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Investigation of intracochlear dual actuator stimulation in a scaled test rig

Abstract: For patients suffering from profound hearing loss or deafness still having respectable residual hearing in the low frequency range, the combination of a hearing aid with a cochlear implant results in the best quality of hearing perception (EAS - electric acoustic stimulation). In order to optimize EAS, ongoing research focusses on the integration of these stimuli in a single implant device. Within this study, the performance of piezoelectric actuators, particularly the dual actuator stimulation, in a scaled uncoiled test rig was investigated.

Keywords: Cochlea, EAS, EMS, intracochlear stimulation, cochlear implant, piezoelectric bending actuator, basilar membrane model.

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1 Introduction

It is known that cochlear implants (CI) can restore a certain degree of auditory impression for patients suffering from profound hearing loss or deafness. In an increasing number of cases these patients still have a respectable residual hearing. Those patients gain most from a combination of a hearing aid with a CI [1-4]. In order to optimize EAS, ongoing research focusses on the integration of these stimuli in a single implant body. Different inner ear actuator concepts were analysed before yielding piezoelectric bimorphs as the most feasible one for basilar membrane stimulation from within the scala tympani (ST). For the experimental validation of the numerical findings [5, 6] that non-resonant intracochlear (IC) stimulation induces two local basilar membrane oscillation maxima, one at the position of the piezoelectric actuator and the other oscillation peak at the position according to the tonotopic cochlea map for round window (RW) stimulation, and that the peak at the tonotopic position regarding its amplitude can be optimized using multiple actuators, the intracochlear dual actuator stimulation in the scaled up uncoiled cochlea test rig was evaluated within this study.

2 Experimental setup

The test rig used for the experimental validation of the numerical results is an up-scaled (4:1) and slightly modified version of the cochlea model proposed in [7] and build out of acryl. It consists out of two fluid-filled chambers, the scala vestibule (SV) and the ST, separated through a silicone membrane (BM) and connected to each other at the distal part, i.e. the helicotrema (HT). Both ST- and SV-chambers have a height of 8 mm and a length of 146.6 mm. The width of the chambers is 8.7 mm at the HT, increasing linearly to 20 mm at 135.8 mm from the HT followed by a linear decrease to 12.8 mm at the oval (OW) respectively RW. Both the OW and the RW have a diameter of 4 mm. The BM, starting at 9.7 mm from the HT, has a length of 124.9 mm and a width of 5 mm which decreases linearly to 1 mm. The HT has a length of 7.1 mm and a width of 8.7 mm. Since hearing loss is often originated by the loss of the active cochlear mechanism as well, this model only represents the passive cochlear mechanism, i.e. without the active amplification performed by the Organ of Corti [8] necessary

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for what is considered normal hearing perception. The BM, OW and RW membranes which were used in the test rig experiments had a fixed thickness along their length and were fabricated out of Sylguard 184 (Dow Corning) using a mould of 0.4 mm depth. After 1.5 hours of curing at 93°C at a pressure of 0.4 bar, the membranes had final thickness of 0.35 mm with a standard deviation of 0.02 mm and were mounted in the test rig by means of double sided adhesive tape (Tesa 5 mx50 mm). In order to enhance the reflection of the BM, the backside of the membrane (towards the ST) was coated with a reflective spray (Ardrox Ax 9D1B). Tap water at room temperature was used as a substitution for the perilymph. Great care was taken that no air was trapped in the setup.

The oscillations of the BM were measured by means of Laser Doppler velocimetry (Scanning LDV, MSA-100-1D at a wavelength of 532 nm, Polytec). To compensate for the optical density difference between air and water n_{air}/n_{water}(@589 nm)=0.983/1.333 [9], the amplitude of the BM oscillation was multiplied by this factor.

For the characterization of the test rig with respect to its tonotopic character, i.e. frequency dependent oscillation profile of the BM, a shaker (Model 4810, Brüel & Kjær) in combination with amplifier (IMG-STA 1508 Pro Power) was used and connected to the LDV frequency generator output. The RW-membrane was driven through a steel rod with diameter of 3.8 mm connected to the shaker (Fig. 1). Great care was taken that the driving rod did not touch anything but the RW membrane. To ensure a stable contact between the rod tip and the RW, the rod was positioned in such a way that the RW membrane was pushed inwards by around 0.5 mm. By means of the acceleration sensor (PCB M353B17 with 96.71 g/V), mounted on the driving rod, the acceleration value for different frequencies was checked and controlled.

