Fully Integrated CMUT-Based Forward-Looking Intracardiac Imaging for Electrophysiology

Amin Nikoozadeh[1], Ira O. Wygant[1], Der-Song Lin[1], Ömer Oralkan[1], Arif Sanli Ergun[1], Kai Thomenius[2], Aaron Dentinger[2], Douglas Wildes[2], Gina Akopyan[3], Kalyanam Shivkumar[3], Aman Mahajan[3], Douglas N. Stephens[4], David Sahn[5], and Pierre T. Khuri-Yakub[1],

1 Stanford University, **2** General Electric Corporate Research & Development, **3** University of California, Los Angeles, **4** University of California, Davis, **5** Oregon Health and Science University

Abstract — **Minimally invasive percutaneous electrophysiological mapping of the heart chambers is becoming a standard procedure to diagnose and treat cardiac arrhythmias. Due to advances in technology that enable small feature sizes and a high level of integration, non-fluoroscopic intracardiac imaging is attracting more attention to better guide electrophysiologal (EP) interventions. In this effort, we are developing a forward-looking intracardiac ultrasound imaging catheter, which is also equipped with several EP electrode sensor bands and a metal RF ablation tip enclosure.**

A 24-element fine-pitch (63 μm) 1-D array, based on capacitive micromachined ultrasonic transducer (CMUT) technology, has been fabricated for high-frame-rate imaging. Through-wafer vias are incorporated in the device to connect the signal and ground electrodes to the flip-chip bond pads on the backside of the array. The total footprint of the array measures 1.73 mm × 1.27 mm. Also a custom-designed integrated circuit (IC) has been fabricated to be closely integrated with the CMUT array for improved SNR. This IC comprises some of the important front-end electronics of an ultrasound imaging system. It measures 2 mm × 2 mm and is composed of 24 individual transmit/receive blocks. The transmit circuitry is capable of delivering 25-V unipolar pulses. The receive circuitry includes a transimpedance preamplifier followed by a line driver buffer.

A CMUT array was flip-chip bonded directly on to the IC for initial testing. All of the 24 elements of the array and the IC are functional. Array uniformity was tested by measuring the resonant frequency in air. A standard deviation of 0.37 percent was measured around the mean value of 17.9 MHz. The same array operates at 9.2 MHz in immersion with a 104 percent fractional bandwidth. Imaging performance of the described front-end was tested on a commercial phantom and also in ex-vivo environment on an isolated perfused rabbit heart (Langendorff).

The final goal is to integrate the CMUT array and the front-end electronics at the tip of a 10-F catheter. A flexible printed circuit board (PCB) has been designed and the first sub-assembly is ready for cable attachment and final catheter integration.

Keywords – forward looking; electrophysiology; intracardiac echo; guidance; ultrasound; capacitive micromachined ultrasonic transducers; CMUT

Work supported by NIH grant: NHLBI 5-R01-HL67647, USA

e-mail contact: aminn@stanford.edu

I. INTRODUCTION

The use of intracardiac echo (ICE) imaging catheters for the guidance of EP interventions to diagnose cardiac arrhythmias is attracting increasing attention. It offers improved visualization in real time and enhanced procedural guidance when compared to fluoroscopy alone [1]. The patient population for atrial fibrillation, the most common type of cardiac arrhythmia, is over 2.2 million in the United States alone with over 60,000 new cases each year [2].

The most common method of catheter positioning during therapeutical EP interventions today is fluoroscopy. The radiation exposure involved with fluoroscopy is undesirable and hazardous for both the patient and the practitioner. With long EP mapping studies and the subsequent ablation procedure, methods to reduce the fluoroscopy time are needed. EP ablation procedures can easily take as long as 3 hours [3, 4] and an average exposure time of 30 minutes or longer could easily be expected [5, 6].

In this effort we are concentrating on developing a multifunctional intracardiac forward-imaging catheter that is EP capable and is equipped with RF ablation capability.

