Large-scale surface-micromachined optical ultrasound transducer (SMOUT) array for photoacoustic computed tomography

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Abstract: This paper reports a new 2D surface-micromachined optical ultrasound transducer (SMOUT) array consisting of 350 × 350 elements with highly uniform optical and acoustic performances. Each SMOUT element consists of a vacuum-sealed Fabry-Perot (F-P) interferometric cavity formed by two parallel partially reflective distributed Bragg reflectors (DBRs). Optical mapping in the 4 cm × 4 cm center region of the SMOUT array shows that the optical resonance wavelength (ORW) of > 94% of the elements falls within a narrow range of ≤ 10 nm. The center frequency, acoustic bandwidth and noise equivalent pressure (NEP) of the elements are determined to be 5 MHz, 5 MHz, and 20.7 Pa (with 16 times of signal averaging) or 172.5 Pa (without averaging) over a bandwidth of 10 MHz, respectively. The temperature and temporal stability of the SMOUT elements is also tested, which shows there is little variation in their ORW under large ambient temperature fluctuation and during continuous water immersion. To demonstrate its imaging capability, 2D and 3D PACT based on the SMOUT array is also conducted within a 3 cm × 3 cm field of view (FOV) at a depth of 3 cm with no interrogation wavelength tuning. These results show that the SMOUT array could overcome some of the major limitations in existing ultrasound transducer arrays for PACT and provide a promising solution for achieving high-speed 3D imaging.

1. Introduction

As a hybrid imaging modality, photoacoustic computed tomography (PACT) has gained significant interests in the past decade [1–2]. In PACT, a short-pulsed laser is used to illuminate the target. Upon the absorption of the incident laser pulses, ultrasound waves (i.e., photoacoustic (PA) waves) will be generated inside the target, which can be detected by an ultrasound transducer array for image reconstruction. Because the amplitude and travel time of the PA waves are closely related to the optical absorption property (not scattering) and the location of the source points, PACT can well resolve optical absorption contrasts (e.g., blood vessels, fat and tumors, etc.) well beyond the optical diffraction limit in biological tissues. It outperforms other optical imaging technologies in terms of penetration depth. To enable high-speed 3D PACT, a large-scale 2D transducer array consisting of high-sensitivity elements is needed for simultaneous or parallel detection of the PA signals even from centimeter-deep tissues. Currently, piezoelectric [3–5] and capacitive micromachined ultrasound transducer arrays [6–7] are commonly used due to their readily availability (Figs. 1(a) and 1(b)). However, as they operate upon electrical charge generation, their acoustic sensitivity becomes poor when the transducer element size has to be made small, e.g., for the formation of a 2D array. Secondly, the need for electrically wiring a large number of elements makes the transducer array rather complex and costly. Thirdly, limited by the number of available signal channels, it is impossible to achieve simultaneous or parallel readout of PA signals from all transducer elements (especially in a 2D array). Instead, a multiplexing
approach oftentimes has to be adopted, whose serial nature hampers the overall data acquisition and imaging speed.

Recently, optical ultrasound transducers have been investigated for PA detection. Different from their piezoelectric or capacitive counterparts, optical ultrasound transducers convert the input ultrasound signal into an optical output. Ultrasound waves impinging onto the transducer change its geometric parameter (e.g., the Fabry-Perot (F-P) cavity length) or optical property (e.g., refractive index), and therefore modulate the reflected/transmitted optical signal from the sensor (Fig. 1(c)). Examples of such modulated optical sensors also include fiber optic intensity-variation based sensors for mechanical parameter measurement [8–9]. Because of this, the sensitivity of the optical ultrasound transducers is solely determined by the acoustic pressure level as well as the power level of the interrogation light source and is unrelated to lateral size of the transducer. Therefore, it is possible to maintain high sensitivity while reducing the transducer element size (required for 2D array formation). In addition, the PA signals can be read out in parallel via optical means without massive electrical wiring. It should be mentioned that the main disadvantage is that a more complex optical setup is needed to read out the optical output signal from the transducer element. However, for transducer arrays, this will not be an issue but may even become an advantage because the same optical setup can be used to read the data from all transducer elements in the array.

