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Thin film piezoelectric acoustic transducer for fully implantable cochlear implants

Bedirhan İlik [a,b], Aziz Koyuncuoğlu [b], Özlem Şardan-Sukas [b], Haluk Külah [a,b,*]

[a] Middle East Technical University, Department of Electrical and Electronics Engineering, Ankara, Turkey
[b] Middle East Technical University, MEMS Research and Application Center, Ankara, Turkey

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ABSTRACT

This paper reports the development of a single cantilever thin film PLD-PZT transducer prototype. The device was experimentally characterized by attaching it on an acoustically vibrating membrane resembling the behavior of the eardrum. Acceleration characteristic of the sensor attached on the membrane was obtained by using a Laser Doppler Vibrometer (LDV) as the output voltage was measured by an oscilloscope. A voltage output of 114 mV was obtained, when the device was excited at 110 dB Sound Pressure Level (SPL) at 1325 Hz. This is the highest value for a thin film piezoelectric transducer in the literature to our knowledge. Using the results of a finite element analysis for this single-channel prototype, which are within 92% agreement with the experimental results, we performed an optimization study to propose a multi-frequency acoustic sensor to be placed on the eardrum for fully-implantable cochlear implant (FICI) applications. The proposed multi-channel transducer consists of eight cantilever beams. Each of these beams resonates at a specific frequency within the daily acoustic band (250–5000 Hz), senses the eardrum vibration and generates the required voltage output for the stimulation circuitry. The total volume and mass of the transducer are $5 \times 5 \times 0.2$ mm^3 and 12.2 mg, respectively. High sensitivity of the transducer (391.9 mV/Pa @900 Hz) enables transmission of strong signals to be the readout circuit, which can easily be processed. Expected to satisfy all the requirements (volume, mass, and stimulation signal at the hearing band) of FICI applications for the first time in literature, the proposed concept has a groundbreaking nature and it can be referred to as the next generation of FICIs since it revolutionizes the operational principle of conventional CIs.

1. Introduction

Cochlea, ossicles and eardrum together form one of the most elaborated structures in mammalians. It provides a very large frequency selectivity (20 Hz – 20 kHz) and sound perception (0–140 dB SPL), which makes ear the best acoustic sensor in nature. Unfortunately, this delicate structure of the cochlea also makes is prone to degradation without recovery. Approximately 15% of the world's adult population has some degree of hearing loss according to the World Health Organization (WHO). In total, there are 360 million people living with a hearing loss higher than 40 dB SPL as of 2015, 32 million of these patients are children [1]. The level of hearing loss can be classified as mild, moderate, severe or profound. For mild-to-moderate damage, a hearing aid can be used to restore the hearing loss with sound amplification [2]. Whereas, Cochlear Implants (CIs) can be utilized for the treatment of severe-to-profound hearing loss caused by irreversible damage of the hair cells [3].

CIs recover hearing to a certain extent by directly stimulating the auditory nerves via electrodes. The current commercial CIs has some drawbacks, such as high cost, the need for frequent battery charging/replacement and the requirement of wearing external components. These result in interruption of patients' access to sound, discomfort and an increased the risk of damage, especially when the CI is exposed to an aqueous medium (e.g. shower, pool) [4,5]. Furthermore, conventional CIs eliminate the entire natural hearing mechanism. However, operational parts of the hearing system, such as the eardrum and ossicles, can be utilized by new generation implantable components. As an example, the vibrations of the eardrum can be detected by acoustic sensors eliminating the need for an external microphone.

* Corresponding author at: Middle East Technical University, Department of Electrical and Electronics Engineering, Ankara, Turkey.
E-mail address: kulah@metu.edu.tr (H. Külah).

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Recent advancements in the field point out that solution of the above-mentioned issues lies in the next generation fully-implantable, self-powered and stand-alone CIs. Earlier studies focused mainly on accelerometer-based implantable middle ear microphones for next generation CIs [6–11]. However, such systems require external control and RF battery charging units. MEMS accelerometers can be used as implantable sensors [12] but they lack sensitivity and suffer from high power consumption. Thermal and electromagnetic sensors were also used as alternative approaches [13], however they have a limited output voltage and an undesirably high system noise. Piezoelectric transducers are widely used to convert mechanical vibrations directly into electricity without a need for an external source [14]. Although piezoceramic devices have been proposed as promising alternatives for CI applications [15,16], there are still major scientific and technical challenges. Recently, Jang et al. reported successful use of a piezoelectric cantilever array for stimulation of the auditory neurons in deafened guinea pigs [17]. Yet, the maximum output voltage (<50 μV) is well-below the lower limit that could be detected by an interface circuit without external amplification. Furthermore, the working frequency of the implemented sensor does not cover the audible frequency range for humans.

Here, we propose a multi-frequency thin film piezoelectric acoustic sensor concept to overcome the main bottlenecks of CIs, considering the limitations regarding volume, mass and stimulation signal. The sensor is to be placed on the eardrum for fully-implantable cochlear implant (FICI) applications. The design consists of several thin film piezoelectric cantilever beams, each of which resonates at a specific frequency within the daily acoustic band. The device will exploit the functional parts of the natural hearing mechanism and mimic the function of hair cells, where the signal generated by the piezoelectric transducers will be processed by interface electronics to stimulate the auditory neurons in the cochlea.

In this paper, we developed a single-channel Pulsed Laser Deposited-Lead Zirconate Titanate (PLD-PZT) thin film piezoelectric acoustic transducer to demonstrate the feasibility of the proposed concept. We verified that the voltage output of this single-channel device is sufficiently high to be detected and processed by a readout circuit without any external power source. Acoustic and electrical performances of the developed transducer were characterized, where the experimental results were used to construct a finite element model to be used in the optimization of the design parameters of the final device covering the complete daily acoustic band.

