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FABRICATION, PACKAGING, AND CATHETER ASSEMBLY OF 2D CMUT ARRAYS FOR ENDOSCOPIC ULTRASOUND AND CARDIAC IMAGING

Azadeh Moini, Amin Nikoozadeh, Jung Woo Choe, and Butrus T. Khuri-Yakub
Department of Electrical Engineering
Stanford University
Stanford, CA 94305

Chienliu Chang
Canon, Inc.
Tokyo, Japan

Doug Stephens
Department of Biomedical Engineering
University of California, Davis
Davis, CA 95616

L. Scott Smith
GE Global Research
Niskayuna, NY 12309

David Sahn
Department of Medicine
Oregon Health and Science University
Portland, Oregon 97239

ABSTRACT

Ultrasound is increasingly in demand as a medical imaging tool and can be particularly beneficial in the field of intracardiac echocardiography (ICE). However, many challenges remain in the development of a 3D ultrasound imaging system.

We have designed and fabricated a quad-ring capacitive micromachined ultrasound transducer (CMUT) for real-time, volumetric medical imaging. Each CMUT array is composed of four concentric, independent ring arrays, each operating at a different frequency, with 128 elements per ring. In this project, one ring will be used for imaging. A large (5mm diameter) lumen is available for delivering other devices, including high intensity focused ultrasound transducers for therapeutic applications or optical fibers for photoacoustic imaging.

We address several challenges in developing a 3D imaging system. Through wafer vias are incorporated in the fabrication process for producing 2D CMUT arrays. Device integration with electronics is achieved through solder bumping the arrays, designing a flexible PCB, and flip chip bonding CMUT and ASICs to the flexible substrate. Finally, we describe a method for integrating the flex assembly into a catheter shaft. The package, once assembled, will be used for in-vivo open chest experiments.

INTRODUCTION

In medical applications, 3D, real-time, high-resolution images are increasingly necessary for disease diagnosis and treatment. Catheter-based electrophysiology and endoscopic ultrasound imaging can provide high resolution, real-time images without the ionizing radiation produced by fluoroscopy, which is common in a clinical setting. Therefore, a need for advanced, non-ionizing 3D imaging systems is growing.

Capacitive micromachined ultrasonic transducers (CMUTs) are a MEMS-based technology that offers several unique advantages in this regard. Their batch processing of arbitrary array geometries, low mechanical impedance, and broadband frequency response lend CMUT technology to use in medical imaging systems. Due to the fabrication flexibility of CMUT technology, a ring array with center lumen is easily realizable, allowing the use of other therapeutic or diagnostic devices, such as optical fibers for photoacoustic imaging or high intensity focused ultrasound arrays for tissue ablation, in addition to ultrasound imaging.

Previously, we have reported *in vivo* volumetric imaging of forward-looking CMUT catheters [1,2]. These promising results have led to the design and fabrication of other arrays. To this end, this paper aims to describe the critical fabrication, packaging, and assembly steps required for building CMUT-based forward-looking catheters for use in 3D medical imaging systems.

CMUT FABRICATION

Array specifications

Ring-shaped CMUT arrays with an outer diameter of 10.1 mm and an inner diameter of 5 mm were fabricated. To achieve volumetric images, a 2D imaging array is necessary. The sparse 2D ring array structure reduces the complexity of the array while still providing high quality 3D images. Additionally, the ring array structure allows for a center lumen in the CMUT array. For ultrasound catheters and other imaging devices, the inclusion of this center lumen allows for the introduction of other diagnostic or therapeutic devices, thereby allowing simultaneous imaging and therapy in a clinical session.

A large CMUT array has several advantages in clinical use. One is that a greater number of elements, 128 rather than 64 [1,2], can be used, resulting in higher output pressures. Additionally, the diameter of these arrays was designed such that there would be four separate, concentric rings per device [3], hence a quad-ring structure. Each ring, #1-4, has a different radius (6, 7.2, 8.5, and 9.7 mm) and center frequency (16, 12, 8, and 6.5 MHz respectively) such that their beam profiles and natural foci align.

