Exploring the Potentials of Polymer-based CMUTs for 3D Ultrasound Computed Tomography

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***Abstract*—This work presents a quantitative evaluation of polymer-based Capacitive Micromachined Ultrasonic Transducers (polyCMUT) for 3D Ultrasound Computed Tomography (3D USCT). The study was motivated by limitations of the currently used PZT fiber technology in terms of bandwidth and transmit sensitivity. We developed large finite element models to predict the acoustic performance of polyCMUT elements consisting of 127 cells. We fabricated prototype transducers using a novel method for microstructuring polymer layers. The produced samples reach a fractional bandwidth of 116%, an opening angle of 47° and increase the transmit sensitivity by 54%, compared to the PZT fiber transducers. The developed models allow for accurate predictions of the acoustic field over a large range of angles and frequencies. More work is required to improve the reliability and reduce sample-to-sample variations. Based on the measured performance and the general properties of the technology, polyCMUTs are very promising for future 3D USCT systems.**

***Index Terms*—polymer-based CMUTs; 3D Ultrasound Computed Tomography; large model FEA; acoustic field characterization;**

I. Introduction

Ultrasound Computed Tomography (USCT) is an emerging imaging method for breast cancer screening ([1]. When compared to mammography and magnetic resonance imaging, ultrasound presents several advantages: it is considered safe, cost-effective, and does not necessitate the use of potentially harmful contrast agents [2], [3]. In most USCT designs, the breast is scanned by a moveable ring of transducers (see Fig. 1a), leading to cross-sectional images. These images can be stacked together to obtain volumetric structures. The fundamental concept behind a full 3D USCT is to surround the breast with numerous ultrasound transducers in a stationary 3D configuration (see Fig. 1b), in most cases a hemisphere. This allows to obtain directly volumetric information (e.g. out-of-plane scatters) of the breast [4]. Unfocused ultrasound waves are emitted for measuring transmission and reflection characteristics of the tissue. The images are reconstructed using synthetic aperture focusing techniques [5].

The image reconstruction algorithms used for 3D USCT requires acoustic fields with specific characteristics. Suitable transducers must exhibit a large bandwidth, wide opening angle and isotropic emission characteristics, with frequencies in the lower MHz range (approximately 0.5 MHz to 5 MHz) [6]. In previous work, we presented a custom-made transducer array design based on single-PZT-fibers [7]. These arrays showed a suitable performance, but could not fully comply with the ideal design goals, especially with respect to bandwidth, transmit sensitivity and opening angle [8]. A detailed analysis showed fundamental limitations of the utilized transducer technology, leading to the exploration of alternative transducer technologies.

Capacitive Micromachined Ultrasonic Transducers (CMUTs) are an emerging technology with the potential to outperform piezoelectric transducers in terms of ultrasound performance [9]. In CMUTs, the ultrasound is generated by the movement of thin silicon membranes, driven by electrostatic forces. These membranes, and the underlying cavities, are produced by means of microfabrication, such as thin film deposition and photolithography. A special CMUT fabrication approach was presented by researchers from the University of British Columbia (UBC), Vancouver, in 2018 [10]. In this approach, polymer instead of silicon layers are structured. These polymer-based CMUTs (polyCMUTs) allow for low-cost and rapid fabrication, making this technology very attractive for 3D USCT.

The main objective of this work was to investigate if the polyCMUT technology is a suitable transducer technology for future 3D USCT systems. We chose the following approach: First, suitable designs were investigated using large finite element (FE) models. These designs were then produced using the facilities available at UBC. Finally, the acoustic performance was characterized to enable quantitative comparison with the currently utilized transducer technology in 3D USCT.

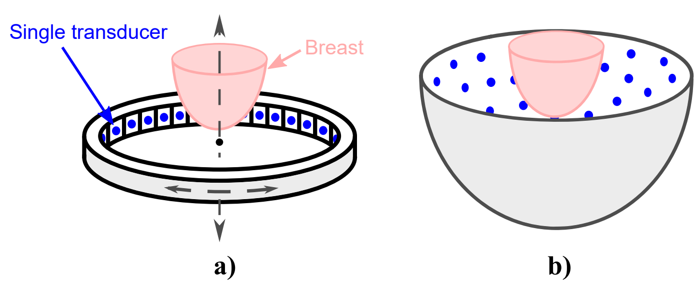

Fig. 1. Two basic concepts of USCT: A 2D configuration with a moveable ring of transducers (a), and a full 3D configuration using a hemispherical aperture (b).