2. Design and modelling

The feasibility of piezoelectric transduction due to the eardrum vibrations has already been demonstrated in the literature [17]. However, implementation of the method to next generation CIs has significant challenges and requires an advanced design procedure should be followed considering several limitations.

In the proposed system, an array of piezoelectric cantilever is to be placed on the eardrum or ossicles to provide the signal for neural stimulation. The major challenge in the design of the piezoelectric cantilevers is covering the daily acoustic band with an adequate number of channels within the small volume of the middle ear. The piezoelectric sensor output is to be processed by an interface circuit and converted into stimulation pulses. The quality of sound perception will typically be improved as the number of the channels increases, due to the increased resolution of stimulation frequencies. However, this also increases the hardware complexity and the power consumption. Therefore, a balance should be sought between the sound perception level and the power consumption.

It has been reported that the average hearing performance increases up to 8 channels, and no further improvement is observed with higher numbers of electrodes (10–20) [18,19]. Furthermore, a recent study from our group demonstrates that 8-channel FICI systems can operate with sub-mW power dissipation [20]. Considering these, an 8-channel transducer is considered to be good enough for the proposed design to cover the daily acoustic band (250–5000 Hz) and to provide adequate spectral resolution [20–22].

The limited volume (<0.1 cm^3) [23] in the middle ear, the mass tolerance (<25 mg) [24] and the size (9 mm × 10 mm) [25] of the eardrum are the main limitations for obtaining an adequate voltage output for neural stimulation. Beker et al. reported that use of bulk piezo-ceramics can generate a significant amount of energy [26], where a single-channel device occupies a minimum of 5 × 5 mm^2 footprint due to impact of the relatively thick bulk piezoelectric layer on the design parameters. It is obvious that such a device in multi-channel configuration cannot satisfy the eardrum dimension limitation. The footprint may be reduced by stacking the single-channel bulk transducers on top of each other, which results in an increased mass. This results in an increase in the level of challenge due to the coupled motion with eardrum and leads to a lower vibration amplitude at the same acoustic input level [27]. Therefore, the usage of bulk piezoelectric transducers as multi-channel acoustic sensors is not convenient for FICI applications.

Thin film piezoelectric materials can be integrated with MEMS in the desired volume [23], which makes them a promising alternative for the application. Pulsed Laser Deposited (PLD) Lead Zirconate Titanate (PZT) is preferred among other thin film piezoelectric alternatives due to their superior ferroelectric and piezoelectric properties for acoustic sensing [28]. Using PLD-PZT, a more compact multi-channel piezoelectric acoustic sensor can be designed, where all cantilevers can be placed on a single layer. This is possible since the thin film fabrication procedure allows for reduction of both the cantilever size and the distance between individual cantilevers. Consequently, mechanical filtering for an adequate number of channels to cover the daily acoustic band within the 25 mg maximum loading requirement will be facilitated, ensuring that there is no significant effect on the ear drum acceleration.

Fig. 1 shows the proposed system for sensing the incoming sound with a close-up view of the piezoelectric cantilever beams, each of which resonate at a specific frequency within the hearing band. The cantilevers are placed facing each other with increasing/decreasing lengths to result in a smaller footprint. When an acoustic sound pressure impinges on the eardrum, the cantilever beam with the resonant frequency matching the excitation frequency starts to resonate. Consequently, the proposed system provides mechanical filtering and shows a frequency selectivity mimicking the natural operation of the cochlea. Center frequencies of the 8-channel filter are distributed between 250 and 5000 Hz as it is in the existing CIs, where frequencies below and above 1200 Hz are spaced linearly and logarithmically, respectively [29,30].

Fig. 2 illustrates a single cantilever beam structure. Lengths of the cantilever beam, the thin film PZT layer and the tip mass are critical optimization parameters to obtain a sufficiently high signal within the volume and weight limitations. Beam length is varied for obtaining the desired center frequencies and the cantilever beams arranged to face each other in such a way that the array fits into 5 × 5 mm^2 footprint. Tip mass usage is inevitable in order to tune the resonance frequency of the transducers and the induced stress level, which increases the vibration level and voltage output. Tip mass thickness is optimized and fixed at 150 μm to obtain the lowest center frequency (300 Hz). PZT length is another critical design parameter to maximize the voltage output. Results show that maximum output occurs when % 44 of the cantilever length is covered with the active piezoelectric layer [31]. For this reason, we also use this value for our cantilevers.

