ON THE IMPLANTABLE HYDROACOUSTIC MICROPHONE

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ABSTRACT

On the way towards the fully implantable cochlear implant (CI), the development of a suitable implantable microphone is one of the remaining challenges. An interesting approach to this problem is the idea of using a hydrodynamic coupling between the vibrating ossicles and a piezoelectric sensor, which is particularly promising from the medical point of view. In order to assess the performance of sensors based on this principle, temporal bone measurements were carried out and transfer matrix models were developed. Based on the results of these measurements and the model predictions, the feasibility of an implantable hydroacoustic microphone is discussed together with implications for further research.

1 INTRODUCTION

Cochlear Implants (CIs) are prostheses for damaged or non-functional inner ear sensory cells. Inner ear dysfunction accounts for the majority of all deafness cases in newborns. Over the past 20 years or so, CIs have proven successful to reestablish acoustic communication for these patients up to the point that otherwise deaf people could play musical instruments.

In principle, a CI consists of a microphone to pick up the sound, a signal processing unit to convert the microphone voltage into electrical nerve stimuli, an electrode to apply these stimuli to the acoustic nerve, and a power supply (i.e. a battery). The currently common design of CIs requires the whole device be split up in two parts, one of which is implanted while the other one remains outside the body. The drawbacks of this design include fussy handling due to external parts and cables, poor directional hearing (the directional filter provided by the outer ears is not used), and psychological/acceptance issues.

Recent advances in technology, in particular in digital signal processors and battery technology make a totally implantable CI feasible, provided a suitable implantable microphone was available. Possible solutions published so far include subcutaneous electret microphones [1, 2], piezoelectric bending beams contacting the ossicles [3, 4] and cochlea hydrophones [5]. Although both subcutaneous microphones and piezoelectric beams have already been implanted in humans successfully [6, 7], none of the concepts mentioned has gained universal acceptance so far. This motivates the continuing quest for alternative solutions.

One such alternative could be a device that uses a fluid to transmit ossicle vibrations to a receiving unit which then performs the acoustical-electrical transduction. Benefits of this design which was recently proposed by HÜTTENBRINK et al. [8], and which will be referred to as “hydroacoustic” microphone in what follows, include 1) the use of outer and middle ears as acoustical pre-filters, 2) the avoidance of hard coupling to ossicles, thereby greatly reducing the risk of bone deterioration, 3) the avoidance of ear canal irritation, 4) the placement of a receiving unit distantly from the middle ear cavity, facilitating the implantation.

The purpose of this paper is to explore the performance of devices based on this principle.

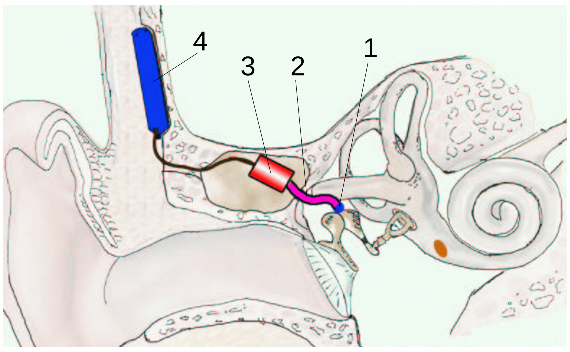


Figure 1: Principal design of the hydroacoustic microphone: Ossicle vibrations are picked up by a diaphragm (1) coupled to a fluid-filled tube (2) with subsequent transduction of the fluid sound field into electrical signals by a piezoelectric sensor device (3) placed in the mastoid. Also shown is the CI processor unit (4).

2 PRINCIPAL DESIGN OF THE HYDROACOUSTIC MICROPHONE

The principal design of the implantable hydroacoustic microphone is shown in figure 1. The natural sound receiving apparatus of the outer ear serves to capture the incoming sound, resulting in vibrations of the ossicles. These are then picked up at the head of the malleus or that of the incus by a diaphragm (1) coupled to a fluid-filled tube (2). The fluid transmits the sound information to a piezoelectric sensor (3) located at the end of the tube, where it is converted into an electrical voltage that can be input into a CI processor (4).

