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Characterization of a Broadband All-Optical Ultrasound Transducer—From Optical and Acoustical Properties to Imaging

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Abstract

A broadband all-optical ultrasound transducer has been designed, fabricated, and evaluated for high-frequency ultrasound imaging. The device consists of a 2-D gold nanostructure imprinted on top of a glass substrate, followed by a 3 μm PDMS layer and a 30 nm gold layer. A laser pulse at the resonance wavelength of the gold nanostructure is focused onto the surface for ultrasound generation, while the gold nanostructure, together with the 30 nm thick gold layer and the PDMS layer in between, forms an etalon for ultrasound detection, which uses a CW laser at a wavelength far from resonance as the probing beam. The center frequency of a pulse-echo signal recorded in the far field of the transducer is 40 MHz with -6 dB bandwidth of 57 MHz. The signal to noise ratio (SNR) from a 70 μm diameter transmit element combined with a 20 μm diameter receive element probing a near perfect reflector positioned 1.5 mm from the transducer surface is more than 10 dB and has the potential to be improved by at least another 40 dB. A high-frequency ultrasound array has been emulated using multiple measurements from the transducer while mechanically scanning an imaging target. Characterization of the device's optical and acoustical properties, as well as preliminary imaging results, strongly suggest that all-optical ultrasound transducers can be used to build high-frequency arrays for real-time high-resolution ultrasound imaging.

I. Introduction

The use of high frequency (>30 MHz) ultrasound for high-resolution imaging has greatly increased during the past decade, with various applications in ophthalmology [1]-[4], dermatology [5]-[7], intravascular imaging (IVUS) [8] and small animal imaging [9]-[11]. All these applications heavily rely on the performance of ultrasound transducers, which play a crucial role in determining the resolution and overall quality of the images obtained. Currently,

all commercially available high-frequency ultrasound systems use mechanical scanning of a single element piezoelectric transducer, where the low f/λ numbers of these transducers severely restrict the depth of field, making it very difficult to make high-quality images of a 3-dimensional (3-D) target at real-time imaging rates. Therefore, 2-D arrays are highly desired because they can be steered and dynamically focused in the image plane and can achieve higher frame rates.

Despite mature technology available to build low-frequency arrays, as well as significant progress by several investigators [12]-[18] on higher frequency devices, piezoelectric array systems are not yet available for routine use at high frequencies. Major problems affecting high-frequency piezoelectric arrays include dicing piezoceramics to micron scale elements, electrical connections, crosstalk between elements, as well as lack of quality high-frequency materials and electronics. Recently, the capacitive micro-machined ultrasound transducer (CMUT) has been developed as an alternative [19], [20] and has proved effective in fabricating high-frequency arrays up to 45 MHz [21]. However, these devices require integrated front-end electronics for each independent array element. For applications where high-frequency and broadband arrays with large element counts are required in simple packages, a new transduction technology is still greatly needed.

An attractive alternative to conventional piezoelectric technology is an optoacoustic array relying on optical generation and detection of ultrasound. Two laser beams are used as input/output instead of electronic signals, one for ultrasound generation and the other for ultrasound detection. The most significant advantage over piezoelectricity is that the size and spacing of each transmit/receive array element is defined by the focal spot of a laser beam and can be easily reduced to several microns using conventional optics, which is suitable for synthetic aperture imaging at frequencies over 100 MHz. Also, an array can be easily formed and is conveniently configurable by splitting the primary laser beam and focusing the resultant secondary beams onto a programmable array of spots, avoiding the trouble of dicing the transducing surface or making any electrical connections.

Optical generation of ultrasound has been studied for several decades and is possible through a variety of mechanisms. The most common and efficient mechanism is the thermoelastic effect [22]-[25], where the key component is a light-absorbing film deposited on a transparent substrate. A typical configuration focuses a laser pulse onto the film surface, and optical absorption rapidly causes a temperature rise in a localized volume in the film, where thermal expansion launches an acoustic wave into the overlying sample. The biggest advantage of the thermoelastic effect is that the acoustic pressure in the far field is proportional to the time derivative of the laser pulse [26], meaning that the center frequency and bandwidth of the generated ultrasound is determined by the incident laser pulse; thus, acoustic waves with desired properties can be achieved simply by modulating the optical input. The major drawback has always been poor conversion efficiency, which can be overcome by choosing polydimethylsiloxane (PDMS) as the film bulk material because of its high thermal expansion coefficient.

In our previous work, we have shown that the optoacoustic conversion efficiency can be increased by more than 30 dB using a spin-cast film consisting of a mixture of carbon black and PDMS instead of a metallic one [26]-[29]. A 2-D gold nanostructure has also been developed as optical absorber [30], not only improving bandwidth, but also offering a convenient method for integration with optoacoustic receivers. These devices can easily operate at above 50 MHz and can generate acoustic surface pressure exceeding 100 MPa, making them realistic tools for real-time biomedical imaging applications.

Optical detection of ultrasound has also been a subject of interest for several decades. The most effective method uses an etalon structure [31]-[38], also known as a Fabry-Perot interferometer. It is an optical device that consists of a transparent slab placed between 2 partially reflecting mirrors. Light incident from an external source undergoes multiple beam interference in the etalon and produces a reflected signal intensity that depends on the optical path length and wavelength. A resonance condition occurs when the optical path of 2-way travel in the etalon bulk equals an integer multiple of the wavelength. A sharp drop in the reflected intensity is observed at or near resonance, creating a mechanism for sensitive ultrasound detection.