Moreover, these measurements enable the validation of larger FE models for future design optimization.

II. Methods

A. Prototype Design

The design of the polyCMUT test transducer was guided by the acoustic characteristics defined in Tab. I. These characteristics were derived from the general requirements of a full 3D USCT system for breast cancer imaging as described in [8].

We chose a conventional circular cell design due to the available experience at UBC and isotropic emission characteristics of circular radiators [11]. For the initial design of the cells we followed the design guidelines in [12]. In order to reach the desired frequency range, cell diameters smaller than 100 μm are required, leading to a small output sound pressure. Hence, multiple cells have to be connected in parallel to increase the transmit sensitivity.

We chose an element design with 127 cells in a hexagonal arrangement to obtain a quasi-isotropic acoustic field in azimuth and elevation, while keeping the microfabrication design simple and the cell-to-cell distance constant. A cell-to-cell pitch of 132 μm was defined to increase the acoustic radiation resistance [13], while limiting the outer dimensions. This ensures a large opening angle. The cells have a diameter of 80 μm, a membrane thickness of 7.3 μm, and a gap size of 430 nm. Figure 2 shows a schematic cross section of a single cell, including the substrate, isolation and polymer layers as well as the vacuum-filled gap.

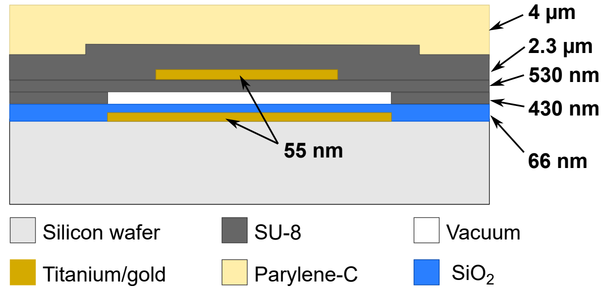

Fig. 2. Layer-by-layer schema of a polyCMUT cell including the layer thicknesses.

B. Finite Element Modeling

The acoustic performance of CMUTs can be accurately studied by means of FE analysis, since direct coupling of electromechanic and acoustic physics can be realized [14]. However, due to the small size of the cells, fine meshing is required, leading to large models and long computation times. However, by introducing symmetries, reducing the model complexity and utilizing computing clusters, solving larger models is still feasible.

We used COMSOL v6.1 to predict the acoustic field of the polyCMUT test transducers. Single cells were embedded as geometry parts and arranged in an array configuration. A 90° segment and only the two top layers (top SU-8 and Parylene-C) are foreseen in order to reduce the model complexity. The acoustic domain is modeled as a quarter-sphere, with a perfectly matched boundary condition and six mesh elements per wavelength. The resulting model is shown in Figure 3.

A DC bias of 40 V is applied to the cells using a stationary study, followed by a frequency domain pertubation step to obtain the performance over frequency. The used model exhibits more than $2{\cdot}10^5$ mesh elements, leading to approximately one million degrees of freedom. The model was solved on two computing clusters (BwUniCluster 2.0 and CMC CAD cluster), using 32 cores and 48 GB of maximum memory. The computation time varied between 1.5 and 3 hours, depending on the available resources. Figure 4 shows the cell displacement at 2.7 MHz in water from the FE analysis, on top of a microscope-photography of one fabricated hexagonal element.

C. Fabrication

The polyCMUTs tested in this work were fabricated in a process similar to the ones described in [10] and [15]. A 4 inch silicon wafer with 500 nm coating of thermal oxide was chosen as substrate. Only two distinct processes are used to create the different layers needed: thin-film lift-off and polymer patterning. For the metal lift-off process, AZ nLOF 2020 (Merck Performance Materials GmbH) is spin coated onto the

TABLE I
Design goals of 3D USCT in comparison with the PZT fibers and the polyCMUT test transducer (mean of three transducers).

Parameter	**Goal** (from [7])	**PZT fibers** (from [8])	**polyCMUT** FE model	Meas.
Center frequency	2.5 MHz	2.6 MHz	3.1 MHz	3.2 MHz
Bandwidth (-6 dB)	>100%	83%	146%	116%
Opening angle (-10 dB)	60°	43°	47°	44°
Transmit sensitivity*		47 Pa/V	74 Pa/V	87 Pa/V

*Peak sound pressure at 4 cm axial distance.