As mentioned above, this design has a number of benefits and seems to be appealing in particular from the surgeon's point of view. However, due to the extended transmission path with quite some potential to introduce losses, these benefits may come at the price of a reduced electro-acoustical sensitivity. It is therefore desirable to know whether the reduction in sensitivity remains at an acceptable level or not.

To this end, models characterizing the transfer behavior of the hydroacoustic microphone were developed and are discussed in the following section.

3 TRANSFER MODELS

The four-pole transfer matrix approach was used to assess the acousto-electrical properties of the hydroacoustic microphone. In this approach, single elements are modeled by four-pole transfer matrices. As an example,

$$\begin{pmatrix} p_\mathrm{D} \\ q_\mathrm{D} \end{pmatrix} = E \cdot \begin{pmatrix} F_\mathrm{I} \\ v_\mathrm{I} \end{pmatrix} = \begin{pmatrix} e_{11} & e_{12} \\ e_{21} & e_{22} \end{pmatrix} \cdot \begin{pmatrix} F_\mathrm{I} \\ v_\mathrm{I} \end{pmatrix}. \tag{1}$$

represents the transfer behavior of the outer and middle ears. Here, p_D and q_D are sound pressure and volume velocity at the ear drum, and F_I and v_I are force and velocity at the incus, respectively. The matrix E contains transfer functions which will be discussed below. All quantities usually depend on frequency.

The use of transfer matrices permits a very quick assessment of the overall transfer behavior of a chain of elements, which is obtained by simple multiplication of the element transfer matrices,

$$\begin{pmatrix} p_\mathrm{D} \\ q_\mathrm{D} \end{pmatrix} = H \cdot \begin{pmatrix} U \\ I \end{pmatrix} = E \cdot C \cdot T \cdot P \cdot \begin{pmatrix} U \\ I \end{pmatrix}. \tag{2}$$

In eq. 2, C, T and P are transfer matrices similar to E, which represent the transfer behavior of the coupling diaphragm, the tube and the piezoelectric sensor, respectively. H is the overall transfer matrix and U and I are voltage and current at the electrical terminals of the piezoelectric sensor. Assuming a high-impedance input of the electronic unit receiving the microphone signal (i.e. $I = 0$), the sensitivity of the microphone is given by the reciprocal of the (1,1)-element of the overall transfer matrix,

$$\frac{U}{p_D} = \frac{1}{H(1,1)}. \tag{3}$$

A typical requirement for an implantable microphone is a sensitivity of $1m$V per Pa, over a frequency range of approximately 100…$6k$Hz.

Outer and middle ears: E-matrix. Transfer models for the parts of the outer and middle ears considered here (i.e., from the ear drum to the incus) were taken from [9]. When using these models one must keep in mind that 1) they represent a mean behavior from which an individual may differ, and 2) that they must be slightly modified to account for the fact that one can usually not contact the processus lenticularis, but rather the head of the malleus or that of the incus.

Item 1) is actually what we want (a microphone that works for the average patient). Still, it is a good idea to check particular temporal bone preparations for consistency with the models. This was done by exciting the temporal bone preparation acoustically, loading the processus lenticularis with small masses and measuring the transfer functions between ear drum sound pressure and velocity of the loads. Based on the measured transfer functions, the transfer functions from ear drum sound pressure to incus (processus lenticularis) free velocity and from ear drum sound pressure to incus (processus lenticularis) blocked force can be solved for with a least squares approach.

Item 2) was accounted for by introducing an ideal lever (lever ratio 1:5) which acted as to "undo" the leverage by the processus lenticularis.

Coupling diaphragm: C-matrix. Out of all chain elements, the coupling diaphragm is the one that is most challenging in terms of modeling. This is partly due to the choice of material (latex or silicone), and partly due to the complicated coupling geometry which results from the combined effect of the diaphragm pre-tension (which in turn is due to the fluid static pressure) and the plastic deformation which is obtained when contacting the ossicle by pressing the whole device against it with a certain static force.

Despite the complications described here the authors believe that for the purpose of a coarse assessment of the overall transfer behavior of the implantable microphone, a simplified model of the coupling diaphragm is sufficient.