II. ULTRASOUND-GUIDED ELECTROPHYSIOLOGY

The "MicroLinear" (ML) catheter is an intracardiac echo (ICE) imaging catheter designed specifically for use in EP therapy guidance. This is an EP capable catheter with a 1-D CMUT array mounted at the tip for high-definition, high-frame-rate, forward-looking imaging. The final catheter incorporates several EP sensor bands near the distal tip and a metal ablation tip enclosure that allows simultaneous RF ablation and imaging. The early assembly, testing and imaging results of the CMUT ML array are the primary subjects of this paper.

III. METHODS

A. CMUT Array Design

The fabricated CMUT ML array dimensions are summarized in Table I.

TABLE I. CMUT ML ARRAY DESIGN PARAMETERS

Array Design Parameters	
Number of elements	24
Center frequency (MHz)	10
Array pitch (µm)	63
Element length in elevation (mm)	1.2
Cell Design Parameters	
Membrane thickness (µm)	0.50
Gap height (µm)	0.10
Insulator layer thickness (µm)	0.18

The ML array is a fine-pitch 24-element 1-D transducer array. It was fabricated using the standard poly-silicon sacrificial layer process with through-wafer interconnects [7]. Through-wafer vias connect the individual signal electrodes and the common reference electrode to the backside flip-chip bond pads. These pads facilitate the tight integration with supporting electronic circuits through flip-chip bonding. The transducer membrane is made of 0.5-µm thick silicon nitride. A number of different configurations with different membrane sizes and shapes (circular, rectangular) were fabricated on a single wafer to allow for extensive testing. Fig. 1(a) shows an optical picture of an array with rectangular membranes with 16-µm width. Fig. 1(b) shows two SEM pictures of a device with circular membranes. These devices were fabricated on a 400-µm thick silicon wafer and were singulated using deep reactive ion etching process. The total footprint of the array measures 1.73 mm × 1.27 mm.

B. Custom-Designed Front-End Electronics

Each element of the CMUT ML array has a capacitance of approximately 2 pF, which is quite small compared to the cable capacitance of about 300 pF in a typical catheter assembly. This means that direct connection of the transducer elements to the imaging system will result in severe signal degradation and in turn poor image quality. Therefore we designed an application-specific IC comprising the important parts of the front-end electronics of an imaging system to be integrated closely with the CMUT array at the tip of a catheter. The custom-designed IC has 24 identical blocks each containing a 25-V unipolar pulser, a TX/RX switch, a transimpedance amplifier and a buffer. A singulated IC die measures 2 mm × 2 mm in size. The inside pads match the pads on the backside of the CMUT ML array for direct flip-chip bonding (Fig. 2).

C. Test and Imaging Methods

For initial testing a CMUT ML array was flip-chip bonded to the custom-designed front-end electronics. Then the IC pads were wire-bonded to a pin grid array (PGA) ceramic package. Bond wires were then buried under UV-cured epoxy for protection and easier handling. After testing the array in air, the device was coated with a 150-µm layer of Polydimethylsiloxane (PDMS) for under-water tests and also for imaging in contact with tissue.

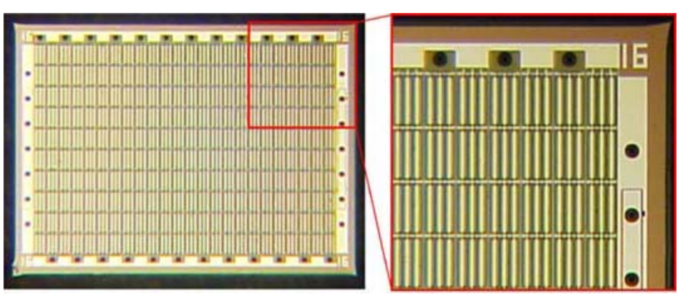
A PC-based data acquisition system was used to characterize the performance of the device (i.e. resonant frequency, pulse echo frequency response etc.). A commercial portable imaging system (Vivid-i, General Electric Co.) was used for imaging experiments. A special interface box (I-BOX) was designed to allow the imaging system to communicate with the front-end electronics. Since the custom-designed IC has built-in pulsers, the output pulses from the imaging system were converted to trigger pulses in the I-BOX.