The arrays were fabricated using a sacrificial release process [4-7]. All elements share a top electrode through which the CMUT DC bias is applied. Due to the circular shape of the arrays, singulation was achieved through DRIE etching [8].

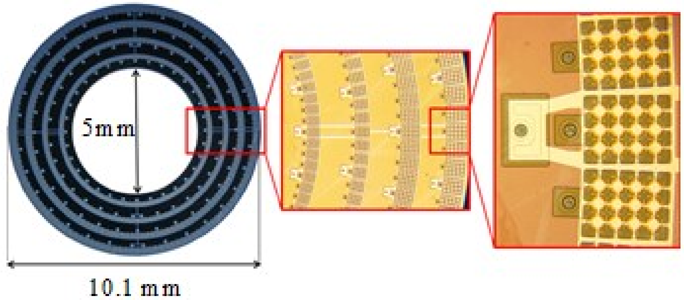

Figure 1. Optical images of the QuadRing CMUT array. The image on the left shows the entire QuadRing array. The element structure of the four concentric rings and individual elements are shown in the right two images.

Through wafer vias

One defining feature of these arrays is their backside electrical contacts. By making use of back-side contacts to provide the CMUT DC bias, transmit pulse, and receive signal, the need for front-side contacts is eliminated, thereby saving real estate that would otherwise be dedicated to wire-bonding pads and reducing the overall size of the array. Furthermore, back-side contacts allow for the use of solder bumps to connect the array to the substrate rather than using wire-bonds.

In order to make an electrical connection between the front and back sides of the wafer, through wafer vias are incorporated in the process [4,5]. The transmit pulse, and similarly received signal, for each CMUT element is brought in through the via, so one via is needed for each element on the array. Additionally, the CMUT bias will be applied through the vias, so each array will have 32 vias dedicated to the DC bias voltage. In total, each array will have 160 vias per ring and 640 vias across the 4 rings of the device.

Several factors are taken into consideration when determining via diameter. The vias, once etched, will be electrically isolated from the wafer substrate with thermal oxide. To provide electrical contact from the back-side pad to the front-side element, the via is subsequently filled with doped polysilicon. Therefore, the via diameter should be large enough to accommodate both an insulating oxide as well as enough doped polysilicon to ensure a low resistivity contact. Ultimately, however, the lower bound of the via diameter is limited by DRIE aspect ratio. Given a 10:1 etching aspect ratio and a wafer thickness of 400μm, a 20μm via diameter was achieved by etching through 200μm of the substrate on each side using a Bosch process (Fig. 2). The via diameter is upper-bounded by the necessity to fill the via enough to allow unflawed resist coverage in future lithography steps. As the polysilicon depositions add stress to the wafer, it is necessary to minimize this step. Further, as the via-filling process results in a non-flat surface, via locations are chosen to be outside the element space. Therefore, to minimize the array space the vias will consume, it is desirable to minimize their diameter. It is important to note that some via expansion is seen during the DRIE etching process. With that in consideration, masks are designed with a sub-20μm diameter to achieve the final desired via diameter.

The vias are filled with highly doped polysilicon to provide electrical contacts to the front-side of the array. A final via resistance of 70 ohms is achieved; a high dopant concentration in the via is desirable to minimize the parasitic series resistance each via adds to the CMUT circuit. It can then be seen that the substrate, thermal oxide insulator, and highly doped polysilicon in the via form a metal-insulator-silicon (MIS) junction. In addition to providing an electrically isolated, low resistance path to the CMUT, when biased properly, the MIS junction is reverse biased, generating a large depletion region around the via, thereby reducing the parasitic parallel capacitance [9].

