Micromachined transducers enable real-time three-dimensional imaging

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Large, high-performance transducer arrays for medical diagnostics and therapeutic applications can now be realized using capacitive micromachined ultrasonic transducers, built using well-established microfabrication techniques.

For many medical applications, real-time 3D images would allow a complete view of internal structures and an accurate assessment of volumes. In 3D sonography, for example, diagnostic interpretations would become less operator dependent because arbitrary cross-sections could be extracted from volume images. The determination of disease mechanisms would also be eased: real-time 3D images could provide simultaneous structural and functional information, such as the structure of blood vessels and their oxygenation levels. A suitable imaging system in this case might combine ultrasonic reflection-mode imaging (which maps mechanical properties using sound-wave reflections) with photoacoustic imaging (which infers oxygenation levels using the optical absorption image made from sound waves generated by the absorption of laser pulses).

However, several obstacles stand in the way of developing a combined ultrasonic-photoacoustic imaging system. First, real-time 3D ultrasonic imaging is still under development and has not yet been widely accepted in the clinic, in part due to the difficulty of manufacturing large-area 2D transducer arrays using piezoelectric materials. Also, the effective implementation of photoacoustic imaging, which is a very promising tool for the early detection of cancer and for understanding other disease mechanisms, is impeded by the lack of high-sensitivity transducer arrays in different shapes and sizes. Many photoacoustic imaging systems today rely on mechanically-scanned single-focus transducers or, at best, a 1D array of transducers: neither alternative can provide 3D images in real time.

Capacitive micromachined ultrasonic transducers (CMUTs) can provide solutions to these problems. The recent advent of the CMUT stems from a fundamental change in how ultrasonic transducers are fabricated. The meticulous and labor-intensive steps used to manufacture piezoelectric transducer arrays have been replaced by standard microfabrication techniques (such as photolithography and thin film deposition) in the CMUT fabrication process. The device’s basic building block is a thin membrane suspended over a sub-micron cavity (see Figure 1). Electrostatic actuation of this structure generates ultrasound, and capacitive sensing allows ultrasound detection. Transducer arrays in a variety of shapes and sizes can be realized by interconnecting many of these tiny capacitors (see Figure 2). We have demonstrated 2D arrays with as many as 16,000 elements in a $5 \times 5$ cm area. We have also designed and fabricated a 64-element annular ring array to fit at the tip of a 2mm catheter for intracardiac imaging.

In addition to advantages in manufacturing, CMUTs also provide wide bandwidth, which translates into improved resolution in ultrasound images. Over the last decade, we have demon-
Figure 2. Shown are three CMUT arrays. (a) A 64-element 30MHz 1D array with 36µm element pitch. (b) A 128×128-element 3MHz 2D array with 420µm element pitch. (c) A 64-element 10MHz annular ring array with 100µm element size.

Figure 3. This 16×16 2D CMUT array has through-wafer interconnects and flip-chip bonding to a custom front-end integrated circuit.

strated many different CMUT arrays operating in a frequency range from 10kHz to 70MHz, with fractional bandwidths in excess of 100%. We have also designed and implemented several generations of front-end integrated circuits for CMUTs used to build compact probes with a minimum number of cables connecting them to the imaging console (see Figure 3).

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We recently demonstrated simultaneous reflection-mode and photoacoustic imaging using CMUTs.\(^4\) We used a 1064nm Q-switched Nd:YAG laser to apply a 2.3mJ, 12ns full-width-at-half-maximum pulse to a 2×2×3cm block of tissue-mimicking material (ATS Laboratories, Bridgeport, CT). A 16×16 2D CMUT array with integrated front-end electronics was used to receive the light-induced ultrasound signals. The phantom consisted of three 0.86mm inner diameter, 1.27mm outer diameter, polyethylene tubes embedded in the tissue-mimicking material. The center tube was filled with India ink to provide optical contrast (see Figure 4). To simulate the effect of a larger aperture, we also stepped the 16×16 array mechanically in the x and y directions to effectively obtain an array of 48×48 elements. A 3D image obtained by photoacoustic techniques overlaid on the conventional reflection-mode image is shown in Figure 5. The photoacoustic image is shown in red, while the pulse-echo image is rendered in grayscale. The images from the 48×48 array shown in Figure 5(b) clearly illustrate the advantages of larger arrays, which include improved lateral resolution, larger field of view, and better signal-to-noise ratio.

These preliminary results show that CMUT technology is capable of providing transducer arrays for real-time structural and functional ultrasound imaging that can potentially revolutionize many medical diagnostic applications. The kind of probes described here can also be used to provide therapeutic modalities such high-intensity focused-ultrasound ablation and the ability to monitor lesion formation during the ablation procedure. An additional advantage is the potential to make flexible arrays by using fabrication techniques such as wafer thinning and deep trench formation. A flexible array could be used to implement a true 3D tomographic imaging system, as illustrated in Figure 6, for imaging the breast. Our current research is focused on implementing larger transducer arrays with integrated electronics, and on improved image-reconstruction techniques that translate the results of our phantom studies to clinical use.

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References


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