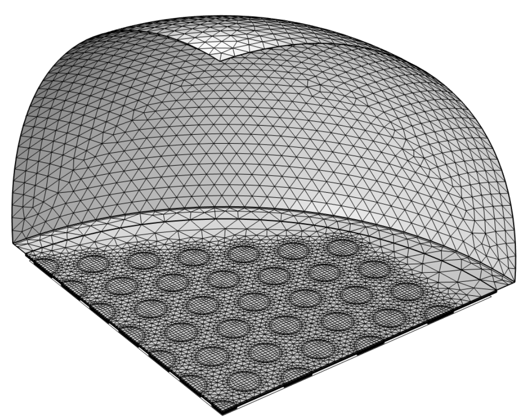

Fig. 3. FE model of the polyCMUT test transducer, implemented as quarter segment and hexagonal cell arrangement.

wafer, exposed in a maskless lithography system (MLA 150, Heidelberg Instruments) and developed in AZ 300 MIF (Merck Performance Materials GmbH). The desired thin film is then deposited via electron beam physical vapor deposition (AJA UHV Hybrid Evaporator System, AJA International) and the resist is removed in an acetone bath.

For the polymer layers, different variants of SU-8 2000 (Kayaku Advanced Materials) are spin coated according to their desired thickness, exposed in a physical mask aligner (NxQ 4006, Neutronix Quintel) and developed in SU-8 developer. More accurate fabrication parameters can be found in [10] and [15].

The first three layers are thin film processes, first a bottom electrode (5 nm titanium (Ti), 50 nm gold (Au)), then an insulation layer (5 nm Ti, 60 nm Silicon Dioxide (SiO_2)) and finally the sacrificial layer (430 nm aluminium (Al)). The bottom and top electrodes are patterned to overlap only at the cells to minimize parasitic capacitance. The insulating layer is just slightly larger than the lower electrodes to facilitate adhesion of the following layers. The SiO_2 insulation layer is necessary to avoid short circuits due to damaged polymer and to increase the electric field strength within the polyCMUT. Next is a polymer planarization layer of 430 nm SU-8 2000.5 in between the different cells as described in [15]. This allows for better clamping of the membrane and avoids different layer thicknesses in subsequent spin coatings.

The first membrane layer is also SU-8 2000.5, with a thickness of 500 nm. The thin film top electrode is patterned out of 5 nm Ti for adhesion and 50 nm Au, and the final membrane layer is patterned from SU8-2002 with a thickness of 2.3 µm. The wafer is then diced, and the Al sacrificial layer is edged with AZ MF 24-A developer (Merck Performance Materials GmbH). The membranes are released in a critical point dryer (Autosamdri-815B, tousimis), the chips mounted on a PCB and sealed in 4 µm of Parylene-C (SCS Labcoater 2, Speciality Coating Systems). The entire microfabrication was completed within four days. The produced samples were

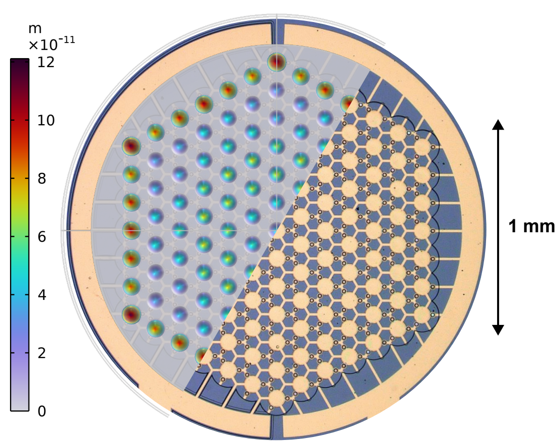

Fig. 4. Photography of the fabricated polyCMUT test transducer, overlayed with the results from the FE analysis of half the element.

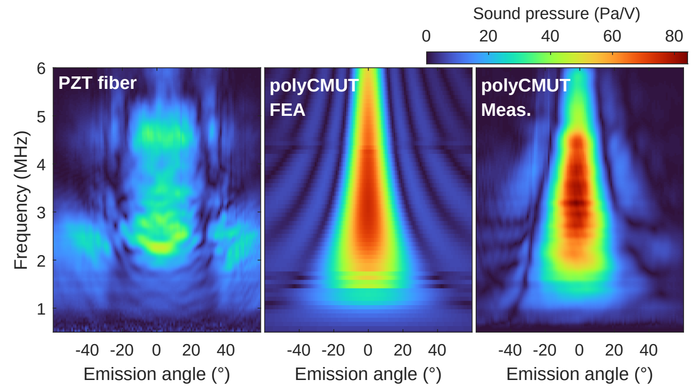

Fig. 5. Comparison of the FASP (120° circular segment at 4 cm distance) between one representative PZT fiber measurement, the modeled performance as well as one measurement of a polyCMUT test transducer.

finally embedded in a plastic housing to be characterized in an ultrasound test setup.