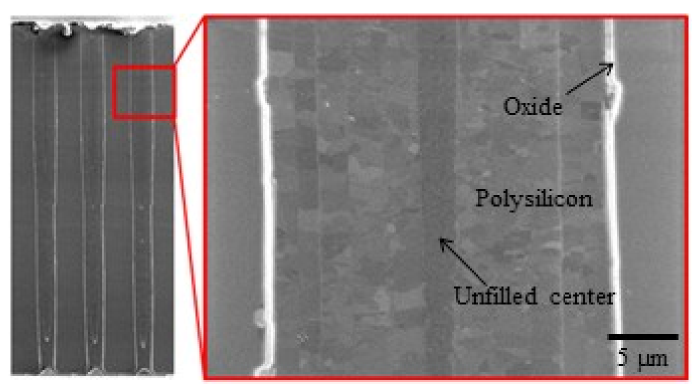

Figure 2. SEM of the cross-section of a through wafer via.

INTEGRATION WITH ELECTRONICS

Flexible PCB design

For *in vivo* testing, the QuadRing arrays will be integrated into a catheter assembly. In the first version of this catheter, only one of the four rings will be in use. This will both simplify the assembly process and reduce the necessary cabling, increasing catheter flexibility.

The arrays are integrated into a catheter assembly by use of a custom-designed flexible PCB (flex). The PCB substrate is flexible 25µm polyimide film which can easily be re-shaped. Two metal layers provide electrical contacts along the length of the flex. Trace width and spacing is limited to 30µm but is maximized where possible. The flex has a center ring, on which the array will be situated, and eight legs that stem radially from the center portion, which are folded in the final assembly process around a catheter shaft, as described later. The flex layout is shown in Figure 3.

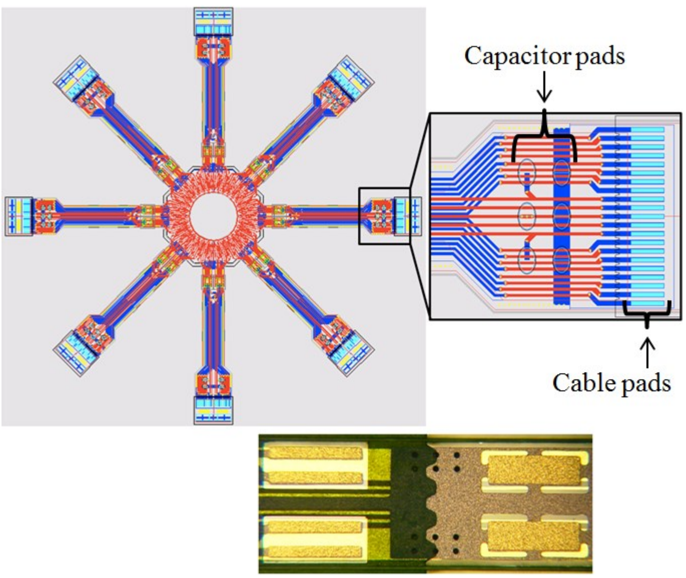

Figure 3. QuadRing flex design is shown on the top left. The top right image shows the distal end of the flex. The bottom image shows the ridged soldermask edge from a fabricated flex of a previous build.

In order to drive the 2m-long cables of the catheter, custom-designed ASICs [1,2] are integrated on the flex in close proximity to the QuadRing CMUT array. Each leg is designed for integration with two 8-channel custom ASICs to serve the 16 channels on that leg. Additionally, the flex is designed for placement of three bypass capacitors on each leg for the ASICs' DC power supplies to reduce noise in the catheter system. Each flex leg carries traces to 16 elements in addition to providing the supplies for the ASICs for a total of 21 traces per leg.

A thin soldermask film covers both sides of the flex and serves to protect the traces during the manual assembly process. The soldermask is removed over certain parts of the flex to allow soldering of devices – namely the QuadRing array, ASICs, bypass capacitors, and cables. Due to the fragility of the flex, the soldermask patterning mask is designed with curved edges rather than sharp 90° edges (Fig. 3). With a sharp soldermask definition, a high-stress area is created on the flex which tends to cause breakage, particularly when the soldermask removal is only on one side of the flex. Previous builds have seen a reduction in mechanical stress by using a non-abrupt edge to distinguish the soldermask areas and therefore an increase in catheter assembly yield.

