A CMOS 2D Transmit Beamformer With Integrated PZT Ultrasound Transducers For Neuromodulation

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***Abstract*—While the mechanisms are not yet completely understood, ultrasound-based neuromodulation has been emerging as a noninvasive modality for interfacing to both the central and peripheral nervous systems, due to its high penetration depth and good spatial resolution. Commercially available ultrasound transducers for neuromodulation applications are typically single-element focused transducers with a bulky form factor and off-the-shelf electronics for drive. Changing the focal position requires mechanical movement of the transducer itself. High-density ultrasound phased arrays allow for electronic focusing. Here, we present a CMOS 2D beamformer with integrated lead zirconate titanate (PZT) ultrasound transducers for neuromodulation of the peripheral nerves. The proposed prototype can achieve a maximum focal pressure of approximately 100 kPa with a 5 V supply at 0.5 cm depth without including an acoustic matching layer.**

Keywords—ultrasound neuromodulation; beamforming; phased-array; CMOS ultrasound interface; PZT

I. Introduction

Neural interfaces have been widely developed with the purpose of providing a path for communication with the central (CNS) and peripheral nervous systems (PNS), for both basic science research and clinical applications. The ideal interface has high spatial resolution and depth of penetration while being non-invasive. These characteristics, however, are typically mutually exclusive: non-invasive techniques such as transcranial direct-current stimulation (tDCS) and transcranial magnetic stimulation (TMS) suffer from reduced spatial resolution, while high-resolution interfaces require surgically implanted electrode arrays, as in the case of deep brain stimulation (DBS) for management of Parkinson's disease or electrocorticography (ECoG) for epilepsy. Focused ultrasound stimultion (FUS) has been emerging as a way of achieving both high spatial resolution (sub-mm) and high depths of penetration (several cm) while remaining non-invasive, which can potentially provide surgery-free therapeutics.

Studies have shown ultrasound-evoked neuronal activity in mice with pressure ranges in the few MPa [1], hundreds of kPa [2], [3] and even below 100 kPa [4]. The human primary visual cortex has also been excited with pressures in the MPa range [5]. However, all of these studies are still executed with single-element focused transducers with bulky form factors, interfaced with off-the-shelf electronics. As such, they must be mechanically moved (e. g. with motorized stages, as shown in Fig. 1a) to change focal positions and do not translate to wearable form factors for human therapeutics. Two-dimensional (2D) phased arrays are required to create electronically steerable

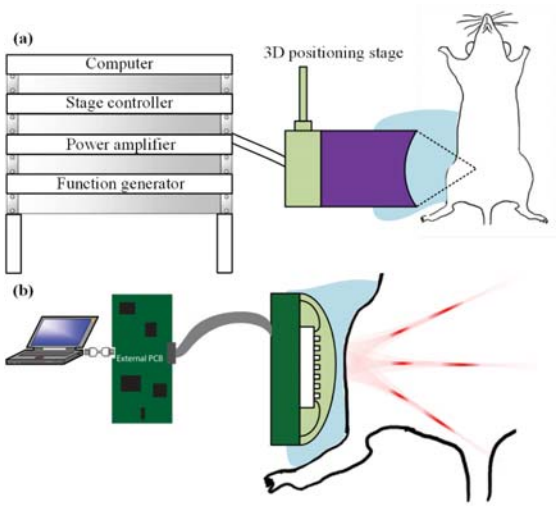

Fig. 1. Ultrasound neuromodulation example. (a) With conventional ultrasound setup from literature and (b) conceptual setup with proposed ultrasound chip for peripheral nerve stimulation in mice.

FUS pressure spots for these applications. The monolithic integration of a 2D array of ultrasound transducers, that is, directly on top of a CMOS chip, is essential to achieve such a phased array design of any considerable size, due to the high density of interconnections required. Close integration also allows parasitic capacitances between electronics and transducers to be reduced substantially. The focus of CMOS-integrated ultrasound array prototyping efforts thus far has been imaging [6]–[8] rather than therapeutics, where typically the receiver channel count largely exceeds the transmitter count. In this work, we present an ultrasound 2D beamformer on CMOS with integrated lead zirconate titanate (PZT) piezoelectric transducers targeting neuromodulation of the peripheral nervous system, as illustrated in Fig.1 b. Section II describes the system architecture and design details. Section III presents the experimental results and Section IV concludes.