Based on their optical modulation mechanism, the optical ultrasound transducers can be either refractometric [10] or interferometric [11]. Especially, interferometric devices are generally considered a better choice due to their higher sensitivity. For example, extrinsic F-P interferometer (EFPI) fiber optic sensors have been developed for high-sensitivity biochemical testing [12–14] and pressure and acoustic testing [15–18]. They are typically fabricated by bulk etching and wafer bonding [15–16] (Figs. 2(a) and 2(d)) or manual assembly (Figs. 2(b) and 2(e)) [17–18]. While they are effective in producing single devices, current fabrication techniques lack the accuracy for creating large-scale (2D) array of transducer elements with uniform optical and acoustic performance, which is essential for parallel signal readout required in high-speed imaging applications. For example, it is rather difficult to precisely control the bulk etching depth (in wafer-bonded sensors) and the distance between fiber tip and diaphragm (in manually assembled ones). As a result, key dimensions of the transducer elements in an array, such as the length of the F-P cavity, could scatter in a wide range.

To address this issue, we report a new surface-micromachined optical ultrasound transducer (SMOUT) array, where transducer element consists of an F-P interferometric cavity formed by surface micromachining [19]. As shown in Figs. 2(c) and 2(f), a “sacrificial” layer is first deposited between the top diaphragm and substrate, and subsequently etched away to create the F-P cavity. The cavity length will be directly determined by the thickness of the sacrificial layer. Formed by thin-film deposition, the thickness variation of the sacrificial layer can be
controlled within a few nm. As a result, highly uniform F-P cavities can be fabricated on a large substrate to form large-scale and high-density 2D arrays. Experiments are conducted to characterize both the optical and acoustic performances and stability of the SMOUT array. To demonstrate its imaging capability, PACT based on the SMOUT array is also conducted with one or multiple point targets. Results showed that the SMOUT array has excellent optical and acoustic uniformity, high acoustic sensitivity, and good temperature and temporal stability, which is suitable for parallel data acquisition to enable high-speed 3D PACT.

2. Sensor design

As shown in Fig. 3(a), a SMOUT element consists of a vacuum F-P interferometric cavity formed by two parallel partially reflective distributed Bragg reflectors (DBRs). Ambient pressure changes due to impinging ultrasound waves vibrate the top flexible DBR diaphragm and tune the cavity length of the F-P cavity. The wavelength at which the reflection spectrum has the lowest reflectivity (defined as the optical resonance wavelength (ORW) ($\lambda_0$ in Fig. 3(b)) is changed because of the ultrasound-induced cavity length variation. Consequently, optical power reflected by the SMOUT elements at wavelength close to the ORW is modulated by the acoustic wave. To maximize the linearity between the optical signal and the acoustic pressure, the SMOUT elements are interrogated at a wavelength that is located around the middle point ($\lambda_{bias}$) between the ORW and the wavelength giving highest reflectivity (Fig. 3(b)). Each DBR consists of multiple silicon oxide and nitride pairs deposited by PECVD (plasma enhanced chemical vapor deposition). The refractive indexes of the silicon oxide and nitride films could change slightly from batch to batch. Before each DBR deposition, a dummy deposition run is conducted, and the refractive indexes of the deposited films are measured. Based the measurement values, the thicknesses of the silicon oxide and nitride layers in the DBR are adjusted accordingly to ensure each layer have an optical length (the product of thickness and refractive index) equal to one quarter of the DBR’s center wavelength (e.g., 785 nm). The initial F-P cavity length is tuned to half of the DBR’s center wavelength by adjusting both the deposition thickness of the sacrificial layer and the deformation of the top DBR diaphragm after the vacuum sealing of F-P cavity [20]. The top diaphragm has a diameter of 70 µm and an overall thickness of $\sim$3 µm (Fig. 3(c)). Its dynamic (flexural-mode)
response in water is simulated with COMSOL Multiphysics, which shows a resonance frequency of 4.3 MHz. To detect the impinging ultrasound wave, the SMOUT elements need to be immersed in water, which serve as a coupling medium for ultrasound propagation. Water also increases damping on the top diaphragm and therefore changes its dynamic response. Therefore, the simulation needs to be conducted with water serving as the ambient condition. A Parylene layer is coated on top of the diaphragm for two reasons. As a soft polymer material, a Parylene coating helps to damp diaphragm vibration and broaden acoustic bandwidth of the SMOUT elements. Also, the Parylene coating better protects the SMOUT array from abrasion and moisture. The element pitch is 140 µm, which is smaller than half of the acoustic wavelength (e.g., ∼350 µm at a frequency of 4.3 MHz) in water. This helps to eliminate the grating lobes in directivity profile and aliasing in spatial sampling. Grating lobes are one type of artifacts generated when a transducer array has element spacing greater than half wavelength of the ultrasound beam. Spatial aliasing means insufficient spatial sampling of the ultrasound beam (sampling rate lower than twice of the ultrasound frequency) that leads to ambiguous image reconstruction. Both effects can be avoided by setting the pitch size of the SMOUT array (spacing between two adjacent elements) smaller than half wavelength of the ultrasound wave. Figure 3(d) shows a 5 cm × 5 cm SMOUT array fabricated on a glass substrate consisting of a total number of ∼350 × 350 elements.