Solder bumping

Electrical contacts between the CMUT array and ASICs to the flex are to be provided through solder bumps through a flip chip bonding process. Flip chip bonding provides several advantages over wire-bonding, including tighter integration between with the device and substrate, as well as a reduction in manual assembly, as flip chip bonding can be used to provide electrical connections.

In the final steps of the fabrication process, an under-bump metallization stack (UBM) is deposited onto the 70µm-diameter back-side pads using evaporation. The UBM stack, composed of 10nm Ti, 250nm Ni, and 150nm Au, is designed to provide proper wetting to the solder bumps [6]. The 100µm eutectic Sn/Pb solder balls are jetted onto the arrays (Fig. 4) and their mechanical strength quantified through shear tests. In previous builds, shear strength of 40MPa has been achieved using this technique.

Flip chip bonding

Solder bumped CMUT arrays are integrated with electronics through the flex using the flip chip bonding method. During flip chip bonding, the solder bumps on the devices are visually aligned to the flex pads. Once alignment is achieved, the devices are brought into contact with the flex and are temporarily held in place with tacky flux.

Once all devices have been aligned and placed on the flex, the assembly is placed in an oven where the solder balls reflow and adhere to both substrates. Nitrogen flow in the oven prevents oxidation of the solder during reflow. Once this is accomplished, underfill epoxy is applied to the cavity between the two substrates generated through flip chip bonding. This single-component, non-conducting epoxy acts as mechanical reinforcement to the assembly. Additionally, it has a very low coefficient of thermal expansion to minimize additional stress on the substrates. The chips are supported by the epoxy on the substrate; therefore, the solder balls only need to provide electrical connections rather than both electrical and mechanical support.

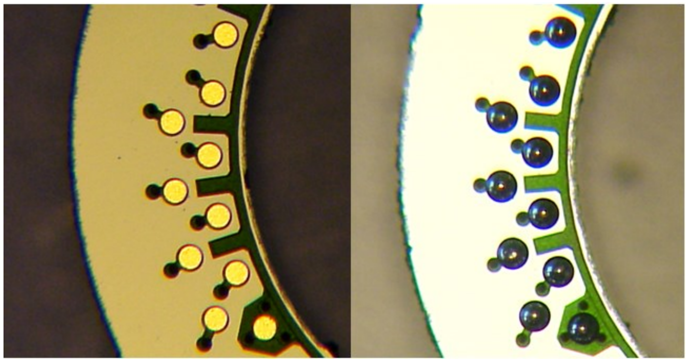

Figure 4. The figure on the left shows the back-side of a 64-element Ring array. The pads and UBM are shown. The figure on the right shows the same type of array, this time with solder bumps on the back-side pads.

CATHETER ASSEMBLY

Probe integration

Once the ASICs and CMUT have been bonded to the flex substrate, integration of the flex assembly with the catheter shaft can begin. For integration with the catheter shaft, the following steps are necessary: 1) the flex leg must be re-shaped and adhered to an underlying substrate; 2) the outer face of the flex assembly should be protected, especially around the ASICs; 3) a reference surface for the passivation layer should be established and the passivation layer deposited onto the CMUT face. In previous builds, two separate, custom-designed parts were used for these steps: one to serve as the underlying substrate support for the flex legs, and one catheter tip that would both protect the outer portion of the assembly and provide a reference for the passivation layer. Figure 5 shows a previous catheter build in two stages. Due to the detailed, manual nature of this assembly process, it is preferable to minimize the number of components and incorporate self-aligning parts where possible. Therefore, in the current version, a single piece has been designed to replace the described parts.