II. System Architecture

Our chip is designed to drive a 26×26 array of PZT transducers to generate ultrasound focal spots with digitally controlled, three-dimensional beamforming, operating in a continuous wave modality. The circuit block diagram is shown in Fig. 2a and can be divided into two different sections: a central coarse phase control, implemented by a delay-locked loop (DLL), and an array of 26×26 transmit channels, which

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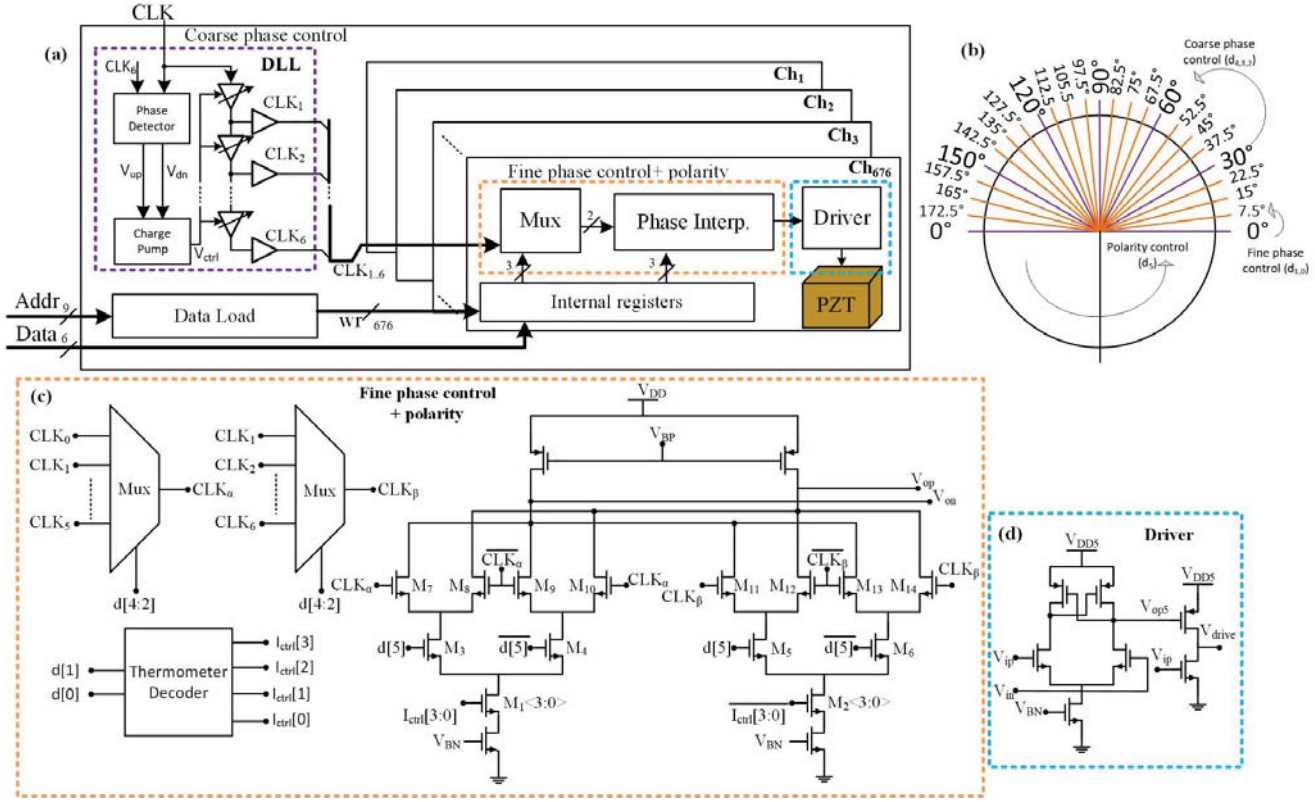


Fig. 2. Architecture of proposed chip. (a) Block diagram of proposed transmit beamformer circuit. (b) Phase map illustration employed by the beamformer. (c) Fine phase control transistor level schematic (d) driver transistor level schematic.