III. Results

We measured the frequency- and angle-dependent sound pressure (FASP) of three polyCMUT test transducer over a 120° circular segment at a radial distance of 4 cm. The generated ultrasound was recorded with a needle hydrophone (Onda HNC0400, Onda Corp.), mounted on an automated XYZ-stage in a water tank. The transducers were excited with a linear chirp signal from 0.5 MHz to 6 MHz to measure the broadband behavior. The measured ultrasound signal was normalized and bandpass filtered to obtain the transmit sensitivity in Pascal per Volt.

A direct comparison of the FASP between an exemplary measurement of a PZT fiber, the results from the FE analysis and one polyCMUT test transducer measurement is shown in Fig. 5. The polyCMUT test transducer shows a higher sound pressure than the PZT fiber transducer as well as a larger bandwidth. In addition, the side lobes, especially at lower frequencies, are less pronounced. When comparing the modeling results with the measurement, a good fit in terms of the frequency range, directivity behavior and generated sound pressure can be observed.

For a better comparison of the acoustic characteristics, the axial frequency response (vertical cut at 0° in Fig. 5) of the PZT fiber measurement, the model and the three polyCMUT test transducers are shown in Fig. 6. All three measured samples exceed the PZT fiber transducer in terms of generated sound pressure and bandwidth. However, while their frequency behavior is very similar, a large variation in sound pressure can be seen.

Table I gives a quantitative comparison of the ultrasound characteristics between the PZT fiber and the polyCMUT technology, based on the conducted tests. The direct comparison shows that improvements can be achieved with the polyCMUTs for all characteristics. Reaching the design goal in terms of the center frequencies only requires an adjustment in the cell dimensions. However, the in Section I introduced

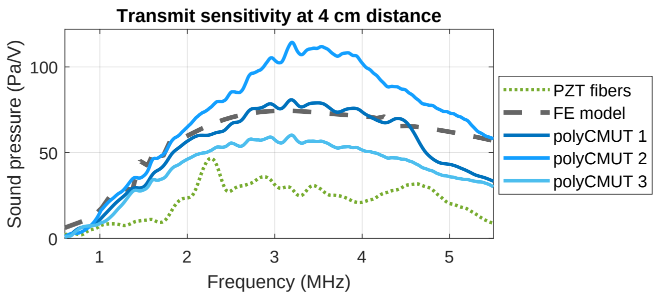

Fig. 6. Comparison of the transmit sensitivity sound pressure between a PZT fiber transducer, the results from the FEA and three polyCMUT test transducers at 4 cm axial distance.

frequency range from 0.5 MHz to 5 MHz can be well covered with the produced test transducer.

IV. Discussion and Conclusion

The modeled and produced polyCMUT test transducer presented in this work enables a quantitative evaluation of the technology in terms of requirements for 3D USCT. The measured characteristics (see Tab. I) better match the ideal design goals and exceed those of the currently used transducer technology (see [8]).

The large variation in transmit sensitivity of the polyCMUTs indicates that the manufacturing and assembly processes require additional quality control measures. Problems with the uniformity and mechanical stability of the Parylene-C layer led to the failure of several samples during testing. However, this does not explain the large variation in sensitivity, since this would introduce a shift in center frequency. Interesting is that one of the measurements significantly exceeds the modeled performance. The cause of this effect is currently being investigated in order to draw conclusions for further improvements.

In a possible use case, an additional protective layer, e.g. several 100 μm of PDMS, would be required for protection and isolation. This shifts the center frequency, reduces the transmission sensitivity and increases the bandwidth [16]. For example, previously conducted tests with thick layers of PDMS ($>$1 mm) showed a reduction in peak sensitivity by up to 39% [6, ch.8], which has to be considered for future designs.

In summary, the polyCMUT technology is very promising to be used in future 3D USCT systems due to the measured acoustic performance, the design flexibility (from microfabrication) as well as the low-cost and rapid fabrication. Future work will focus on the optimization of polyCMUT elements with respect to cell design, arrangement, shape and cell-to-cell pitch. However, this requires the reduction of modeling complexity since a large number of parameters have to be investigated.