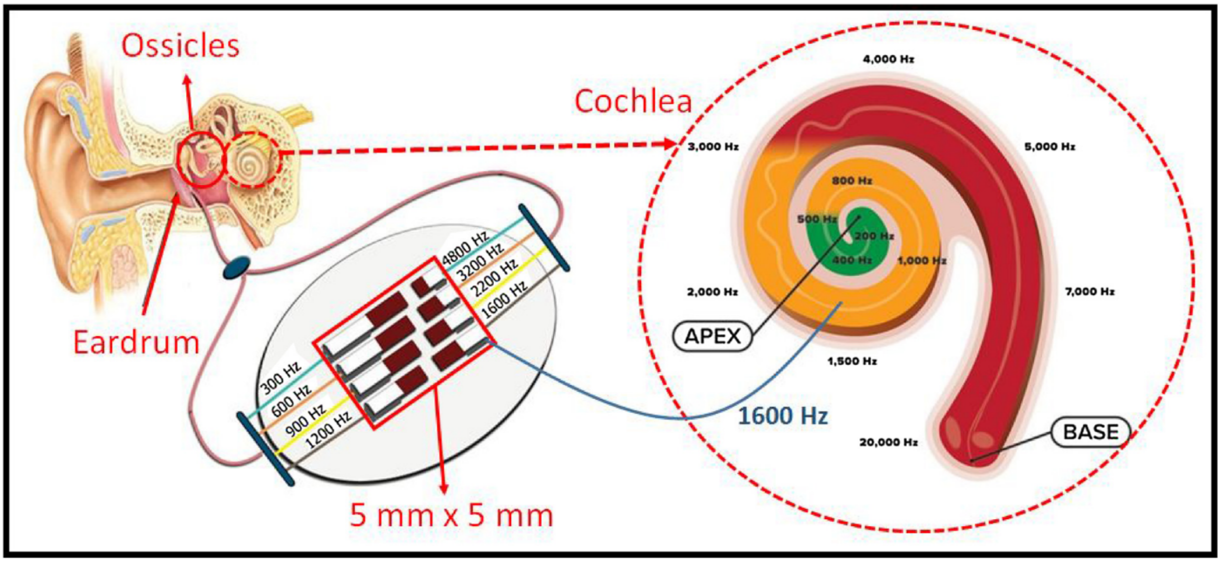

Fig. 1. The proposed system for sensing the sound with close-up views of the cochlea and the cantilever array.

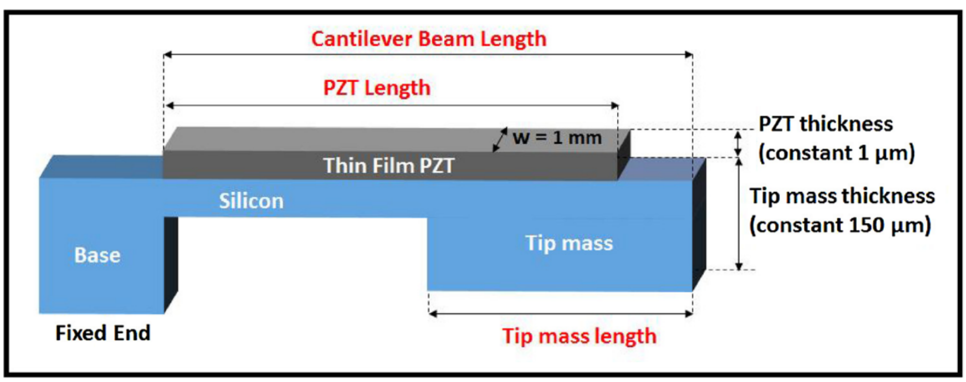

Fig. 2. Schematic view of the transducer.

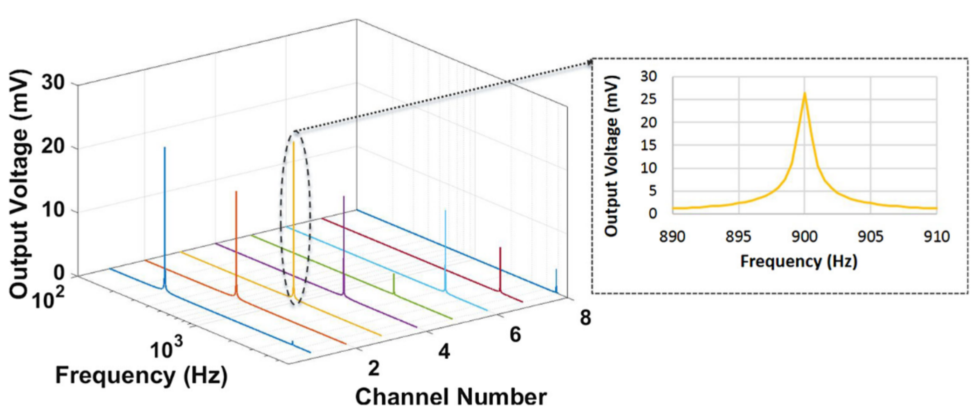

Fig. 3. Simulation results showing the frequency response of all channels with a close-up view of channel 3 (operating at 900 Hz).

A finite element model is established using the Optimization Module of COMSOL Multiphysics to design the 8-channel transducer within weight (<25 mg) and volume (<0.1 cm^3) constraints. The total volume and mass of the device were $5 \times 5 \times 0.2$ mm^3 and 12.2 mg, respectively, which are much lower than the limitations of the system. In order to characterize the device properties accurately, mechanical, electrical and squeeze film damping parameters are inserted to the COMSOL finite element simulations [32]. Fig. 3 shows the proposed 8-channel multi-frequency structure, where each cantilever corresponds to a selected frequency band in cochlea.

Table 1 lists the obtained piezoelectric output voltage, sensitivity to sound and quality factor for each frequency. Results demonstrate that the proposed design has a clear frequency selectivity with a minimum quality factor of 1285 and mimics the natural operation of cochlea. Both the sensitivity and the quality factor of the proposed system are higher than the state-of-the-art piezoelectric transducers [17]. Results show that the device generates