The approach eventually adopted builds on the model of a lossless, pre-stressed diaphragm excited by a circular area force as given in [10]. This model essentially is an ideal mechanical-acoustical transducer (i.e. a massless rigid piston) of an effective transducer area smaller than the tube cross section (by a factor depending on the ratio of radius corresponding to the the excitation area to that of the tube) and both a shunt mechanical compliance and a shunt acoustical compliance which not only depend on the above mentioned radius ratio but in addition on the diaphragm pre-tension and its thickness.

In order to get an estimate of the diaphragm pre-tension, Finite Element calculations were performed. For a static fluid pressure of normal plus $2k$Pa and a latex diaphragm of 50μm thickness, the pre-tension is about 30N/mm^2. In the calculations, this pre-tension was varied from one third to three times this value. Also, the ratio of the radius of the coupling region to that of the tube was varied from 0.1 to 0.5.

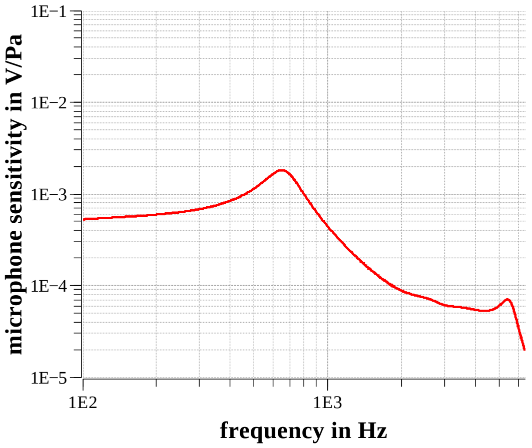
Tube: T-matrix. Transfer models for the fluid-filled tube were based on measurements made by HUDDE [11]. The main result of these measurements was that the compliance of the tube wall, which is high compared to that of the fluid (water), results in a decrease in the speed of sound in the fluid. This in turn means that it may become necessary to consider wave phenomena, i.e. the general form of the T-matrix is that of a one-dimensional wave-guide,

$$T = \begin{pmatrix} \cosh\gamma\, l & Z_\mathrm{w}\sinh\gamma\, l \\ \sinh\gamma\, l\,/\,Z_\mathrm{w} & \cosh\gamma\, l \end{pmatrix}, \tag{4}$$

with l being the length of the tube and the wave impedance Z_w and propagation constant γ given by

$$\begin{aligned} Z_\mathrm{w} &= \varrho c\,/A_T && (5)\\ \gamma &= \alpha + \mathrm{j}\,\omega/c. && (6) \end{aligned}$$

Here, j is the imaginary unit ($\mathrm{j} = \sqrt{-1}$), ω is the frequency in radians/s, A_T is the cross-section of the tube, and ϱ and c are the mass density of and the speed of sound in the fluid, respectively. For the latter, values between the theoretical limit of $c = 1440\mathrm{m/s}$ (ideally rigid walls) and $c = 400\mathrm{m/s}$ (which is about twice the value found for thin Teflon tubes in [11]) were used in the model calculations. The damping constant α was set to 4.6/m (corresponding to 40 dB/m).



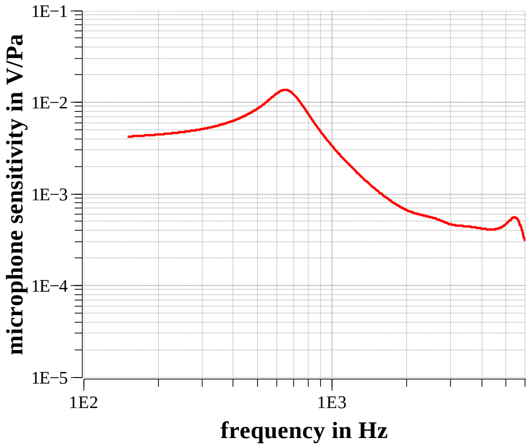


Figure 2: Numerical simulations: predicted best performance of the implantable microphone. **Left:** PZT hollow cylinder. **Right:** PZT coated bending plate.

Piezoelectric sensor: P-matrix. Two types of piezoelectric elements were used in this investigation: a radially polarized hollow cylinder of Lead Zirconate Titanate (PZT) and a clamped circular bending plate of titanium with a circular layer of PZT polarized in the thickness direction.