Fig. 3. (a) Schematic showing the cross-section and ultrasound detection mechanism of a SMOUT element. (b) Schematic showing the modulation of reflected amplitude at interrogation wavelength λbias by acoustic-induced reflection spectrum shift. (c) Zoom-in view of a 5 × 5 SMOUT elements (the scale bar is 200 µm). (d) Zoom-out view showing the entire SMOUT array (the scale bar is 2 cm).
3. Testing and characterization

3.1. Optical reflection spectrum

The optical reflection spectrum of the F-P cavity in each SMOUT element determines the optical wavelength for reading out the PA or ultrasound signal. It should be made as uniform as possible, such that no wavelength tuning is required to speed up the data acquisition from the entire SMOUT array. The experimental setup for characterizing the optical reflection spectrum of the SMOUT elements is shown in Fig. 4. The light sources consist of a 765 – 815 nm CW tunable laser for the characterization and a halogen lamp for illuminating the measured element. To simulate its actual working conditions, the SMOUT array is immersed in water with the backside facing toward the incident light beam. The laser and the collimated white light beams are combined with a dichroic mirror and focused onto the center region of a SMOUT element through the glass substrate by a 10× objective lens. Reflected light from the SMOUT element is coupled into a single mode (SM) fiber coupler and received by a photo detector. Part of the reflected white light is split by a beam splitter and projected onto a CCD camera for monitoring the location of the SMOUT element under testing. The output of the photodetector is amplified and recorded by the data acquisition card (DAQ), which is synchronized by the trigger signal from the tunable laser. The recorded time-domain signal during one sweeping cycle of the tunable laser is converted into the reflection spectrum (Fig. 5(a)). To characterize the optical uniformity, the interrogation laser spot is scanned over the SMOUT array substrate with a motorized two-axis stage. The scanning step is 0.98 mm × 0.98 mm (every 7 elements) and the scanning range is 40 × 40 steps (~ 4 cm × 4 cm). Figure 5(b) and 5(c) show a mapped image and a histogram of the distribution of the ORW, respectively. More than 94% of the elements (except those at the corners) have an ORW between 802 nm and 812 nm. It should be noted that the measured ORWs are somewhat larger than the design value (e.g., 785 nm), possibly due to the slight deviation of the deposition thickness of the sacrificial layer and the small bending of the top DBR diaphragm.