Rather than separating the flex leg support and outer shell into two separate pieces, the larger size of the QuadRing arrays allows for a new tip design that incorporates all these needs in one custom-designed piece (Fig. 6). The QuadRing array is introduced to the tip through eight slits around the tip's circumference. Once the flex legs are passed through these slits, the array sits on the octagonal tip base. One flex leg is aligned to each face of the octagonal base and is held in place with epoxy to ensure the flex legs will not overlap and damage one another during catheter use. This base has a circular opening through the center which aligns with the center lumen of the QuadRing array to allow full use of the device.

The outer portion of the tip extends just below the ASIC region on the flex; it is long enough to encase the critical assembly components but not so long as to impede the device flexibility. On the proximal end of the catheter, the tip extends 100µm beyond the CMUT face to serve as a reference for the passivation layer.

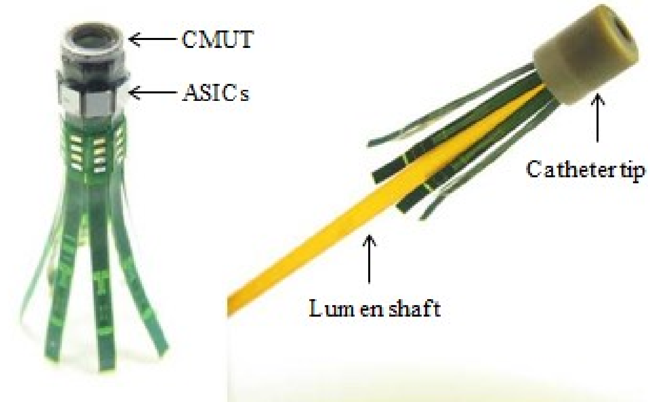

Figure 5. Left: A 64-element Ring array assembly with flex legs folded and adhered with epoxy to the underlying support substrate. Right: Assembly with tip integrated to the flex. The next phase of catheter integration, device encapsulation, has also been completed. Also shown in yellow is the inner lumen shaft.

The tip material is another point of consideration. It should be easily machinable to the required precise specifications. Additionally, it is preferable to use an insulating material to prevent electrically shorting traces or otherwise. If an insulating material is not used, an electrically insulating layer should encapsulate the tip prior to integration with the flex assembly. These parts were 3D printed with 16um accuracy. For clinical use, a biocompatible material should be used.

In order to maintain the inner lumen clear of cables throughout the length of the catheter, a separate hollow shaft is inserted through the assembly and secured to the back-side of the array.

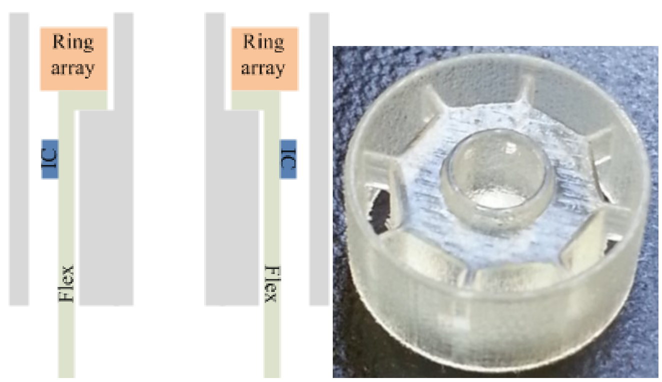

Figure 6. Cross-sectional design (left) and 3D printed tip (right) for the new QuadRing array assemblies. The cross-sectional view shows the flex assembly supported by the inner and outer parts of the structure, as well as the 100µm extension of the structure beyond the CMUT surface. The top-down view shows the octagonal structure of the inner support structure. Each flex leg will be folded through a slit and adhered to one side of the octagon.

