympanic membrane. It is practically invisible for an outside observer. In the HCL, the loudspeaker used in the conventional hearing aids is replaced by a new type of a piezo-electric actuator that acts directly with a mechanical displacement excitation on the tympanic membrane and can transmit frequencies of up to 12.000 Hz. The conventional hearing aid devices usually reach frequencies up to 6.000-8.000 Hz. Thus, the HCL could principally amplify higher frequencies, which is generally important for spatial hearing and the understanding of speech, especially in more complex listening situations involving, for example, load background noise [2].

The goal of the research presented in this paper is the development of a FE model of the human middle ear coupled with the HCL. The research questions concern the mechanical coupling of the actuator to the tympanic membrane, its internal mechanical structure and the investigation of the stability limit with regard to the feedback of the reflected sound. The development of the FE model involves LDV measurements of the HCL alone and coupled to temporal bones and parameter fitting on different assembly levels of the HCL. By coupling the HCL model to an already existing middle-ear model [3], aspects like feedback and coupling of the HCL to the tympanic membrane are investigated.

2 HCL model

The PEA consists of three main layers: a piezo-electric layer, a handle layer, and a device layer, see Figure 1. Both the device layer and the handle layer are made of the standard (100) silicon wafer.

The piezoelectric layer is made of lead zirconate titanate and modelled using the coupled equations for the inverse piezo-electric effect:

$$\{T\} = [c]\{S\} - [e]\{E\}$$
$$\{D\} = [e]^T\{S\} + [e]\{E\}, \quad (1)$$

with T the stress vector, S the elastic strain vector, D the electric flux density, E the electric field intensity, c the elastic stiffness matrix, e the piezoelectric matrix and ε the dielectric permittivity.

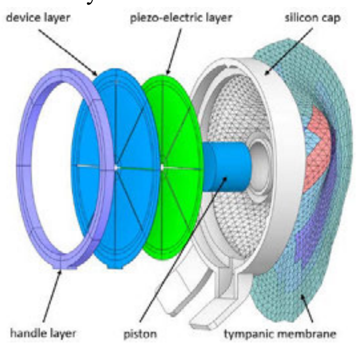

Figure 1: Piezo-electric actuator design.

Different meshing technics were tried out in order to obtain a mesh of desirable quality. At the end, to ensure the best fit to the measured data, it was decided to mesh the actuator symmetrically with quadratic solid elements.

The vibration measurements on the PEA were performed at the Laser Doppler Vibrometer laboratory at Reutlingen University to validate the actuator model. The measurement setup is shown in Figure 2. The PEA is hung by the wires supplying the voltage to the electrodes bonded to the piezo-electric layer. The voltage signal is generated in MATLAB and given out using the NI 9263 (National instruments) analog output module. The vibration of the PEA's segments is measured with a 3D-Laser Doppler vibrometer system which consists of 3 compact laser vibrometers (NLV-2500-5, Polytec GmbH, Germany). The LDV system is mounted on a Fanuc robot arm (FANUC LRMate 200iD, FANUC Deutschland GmbH, Neuhausen, Germany). The arm serves as the means of positioning the focal point of the LDV lasers at the specific location on the surface of the PEA's segments allowing to scan the segments at multiple points during the measurement cycle. The LDV velocity signal is measured using the NI 9239 (National instruments) analog input module. Then the frequency response is calculated from the velocity signal measured at the specific measurement point on the PEA. The PEA was excited with a linear sweep voltage signal with the amplitude of 1 Volt in the frequency range of 100 – 20000 Hz.

In order to ensure the accuracy of the results, preliminary measurements were performed to confirm the linearity of the measurement setup and the repeatability of the measurements.

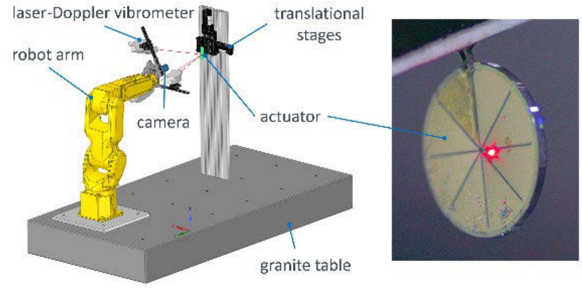

Figure 2: 3D-LDV scanning measurement setup at Reutlingen University.

The actuator was measured at all 8 segments at different radii: 0.3, 1 and 1.5 mm. The measured data has been compared to the FE results and some adjustments were made to the model to ensure a good fit, namely, the thickness of the device layer was reduced by 7 %. This made the main resonant frequency go down from 6299 Hz to about 5985 Hz and the amplitudes in the static range went up, since the segments became more compliant. For this resonant frequency MAC criterion of 0.97 was calculated considering all 24 measurement points. The fit between the simulation and measurements is shown at three points on the HCL exemplarily in Figure 3.