A prototype CMUT ML array in the PGA package was used to image point and contrast resolution test phantoms. To further demonstrate the imaging capability of the device the same assembly was used in contact with an ex-vivo rabbit heart Langendorff model, aiming to image the left atrial appendage.

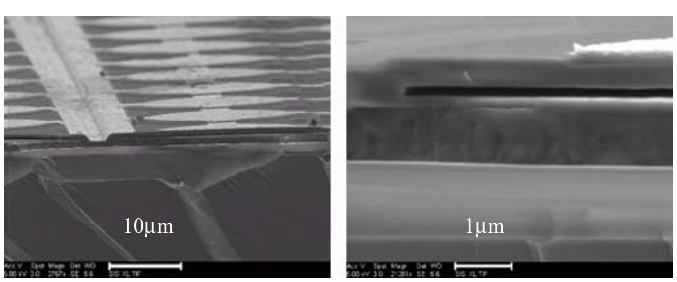
IV. RESULTS

A. Benchtop Testing of the ML Array

We first measured the resonant frequency of each element of the ML array in air before coating the array with PDMS. The results showed a mean center frequency of 17.9 MHz with a standard deviation of only 0.37%, when the array was biased at negative 30 V. This is not only an indication of good array

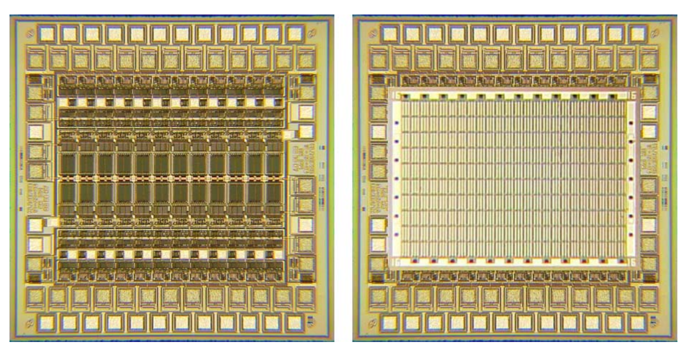


(a)



(b)

Figure 1. (a) Optical picture of a CMUT ML array composed of rectangular membranes, (b) SEM pictures of a CMUT ML array with circular membranes.



(a) (b)

Figure 2. (a) Photograph of the custom-designed front-end electronics, (b) Photograph of a CMUT ML array flip-chip bonded to the front-end electronics.

performance, but also a verification of the full functionality of the IC and 100% yield of flip-chip bonding. We also measured the acoustic pressure as a function of DC bias using a calibrated hydrophone (Model HNV-0400, Onda Corporation, Sunnyvale, CA). The unipolar pulse amplitude supplied by the custom-designed IC was 24 V. Pressure was measured at about 5-mm distance from the array immersed in water and then corrected for attenuation and diffraction losses. Fig. 3 shows the pressure at the face of the ML array for various DC bias voltages.

Pulse-echo performance of the ML array in water was also tested with negative 30-V DC bias. Fig. 4 shows the echo signal from a total plane reflector (water-air interface) and the associated frequency spectrum. The array has a center frequency of 9.2 MHz with a fractional bandwidth of 104%. The unwanted second echo after the main one is due to the PDMS passivation layer. As mentioned earlier, we coated the ML array with this layer to be able to perform under-water experiments. By slightly modifying the material properties of this layer, a better acoustical match can be achieved to eliminate artifacts in the pulse-echo response and as a result in final images.