3.2. Acoustic response

Upon the illumination of acoustic pressure, the top reflector (diaphragm) of the F-P cavity, as a mechanical resonator, will vibrate under flexural mode. The acoustic response of the SMOUT element is determined by the material properties (e.g., Young’s modulus, Poisson’s ratio, density, viscosity, etc.), lateral dimensions and thickness, and the shape of the diaphragm [21]. Typically, the SMOUT elements have a narrow acoustic bandwidth, which is due to the high Young’s modulus and extremely low or zero viscosity of the silicon oxide and nitride films. The coating of a viscoelastic polymeric Parylene layer helps to enhance the damping to the diaphragm vibration and broaden acoustic bandwidth of the SMOUT elements. When immersed in water (for ultrasound coupling), the center frequency of a diaphragm drops, and the frequency bandwidth of the resonance becomes wider due to increased inertia and viscous damping [22]. PA testing is conducted to characterize the acoustic response of the SMOUT elements. A 1-mm-long pencil lead with a 0.5-mm diameter serves as the target. The excitation light source is a 532-nm pulse laser with output pulse energy of 144 µJ, a pulse duration of ~3 ns, and a 1-kHz repetition rate. Triggered by a function generator, the excitation laser beam is coupled into a multi-mode optical fiber (200 µm core, 0.22 NA) through a 10× objective lens mounted on a fiber launch stage. The excitation laser pulse is delivered onto the target at a distance of 7 mm from the optical fiber tip. The generated PA signals propagate upward to the SMOUT array. The optical output of the interrogated SMOUT element is recorded with the same setup shown in Fig. 4. As shown in Fig. 5(a), the SMOUT elements have an ORW around 805 nm, therefore the interrogation wavelength is set at 808 nm. Figures 6(a) and 6(b) show the received PA signal and its acoustic frequency spectrum after FFT (Fast Fourier Transformation), respectively. The center frequency and the 6-dB bandwidth are around 5 MHz and 2 MHz, respectively. The relatively narrow
Fig. 4. Experimental setup for optical and acoustic characterization of the SMOUT array.

Fig. 5. (a) Representative optical reflection spectrum of the SMOUT elements. (b) Optical resonance wavelength mapping in the 40 mm × 40 mm center region of the SMOUT array. (c) Histogram showing the distribution of the optical resonance wavelength (the vertical axis indicates the percentage of the elements).
bandwidth could be partially due to the acoustic resonance within the pencil lead. To better evaluate the acoustic bandwidth, a second PA testing is performed by using a piece of black electrical tape as a wideband PA target. Figures 6(c) and 6(d) show the received PA signal and its acoustic frequency spectrum after FFT, respectively. The center frequency and the 6dB bandwidth are around 3.5 MHz and 5 MHz, respectively. Such acoustic properties are comparable with those of other micromachined transducer arrays whose elements have hollow-cavity structure, such as CMUT. Based on the existing work, the typical acoustic bandwidth of CMUT is from 75% [23] to more than 100% [7,24] of the center frequency, which is close to the acoustic bandwidth of the proposed SMOUT array (5MHz bandwidth with 5MHz center frequency).

Fig. 6. (a) and (b) Representative PA signal received from the pencil lead target and its acoustic frequency spectrum after FFT. (c) and (d) Representative PA signal received from the black tape target and its acoustic frequency spectrum after FFT.

3.3. Noise equivalent pressure (NEP)

Sensitivity of an optical interferometric ultrasound sensor is determined by both the optical and mechanical sensitivity. The optical sensitivity is mainly affected by the finesse of the interferometer, and the mechanical sensitivity mostly depends on the stiffness and acoustic resonance behavior of the top diaphragm. The overall sensitivity of the SMOUT element and the optical read-out setup can be evaluated by the noise equivalent pressure (NEP). NEP is defined as the detected acoustic pressure level when the signal to noise ratio (SNR) reaches one in the low-frequency limit (when acoustic wavelength is much larger than F-P cavity length), which can be calculated as $\text{NEP} = P / \text{SNR}$, where $P$ is the acoustic pressure (Pa) received by the sensor.