For the intracochlear (IC) stimulation from within the ST, piezoelectric bending actuators (PICMA-Bender from PI Ceramic GmbH) were used and mounted within the ST (Fig. 4 inset). These piezoelectric bending actuators, sizing 10 mm x 5 mm and having a total thickness of 0.21 mm, were built of modified lead zirconate titanate (PIC252) and were driven by the same amplifier used for the RW stimulation experiments. Whereas for the single actuator stimulation experiments the build-in LDV frequency generator was used, the extra one, Aim-TTi TGA12104 100 MHz Arbitrary Waveform Generator was used for the double actuator stimulation experiments.

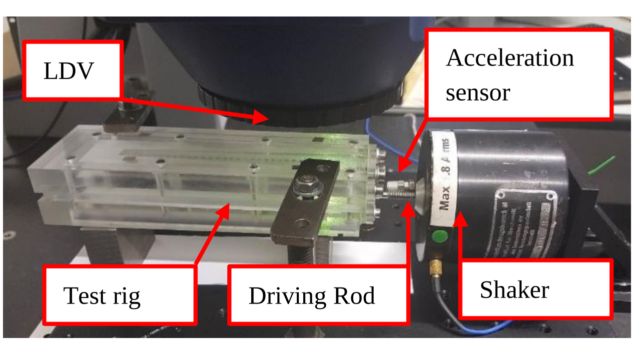

Figure 1: The experimental setup characterization of the test rig with respect to its tonotopic character.

3 Measurements and results

To characterize the test rig with respect to its tonotopic character, a grid of 47 LDV scan points was placed at the BM and 4 control points to measure the vibration of the test rig housing were placed at the test rig. The acceleration sensor was connected to the LDV reference 1 input whereas the generator output was connected to the shaker amplifier input. The measurements were performed for 10 different frequencies ranging from 400 to 3000 Hz for a constant acceleration value of 4.2±0.6 m/s^2. For each LDV scan point, the amplitude and the phase shift with respect to the RW stimulation (i.e. acceleration sensor) was averaged 3 times. An example of the BM oscillation for a RW stimulation at 600 Hz is given in Figure 2. On the vertical scale the BM oscillation for four chosen phases is shown.

For each oscillation profile (i.e. frequency), the position of the maximum amplitude on the BM (x_{ton}) was determined by means of a 6th order polynomial fit, which was found to empirically describe the oscillation profile the best. The results, summarised in the tonotopic plot (Fig. 3), show that the position of the maximum of the BM oscillation moves with increasing frequency towards the base of the BM. For comparison, the Greenwood function describing the human cochlea [10] but transferred to our membrane length is given in the plot as well. Apart from the offset, the tonotopic plot of the test rig shows a similar trend despite some simplifications such as the BM material and viscosity of the fluid. The

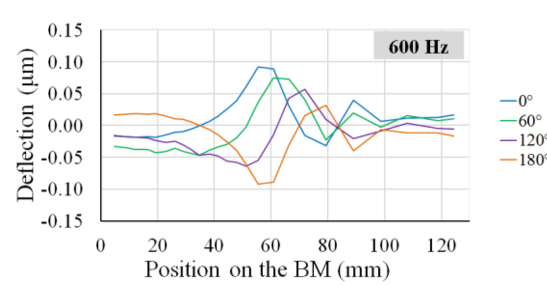

Figure 2: The BM oscillation for RW stimulation at f=600Hz for different phases.

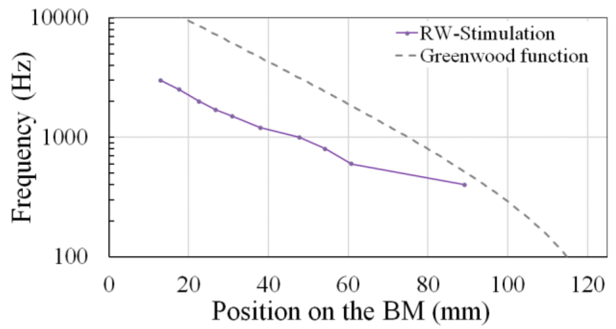

Figure 3: Line: The tonotopic character of the test-rig. Dashed line: Greenwoods function describing the human cochlea adapted to our membrane length.

fundamental assessment is not affected from this difference.

