ABSTRACT Two kinds of 128 channels pMUT-based phased array ultrasound transducers are described in this paper, one with a high frequency of around 6 MHz and another with a low frequency of around 1.5 MHz. The active area of the transducer is around 25 mm long and 10 mm wide. There are in total 6270 pMUT elements in the reported arrays. For the transducer with a center frequency of 6 MHz, each pMUT has a membrane diameter of 85 µm and the pitch between every two channels is 200 µm. The transducer benefits from a high transmission and receive sensitivity of 44 kPa/V/Channel @ 3 cm and 204 mV/MPa, respectively. For the transducer with a center frequency of 1.5 MHz, each pMUT has a membrane diameter of 160 µm and the pitch between every two elements is 214 µm. The proposed transducer obtained a transmit and receive sensitivity of 430 Pa/V/Channel @ 3 cm and 190 mV/MPa, respectively. The transducer has a −3dB and −6dB bandwidth of 118% and 184%, respectively. The bandwidth is higher than any previously reported transducer in any technology, such as bulk-PZT, pMUT, or capacitive MUT (cMUT). The functionality of the transducer arrays is confirmed by obtaining B-mode images in water medium.

INDEX TERMS Piezoelectric micromachined ultrasound transducer (pMUT), medical imaging, phased array, ultrasound.

I. INTRODUCTION

ULTRASOUND has found applications in nondestructive evaluation (NDE) [1], ultrasonic actuation [2], medical imaging [3], drug delivery and other therapeutic applications [4], particle and cell manipulation [5], and object recognition [6]. Especially, ultrasound medical imaging has had a considerable impact on our life since its advent in the 1940s. Because of the advances in electronics, medical ultrasound imaging systems have seen remarkable progress over the last years. Similar to other electronic systems, ultrasound devices are becoming more compact, portable, smarter, more power-efficient, and cheaper. Therefore, currently, there are many emerging ultrasound imaging applications in medical diagnostics, such as handheld probes for point-of-care ultrasound (POCUS) applications [7] and intravascular ultrasound systems (IVUS) [8]. Currently, almost all imaging systems on the market are using bulk piezoelectric materials to convert electrical excitation to ultrasound waves or vice-versa. However, these conventional technologies face various limitations. The elements in a transducer array require mechanical dicing of the bulk layer by a saw, which limits the pitch by the kerf of the saw [9]. Also, the transducers use high actuation voltages in the range of 70-140 V during transmission, which limits their usage or performance in a compact, battery-based devices. Moreover, the manufacturing of conventional transducers is very costly and labor-intensive.

To overcome the aforementioned drawbacks and align ultrasound imaging systems with the market trend of electronic devices, micromachined ultrasound transducers (MUTs) have been developed in the last decade. MUTs benefit from a lithography-based MEMS fabrication process, ultra miniaturization, potential integration with CMOS technology, and a low-cost fabrication process. There are two types of micromachined technologies, namely piezoelectric micromachined ultrasound transducer (pMUT) with a piezoelectric thin film typically in d31 mode [10], [11] and capacitive micromachined ultrasound transducer (cMUT) also in the flexural mode with electrostatic forces [12], [13], [14].

A cMUT element is, in essence, a miniaturized capacitor that consists of a thin metalized suspended membrane.
CMUTs can receive and transmit acoustic signals and are currently used in several ultrasound probes from different manufacturers [15]. However, cMUT technology also has some critical drawbacks that prevent its use as an alternative to conventional ultrasound technology. CMUTs rely on a small gap (100 - 300 nm) between the membrane and electrode, which can easily be contaminated. For high performance operation of cMUT, a high DC bias voltage (30-100 V) is applied near the collapse voltage. This limits their usage for three reasons: (i) increased power consumption, (ii) higher noise levels, therefore, low SNR, and (iii) relatively high probability of device failure. Moreover, the membrane thickness in cMUTs is limited to a thickness of maximum of about 2 μm, first because they have relied on surface micromachining fabrication techniques, and second because the weak electrostatic should be able to bend the membrane. As a result, the lateral dimension of a cMUT is often very small to achieve a certain resonance frequency, which results in a low output acoustic pressure and sensitivity [16]. Consequently, cMUTs suffer from low image quality, especially in applications that require a high depth of penetration.