include fine phase and polarity control and a driver circuit to level shift the voltage to interface with the PZT transducers. The phase map implemented by the circuit is illustrated in Fig. 2b. The DLL is locked at a 180° phase shift and uses a six-element voltage controlled delay line (VCDL) to generate six output clocks with phases spanning 0° to 180° in steps of 30°. In each channel, a multiplexer selects two input clocks with consecutive phases, and a digitally controlled phase interpolator outputs a clock with a phase resolution of 7.5° while also allowing the addition of a 180° phase offset to provide complete coverage of the 360° phase map. Fig. 2c depicts the phase-fine-control transistor-level diagram. The input multiplexers select (bits d[4:2]) two consecutive clocks with a 30° phase shift, CLK_α and CLK_β, that are the inputs of the next stage phase interpolator. Control bits d[1] and d[0] are thermometer decoded into four outputs ($I_{ctrl}[3:0]$) that control two symmetrically weighted current sources that bias two different sets of differential pairs (M_7-M_{14}). The least significant bit (LSB) of the phase interpolator corresponds to a phase difference between CLK_α and CLK_β of 7.5°. Bit d[5] and its complement control the transistors M_3-M_6, which steer the biasing current between the differential pairs to control the polarity of the output clock. The output of the phase interpolator is then level shifted to a voltage supply up to 5 V and is used to drive the PZT transducers, according to the driver schematic of Fig. 2d. The biasing circuit that generates V_{BN} and V_{BP} includes a power-down mode controlled by an external enable signal, which also allows the ultrasound output to be duty-cycled.

III. System Electrical Chacterization

The proposed CMOS IC was fabricated in a 0.18 µm 1.8/5 V process and occupies an area of 4 mm × 5 mm, as seen in the photo of Fig. 3. The CMOS chip was post-processed for the integration of the 2D array of 26 × 26 PZT transducers directly on top of the chip. PZT was chosen due to its high electromechanical coupling coefficient, obviating the need for the high biasing voltages (> 10 V) characteristic of capacitive micromachined ultrasound transducers (CMUT) for the pressure levels required here. The transmiter circuits have a pitch of 135 µm, which corresponds to approximately 0.9 times the wavelegth of ultrasound waves in soft tissue at the working frequency of 10 MHz, minimizing unwanted grating lobes during beamforming. A PZT film (PZT5A, Piezo Systems) with a thickness of 267 µm was utilized for the CMOS integration according to the fabrication processes illustrated in Fig. 4 [9]. The PZT film is adhered to the chip by using an anisotropic conductive adhesive (TFA22023, H&S HighTech), making electrical connection between each transmiter pad and the PZT. The film is subsequently patterned through a mechanical dicing process to form individual transducer elements. The kerfs are filled with an epoxy (Epotek 301-2, Epoxy Technology) to provide mechanical stability to the array. Finally, a copper ground plane is DC-sputtered on the top surfaces of the PZT

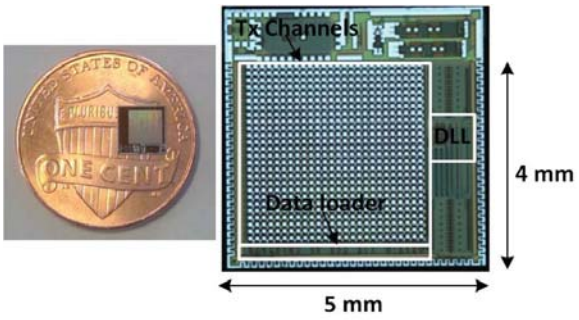
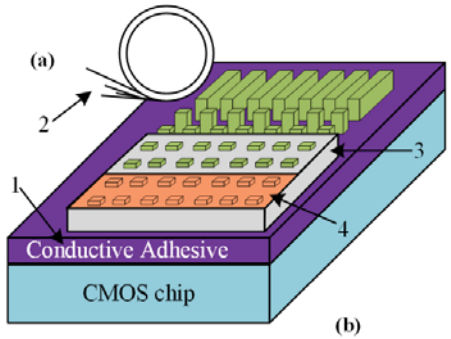


Fig. 3. Chip microphotograph.