To confirm our previously presented numerical findings that IC stimulation induces two local BM oscillation maxima [6], one at position of the piezoelectric actuator (direct stimulation) and the other at the position according to the tonotopic cochlea map for RW stimulation (indirect stimulation), the shaker and driving rod were removed and a piezoelectric bending actuator was positioned within the ST

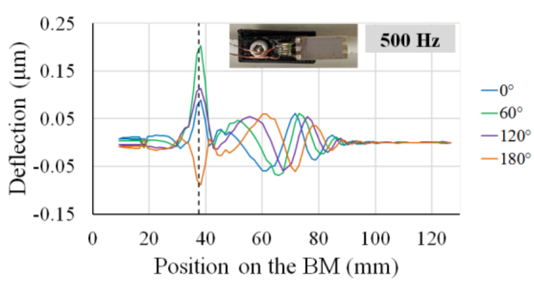

Figure 4: The BM oscillation for IC stimulation at f=500Hz for different phases.

of the test rig at 1.8 mm below the BM. The amount of a LDV scan points was increased to 83 and the LDV generator output was connected to both the reference 1 input and amplifier input. The measurements were performed at 1 V nominal output for three different positions (12 mm, 38 mm and 65 mm) and for at least 3 different frequencies in between of the 250 and 2000 Hz (Fig. 5). For each LDV scan point at the BM, the amplitude and the phase shift with respect to the generator output was averaged 3 times. An example of the BM oscillation for the IC stimulation at 38 mm for f=500 Hz is given in figure 4.

In contrast to the RW-stimulation measurements, the position of the maximum amplitude on the BM for the indirect stimulation peak was determined only for those points where an increase in the absolute phase shift, the characteristic behaviour for a traveling wave, along the BM is observed. The results, summarized in figure 5, show the same behaviour compared to the RW stimulation experiments.

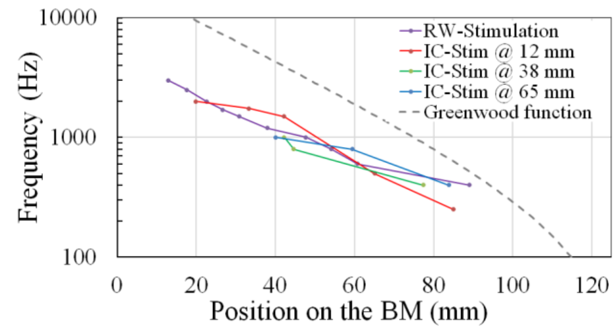

Figure 5: The tonotopic character of the test rig for RW and IC stimulation. Dashed line: Greenwood function describing the human cochlea adapted to our membrane length.

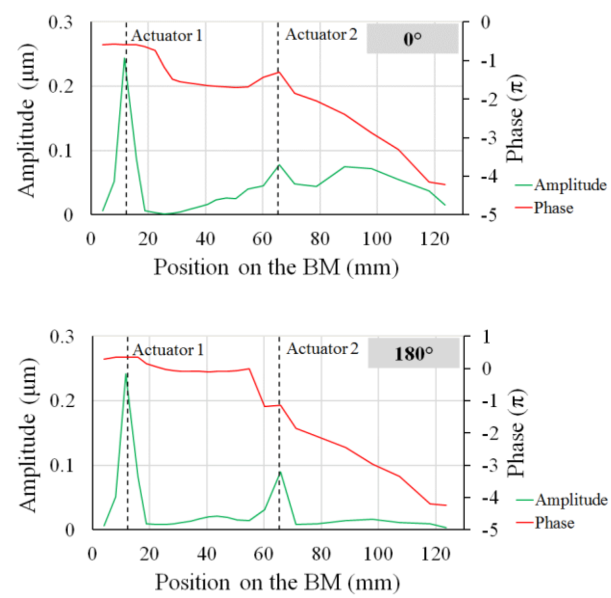

Figure 6: Dual actuator stimulation: BM oscillation envelope curves with simultaneously identical frequency driven piezoelectric actuators when stimulating with f=400Hz for different phases and the corresponding phase response. The grey dashed lines mark the position of the piezoelectric actuators.

Since the amplitude of the BM oscillations at the position of the piezoelectric actuator is larger than the amplitude of the secondary oscillations at the tonotopic position, the feasibility to simultaneously drive an array of actuators and optimize the BM oscillation peak at the tonotopic position with regard to its amplitude by optimizing the phase between the actuators was investigated experimentally as well. For these measurements the investigations with a single intracochlearly positioned piezoelectric actuator were extended to two simultaneously driven piezoelectric actuators, located at 12 mm and 65 mm from the BM base position. The measurements were performed at a frequency of 400 Hz and showed so that the amplitude of the BM oscillation amplitude at the tonotopic position dependents on the phase difference between the actuators (Fig. 6).