PMUT consists of a membrane with a piezoelectric thin-film. The membrane is driven by applying an excitation voltage between the top and bottom electrodes. The applied electric field forms transverse stress in the active piezoelectric layer, which causes out-of-plane membrane displacements generating a pressure wave in the outer medium. In contrast to cMUTs, piezoelectric MUTs (pMUTs) do not require a DC biasing voltage and their membrane geometry can be freely chosen in order to provide sufficient output acoustic power [11]. However, the main issues with pMUTs up to now are: (i) the low frequency bandwidth and (ii) a low transmission response with respect to bulk piezoelectric technology.

In this paper, we propose 1D pMUT phased arrays with 128 channels to address these issues. The design and fabrication process of the pMUT arrays are explained in detail. Moreover, measurement results of imaging prototypes are described, which illustrate the promising potential of the transducer for medical imaging applications. This paper is an extension to our conference paper [17], and beyond what was published before it elaborates more on the design, fabrication process, discussions on the imaging experiments and measurements, benchmarking, and more importantly, the introduction of an extra pMUT array for a broader investigation and comparison.

II. MATERIALS AND METHODS

A. DESIGN OF THE ARRAY

pMUT design typically involves maximizing transmission sensitivity and bandwidth. Transmission sensitivity is related to the output acoustic power per unit of input electrical power. It has been previously demonstrated that in order to have sufficient output acoustic power, the lateral dimension of the pMUT membrane should be equal to or bigger than half of the transmitted wavelength [16]. Therefore, based on the desired working frequency, lateral dimensions and the thickness of the membrane are optimized. On the other hand, the bandwidth of each pMUT can be tuned by utilizing the damping effect of the medium on the pMUT membrane. The mass of the medium damps the membrane vibration more effectively when the lateral dimension of the membrane is smaller than the wavelength. Therefore, the lateral dimension of the membrane should be a compromise between the output acoustic power and the bandwidth. Due to the inverse square relationship of frequency to aperture, a higher resonant frequency is likely to have a higher aperture to wavelength ratio and hence higher acoustic power and transmission response. In contrast, low frequency designs normally have a lower aperture-wavelength ratio and are exposed to higher damping resulting in higher bandwidths. Based on these aspects, two designs are proposed in this work.

For the following discussion, the pMUT array is considered to be in the x-z plane with the y-axis perpendicular to the surface of the array. The azimuth angle is placed in the x-y plane and any azimuth and elevation angle on the y-axis is zero. The elements are placed along the azimuth x-axis and each channel is elongated along the elevation direction (z-axis). Since there is no significant beamforming in the elevation direction, a pitch larger than half of the wavelength can be chosen between the pMUTs of each channel. This helps to have a sharper beam pattern in the elevation angle. The number of pMUTs in each channel is also an important factor in enhancing the output pressure and the elevation directivity [18]. However, if the size of the transducer in the elevation direction is too large, the lateral spatial resolution will be negatively affected. Depending on the application and depth of penetration, the elevation size of the transducer should be chosen. Also, a larger number of pMUT cells in a channel implies larger parasitic capacitance which leads to higher power consumption and lower receive sensitivity and signal to noise ratio (SNR).

The proposed high frequency pMUT array consists of 128 channels with a pitch of 200 μm. Each channel comprises several pMUTs that are divided into three groups (aligned with the elevation direction), in order to have access to each group separately. The pMUTs in each group are connected to each other and are accessible via a bond-pad. In this way, the beam pattern in the elevation direction can be steered and focused up to a certain level and make the array independent of an acoustic lens. Fig. 1 shows a microscopic image of the array, in which the inset shows a magnified view of some of the pMUTs. Each pMUT has a diameter of 85 μm and a Si membrane thickness of 6 μm. For low frequency pMUT array with 128 channels with a pitch of 214 μm. Each pMUT cell has a membrane diameter of 160 μm making this a high fill factor design. Each channel consists of a column of several pMUT elements connected to each other in line with the z-axis.