The transfer characteristics of the PZT cylinder were derived analytically: Neglecting mechanical losses and assuming the length of the cylinder is large compared to its radius and further neglecting shear and mechanical radial components, some calculus [12] eventually leads to

$$P = \begin{pmatrix} 2\,/\,(3\,g_{31}\,R) & \mathrm{j}\,h\,/\,\left(3\,\pi\,\omega\,R^2\,l\,\varepsilon_{33}^T\,g_{31}\right) \\ \mathrm{j}\,2\,\pi\,\omega\,R^2\,l\,\left(2.5\,s_{11}^D + 2\,s_{12}^D\right)\,/\,(3\,g_{31}\,h) & 1.5\,g_{31}\,R - R\,\left(2.5\,s_{11}^D + 2\,s_{12}^D\right)\,/\,\left(3\,\varepsilon_{33}^T\,g_{31}\right) \end{pmatrix}, \tag{7}$$

In this equation, R is the mean radius, l the length and h the wall thickness of the cylinder and s_{11}^D, s_{12}^D, $\varepsilon_{33}^{\mathrm{T}}$ and g_{31} are standard constants of the PZT material used here (indices indicate directions: 1-lateral, 2-tangential, 3-radial).

The transfer characteristics of the bending plate device was estimated using a Finite Element model.

Prediction of microphone sensitivity. The above considerations showed that there is a number of parameters that can be varied within certain limits, including a) the radius of the coupling region, depending mainly on the static force with which the IM can be pressed against the ossicle, b) diaphragm pre-tension and thickness, depending mainly on the fluid static pressure, c) the speed of sound in the tube, depending on the tube wall compliance, and d) the geometry of the two types of piezoelectric elements considered here. The latter is limited by the space requirements, with maximum dimensions of 30mm (length) $\times$ 3mm (radius) for the cylinder and of 6mm (radius) $\times$ 5mm (height) for the device based on the circular bending plate.

The variation of these parameters in numerical simulations indicate that they contribute to the overall transfer behavior independently. A high microphone sensitivity is achieved by

1. a large radius of the coupling region,
2. a high diaphragm pre-tension,
3. a tube wall compliance that is tuned such that the first tube resonance is in the 4...6kHz region.

The best design of the piezoelectric devices are thin-walled hollow cylinders with the maximum length and radius and bending plate devices of the maximum radius.

In figure 2, the predicted highest microphone sensitivity for both piezoelectric devices (i.e. a hollow cylinder of 6mm outer diameter, 0.5mm wall thickness and 30mm length and a titanium bending plate of 5mm radius and 100μm thickness with a 100μm thick layer of PZT of 9mm diameter) is shown. It is seen that for the given space requirements the bending plate design performs better and is actually the only design that could theoretically achieve the required sensitivity, although even for this design and even under the quite optimistic assumptions regarding the radius

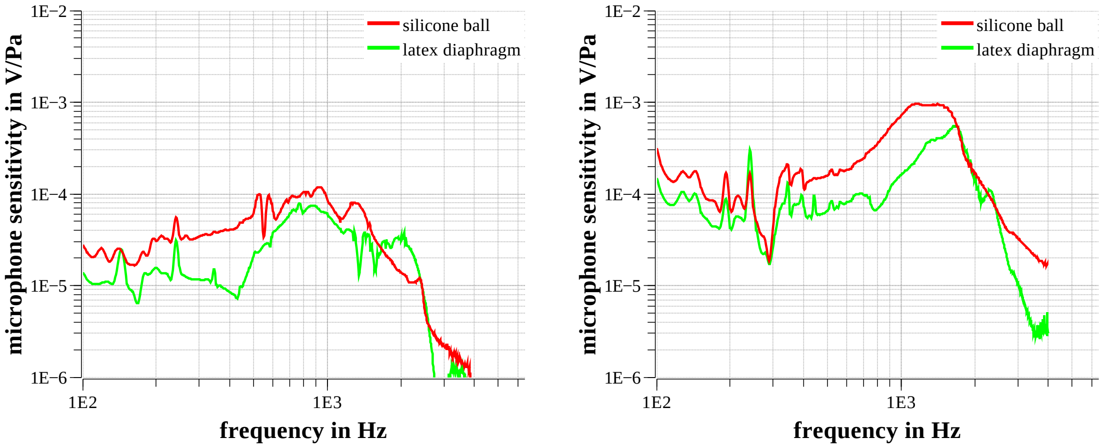


Figure 3: Temporal bone experiment: Sensitivity of implantable microphone prototypes. Note that the piezoelectric elements were of non-optimal dimensions. **Left:** PZT hollow cylinder. **Right:** PZT coated bending plate.

of the coupling region and the diaphragm pre-tension, the sensitivity requirement is missed by some dB for frequencies above 2kHz.