B. Imaging Tests

The initial array prototype assembled in a PGA package was tested using commercial phantoms and also with ex-vivo tissue. Fig. 5 shows an image of a standard point resolution test phantom (Model RMI 404GS LE gray scale phantom, Gammex, Inc., Middleton, WI) [Fig. 5(a)], an image of a contrast resolution test phantom (Rubber-based soft tissue mimicking material, ATS Laboratories, Inc., Bridgeport, CT) [Fig. 5(b)], and an image of the left atrial appendage in a Langendorff isolated perfused rabbit heart model [Fig. 5(c)]. For these experiments the device was biased at negative 60 V and the pulse amplitude was 24 V. The addition of an elevational lens is expected to improve the image quality even further.

V. Ongoing Work

Our preliminary experiments with the CMUT ML array in the PGA package demonstrated the functionality and the

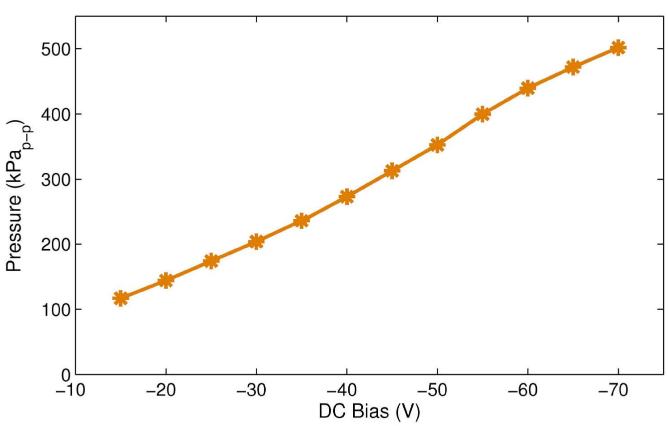


Figure 3. Acoustic peak-to-peak pressure at the face of the ML array as a function of DC bias. The AC pulse amplitude was 24 V.

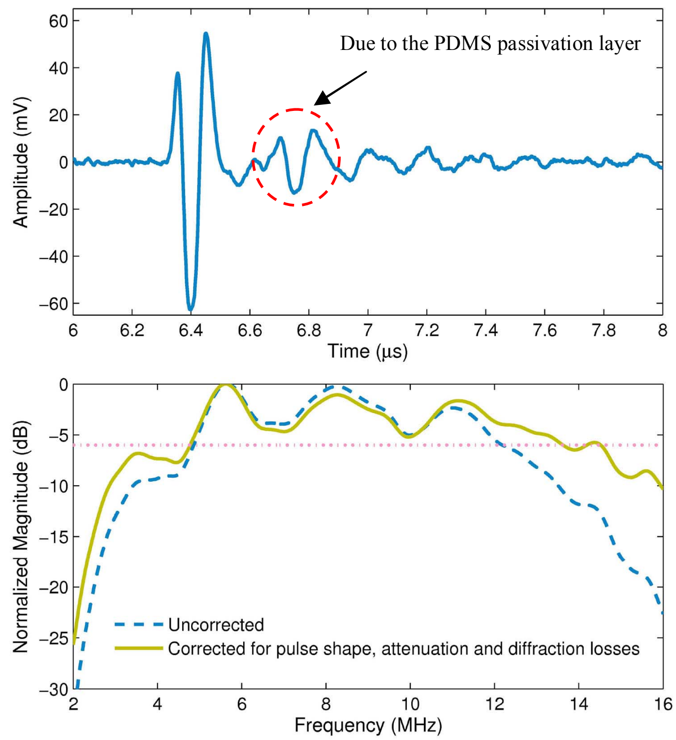
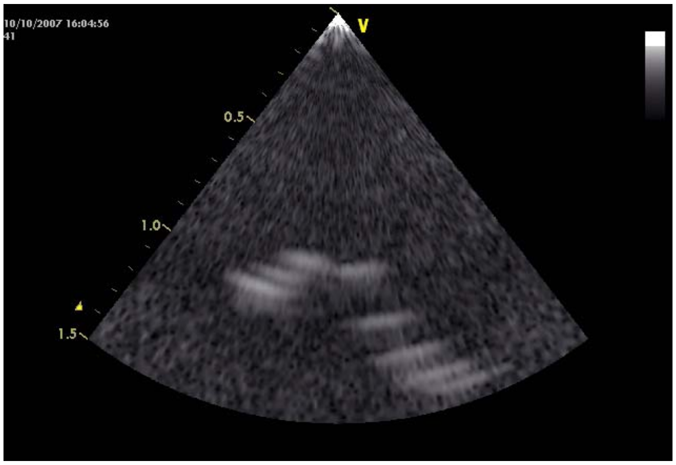