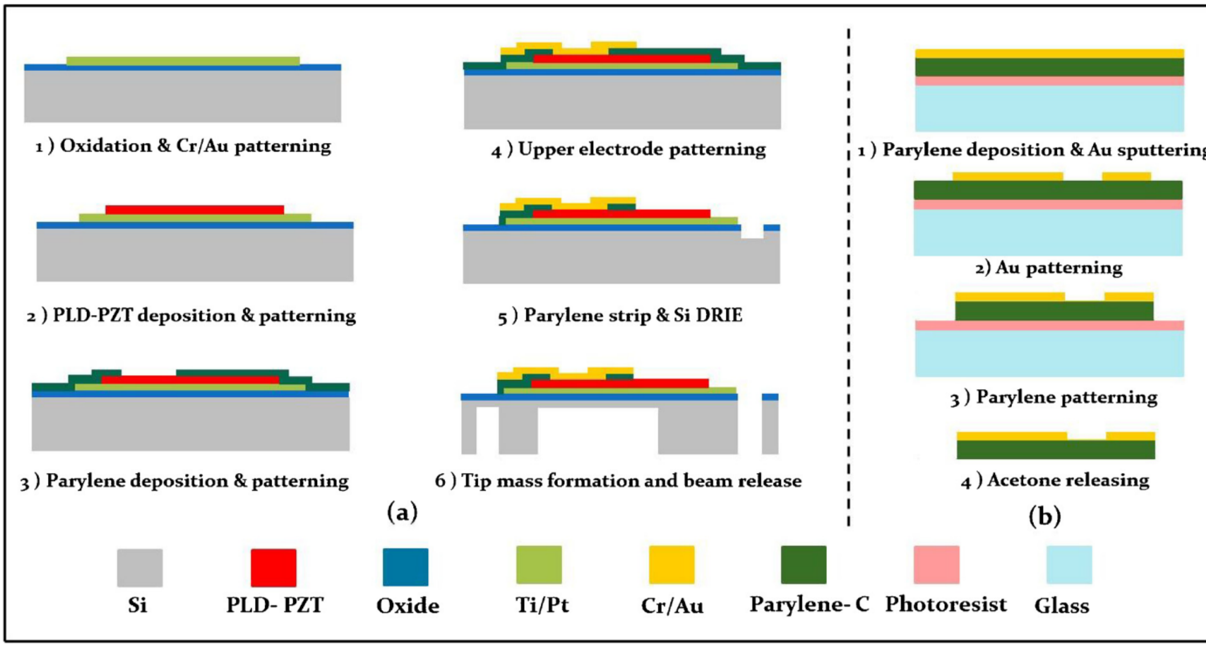

Fig. 4. Fabrication flow of the thin film sensor (a) and the flexible Parylene carrier (b).

Table 1
Specifications of 8-channel structure obtained through finite element simulations.

Frequency (Hz)	Beam length (mm)	Output Voltage (mV)	Sensitivity (mV/Pa)	Quality Factor
300	3.4	22.98	363.34	984
600	2.4	16.87	265.4	1012
900	1.9	24.79	391.9	1285
1200	1.7	15.88	251.2	1196
1600	1.4	22.71	358.94	976
2200	1.2	13.12	204.6	1043
3200	1	7.21	114.1	996
4800	0.8	3.75	59.3	1121

sufficient output voltage considering the reported sensing voltage requirement of a neural stimulation circuitry for auditory neurons [30]. The proposed system replaces the electrical filters with mechanical ones, which results in a decreased power dissipation by the circuit, by at least 40 μW [33]. The power dissipation further decreases due to the higher signal-to-noise ratio of the thin film PZT transducers, which result in a lower power requirement by the interface circuitry for detecting the signal. Therefore, it can be anticipated that the battery lifetime of an implantable system increases and the frequency of battery charging decreases.

3. Materials and methods

A 6-mask process was developed for the microfabrication of the thin film PLD-PZT transducer. Fig. 4(a) shows the detailed flow diagram of the fabrication process. Initially, a 500 nm-thick PECVD SiO_2 layer was deposited on the front side of a 4-inch (100) silicon wafer to provide isolation of the transducer electrodes. Then, 10 nm Titanium (Ti) and 100 nm Platinum (Pt) layers were sputtered as the seed layer for the PLD-PZT deposition. The PLD-PZT (1 μm) layer was deposited at Solmates BV (SMP-700 PLD) and later, patterned using the PZT etchant provided by the same company. The Ti/Pt layer was utilized as the bottom electrode of the transducer and patterned in a hot aqua-regia solution [34]. Following the formation of the bottom electrode, a 5 μm Parylene-C layer was deposited (SCS PDS2010) for insulation between the bottom and the top electrodes and patterned using a reactive ion etching (RIE) process. Chromium/Gold (Cr/Au) (30/400 nm) was used as the top electrode and patterned using a lift-off process. The remaining Parylene-C layer around the top electrode pattern was stripped using RIE. Finally, the cantilever beam and the tip mass structure were formed with front and back-side deep reactive ion etching (DRIE) processes.

The thickness of the tip mass defines the resonance frequency of the transducer. For the backside DRIE step, spray coating was used for patterning to provide a conformal coating. During this step, a frame was also patterned around the chips to allow release of individual devices without dicing. All processes were realized at low temperatures (<120 °C), which are well below the poling temperature (200 °C) of PLD PZT [35].

Fig. 4(b) illustrates the process flow of the flexible carrier with contact pads/lines, which will be used during the acoustic experiments. The carrier should allow for easy placement of the device on the membrane, while providing electrical connections. Hence, a flexible and biocompatible material was required. For this purpose, a 20 μm thick Parylene-C was coated on a glass wafer to ensure biocompatibility (approved by FDA). Cr/Au (30/500 nm) was deposited on the Parylene layer as the metallization layer and patterned using wet etching. Finally, the Parylene was peeled-off after pattering by RIE, to form the flexible carrier system.

A Parylene-C elliptical membrane with a thickness of 40 μm was fabricated to obtain a vibration response similar to that of a combined eardrum-cantilever system during the acoustic experiments [36]. Diameters of 9 mm and 10 mm were used considering the typical eardrum dimensions [25].