A further difference between the design of the two arrays is the presence of a thin polyimide layer in the high frequency
array between top electrode interconnections and PZT. This reduces parasitic capacitance, which defines the electrical impedance of the transducer at the working frequency. The electrical impedance of the transducer should be matched to the front-end system to guarantee the maximum power transfer and minimize the signal reflection at the interface of the transducer and front-end. Therefore, a matched impedance is also important for a wide bandwidth response. Furthermore, capacitance is an important parameter in defining the power consumption of the transducer. Basically, everywhere in the pMUT chip, where the PZT layer is sandwiched between the top and bottom electrode a parasitic capacitor is made. The capacitance of an element together with the parasitic resistors form a low pass filter with a bandwidth of 1/2RC. If the resonance frequency of the transducer is higher than this value, most of the input electrical power is consumed to charge and discharge the capacitor [16]. Hence, for the high frequency pMUT an additional polyimide dielectric layer is introduced on top of the PZT layer, which works as an insulating layer, to reduce the parasitic capacitance.

B. FABRICATION PROCESS

The pMUT arrays were fabricated based on the bulk micromachining process, as shown in Fig. 2 [16]. In the first step, (a) about 250 nm Ti/Pt was deposited on an SOI wafer as the bottom electrode. Then, (b) a 1 μm PZT layer was deposited by a sol-gel process [19]. The PZT layer was patterned to access the bottom electrode. For the high frequency pMUT array, an additional step (c) is performed. In step (c), in order to reduce the parasitic capacitance, a thin layer of polyimide was deposited all over the wafer, which was partially etched away to expose the pMUT membranes to the top electrode. As the next step (d), a multilayer of Ti/Ag/Ti/Pt with the respective thickness of 30, 170, 30, and 110 nm was deposited and lifted-off as the top electrode. The radius of the top electrode was chosen to be about 70% of the membrane radius to maximize the induced lateral stress and thus maximize the displacement response [20]. Afterward, (e) a 1 μm polyimide layer was deposited in order to (i) introduce more damping on the pMUT membrane to widen the bandwidth and (ii) serve as a protective layer since the array will be submerged in ionized water or ultrasound gel. The polyimide layer is then patterned to have access to the bottom and top electrodes, which is required for wire bonding at a later stage. Then, (f) the membrane was realized by a deep reactive ion etching (DRIE) process on the backside of the wafer. This is followed by HF vapor phase etching of the buried oxide (BOX) layer. The cross-section SEM images of the realized membranes are shown in Fig. 3.

Finally, in order to have electrical access to channels, the transducer was wire-bonded to a PCB, where each channel is connected to a coaxial cable. A part of the transducer wire bonded to the PCB is shown in Fig. 4. The wire bonding is later covered with an epoxy layer (EPO-TEK H54) for the protection against water, humidity, and physical contact.

III. RESULTS AND DISCUSSION

A. ELECTRICAL CHARACTERIZATION

The measured capacitance of each individual channel is about 8.5 nF for the low frequency design. However, for the high frequency array design, in which the low pass filtering effect is more prominent, we could successfully reduce the parasitic capacitance from 8.5 nF to 1.8 nF per channel due to the
polymide layer between the PZT and top electrode layers. Therefore, the performance of the array is enhanced in three aspects: (i) the power efficiency: less input electrical power is needed to generate the same output acoustic power since there is less parasitic capacitance; (ii) it increases the SNR of the array as a receiver and (iii) the input electrical impedance of each channel has a higher value with respect to the case that no polymide isolation layer was used, which makes it easier to be matched with the analog front-end, (15 Ω and 3 Ω for the case of with and without the polymide layer, respectively).

