ALN PMUT-BASED ULTRASONIC POWER TRANSFER LINKS FOR IMPLANTABLE ELECTRONICS

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ABSTRACT

The present work reports on the first demonstration of acoustic power transfer through the use of Aluminum Nitride (AlN) Piezoelectric Micro Machined Ultrasonic Transducer (PMUT) arrays at transmission distances suitable for intra-body powering applications. In contrast to Lead Zirconate Titanate (PZT), used as the piezoelectric in state-of-the-art devices, AlN offers higher biocompatibility and the possibility of single chip integration with Complementary Metal Oxide Semiconductor (CMOS) technology. An output power of 1 μW, from a minute (8 mm x 8 mm x 300 μm) PMUT array chip, was achieved on an optimal 330 Ω load. The operating frequency was 2 MHz, through a distance of 4 cm, in an oil medium resembling intra-body conditions. This implementation is a critical step towards increasingly miniaturized, highly integrated implantable circuits.

KEYWORDS

PMUT, Piezoelectric Micromachined Ultrasonic Transducers, Power Transfer, Implantable, Aluminum Nitride

INTRODUCTION

Deep Brain, Vagus Nerve, and Peripheral Nerve Stimulation are therapeutic techniques that have been shown over the last years to aid in the treatment of complex diseases such as epilepsy, depression and chronic pain [1]. Similarly to pacemakers, they operate by stimulating areas of the nervous system with designed sequences of electrical pulses. The systems are currently battery operated, which makes the implanted devices bulky and increases risk of infection as replacement of depleted batteries has to be done surgically.

There has also been a trend over recent years towards the vision of the Internet of Medical Things in which, increasingly miniaturized, wearable or implantable devices gather information from bodily variables such as blood flow, oxygen concentration or electrocardiogram (ECG) and transmit it securely over the internet for chronic patient monitoring or research [2].

Considering these advancements, it is possible to envision a system such as the one shown in Figure 1 in which the batteries of the described implantable devices can be wirelessly recharged or even run completely powered by transducers that are placed externally to the body where constraints on batteries are less demanding.

Within the wireless powering strategies for implanted circuits implementing this functionalities, such as radio frequency, inductive or optical power transfer, acoustic powering offers the least medium attenuation as well as the highest allowed intensity threshold for safe operation.

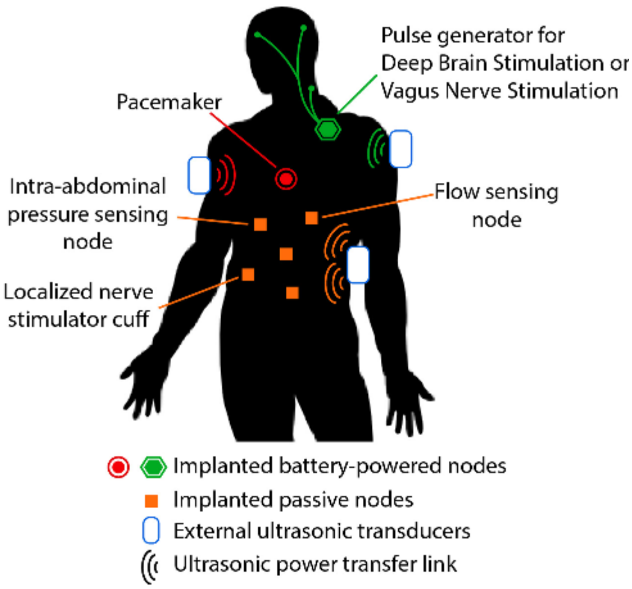

Figure 1: Envisioned application: AlN PMUT-based ultrasonic receivers enable wireless acoustic powering of miniaturized implantable sensors and stimulator nodes. Larger transmitters and batteries are placed out of the body, where size constraints are less stringent.

Previous work towards ultrasonic powering of implantable devices demonstrated a peripheral nerve stimulator [3], an optogenetic stimulator [4], a piezoelectric energy harvester [5], a micro-oxygen generator [6] and a data link [7]. These systems are all powered by an external ultrasonic transmitter, are within mm scale dimensions, and are based on PZT as the piezoelectric in the receiver.

Even though PZT has a higher electromechanical coupling coefficient, AlN offers higher biocompatibility and can be integrated with CMOS processing, empowering dramatic miniaturization. Current demonstrations of power transfer based on AlN both operate at short distances and utilize air [8] or chip packaging [9] as the propagation medium.