The experimental setup used for NEP characterization is similar with that shown in Fig. 4. The SMOUT array is placed onto a holder with the device side facing down toward a lead zirconate titanate (PZT) plate serving as the acoustic source, which has a thickness of 0.4 mm and a thickness-mode resonance frequency of 5 MHz. Acoustic pressure generated in the center region
of the SMOUT array by the PZT plate is measured by a needle hydrophone. The averaged signal from five different locations is 575 mV (Fig. 7(a)). With the hydrophone responsivity of 60 mV/MPa and 50-dB amplification, the acoustic pressure is estimated to be 30.2 kPa. Twenty-five elements within the 4 cm × 4 cm center region of the SMOUT array are measured, and the averaged signal amplitude is 7.3 V (Fig. 7(b)). The peak-to-peak noise amplitude is measured at the time range between trigger and ultrasound signal pulse, which is mainly from the CW laser, photo detector and the amplifier. The noise amplitude is determined to be 41.6 mV and 5.0 mV with no and 16 signal averaging, respectively. The averaged NEP is calculated as 20.7 Pa (with 16 signal averaging) or 172.5 Pa (without signal averaging) over a bandwidth of 10 MHz. The NEP (with 16 signal averaging) of the 25 elements ranges between 19.5 Pa and 22.5 Pa (Fig. 7(c)), which means that the sensitivity of the elements is quite uniform across the entire SMOUT array. The NEP of the SMOUT elements is also much lower than those of piezoelectric needle hydrophones (∼6 KPa) [25]. The finesse of the SMOUT elements, as implied by the small slope of reflection spectrum (Fig. 5(a)), is low possibly because of the bent top DBR diaphragm. However, the SMOUT element still shows high acoustic sensitivity due to the flexural-mode vibration of the top diaphragm. The hard dielectric materials which constitute the top diaphragm make the SMOUT sensor stable against high level of laser power; thus, the NEP of the SMOUT sensor can be further improved with a stronger interrogation light source.

\[ \text{3.4. Temperature and temporal stability} \]

Good device stability ensures consistent sensing performance for an extended period of operation. For PA imaging, long-time water immersion of the transducer array is required for acoustic coupling. Also, the ambient temperature could fluctuate with different operation parameters. Therefore, drift in the ORW of the SMOUT elements due to water immersion and ambient temperature change could be a potential issue. To evaluate the stability of the SMOUT elements, two sets of testing are conducted. Firstly, the ORWs of three SMOUT elements are acquired.
under an ambient temperature ranging from 25°C to 55°C with a step of 5°C. The ambient temperature of the SMOUT array was controlled with a hotplate and monitored with an infrared thermometer. The mean values of the ORW vs. the ambient temperature is shown in Fig. 8(a). The standard deviation (STD) of the ORW is 0.18 nm, which is only 0.022% of the mean value (Relative STD). Secondly, the ORWs of three SMOUT elements immersed in water under room temperature are continuously monitored for one week. The mean values of the optical resonance wavelengths in each day are plotted in Fig. 8(b), which show a STD of 0.33 nm and a relative STD of 0.04%. Error bars are included in both plots to show the deviation of the ORW.

Fig. 8. Average value of ORWs of three tested SMOUT elements (a) at different ambient temperature and (b) immersed in water for seven consecutive days.

4. PACT experiment

4.1. Imaging setup

To demonstrate the imaging capability of the SMOUT array, PACT experiments are conducted to evaluate the field of view (FOV), imaging depth, and contrast-to-noise ratio (CNR). The PA data acquisition setup is similar with the one shown in Fig. 3. The interrogation wavelength is set to 808 nm for the entire imaging process, and the laser spot is aimed at the center part of each SMOUT element. The PA excitation condition is the same as that for the acoustic response characterization. The generated PA signals propagate upward and are detected by the SMOUT array. The imaging targets consist of three dot-shape pencil leads (1 mm long, 0.5 mm in diameter) fixed sparsely below the SMOUT array (Fig. 9(a)). The distance between two adjacent pencil leads is 10 mm along X and Y axes (lateral plane) and 5 mm along Z axes (depth), as shown in Fig. 9(b). The SMOUT array is located at x-y plane (z = 0), and the three targets are located 20 mm, 25 mm, and 30 mm below the SMOUT array, respectively.