B. IN-AIR CHARACTERIZATION
The fabricated pMUT arrays were first characterized in air by laser doppler vibrometer (LDV) (Polytec MSA-600, Germany) to determine the performance of a single pMUT. Fig. 5 (a) shows the frequency response and the time domain response of the high frequency pMUT. It has a resonance frequency of 8 MHz in air and a displacement response of 2.6 nm/V, which is equal to 130 mm·s⁻¹·V⁻¹. Fig. 5 (b) shows the frequency and time domain responses of a low frequency pMUT. It has a resonance frequency of 2.35 MHz in air and a displacement response of 15 nm/V. It should be noted that the frequency responses were measured by applying a chirp signal where the resonance frequency of the pMUT is only a very small part of it. Therefore, the input energy of the desired working frequency to the pMUT is very small compared to the time domain signal, where only one frequency is used to excite the pMUT. This is the reason why the peak amplitude of the frequency response is much lower than the time domain response.

C. UNDERWATER CHARACTERIZATION
The transducer was also characterized underwater, as water can be considered as a similar medium to a human body, which gives a good indication of its performance for medical imaging applications. Firstly, transmitted acoustic pressure by the transducers was measured. Fig. 6 shows a graphical illustration of the underwater experiment. A commercial 1mm needle hydrophone (from Precision Acoustics, UK) and a single element transducer (PVDF, 19 mm diameter with 10MHz
center frequency from Precision Acoustics, UK) were used to characterize the transmission and receive performance of the pMUT array. In this measurement, the transducer was directly connected to a waveform generator, terminated with 50 Ω.

Fig. 7 (a) shows the underwater sound pressure level of one randomly chosen channel of the high frequency array measured by the hydrophone at a distance of 3 cm. The pMUT array has a center frequency of 6 MHz underwater and shows a transmission sensitivity of 45 kPa/V @ 3 cm per channel and a −6 dB bandwidth of 18% with respect to the center frequency. In order to compare the performance of our pMUT array, we have ordered a custom design transducer from Olympus (based on the classical bulk PZT technology) with similar possible specifications to our pMUT arrays. The customized Olympus transducer comprises 128 channels with a center frequency of 3 MHz, 200 µm pitch, −6dB bandwidth of about 60%, and 100 Ω electrical input impedance. The obtained transmission sensitivity by our high frequency pMUT array is about one order of magnitude higher than Olympus transducers. Consequently, the proposed pMUT array has the potential to exhibit a higher depth of penetration and image resolution in medical applications.

The frequency response of the low frequency transducer is shown in Fig. 7b, which was measured by sweeping the frequency with the waveform generator and recording the acoustic response by the hydrophone. As shown, the transducer has a center frequency of 1.5 MHz and a −3dB and −6dB bandwidth of 1.76 MHz (118% of the center frequency) and 2.76 MHz (184% of the center frequency), respectively. Moreover, it has a transmission response of 430 Pa/V @ 3 cm per channel, which is higher than most bulk-PZT based ultrasound transducers [21]. By considering the capacitance and the center frequency, the impedance of the transducer was estimated to be $1/2\pi f C = 8 \Omega$. Table 1. shows the comparison between specifications of the fabricated low frequency array and other previously reported pMUT arrays in the literature. It was tried to select the pMUT arrays that were intended to be used in medical ultrasonic imaging applications.

There are three reasons why the high frequency array has higher transmission sensitivity with respect to the low frequency array. First, the high frequency array has a higher input impedance. Therefore, since our signal generator has an impedance of 50 Ω, more voltage is transferred over the pMUT. Second, the high working frequency, results in a higher velocity of the membrane, which causes a higher output pressure. And third, the low frequency array has a very wide bandwidth, which means that its vibration is damped significantly. Therefore, a lower vibration amplitude, and consequently, a lower output pressure is obtained.