Figure 4. The pulse echo and its frequency spectrum from an element of the CMUT ML array immersed in water at a 4.7 mm depth. The device was biased at negative 30 V. The second echo in the top figure is because of the rather large impedance mismatch between the PDMS layer and water.

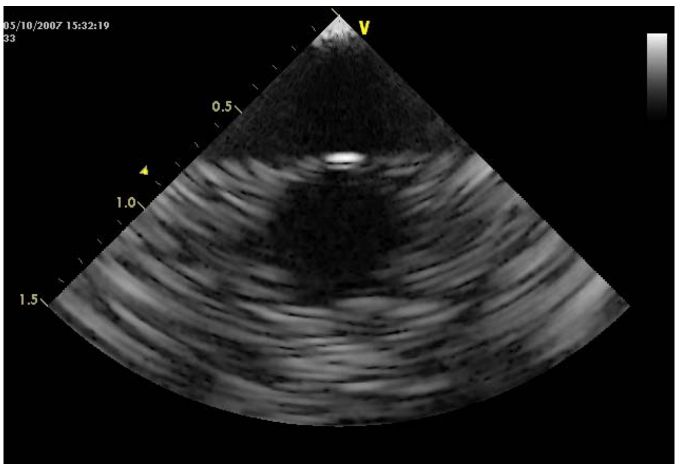
performance of all the system components integrated together. Currently we are working on the final catheter assembly. To accomplish this, we have designed a flexible PCB that measures 2 mm in width, the same as the front-end electronics die. It is long (110 mm) in the other direction for easier catheter steerability. The center portion of the flexible PCB, on one side has pads that match to the IC, and on the other side, to the backside of the CMUT ML array. This configuration makes it possible to flip-chip bond the IC to one side of the flexible PCB and the CMUT array to the other side. Fig. 6 shows the first successful integration of these three components. The flexible PCB will then be folded around the IC to fit within the distal tip housing of the catheter. After this step the flex will be ready for coaxial cable attachment. The catheter cable pads are located at the two ends of the flex. Once the coaxial cables are attached the sub-assembly will be ready for the final catheter integration.

VI. Conclusions

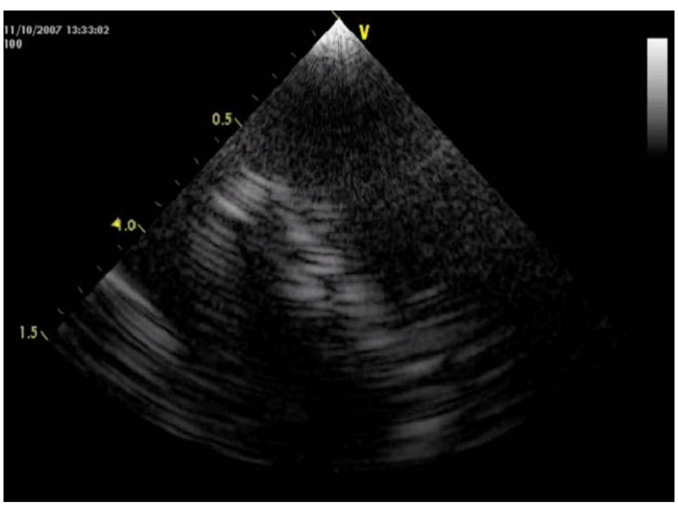
We successfully tested a CMUT ML array assembly fully integrated with custom-designed front-end electronics. We demonstrated ex-vivo images of a beating rabbit heart (Langendorff model). These results show that the CMUT technology is ideal for implementing miniature transducer arrays with integrated electronics for catheter-based intracardiac and intravascular applications.