Fabrication steps
1 – PZT die adhesion
2 – Dicing
3 – kerf filling
4 – ground plane sputtering

(b)

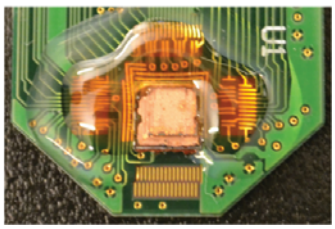


Fig. 4. (a) PZT Fabrication process illustration. (b) Photo of post-processed chip mounted on a PCB socket.

structures. The post-processed CMOS chip mounted on a PCB socket is shown in Fig. 4. The chip socket is connected to a test board, which includes a Spartan-6 FPGA (embedded in an Opal Kelly XEM6310 board), to digitally program the chip and allow communication with a computer.

Fig. 5 shows the measured output of one channel for all available phases with a drive voltage of 1.8 V. To determine the linearity of the beamforming circuits, 16 channels spreading across the 2D array (rows and columns 3, 10, 14 and 24) were measured, with the transfer function shown in Fig. 6, revealing monotonicity; the corresponding integral and differential non-linearity (INL and DNL) are also plotted in Fig. 6.

The post-processed chip was immersed in a tank filled with distilled water, which presents similar acoustic properties to soft tissue, and configured to generate different ultrasound beam patterns. A pre-calibrated (~200 nV/Pa) hydrophone (HGL-200, Onda) connected to an amplifier with 20 dB of gain (AH-2010, Onda), is mounted on a three-axis motorized stage (MTS50A-Z8, Thorlabs) to record the ultrasound beam-profiles, as shown in the photo of Fig. 7. The input clock frequency was set to 10 MHz, with a pulse duration of 1 µs, and the high voltage supply was set to 2 V. The The hydrophone location was swept over an

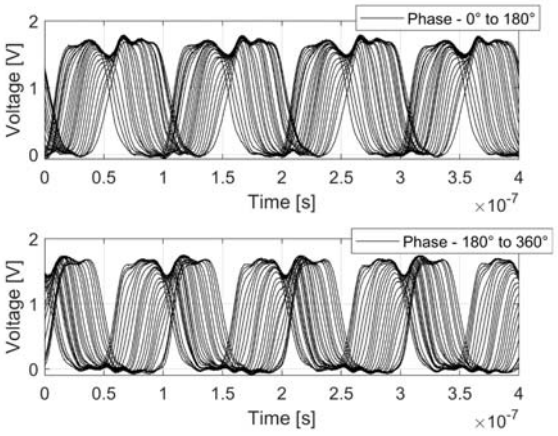


Fig.5. Measured phase vairation from 0 to 380 degrees of a single channel.

area of 5 mm × 5 mm in the XZ plane (steps of 100 µm) with the Y location at the center of the chip. The start for the Z axis was 2 mm away from the chip, to avoid data contamination due to the electromagnetic interference generated from the transmitters and picked up by the hydrophone tip. Data was obtained by measuring the peak-to-peak voltage of the hydrophone signal at each location. The chip was initially configured to output a planar wave and afterwards configured to

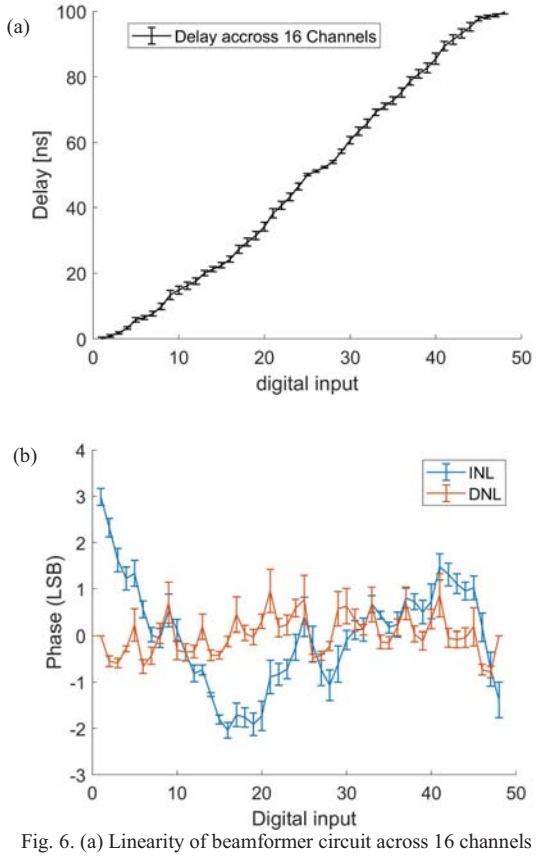


Fig. 6. (a) Linearity of beamformer circuit across 16 channels spread across the 2D array. (b) INL and DNL.