To measure the receive sensitivity of the pMUT transducer, a single element wide-band transducer (Precision Acoustic, UK) was used to transmit a burst signal. In this experiment, an acoustic pressure wave was transmitted by the single element transducer and received by one channel of the pMUT transducer, which was connected to an oscilloscope terminated at high impedance. The transmitted signal was measured by both the hydrophone and the pMUT transducer. Receive sensitivities of 204 mV/MPa and 190 mV/MPa per
TABLE 1. Comparison between the fabricated low frequency pMUT array in this paper and previously reported pMUT arrays in the literature, which were intended to be utilized for medical ultrasonic imaging applications.

|                         | This work | [22] | [23] | [24] | [25] | [26] | [27] |
|-------------------------|-----------|------|------|------|------|------|------|
| Center frequency        | 1.5 MHz   | 5 MHz| 1.2 MHz| 2.25 MHz| 7.5 MHz| 1.63 MHz| 4 MHz |
| Number of channels      | 1 x 128   | 1 x 64| 1    | 8 x 8 | 1 x 15 | 1    | 5 x 17|
| -3dB Bandwidth          | 118 %     | 30 % (-6dB)| 29 % | 6.25 %| --- | 38 % | 87 % |
| Transmit sensitivity    | 430 Pa/V @ 3 cm | --- | 950 Pa/V @ 2 cm | 7 kPa/V @ surface | 1.2 kPa/V @ 0.5 mm | --- | 960 Pa/V @ 1 mm |
| Receive sensitivity     | 190 mV/MPa | --- | 1.35 μV/Pa | --- | 6 μV/Pa | --- | 0.016 pC/kPa |
| Depth of performed imaging | 12 cm        | 4 cm | N/A | --- | 0.5 cm | N/A | 0.3 cm |

FIGURE 8. For the high frequency pMUT array, (a) underwater time domain transmission responses of one channel to one cycle burst sinusoidal signal. (b) Underwater time domain receive response of one channel to one cycle burst sinusoidal signals.

channel were obtained for the high frequency and low frequency pMUT transducers, respectively.

The underwater time domain responses of one channel in the arrays are shown in Fig. 8 and 9. In transmission mode, a channel of the high frequency array was excited with a one-cycle sinusoidal burst signal, and its response was measured with the hydrophone, shown in Fig. 8a, whereas Fig. 8b shows the receive response of one channel of the pMUT array to one cycle sinusoidal burst signal and 5 cycles sinusoidal burst signals. Fig. 9 shows the time domain transmitted and received signals by a channel of the low frequency pMUT transducer. As shown in Fig. 9a, the short transmission time response without any ringing is a verification of the transducer’s ultrawide bandwidth. This also applies to the received signal by a channel of the pMUT transducer, which is shown in Fig. 9b. Unfortunately, the pulse response of the single element transducer, utilized in this experiment, is not short and has some ringing. However, a very similar response was captured by the pMUT channel. The reflection shown in the received signal by the pMUT channel corresponds to the reverberations bouncing back and forth between the pMUT array and the single element transducer.

D. IMAGING EXPERIMENTS

The functionality of the pMUT arrays for imaging was demonstrated by an imaging experiment, again performed...
underwater. All of the 128 channels of the array were connected to an ultrasound research system (HD Pulse, Diagnostic Sonar LTD, UK) for transmission and acquisition. As the transmission method, single-line transmission (SLT) with a focus point at 60 mm was used to scan a 90° wide and 12 cm deep area. The azimuth scan resolution was selected to be 1°. All the steps taken in the imaging experiments are as follow:

1) Each channel in the array is excited with a transmission pulse. The delay between channels in excitations is calculated based on our SLT algorithm.
2) After each excitation, the signals are recorded via HD Pulse to cover a depth of 12 cm.
3) The excitation through the array and recording the signals are repeated for 90 times to cover a 90° wide angle with steps of 1°.
4) A delay and sum (DAS) beamforming algorithm is applied on the recorded signals for each scanned angle.
5) At the end, the beamformed data are used to construct the image.