The present work brings the AlN PMUT array technology to the intra-body powering realm by utilizing these devices as power receivers in a liquid medium covering link distances enough for powering implantable circuitry.

PMUT ARRAY DESIGN

The first parameter to be defined in order to achieve an efficient power transfer link through the chosen medium is the frequency at which the ultrasonic waves will propagate. Besides the inevitable $1/r$ geometric spreading, ultrasonic wave propagation suffers from a medium attenuation factor which scales with the square of the operating frequency [10]. From this point of view, a relatively lower frequency of operation is desirable.

PMUTs consist of a membrane including the piezoelectric material, with electrodes on its top and bottom faces, on top of a passive layer (Figure 3b). When using PMUTs as receivers, an incoming acoustic wave produces deflection of the membrane and generates a voltage based on the piezoelectric effect. This displacement is enhanced at the specific resonance frequency of the device. If the thickness of the membrane stack is set, for operating at a lower resonance frequency, the cavity size needs to be larger. This results in a lower flexural rigidity , which increases the displacement sensitivity to incoming pressure waves as well as the increased produced voltage.

For the transmitting piezoelectric element, however, the magnitude of the generated pressure increases with frequency [11], so a trade-off exists. A frequency of the link of 2 MHz was chosen to avoid excessive attenuation, and both achieve acceptable receiving sensitivity on the PMUT array and generated pressure amplitude on the transmitter. The corresponding PMUT radius to achieve this resonance frequency can be then calculated based on:

$$f_0 = \frac{1}{2\pi}\sqrt{\frac{3.2D}{a^4\rho}} \tag{1}$$

where ρ is the area mass density, D is the flexural rigidity and a is the PMUT radius [12].

To further increase the output power, it is desirable to increase the number of elements connected in parallel in the array, but considering th application of miniaturized implantables, the maximum size has to be constrained. Directivity of the array also increases with its size, producing a beam that is more focused and allows for a higher gain as a receiver. However, due to body motions and the inability to directly see and point to the implanted devices, some angle tolerance should be allowed in the design for the envisioned system. It was then settled on an 70 x 70 array whose directivity calculated from an analytical model [11] is shown in Figure 2.

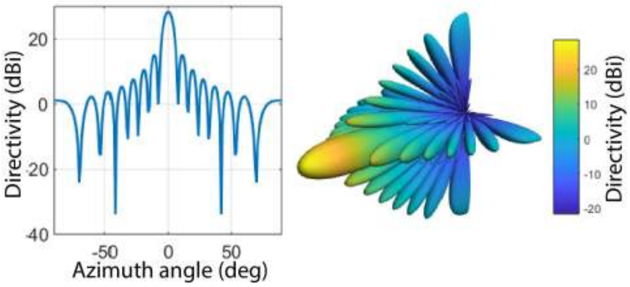

Figure 2: 2D and 3D directivity for the array. The beamwidth is approximately 16 degrees.

FABRICATION

A double-side polished, 300 μm silicon wafer was used as a substrate and an 850 nm low temperature silicon dioxide layer was then deposited through Plasma Enhanced Chemical Vapor Deposition (PECVD). A 5 nm Ti stiction layer and a 95 nm Ti bottom electrode were then deposited by electron beam evaporation and patterned through a photo lithography and liftoff process.

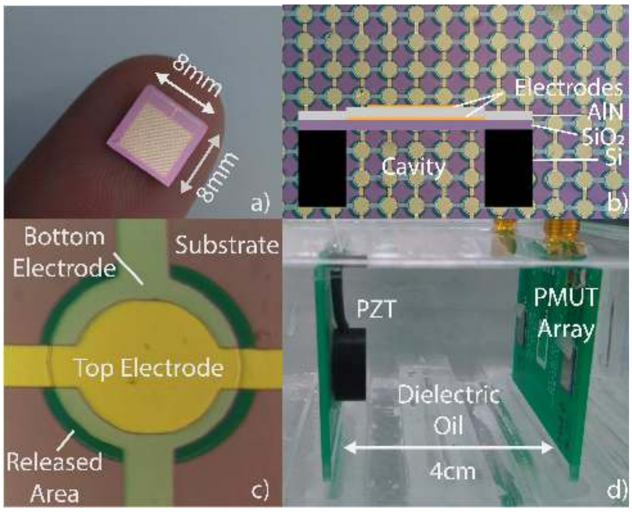

Figure 3: PMUT Array; a) Fabricated chip on a human finger, b) Optical image of array elements and connectors, inset detailing device structure, c) Individual PMUT optical image, d) Device in test setup.