Device encapsulation

Device encapsulation is critical for a transducer or any medical device that is to be used in a clinical setting. The passivation layer serves not only to protect the face of the transducer against damage upon interaction with the system but also to provide a biocompatible interface to the test subject. Therefore, for this application, the encapsulation material must be biocompatible, minimally attenuating, and easily deposited while providing an impedance match to water. Additionally, the coating must be conformal to avoid air bubbles and any resulting undesired reflections at the air-coating interface. Based on these requirements, Sylgard 160 is a suitable encapsulation material. Sylgard 160 is a two part PDMS which has an acoustic impedance very similar to tissue (1.5MRayl), resulting in minimal signal loss [10].

The catheter tip, described in the previous section, provides a built-in reference 100µm above the transducer face. Prior to device encapsulation in the assembly process, the transducer surface is thoroughly cleaned to allow for good adhesion between the encapsulant and CMUT. The PDMS is then prepared and degassed to eliminate any air bubbles prior to deposition. In deposition, PDMS thickness is limited to 100µm, referenced by the catheter tip, to provide necessary protection while minimizing signal attenuation in this layer. The tip reference has a flat rim and Sylgard is poured in to a flat level to avoid any lensing effects.

Cabling and system integration

The final step in the probe integration, prior to insertion of the assembly into the catheter shaft, is cabling. Each trace on the flex leg terminates on a pad at the distal end of the flex which is then soldered to a 48AWG coaxial cable. For the QuadRing arrays, the number of connections will result in a total of 168 cables in the catheter shaft. The coaxial cable provides noise shielding while the thin gauge maximizes catheter flexibility.

Finally, the probe is integrated with the catheter shaft (Fig. 7). For electrical shielding, the shaft is lined with a wire mesh. Two pulling wires, one on either side of the CMUT array, are also incorporated in the catheter shaft to allow bending of the catheter tip from the distal end. The catheter cables are then terminated on a custom PCB connector, the mate of which is on an interface box [1,2] which brings DC power supplies to the catheter and allows connectivity to an imaging system.

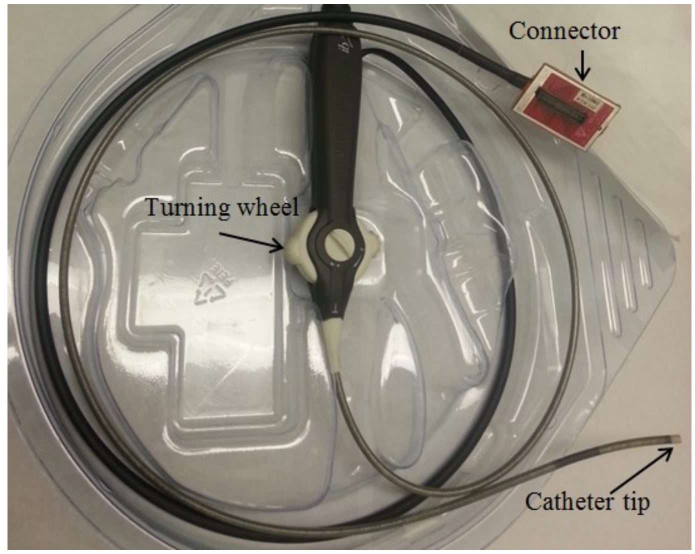

Figure 6. Completed 64-element Ring array catheter. The shaft handle is shown, on which there is a turning wheel for controlling the pulling wires of the catheter shaft. Also pictured are the custom PCB and connector, used to connect the catheter to the power supplies and imaging system hardware.

CONCLUSION

3D medical imaging systems are increasingly desirable, and ultrasound imaging, specifically with CMUTs, is a strong candidate in that field. In this paper, we have described a packaging and assembly method for building CMUT catheters. First, the fabrication of through wafer vias in a 3D QuadRing CMUT array has been described. A flexible PCB design and method for integrating the array and necessary electronics in a catheter package have been detailed. The fabrication, packaging, and assembly techniques presented allow for robust, repeatable catheter assemblies designed for volumetric imaging. Previous builds have been used for intracardiac echocardiology, and future work includes integrating QuadRing arrays into catheters for use in open heart, *in vivo* imaging and endoscopy.