4. Results and discussion

For demonstrating the feasibility of the proposed structure, a single cantilever thin film PLD-PZT prototype was designed and fabricated. The realized device and the experimental setup for performance analysis can be seen in Fig. 5. Footprint of the microchip is optimized to fit within $5 \times 5\ mm^2$ by considering the eardrum dimensions. The PLD-PZT transducer was assembled onto a custom-made PCB, where the electrical connections between the contact lines on the PCB and the device electrodes were established

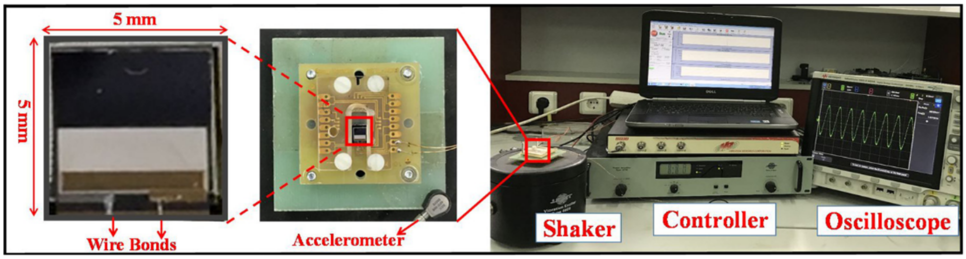

Fig. 5. The fabricated device with a close-up view of the transducer (left) and the experimental setup used for performance evaluation (right).

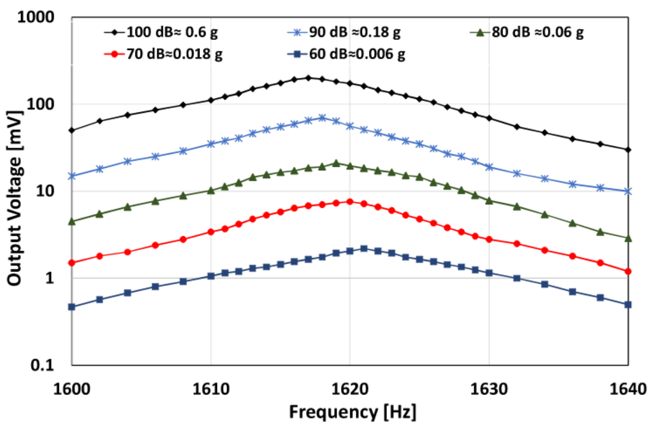

Fig. 6. Frequency response of device at different umbo vibration levels.

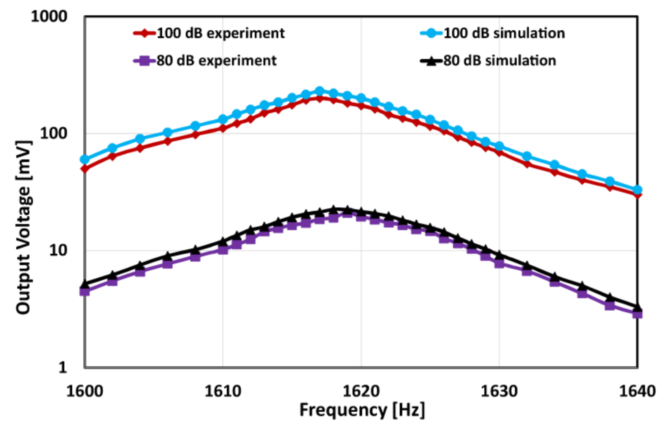

Fig. 7. Simulation and shaker table test results of the device at two different SPLs.

using wire bonding. The PCB assembly was mounted on a shaker table, which can be vibrated at various acceleration and frequency levels set via a controller system. The voltage generated by the PLD-PZT transducer due to the vibration of the shaker table was monitored through an oscilloscope.

The frequency response of the PLD-PZT transducer was obtained using the standard shaker table setup. Fig. 6 shows the generated voltage waveform from the device around its first resonance (1620 Hz) at 0.018 g acceleration. A decrease in the resonance frequency is observed with increased acceleration levels due to the ferroelectric properties of piezoelectric materials [37]. The frequency response of the device were acquired for acceleration levels from 0.006 g to 0.6 g, each of which corresponds to the vibration of the umbo at specific sound levels from 60 dB to 100 dB, respectively [30]. The output of the acoustic sensor is going to be processed through an interface within the dynamic hearing range in the final device. A current generator circuit will convert the sensed signals to current pulses, which will be sent to the corresponding electrodes to stimulate the auditory neurons according to the incoming sound level. The critical point is to generate the high output voltages in order to obtain a higher signal to noise ratio (SNR). Interface circuitry will consume less power for detecting the piezoelectric output signal at higher SNR levels. In a relevant study, average noise level over the sound processor is reported as 1.93 μV [30]. Experimental results show that the prototype device generates 2.6 mV at 60 dB Sound Pressure Level (SPL), which provides high enough SNR when compared with the reported sensing voltage of state-of-the-art neural stimulation circuitry for auditory neurons [17]. High output voltages at higher sound levels facilitate implementation of FICI systems and decrease the power requirement of the system, which is highly critical for FICI applications.

Fig. 7 shows the results of the finite element analysis carried out using COMSOL Multiphysics. Quality factors of experimental and simulation results at 100 dB are 162 and, 154 respectively, while these numbers are found as 135 and 129 at 80 dB. The simulation results are within 92% agreement with the experimental results. Experimental results are slightly lower than simulation results due to the non-uniformity of the beam thickness, which can be reduced by optimizing the process parameters at DRIE stage of the fabrication or using a silicon-on-insulator (SOI) wafer for the fabrication.