4.2. Data processing

The interrogation laser beam is scanned in two orthogonal (X and Y) directions to read out the PA signal from each SMOUT element. A 3D matrix of temporal pressure signals $P_0(x, y, t)$ is recorded by the DAQ card and saved in the PC. The raw signal matrix is band-pass filtered to remove the DC components and high-frequency noise. Hilbert transform is applied to get an enveloped signal matrix $P(x, y, t)$. The enveloped signal matrix $P(x, y, t)$ is processed by the space-domain synthetic aperture focusing tomography (SAFT) method [26] and reconstructed as the image matrix $I(x, y, z)$ of the target.
4.3. Imaging experiments and results

Two PA imaging experiments are conducted: 2D B-mode imaging with one pencil lead (Target #2 illustrated in Fig. 8) and 3D volumetric imaging with the three pencil leads as the target. For the 2D B-mode imaging, the interrogation laser beam acquires the PA signals from the 1D array of SMOUT elements (along the y axis) right above the target. The scanning range and interval are set to be 3 cm and 140 µm (every element), respectively, corresponding to a total number of ~200 elements. The scanning takes around 5 minutes with 128 signal averaging. Figure 10(a) shows the reconstructed B-mode image after applying SAFT and thresholding, which has an overall CNR of 66.4 dB. The location of the pencil-lead target in the reconstructed B-mode image matches well with its actual location in the imaging setup.

For the 3D volumetric imaging, due to the limited optical fluence of the excitation laser beam, each pencil lead is illuminated and imaged individually. Three images corresponding to the three targets are reconstructed separately and combined to form the final image. Orthogonal scanning of the interrogation laser spot on the SMOUT array is performed to read out the signals from
55×55 elements to cover a FOV of 3 cm × 3 cm. The scanning step is set to be 0.56 mm (every five elements) to accelerate data collection and image reconstruction. The total scanning time is around 80 minutes with 128 signal averaging. Figure 10(b) shows the reconstructed PA image of the three targets after applying SAFT and thresholding, which shows a large well-resolved FOV (3 cm × 3 cm) and imaging depth up to 3 cm. The CNR of the reconstructed 3D image is 60.6 dB, which is slightly lower than the 2D image of the one pencil lead target because of the lower spatial sampling rate and lower element count.

5. Conclusion

In conclusion, a new highly uniform and sensitive surface-micromachined optical ultrasound transducer (SMOUT) array has been designed, fabricated, and characterized with both optical and acoustic measurements. The experimental results show that the SMOUT array can have high element density, excellent uniformity, superior acoustic sensitivity, and good temperature and temporal stability. As a result, it requires no interrogation wavelength tuning or incident angle adjustment for different elements, which could drastically improve the data acquisition speed and eventually could enable parallel data acquisition from the entire array. In addition, it can be made optically transparent over a wide range of optical wavelengths to facilitate the delivery of laser pulses for PA excitation. These features make the SMOUT array a promising solution for high-speed 3D PACT applications. One potential disadvantage of the proposed method is that in order to read out the ultrasound signal from a SMOUT element, the interrogation light beam needs to be precisely focused at the center of the element. This would be challenging for parallel data acquisition where multiple elements are interrogated simultaneously. This issue could be addressed by integrating a matching micro lens array onto the back of the SMOUT array substrate. The focal point of one micro lens is aligned to the center of one SMOUT element. A collimated light beam and a 2D imaging camera can be used for parallel readout from all illuminated elements. In the future, integration of the SMOUT array with a micro lens array will be conducted to further improve the optical and acoustic performances and usability of the proposed technology.

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Data availability. Data underlying the results presented in this work are not publicly available at this moment but may be obtained from the authors upon reasonable request.

References