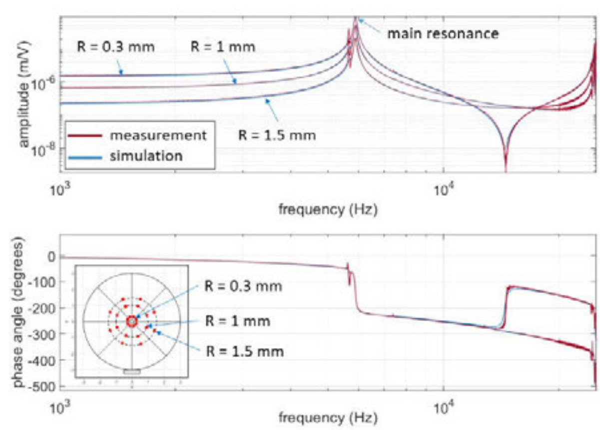

Figure 3: Measurement vs. simulation – HCL only.

It was noticed during the measurements that the phase shifts downwards in higher frequencies. This can be explained by non-linear relation between the strain in the piezo-electric layer and the excitation voltage [4]. In order to correctly simulate the phase angle, the phase shift angle was derived from the measured phase curve. The excitation voltage in the ANSYS model was made frequency dependent to account for the phase shift.

3 Middle-Ear Model

The middle-ear model consists of the tympanic membrane meshed with shell elements of varying thicknesses [3]. The thickness distribution is marked by various colours. The malleus, incus and stapes (the ossicles) are modelled as rigid bodies, and the cochlea modelled as a point mass attached to the stapes (Figure 4). The ligaments connecting the ossicles to the tympanic cavity and to each other are modelled as bushing elements. The ear canal and the tympanic cavity are modelled as fluid bodies with the properties of air.

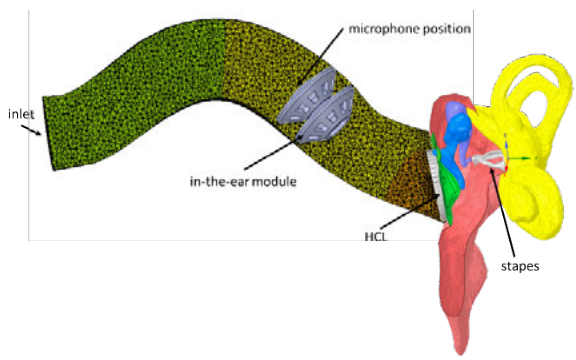

Figure 4: The middle-ear model coupled with the HCL.

4 Coupled Model

Figure 1 shows how the actuator model together with the silicon cap is attached to the tympanic membrane. Initially, the connection between the tympanic membrane and the silicon cap was modelled as a bonded contact assuming that the adhesive forces of the oil between the cap and the tympanic membrane are larger than the dynamic forces driving the membrane.

Both the piston and the cap were modelled as solid bodies. The cap is made of silicone. The piston is made of polymer.

Compared to the measurements, the FE model with full bonded contact between the tympanic membrane and the silicone cap has larger amplitudes in the higher frequency range as seen in Figure 5. If the contact area in the middle of the silicone cap is replaced by a weak bushing element the simulation results fit better to the measurements. This would mean that the adhesive forces in the middle of the silicone cap are weaker than at the edge.

The temporal bone measurements have been carried out at the ENT university hospital in Tübingen to validate the coupled model. The measurement setup is shown in Figure 6a. On the right side, there is an ER10C-OAE probe for measuring sound pressure in the ear canal. To the left from the probe there is the focal point of the LDV beam used for measuring the spatial motion of the stapes. Figure 6b shows a HCL placed on the eardrum and seen through the ear canal. The focal point of the LDV beam is on the tip of the actuator segment.

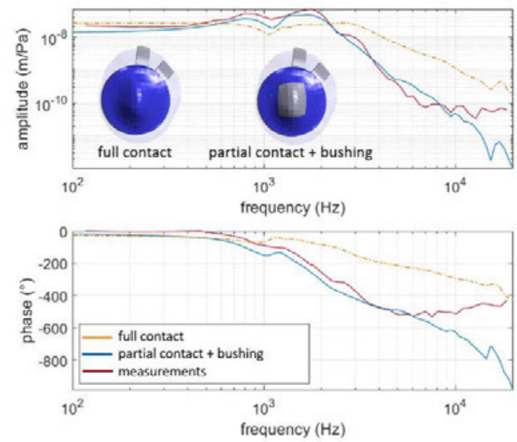

Figure 5: Frequency response measured at the stapes footplate center in the y-direction (HCL excitation).