(a)



(b)



(c)

Figure 5. Preliminary images with the early CMUT ML array assembly in the PGA package using a portable commercial ultrasound system (Vivid-i , GE). (a) An image of a point resolution test phantom (Model RMI 404GS LE gray scale phantom, Gammex, Inc., Middleton, WI). (b) An image of a contrast phantom (Rubber-based soft tissue mimicking material, ATS Laboratories, Inc., Bridgeport, CT). (c) An ex-vivo image of the left atrial appendage in a Langendorff isolated perfused rabbit heart model.

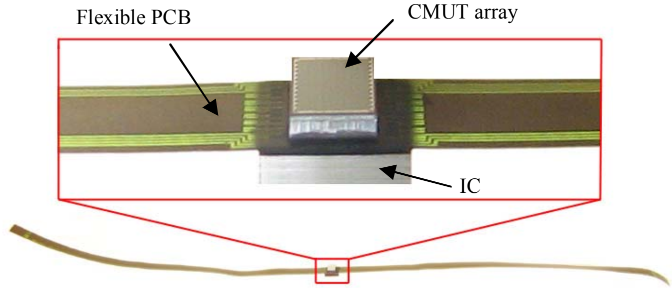


Figure 6. A photograph of the front-end IC and the CMUT ML array flip-chip bonded to two sides of the flexible PCB for final catheter assembly.

ACKNOWLEDGMENT

This work was funded by the National Institutes of Health under grant HL67647. We would like to thank National Semiconductor (Santa Clara, CA) for their valuable support in the design and fabrication of the IC. CMUT fabrication was done at the Stanford Nanofabrication Facility (Stanford, CA), which is a member of National Nanotechnology Infrastructure Network. Solder bumping was provided by Pac Tech USA Inc. (Santa Clara, CA). The flexible PCB was manufactured by MicroConnex (Snoqualmie, WA).

REFERENCES

[1] M. Jongbloed, M. Schalij, K. Zeppenfeld, P. Oemrawsingh, E. van der Wall, J. Bax, "Clinical applications of intracardiac echocardiography in interventional procedures," *Heart*, vol. 91, pp. 981-990, 2005.

[2] Go AS, Hylek EM, Phillips KA, et al., "Prevalence of diagnosed atrial fibrillation in adults: national implications for rhythm management andstroke prevention: the anticoagulation and risk factors in AF (ATRIA) study," *JAMA*., vol. 285, pp. 2370–5, 2001.

[3] M. Earley, et al., "Radiofrequency ablation of arrhythmias guided by non-fluoroscopic cathter location: a prospective randomized trial," *European Heart Journal*, vol. 27, pp. 1223 – 1229, 2006.

[4] T. Hung-Fat, "Nonfluoroscopic magnetic electroanatomic mapping to facilitate local pulmonary veins ablation for paroxysmal Atrial Fibrillation," *Journal of Pacing and Clinical Electrophysiology*, vol. 25, no. 1, Jan. 2002.

[5] H. Kottkamp, et al., "Electromagnetic versus fluoroscopic mapping of the inferior isthmus for ablation of typical atrial flutter, a prospective randomized study," *Circulation*, vol. 102, pp. 2082-2086, 2000.

[6] Perisinakis, et al., "Fluoroscopically Guided Implantation of Modern Cardiac Resynchronization Devices," *Journal of the American College of Cardiology*, vol. 46, no. 12, pp. 2235 – 2239, 2005.

[7] A. S. Ergun, Y. Huang, X. Zhuang, O. Oralkan, G. G. Yarahoglu, B. T. Khuri-Yakub, "Capacitive micromachined ultrasonic transducers: fabrication technology," *IEEE Trans. Ultrason., Ferroelect., Freq. Contr.*, vol, 52, no. 12, pp. 2242 - 2258, Dec. 2005.