The transducer filters the sound mechanically by exciting only the beam with the matching resonance frequency. This system shows clear separation of frequency and provides an accurate excitation signal as an acoustic sensor mimicking the natural operation of the cochlea. The proposed design benefits from eardrum vibrations through frequency selective piezoelectric cantilevers to generate the signals for neural stimulation. This eliminates most of the power-hungry electronics, such as microphone and active band filters, while keeping the healthy portions of the middle ear functional. Hence, the proposed 8-channel thin film multi-frequency cantilever array model shows the feasibility for next generation FICIs, which can stimulate nerves while covering the acoustic band with enough number of channels in limited volume and mass.

Fig. 8 shows the acoustic setup used for characterization of the PLD-PZT transducer prototype on a vibrating membrane, mimicking the behavior of the eardrum. The device was glued onto the flexible Parylene carrier via epoxy bumps. Electrical connections between the device and the contact pads on the flexible carrier were established through wire bonding and conductive (silver) epoxy. Electrical connections between the carrier and the equipment were realized using a zero-insertion-force (ZIF) connector. The assembly was placed on a 40 μm thick Parylene membrane, which mimics the eardrum characteristics. The membrane was fixed at the end of a flexible hollow tube (castermid) using double-sided tape. The inner diameter and the length of the tube are 9 mm and 3.5 mm, respectively, to match the dimensions of the ear drum and ear canal. The stress level of the membrane was adjusted by stretching and loosening it via an adjustable structure on the flexible hollow tube.

Sound signals were applied to the fabricated ear canal model through an insert earphone, ER-2, system. The ER-2 provides a flat response for desired measurement interval. It is driven by an audio amplifier that presents pure tones within the audible frequency

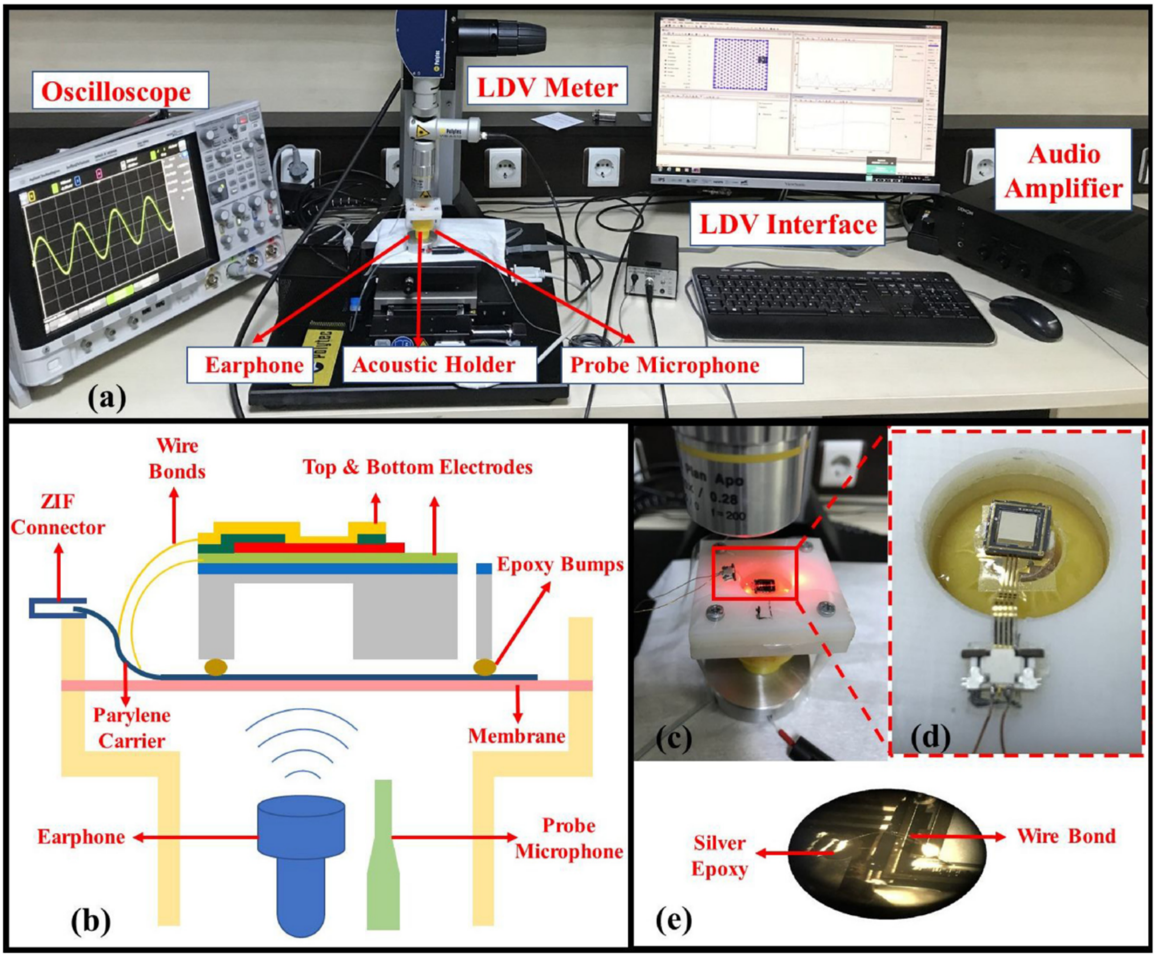

Fig. 8. **(a)** The acoustic characterization setup with **(b)** a schematic view showing the details. **(c)** Flexible hollow tube and the sensor under LDV measurement. **(d)** The sensor and carrier attached on the membrane. **(e)** Tilted view of the glued chip showing the wire bonds and the silver epoxy.

spectrum. The sound is transmitted through the ear canal model to vibrate the flexible membrane. A probe microphone is positioned approximately 4 mm from the membrane for calibration of the incoming sound. The displacement output was measured and recorded with a Laser Doppler Vibrometer (LDV) system. The frequency response of the eardrum-cantilever system was obtained under different SPLs above 60 dB, which is the lower limit of the utilized LDV measurement system. While the piezoelectric transducer vibrates with the incoming sound, the generated voltage was monitored, simultaneously via an oscilloscope.