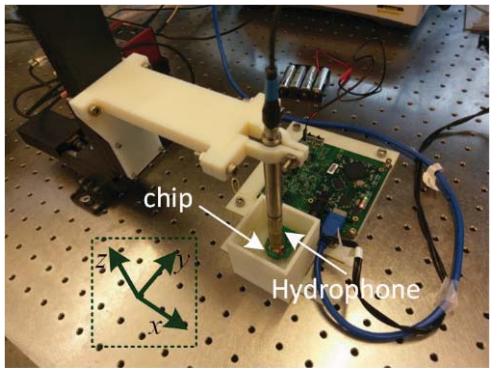


Fig. 7. Ultrasound measurement setup.

generate an ultrasound focal spot with a depth of 0.5 cm and steering angles of -15°, 0° and 15° in the XZ plane. The obtained reconstructed beam profiles are shown in Fig. 8. The results show the correct beam profile shapes and reveal a focal pressure of approximately 40 kPa, translating to a focal spot pressure of 20 kPa/V. With a maximum driver voltage of 5 V, the maximum focal pressure reaches 100 kPa without requiring any matching layer. The addition of an acoustic matching layer would significantly increase the transmission efficiency by mitigating the mismatch in acoustic impedances between PZT and soft tissue. In addition, thinning the chip down to less than 20 µm would allow for a flexible patch-like form factor for the device for non-invasive ultrasound neuromodulation.

IV. Conclusions

This work presents a CMOS 2D transmit beamformer with integrated PZT ultrasound transducers. The proposed prototype can achieve a maximum focal pressure of approximately 100 kPa with a 5 V supply at 0.5 cm depth without the incorporation of an acoustic matching layer. Comparing to other published works with similar technologies (Table 1), the achieved focal pressures are comparable, but voltage requirements are decreased by up to one order of magnitude. Focal pressure can still be boosted significantly using a matching layer. In addition, at 10 MHz, the center frequency is the highest yet achieved for such an array, leading to improved focal spot spatial resolution. This prototype shows promise as a viable tool for ultrasound-based neuromodulation in an integrated form factor.

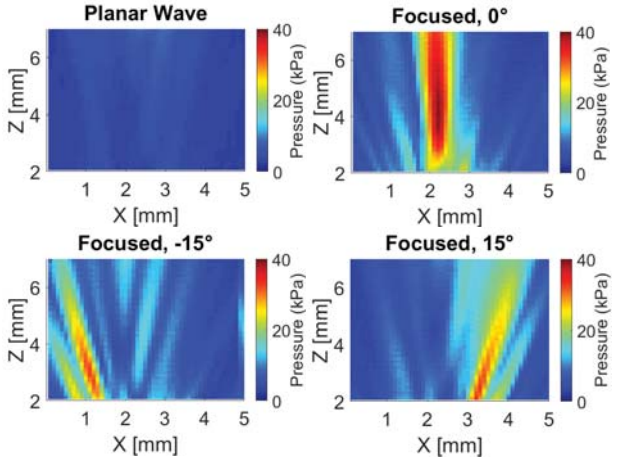


Fig. 8. Meausured ultrasound beam profiles.

TABLE I. Comparison with state-of-the-art

	[6]	[7]	[8]	**This work**
Application	Imaging	Imaging	Imaging	**Neuromod.**
Process	0.25 µm HV	0.18 µm HV	0.18 µm LV	**0.18 µm LV**
Transducer	CMUT	CMUT	PZT	**PZT**
Integration method	Multi-chip stack	Multi-chip stack	Monolithic	**Monolithic**
Acoust. Match.	No	No	Yes	**No**
Chip area	9.2 x 9.2 mm^2	6 x 5.5 mm^2	6.1 x 6.1 mm^2	**5 x 4 mm^2**
Center freq.	5 MHz	5 MHz	5 MHz	**10 MHz**
# of Tx el.	960	256	64 (off-chip circuitry)	**676**
Max. focal pressure	112 kPa (0.24 cm depth)	N/A	300 kPa (5 cm depth)	**100 kPa (0.5 cm depth)**
Required voltage	55 V	30 V	50 V	**5 V**

Acknowledgements

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