750 nm of AlN to form the piezoelectric layer were then reactively sputtered and vias were etched through the layer for access to the bottom electrode by hot phosphoric acid etching. 5 nm Ti as a stiction layer and 95 nm Au as the top electrode were sputtered. Backside alignment photo lithography was then used to pattern a PECVD oxide hard mask through Reactive Ion Etching (RIE) for later Deep Reactive Ion Etching (DRIE) to define cavities and release the device membrane (Figure 3).

METHODOLOGY

A 70 x 70 element PMUT array was micro-fabricated (Figure 3). The array elements were connected in parallel to increase the total output current while the output voltage is determined by the output of a single element. The wire bonded array was used as a receiver in a dielectric oil tank as the medium. The ultrasonic transmitter element used was a bulk PZT transducer with a frequency matching the resonant frequency of the PMUTs. Two experimental setups were implemented for characterizing the resulting power transfer link: one with a Vector Network Analyzer measuring the frequency response and another with an ultrasonic pulser circuit sending high voltage bursts to the PZT (Figure 4). A bursting technique achieves higher output voltage on the receiver as it prevents standing wave effects due to the reflection of continuous waves (Figure 6).

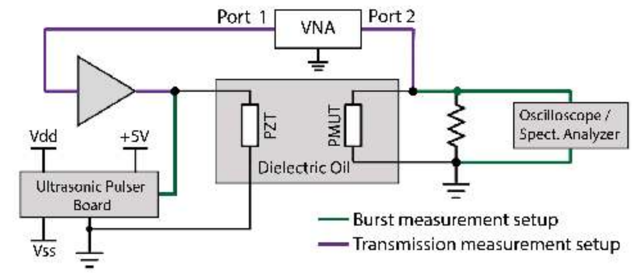

Figure 4: Schematic of measurement setups. The transmitter is pulsed in the burst measurement to avoid standing wave effects. VNA measures the ultrasonic channel's continuous wave frequency response.

RESULTS

Frequency response measurements were first obtained to characterize the channel. A frequency sweep of the PMUT displacement amplitude in air performed with a Digital Holographic Microscope revealed a resonance frequency of 2.46 MHz, which reduces to 2 MHz in oil due to medium loading (Figure 5b). The transmitter element was also verified to operate at 2 MHz through a S_{11} VNA measurement (Figure 5a: lower reflection coefficient means higher coupled power to the acoustic domain). Then, the full channel was characterized by an S_{21} VNA measurement which showed a well-defined transmission band around 2 MHz (Figure 5c). Although the bandwidth of operation of the PMUT array in liquid is relatively large, the full system bandwidth is defined by the PZT.

Once the channel's center frequency was characterized, a 5-cycle burst was sent to the transmitter and the output from the receiving PMUT array was measured (Figure 6). A delay between the burst and the received signal appears matching the propagation delay of the acoustic wave for its specific travel speed. The output of the burst test was also measured with a spectrum analyzer to verify its frequency content (Figure 5d).

The optimum load resistance for maximum power transfer was found in a circuit simulator implementing the channel transmission (S_{21}) parameters and verified experimentally by repeating the burst experiment varying the load, with good agreement between measured and simulated values (Figure 7).

A reference hydrophone was used to measure the ultrasound intensity on the medium and its value is well below the Food and Drug Administration's safety threshold of $720\mathrm{mW/cm^2}$, leaving ample room for output power improvement by increasing the power on the transmitter.