Fig. 9 presents the average peak-to-peak acceleration of the device measured within the audible frequency spectrum under different sound levels. As the membrane vibrates with incoming acoustic sound, the sensor starts to vibrate along with the membrane. It is critical to derive sensor acceleration characteristics along the membrane from the measured vibration behavior by taking a second-order differentiation of the displacement in time domain. It is observed that the membrane acceleration amplitude at a given frequency interval has a 20 dB per decade slope, indicating a linear relationship between the acceleration amplitude and the input sound pressure level.

Results show that the sensor on the membrane has a maximum 0.054 g at its first resonance frequency of 1325 Hz which corresponds to 110 dB input SPL. The measured acceleration demonstrates clear frequency selective characteristics, which is directly related to acoustic properties of the membrane-cantilever assembly. The acceleration and frequency characteristics of the vibrating membrane is highly dependent on the sensor attached on it due to the coupled motion. Acoustic response of the membrane changes as a result of the added mass of the sensor and the diaphragm characteristics, hence leading to deviations in the frequency and damping characteristics [38].

While the membrane transforms the incoming acoustic pressure waves into base vibrations in the ear canal, the piezoelectric sensor starts to deflect according to the acceleration level as in Fig. 9. This deflection creates a polarization within the piezoelectric material and the sensor generates voltage by converting the mechanical motion of acoustic sound into an electrical signal and act as the eardrum itself. Fig. 10 shows the voltage output of the thin film piezoelectric transducer that was attached onto an acoustically vibrating membrane. Results show that the device generates 114 mV_{pp} at 110 dB SPL. Although the membrane suppresses the voltage generation of the sensor, it is still able to generate sufficient voltage for auditory nerve stimulation [30]. Generating higher voltage from the sensor will enable more accurate and sensitive detection by the readout circuit in the final FICI device. Also, higher voltages decrease the noise level and eliminates additional amplification which enables reduction in stage number at readout circuitry. This decreases the power requirements of the overall system and hence eliminate the battery requirement of cochlear implants, which is one of the main bottleneck of conventional CIs.

Recent studies show that critical bandwidth for speech recognition varies around 150–300 Hz in the audible frequency range [39,40]. The study also indicates that required normalized bandwidth are slightly wider for the low frequency channels than for the high frequency channels. The device shows clear mechanical sen-

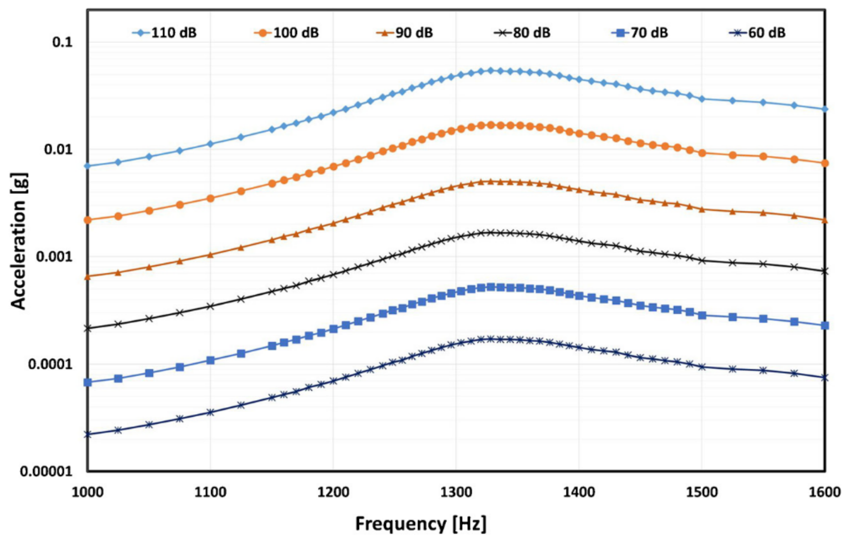

Fig. 9. Membrane on chip acceleration characteristics for sound pressure levels from 60 to 110 dB SPL.

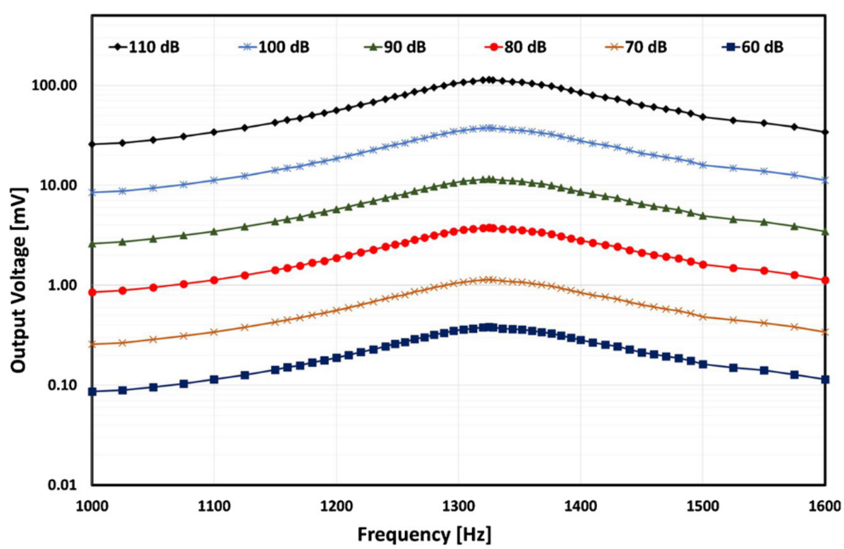

Fig. 10. Measured acoustic results of the sensor for sound pressure levels from 60 to 110 dB SPL.