Furthermore, the possibility of simultaneous stimulation of the cochlea for different frequencies was explored. The actuators were kept on the same position but operated at different frequencies. The frequency generator output with the lowest frequency was connected to the LDV reference 1 input and served as reference value. The BM oscillations were analysed with respect to the corresponding operating frequency components, i.e. f=400 Hz was analysed in the 400 Hz frequency domain. An example of the BM oscillation for IC dual actuator stimulation at f_1=800 Hz and f_2=400 Hz is given in figure 7 and shows that one can stimulate multiple positions on the BM, allowing one to address multiple frequencies simultaneously.

4 Discussion and conclusion

The experimental investigations of IC stimulation from within of the ST was evaluated in this study. The measurements with a single intracochlearly positioned piezoelectric actuator for different stimulation frequencies and different locations showed two local BM oscillation peaks with one oscillation peak at position of the piezoelectric actuator and the other oscillation peak at the position accordingly to the tonotopic cochlea map created for RW stimulation. Thus confirms our previously presented numerical findings that non-resonant IC stimulation induces two local BM oscillation maxima, allowing one to stimulate even at those positions should no actuator is present. Please note that position of oscillation peak at the tonotopic position solely depends on the stimulation frequency of the piezoelectric actuator (Fig. 5) and can be shifted along the BM accordingly.

By operating the array of actuators with identical frequency and optimizing the phase difference between each

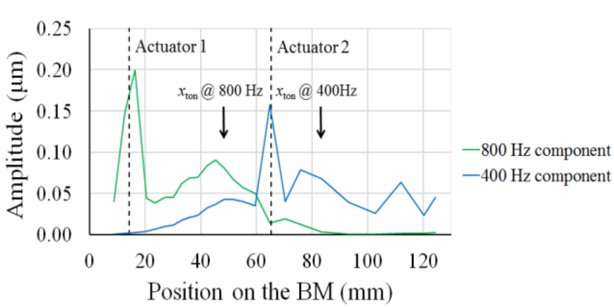

Figure 7: Dual actuator stimulation: BM oscillation envelope curves for simultaneously driven piezoelectric actuators when stimulating actuator 1 with f=800Hz and actuator 2 with f=400Hz. The grey dashed lines mark the position of the piezoelectric actuators.

single actuator, an increase of the BM oscillation peak at the tonotopic position is attained (Fig. 6). Furthermore one can simultaneously stimulate multiple positions on the BM (Fig. 7), thus allowing one to address multiple frequencies at the same time.

Future investigations will address the effect of the distance between the actuator and the BM, the phase correlation for multiple actuators as well as the integration and investigation of the piezoelectric bending actuators in a CI electrode.

Author's Statement

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References

[1] von Illberg C, Kiefer J, Tillein J, Pfenningdorff T, Hartmann R, Stürzebecher E, Klinke R. Electric-Acoustic stimulation of the auditory system. ORL 1999; 61: 334-340
[2] Gantz BJ, Turner C. Combining acoustic and electrical speech processing: Iowa/Nucleus hybrid implant. Acta Otolaryngol. 2004; 124: 344-347
[3] Büchner A, Schüssler M, Battmer RD, Stöver T, Lesinski-Schiedat A, Lenarz T: Impact of low-frequency hearing. Audiol. Neurotol. 2009; 14(S1): 8-13.
[4] von Ilberg CA, Baumann U, Kiefer J, Tillein J, Adunka OF. Electric-acoustic stimulation of the auditory system: A review of the first decade. Audiol. Neurotol. 2011; 16(S2): 1-30.
[5] Schurzig D, Rau TS, Wallaschek J, Lenarz T, Majdani O. Determination of optimal excitation patterns for local mechanical inner ear stimulation using a physiologically-based model. Biomed Microdevices 2016; 18:36.
[6] van Drunen WJ, Schurzig, D, Kiewning M, Schwarzendahl S, Wallaschek J, Lenarz, T, Majdani O. Feasibility Analysis of an Implantable Piezoelectric Hearing Prostheses for the Inner Ear. Computer Aided Medical Engineering 2016; 7:27-32.
[7] Gan RZ, Reeves BP, Wang X. Modeling of sound transmission from ear canal to cochlea. Annals of Biomedical Engineering 2007; 35(12):2180-2195.
[8] Pickles J. An Introduction to the Physiology of Hearing. 3rd edition. Academic Press Limited, London, 2008.
[9] Lide DR. Handbook of chemistry and physics. 76th edition. CRC Press, Inc., Florida, 1995.
Greenwood D: A cochlear frequency-position function for several species - 29 years later. J. Acoust. Soc. Am. 1990; 87(6): 2592-2605.