4 TEMPORAL BONE MEASUREMENTS

In order to verify the model predictions, temporal bone measurements were performed at the middle ear laboratory of the ENT-clinic of TU Dresden. Experiments were carried out in 2 fresh temporal bone specimens obtained within 48 hours after death (age of the patients at death: 65 years). Specimens were kept constantly refrigerated at 4°C in Ringer's solution until studied. All specimens underwent accurate inspection before each experiment, using a surgical microscope (Zeiss OPMI 111) in order to rule out any middle ear pathology which could interfere with the measurements. All experiments were performed within 3 days after death. Specimens were kept moistened constantly by using physiological Sodium chloride solution throughout the experimental investigations. After confirmation that ossicles and tympanic membrane (TM) were normal in appearance, a posterior tympanotomy including removal of the facial nerve was performed. The external auditory meatus was sealed after positioning of a silicone tube (2x1cm) to which a sound source (Praecitronic Co.) was coupled tightly. The sound pressure at the tympanic membrane, which served as reference in these experiments, was measured by a probe microphone (Etymotic Research ER-7C). The voltage at the electrodes of the piezoelectric sensor was fed into a low-noise preamplifier (Brüel&Kjær 2669) driven by a power supply (Brüel&Kjær 5935). Test signal generation, data acquisition and transfer function computation was performed by a dynamical analyzer (Brüel&Kjær 2035).

Both types of piezoelectric elements discussed above were tested. It should be noted however that the main purpose of the measurements was to verify the models, which is why the choice of the geometry of the piezoelectric elements was based on availability rather than on an optimized performance. Because the transfer behavior of these devices can be modeled with a satisfactory precision, the authors feel comfortable in extrapolating to devices of slightly different dimensions. The dimensions of the devices tested were: 20mm (length) $\times$ 1.1mm (radius) for the hollow cylinder, and 3mm/2.5mm (radius of Ti plate and PZT layer, both 100μm thick).

As the numerical simulations indicated that the actual radius of the coupling region should be as large as possible, a different design of the coupling "diaphragm", consisting of a silicone ball embedded in the silicone membrane closing the tube, was also tested. The tube itself was of brass with a length of 28mm and an inner diameter of 1mm.

The implantable microphone contacted the head of the malleus or that of the incus with a suitable static force which was limited by the requirement that no important luxation of the ossicles should result.

In figure 3, the results of these measurements are shown. They agree with the model predictions for values of the unknown parameters (radius of the coupling region and diaphragm pre-tension) close to the mean of the values assumed in the numerical simulations.

Thus, there is some potential of improving the unsatisfactory sensitivity seen in figure 3: An optimized design of the piezoelectric device could improve the sensitivity by an estimated 10dB. An optimization of the coupling diaphragm is certainly difficult but the only choice if one wants to increase the sensitivity further. Its potential effect is estimated at another 10dB. By properly adjusting the wall compliance and the length of the tube, one could partially compensate for the high-frequency roll-off. In summary, applying all these improvements successfully (this means a lot of good luck) one could end up with a sensitivity close to the required $1m$V/Pa for frequencies up to about $2k$Hz, but will miss this requirement for higher frequencies.

5 CONCLUSION

In this paper, it was attempted to characterize the electro-acoustical performance of an implantable hydroacoustic microphone for Cochlear Implants. Although it seems theoretically possible to achieve a microphone sensitivity close to that of competing designs (in particular, subcutaneous and bending beam type designs), temporal bone measurements indicate that practically feasible designs will miss the required sensitivity for frequencies above about $2k$Hz. In order to improve the sensitivity of the otherwise appealing hydroacoustic design, both the diaphragm providing the coupling to the ossicle and the piezoelectric sensor must be optimized.

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