The temporal bones were checked visually under a surgical microscope to exclude pathologies. The medial surface of the footplate was completely coated with retro-reflective beads to improve the reflectivity of the laser beam of the one-dimensional laser Doppler vibrometer (1D-LDV) system. To prevent drying out of the specimen, it was moistened with saline solution and enveloped in plastic wrap with moistened paper towel pieces placed in the plastic wrap beforehand.

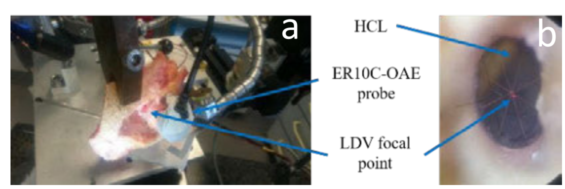

Figure 6: Measurement setup at the University of Tübingen.

5 Feedback and maximum gain before feedback

The maximum possible amplification is limited by several factors such as electrical performance of the actuator, contact forces and the feedback [5]. To understand the amplification limit associated with the feedback, the signal flow through the reconstructed ear is shown in a form of a diagram in Figure 7.

The upper branch (red) represents the transfer function G_{aco} from the pressure at the inlet of the ear canal p to the stapes motion y_s in case of the acoustic excitation. The middle branch (blue) represents the transfer function from pressure p over the in-the-ear module that includes the microphone and the amplifier, and over the HCL to the stapes motion y_s. The pressure difference between the inlet pressure p and the pressure at the microphone p_m is given by G_{pm}. The conversion of the sound picked up by the microphone p_m into the voltage signal U_m is given by G_{mic}. The amplifier amplifies the microphone voltage by the factor G_{amp} which is applied to the HCL. The transfer function G_{act} characterizes the transfer behaviour of the actuator that excites the tympanic membrane. The transfer function between the tympanic membrane motion y_{TM} and the stapes is given by G_{oss}. The lower branch (green) represents the feedback loop from the HCL to the microphone which is given by G_{ref} with p_{ref} being the reflected pressure.

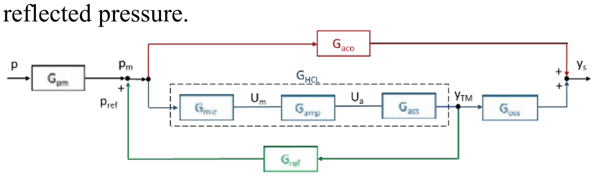

Figure 7: Signal flow in the reconstructed ear.

The transfer function between the inlet pressure p and the motion of the stapes y_s is given by:

$$y_s = \left(\frac{G_{HCL}G_{oss}}{1 - G_{HCL}G_{ref}} + G_{aco}\right) \cdot G_{pm} \cdot p \qquad (2)$$

The stability of the system is ensured as long as the denominator in the equation 2 does not become zero. Thus, the stability condition is

$$|1 - G_{HCL}G_{ref}| < 1, \qquad (3)$$

with the maximum amplification factor amp_{max} equaled to

$$amp_{max} = G_{mic} \cdot G_{apm} = \frac{1}{G_{act} \cdot G_{ref}}, \quad \left(\frac{V}{Pa}\right). \quad (4)$$

The amplification factor amp_{max} can be normalized by the equivalent sound pressure p_{eq}

$$p_{eq} = \frac{G_{act}G_{oss}}{G_{nat}}, \quad \left(\frac{Pa}{V}\right) \qquad (5)$$

wish $G_{act}G_{oss}$ the transfer function between the input voltage of the HCL and the stapes motion y_s, and G_{nat} the middle-ear transfer function of the natural ear. The maximum gain before feedback (MGBF) is defined as

$$MGBF = p_{eq} \cdot amp_{max}, \qquad (6)$$

It can be seen in Figure 8 that at around 2 kHz, die MGBF is below 0dB, whereas at other lower and higher frequencies the gain is well above and reaches a maximum of 56 dB at 100 Hz. Using the FE model, MGBF can be optimized to achieve a better amplification performance of the HCL.

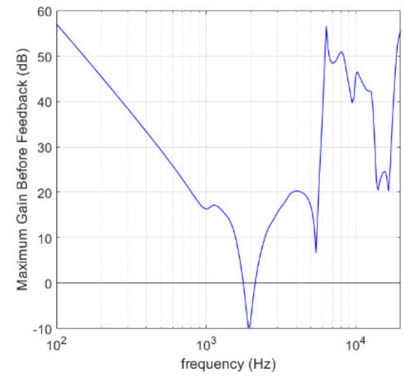

Figure 8: Maximum gain before feedback.

Conclusion

The physical behaviour of a human ear coupled with the HCL have been modelled. The FE model was validated using the LDV measurements. The model was used to estimate the HCL's performance limit. With this information the performance can be optimized by varying different geometrical and material parameters of the HCL to achieve a better amplification capability.

Author Statement

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