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REFERENCES

1. Nikoozadeh, A., Oralkan, O., Gencel, M., Choe, J.W., Stephens, D.N., de la Rama, A., Chen, P., Thomenius, K., Dentinger, A., Wildes, D., Shivkumar, K., Mahajan, A., O'Donnell, M., Sahn, D., and Khuri-Yakub, P.T., 2009, "Forward-Looking Volumetric Intracardiac Imaging Using a Fully Integrated CMUT Ring Array," in *Proc. IEEE Ultrason. Symp.*, Rome, Italy, pp.770-773.
2. Moini, A., Nikoozadeh, A. Oralkan, O., Choe, J.W., Sarioglu, A.F., Stephens, D.N., de la Rama, A., Chen, P., Chalek, C., Dentinger, A., Wildes, D., Smith, L.S., Thomenius, K., Shivkumar, K., Mahajan, A., O'Donnell, M., Sahn, D.J., Khuri-Yakub, P.T., 2011, "Volumetric Intracardiac Imaging Using a Fully Integrated CMUT Ring Array: Recent Developments," in *Proc. IEEE Ultrason. Symp.*, Orlando, FL.
3. Nikoozadeh, A., Chang, C., Choe, J.W., Bhuyan, A., Lee, B.C., Moini, A., Khuri-Yakub, B.T., "An Integrated Ring CMUT Array for Endoscopic Ultrasound and Photoacoustic Imaging," in *Proc. IEEE Ultrason. Symp.*, Prague, Czech Republic.
4. Wygant, I.O., Zhuang, X., Yeh, D.T, Oralkan, O., Ergun, A.S., Karaman, M., Khuri-Yakub, B.T., 2008, "Integration of 2D CMUT arrays with front-end electronics for volumetric ultrasound imaging," *IEEE Trans. Ultrason. Ferroelect. Freq. Contr.*, 55, pp. 327–42.
5. Oralkan, O., Ergun, A.S., Cheng, C.H., Johnson, J.A., Karaman, M., Lee, T.H., Khuri-Yakub, B.T., 2003, "Volumetric ultrasound imaging using 2D CMUT arrays," *IEEE Trans. Ultrason. Ferroelect. Freq. Contr.*, 50, pp. 1581–94.
6. Lin, D.S., Wodnicki, R., Zhuang, X., Woychik, C., Thomenius, K.E., Fisher, R.A., Mills, D.M., Byun, A., Burdick, W., Khuri-Yakub, P., Bonitz, B., Davies, T., Thomas, G., Otto, B., Topper, M., Fritzsch, T., Ehrmann, O., 2013, "Packaging and modular assembly of large area and fin-pitch 2D ultrasonic transducer arrays," *IEEE Trans. Ultrason. Ferroelect. Freq. Contr.*, 60, pp. 1356–75.
7. Ergun, A.S., Huang, Y., Zhuang, X., Oralkan, O., Yaralioglu, G.G., Khuri-Yakub, B.T., 2005, "Capacitive micromachined ultrasonic transducers: fabrication technology," *IEEE Trans. Ultrason. Ferroelect. Freq. Contr.*, 52, pp. 2242–58.
8. Chang, C., Moini, A., Nikoozadeh, A., Sarioglu, A.F., Apte, N., Zhuang, X., Khuri-Yakub, B.T., 2014, "Singulation for imaging ring arrays of capacitive micromachined ultrasonic transducers," *J. Micromech. Microeng.*, 24, pp. 107002.
9. Pierret, R.F., 1996, *Semiconductor Device Fundamentals*, Addison-Wesley, Boston, MA, Chap. 16.
10. Lin, D.S., Zhuang, X., Wong, S.H., Kupnik, M., Khuri-Yakub, B.T., 2010, "Encapsulation of Capacitive Micromachined Ultrasonic Transducers Using Viscoelastic Polymer," *J. Microelectromech. Syst.*, 19(6), pp. 1341-1351.