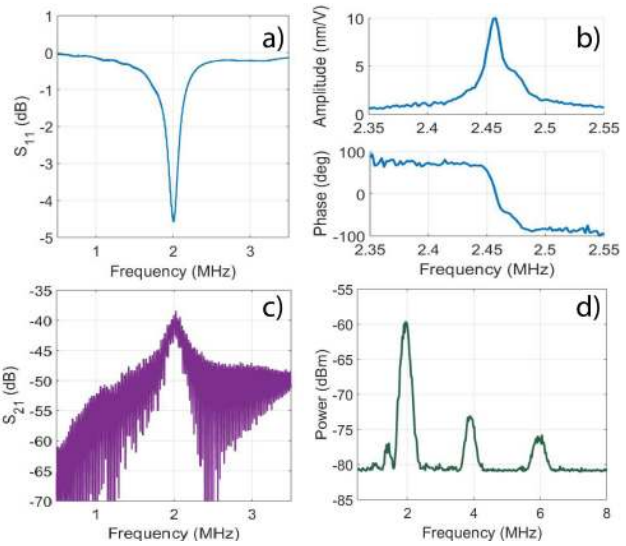

Figure 5: Frequency response characterization; a) PZT transducer reflection coefficient, b) PMUT array Digital Holographic Microscope (DHM) displacement measurement in air, resonant frequency reduces to 2 MHz in oil, c) System transmission measurement (continuous wave), d) Output power spectrum for burst test (50 Ω load due to test equipment standard impedance).

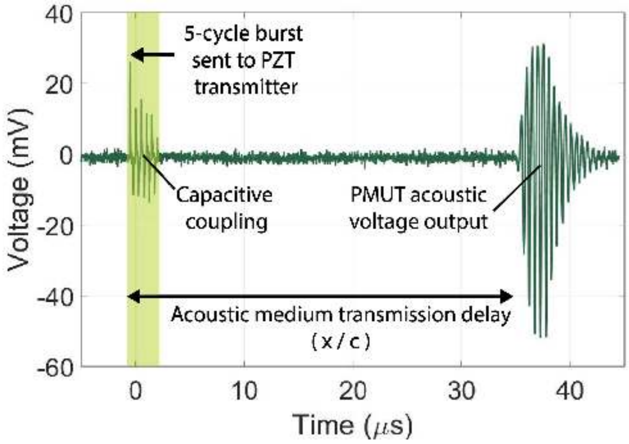

Figure 6: Time domain PMUT array output. Pulser board sends a 5 cycle, 95V square wave burst at 2 MHz to the PZT transmitter. The x/c delay to the appearance of the received signal confirms its acoustic nature.

The power output at 4 cm (1 μW) is adequate for powering Application Specific Integrated Circuits (ASICs) [5] and results in $7.1\mathrm{\mu W/cm^2}$ at a $77\mathrm{mW/cm^2}$ ultrasound intensity. These values are comparable to the state-of the-art but are obtained at a larger medium depth, which shows promising potential for AlN PMUT arrays as implantable power receivers.

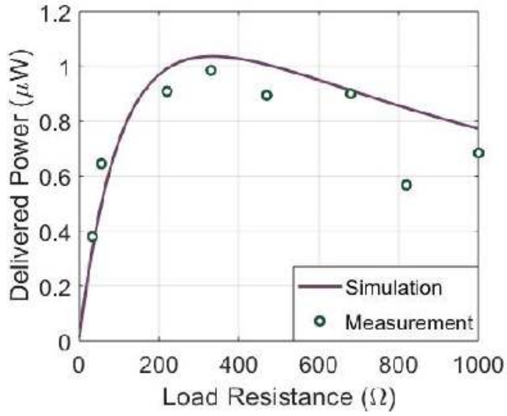

Figure 7: Simulated power transfer load optimization. Simulation implements the transmission values (S_{21}) for a 4 cm distance between transmitter and receiver. Measured power output for distinct load values shows good agreement with simulation.

CONCLUSIONS

An acoustic power transfer link featuring an AlN PMUT array as a receiving element was successfully implemented in a medium resembling intra-body conditions at distances that can enable implantable devices by wireless power transfer. A 70 x 70 PMUT array was able to output $7.1\mathrm{\mu W/cm^2}$, 4cm away from the transmitter. The acoustic pressure on the membrane face of the transducers was measured to be $77\mathrm{mW/cm^2}$, which is far below the $720\mathrm{mW/cm^2}$ FDA safety threshold which leaves room for further power increase. The power transfer channel was first characterized in terms of its frequency response, verifying a 2 MHz band and then burst tests were performed with a varying load to obtain the maximum power transfer condition.

While maintaining minimal dimensions (8 mm x 8 mm x 300 µm), the present implementation achieves an output that is adequate for powering ASICs without utilizing PZT as the piezoelectric which has the advantage of bio compatibility and ease of CMOS integration.

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