sor characteristics, which has a 150 Hz bandwidth around 1325 Hz. Results indicate that, bandwidth and sensitivity of the design are sufficient for detecting the hearing in audible frequency range and better than state of art thin film sensors [17]. Also, a piezoacoustic sensor with such band pass characteristics will simplify the signal processing electronics and provide patients' continuous access to sound at every medium. This is foreseen to outplace the need for using external components, eliminating the aesthetic concerns of patients and creating a paradigm shift in operational principle of the conventional CIs.

5. Conclusions

This study presents modelling, fabrication and characterization of a cantilever-based thin film PLD-PZT acoustic sensor to be used in a proposed multi-channel sensor to be placed on the eardrum for fully implantable CI applications. The proposed model utilizes the natural hearing mechanism and mimics the hair cells via a set of frequency-selective piezoelectric cantilevers to stimulate the auditory nerve. The proposed design is expected to solve the main challenges of cochlear implants by minimizing the mass, footprint, and volume of the transducers so that they can fit into middle ear, without sacrificing the generated voltage levels. As a proof-of-concept prototype, a single channel thin film PLD-PZT chip was designed and fabricated. The realized device was assembled onto a flexible carrier and placed on a Parylene membrane, mimicking the operation of the eardrum. The mechanical, electrical and acoustic properties were characterized by a standard shaker table and a more advanced acoustic setups for the first time. Experimental results show that the voltage output of the device exceeds the minimum required sensing voltage for the neural stimulation circuitry and decreases the required power for readout circuitry. Agreement of simulation and experimental results clearly indicate that the proposed multi-frequency thin film model satisfies all the requirements of a FICI system. Hence, the feasibility of the proposed next generation FICI system, which can stimulate nerves while covering the acoustic band with enough number of channels is verified. The proposed design is also predicted to solve the main challenges of cochlear implants such as damage risk of external components, high cost and aesthetic concerns of patients.

Conflicts of interest

The authors declare no conflict of interest.

Author contributions

Authors, specifying their individual contributions must be provided.

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Biographies

Bedirhan İlik received his B.Sc. degree in Electrical and Electronics Engineering from Middle East Technical University (METU), Ankara, Turkey in 2015. He is currently pursuing his M.Sc.in Electrical and Electronics Engineering in METU and employed as Research Assistant at FLAMENCO Project. His research interests include design and fabrication of piezoelectric sensors, MEMS energy harvesting systems, Integrated circuits, sensors and actuators, the wearable and implantable medical applications.

Aziz Koyuncuoğlu received his B.Sc. and M.Sc. degree in Mechanical Engineering from Middle East Technical University (METU), Ankara, Turkey in 2007 and 2010, respectively. He is currently a Ph.D. candidate in Micro Nano Technology Program in METU and employed as researcher in FLAMENCO Project. His research interests include MEMS energy harvesting systems, piezoelectric sensors for biomedical implants, microfluidics and micro thermal systems.

Özlem Sardan Sukas received her BSc and MSc degrees in mechanical engineering from Middle East Technical University, Turkey, in 2004, and Koç University, Turkey, in 2006, respectively. During her PhD, she worked on "Topology Optimized Electrothermal Microgrippers for Nanomanipulation and Assembly" within the EU project NanoHand. She received her PhD degree from Technical University of Denmark in 2010. Her research activities include development of microelectromechanical actuators and sensors for various applications. She is currently affiliated by the Vrije Universiteit Brussels, where her current research focuses on development of a three-dimensional visualization and tracking system for microparticles and biological samples.

Prof. Dr. Haluk Külah received B.Sc. and M.Sc. degrees in electrical engineering with high honors from METU, Ankara, Turkey, in 1996 and 1998, respectively. He received Ph.D. degree in electrical engineering from the University of Michigan, Ann Arbor, in 2003. During his Ph.D., he worked on micro machined inertial sensors and their interface electronics. His interface electronics designs received several awards in design contests organized/sponsored by prestigious institutions including by DAC (Design Automation Conference), IBM, Analog Devices, Compaq, and Texas Instruments. From 2003 to 2004, he was employed as a Research Fellow at the Department of Electrical Engineering and Computer Science, the University of Michigan. Dr. Külah joined the Electrical and Electronics Engineering Department of METU as a faculty member in August 2004. He is also the Deputy Director of METU-MEMS Research and Applications Center since 2008. His research interests include MEMS, MEMS-based energy scavenging, microsystems for biomedical applications (BioMEMS), and mixed-signal interface electronics design for MEMS sensors. Dr. Külah has more than 130 international publications, 23 national/international patents, and 8 international patent applications. He has been awarded with 2009 Research Encouragement Award by Prof. Mustafa PARLAR Education and Research Foundation, 2013 Research Encouragement Award by TÜBİTAK, 2013 IBM Faculty Award, and 2015 Young Scientist Award by Turkish Science Academy. Dr. Külah has received 2015 ERC Consolidator Grant by FLAMENCO project, which on autonomous and fully implantable cochlear implants.