For the high frequency pMUT array, we used a custom designed reflector matrix with different reflector dimensions as the imaging target, which is shown in Fig. 10a. The width of reflectors varied from 0.5 up to 2 mm by a step of 0.5 mm from right to left of the matrix. A 4-cycle rectangular pulse at 6 MHz with 50 Vp-p was used to actuate each channel in the array. The final reconstructed B-mode image is shown in Fig. 11a.

For the low frequency array, four metal rods were placed in front of the pMUT array, as such that they were perpendicular to the B-mode scanning surface. Two rods were placed closer and two rods further from the array with a diameter of 7.5 mm and 5 mm, respectively. The imaging setup is shown in Fig. 10b. A one cycle rectangular pulse at 1.5 MHz with 25 Vp-p was used to actuate each channel in the array. The final reconstructed B-mode image is shown in Fig. 11b.

IV. CONCLUSION
High frequency (6 MHz) and low frequency (1.5 MHz) pMUT arrays with 128 channels for medical imaging applications are proposed in this paper. The design, fabrication process, and characterization of the array are described in detail. The pMUT arrays are characterized in air by LDV and underwater by using a 1 mm needle hydrophone and a wide bandwidth single element transducer. The functionality of the arrays for imaging is confirmed by an underwater imaging experiment, which shows the high potential of the proposed pMUT array for medical imaging applications. Using a high frequency array is useful to obtain higher spatial resolution images, while the energy of a lower working frequency ultrasound wave is absorbed less by the medium and can
penetrate deeper into the body. The design of the arrays was performed with an emphasis on the frequency bandwidth of the pMUTs. A higher bandwidth in the frequency response results in a shorter acoustic pulse and less ringing on the output of pMUTs, which increases the axial spatial resolution. A wide bandwidth is also necessary during the receive, as it helps to capture other frequencies and harmonics generated by the organs and tissues, which are useful for e.g., doppler or harmonic imaging methods. The meaning of a good quality ultrasound image is always merged the level of signal to noise (SNR) ratio obtain by the transducer. To have a high SNR, a lower parasitic capacitance and higher transmission and receive sensitivities are always preferred. The proposed high frequency array has a transmission sensitivity of 45 kPa/V @ 3cm and a receive sensitivity of 204 mV/MPa. The high transmission sensitivity is also beneficial in having a high depth of penetration and obtaining high resolution images. Whereas the low frequency transducer has a center frequency of 1.5 MHz and benefits from a wide bandwidth of 118%. The transmission and receive pulse responses are without any noticeable ringing, which is beneficial for a high spatial image resolution.

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SANJOG VILAS JOSHI received the integrated bachelor’s and master’s degrees from the Indian Institute of Technology Bombay (IIT Bombay) in 2020. He is currently pursuing the Ph.D. degree in electrical engineering with KU Leuven, Belgium. He is mainly interested in the areas of MEMS and microelectronics. He is currently working on flexible piezoelectric micromachined ultrasound transducers for the Ph.D. project.

CHEN WANG received the B.S. degree in optical information science and technology from Anhui University, China, in 2013, and the Ph.D. degree in measuring and testing technology from Zhejiang University, China, in 2018. He was a joint training Ph.D. Student, with prof. M. Kraft, at the University of Liege, Belgium, from 2015 to 2017. He is currently a Post-Doctoral Researcher with ESAT-MNS, KU Leuven. His current research interests include MEMS inertial sensors, MEMS pressure sensors, the design and optimization of MEMS devices with freeform structures, intelligent feedback systems for MEMS sensors, and electromechanical sigma–delta modulator interface circuits.

MICHAEL KRAFT leads the ESAT Research Division Micro- and Nano-Systems. He joined ESAT as a Full Professor in October 2017 and has more than 20 years of experience in the design, fabrication, and characterization of a wide range of micro- and nanosystems (MEMS), sensors and devices. He has worked on inertial sensors, intelligent interface circuits, and control systems for micro-devices, atom and ion chips, bio-medical and biochemical sensors and devices, energy harvesters, and piezoelectric ultrasound transducers. He is in charge of the Cleanroom and MEMS activities in the Leuven Nanocentre.