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References

[1] N. Duric, P. Littrup, C. Li, O. Roy, S. Schmidt, R. Janer, X. Cheng, J. Goll, O. Rama, L. Bey-Knight, and W. Greenway, "Breast ultrasound tomography: Bridging the gap to clinical practice," *Progress in Biomedical Optics and Imaging - Proceedings of SPIE*, vol. 8320, p. 23, 2012.

[2] L. Lin, P. Hu, J. Shi, C. M. Appleton, K. Maslov, L. Li, R. Zhang, and L. V. Wang, "Single-breath-hold photoacoustic computed tomography of the breast," *Nature Communications*, vol. 9, no. 1, p. 2352, 2018.

[3] K. J. Opieliński, P. Pruchnicki, T. Gudra, P. Podgórski, J. Kurcz, T. Krasnicki, M. Sasiadek, and J. Majewski, "Imaging results of multi-modal ultrasound computerized tomography system designed for breast diagnosis," *Computerized Medical Imaging and Graphics*, vol. 46, pp. 83–94, 2015.

[4] N. V. Ruiter, M. Zapf, T. Hopp, R. Dapp, E. Kretzek, M. Birk, B. Kohout, and H. Gemmeke, "3d ultrasound computer tomography of the breast: A new era?," *European Journal of Radiology*, vol. 81, pp. S133–S134, 2012.

[5] H. Gemmeke, L. Berger, T. Hopp, M. Zapf, W. Tan, R. Blanco, R. Leys, I. Peric, and N. V. Ruiter, "The new generation of the KIT 3D USCT," in *International Workshop on Medical Ultrasound Tomography*, vol. 1, pp. 271–281, KIT Scientific Publishing, 2018.

[6] M. Angerer, M. Zapf, S. Gebhardt, and N. V. Ruiter, "Single-PZT-fiber transducers for 3D ultrasound computed tomography: Characterization and modeling," in *2022 IEEE International Ultrasonics Symposium (IUS)*, pp. 1–4.

[7] M. Angerer, *Transducer Arrays for 3D Ultrasound Computed Tomography*, vol. PhD thesis. Karlsruhe Institute of Technology, KIT scientific publishing, 2022.

[8] M. Angerer, M. Zapf, S. Gebhardt, H. Neubert, and N. V. Ruiter, "Single-fiber transducer arrays for 3D ultrasound computed tomography: From requirements to results," *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*, pp. 1–1, 2023.

[9] E. B. Dew, A. Kashani Ilkhechi, M. Maadi, N. J. M. Haven, and R. J. Zemp, "Outperforming piezoelectric ultrasonics with high-reliability single-membrane CMUT array elements," *Microsystems and Nanoengineering*, vol. 8, no. 1, p. 59, 2022.

[10] C. D. Gerardo, E. Cretu, and R. Rohling, "Fabrication and testing of polymer-based capacitive micromachined ultrasound transducers for medical imaging," *Microsystems and Nanoengineering*, vol. 4, no. 1, 2018.

[11] E. L. Kinlser, *Fundamentals of Acoustics*. John Wiley and Sons, 4th ed., 2000.

[12] I. O. Wygant, M. Kupnik, and B. T. Khuri-Yakub, "Analytically calculating membrane displacement and the equivalent circuit model of a circular CMUT cell," in *2008 IEEE Ultrasonics Symposium*, pp. 2111–2114.

[13] M. N. Senlik, S. Olcum, H. Koymen, and A. Atalar, "Radiation impedance of an array of circular capacitive micromachined ultrasonic transducers," *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*, vol. 57, no. 4, pp. 969–976, 2010.

[14] G. G. Yaralioglu, S. A. Ergun, and B. T. Khuri-Yakub, "Finite-element analysis of capacitive micromachined ultrasonic transducers," *IEEE Transactions on Ultrasonics, Ferroelectrics, and Frequency Control*, vol. 52, no. 12, pp. 2185–2198, 2005.

[15] J. Welsch, E. Cretu, R. Rohling, and C. D. Gerardo, "Ultrathin, high sensitivity polymer-based capacitive micromachined ultrasound transducers (polyCMUTs) for acoustic emission sensing in fiber reinforced polymers," in *2022 IEEE International Ultrasonics Symposium (IUS)*, pp. 1–5.

[16] M. F. l. Cour, M. B. Stuart, M. B. Laursen, S. E. Diederichsen, E. V. Thomsen, and J. A. Jensen, "Investigation of PDMS as coating on CMUTs for imaging," in *2014 IEEE International Ultrasonics Symposium*, pp. 2584–2587.