The strain associated with an ultrasound wave passing through the etalon modulates its thickness and, therefore, changes the resonance condition. This in turn changes the reflected optical intensity; thus, the acoustic pressure can be measured by measuring the intensity change of the reflected signal. Etalon sensitivity has been shown to be comparable to a piezoelectric transducer of equivalent size [38], [39]. Etalon bandwidth increases with a thinner polymer layer given a certain bulk material. Thus, with current polymer fabrication techniques, very thin films from 3 μm to 10 μm can be easily produced, leading to etalon bandwidths of more than 100 MHz.

Both optoacoustic transmitters and receivers have been significantly improved during the past decade and are mature enough to be integrated into a single device for practical imaging applications. Particularly, optoacoustic transmitters using a 2-D gold nanostructure as the optical absorber provide a convenient configuration for pulse-echo operation [30]. We have reported the design and preliminary evaluations of the first all-optical ultrasound transducer for high-frequency array applications [40]. In this paper, we explore details of the design and fabrication of the integrated device, as well as characterization of the optical and acoustical properties. Images obtained from a 1-D synthetic aperture formed by mechanically scanning the imaging target also are presented.

II. Fabrication, Structure, and Mechanism

The most crucial part of the device is the 2-D gold nanostructure, and the first step is to fabricate a SiO_2 mold using laser interference lithography [41]-[43] and nanoimprint lithography [44]-[46]. The mold consists of a 2-D array of SiO_2 pillars with thickness of 200 nm and period of 220 nm on a Si wafer. Each pillar is 128 nm by 110 nm in cross section. It is then imprinted on a 200 nm thick photoresist (mr-I 8020, Micro Resist Technology, Berlin, Germany) layer, which was spin-coated onto a glass substrate. After separation, a 200 nm thick photoresist layer with 2-D arrangements of air holes spaced every 220 nm is obtained, where each air hole is 200 nm deep and 128 nm by 110 nm in cross section. A 20 nm layer of gold is then deposited on top of the structure using an electron beam evaporator, not only leaving a 2-D array of gold nanoparticles at the bottom of the air holes, but also covering the top of the photoresist layer.

Fig. 1(a) shows a scanning electron microscope (SEM) picture of the top view of the gold nanostructure, and Fig. 1(b) shows a sketch of the side view. A mixture of PDMS (Sylgard 184, Dow Corning, Midland, MI) and D4 (Octamethylcyclotetrasiloxane, Gelest Inc., Philadelphia, PA) is then spin-coated on top of the gold nanostructure, where the thickness is about 2.7 μm . A 30 nm thick gold layer is then deposited using a second gold evaporation. As a final step, an additional 0.5 μm thick PDMS layer is spin cast over the entire device for protection. A sketch of the side view of the entire structure is shown in Fig. 1(c).

Optical transmission measurements were done using a Nikon TE300 Eclipse inverted microscope (Nikon Inc., Melville, NY) (20 $\times$ objective; NA = 0.44) with transmitted light coupled into an Ocean Optics SD2000 fiber (Ocean Optics, Dunedin, FL) coupled spectrometer using an achromatic lens. Fig. 2 shows how optical transmission varies with wavelength.

The gold nanostructure serves as efficient optical absorber for optoacoustic transmission because surface plasmons localized around the particles strongly absorb light at a resonant wavelength that depends on the size, shape, and local dielectric environment of the gold nanoparticles [47]-[51]. It has been previously shown that the resonance wavelength for the structure used in our device is generally in the range of 700 nm to 800 nm [30], [45], [52]. Optical reflection from the substrate side of the gold nanostructure has been measured at wavelengths of 780 nm (51%) and 1550 nm (75%). This means that optical extinction, mostly optical absorption, is about 30% at 780 nm and about 10% at 1550 nm, consistent with the previously demonstrated resonance wavelength.

Due to the large percentage of optical absorption around the resonance wavelength, a pulsed laser beam at a wavelength of 780 nm is used for ultrasound generation. The laser pulse is focused onto the gold nanostructure, where the gold nanoparticles absorb a large portion of the optical energy. Heat absorbed is then rapidly transferred to PDMS surrounding the gold nanostructure, causing a temperature rise at the focal spot of the laser. Therefore, thermal expansion in the PDMS layer will launch an acoustic wave, which easily penetrates the thin gold layer and the additional PDMS layer into the overlying samples. High-frequency ultrasound is guaranteed using a laser pulse with duration of several nanoseconds.

An etalon is formed for optical detection of ultrasound with the gold nanostructure, the additional 30 nm gold layer, and the 2.7 μm PDMS layer in between. A high quality factor is achieved if both the gold layer and the gold nanostructure are highly reflective. Optical reflection from the polymer side of the gold nanostructure is measured and fluctuates between 85% and 95% in the wavelength range of 1440 nm to 1590 nm, far from the resonance wavelength, as shown in Fig. 3. Optical reflection from a 30 nm thick gold layer in the same wavelength range is about 90%. Thus, taking both reflection coefficients to be 90%, the quality factor is estimated to be [53]

$$Q = \frac{2nd}{\lambda} \cdot \frac{\pi \sqrt{R}}{1 - R} \approx 150. \quad (1)$$

Here $n = 1.43$ is the refractive index of PDMS; $d = 2.7 \mu\text{m}$ is the thickness of the PDMS layer; $\lambda = 1520 \text{ nm}$ is the resonance wavelength; and $R = 0.9$ is the optical reflection.

As stated in the introduction, optical reflection from a Fabry-Perot etalon shows sharp resonance when the optical path of 2-way travel in the etalon bulk equals an integer multiple of the wavelength ($2nd = m\lambda$, where m is an integer). Fig. 4 shows the experimentally measured optical reflection resonance, which occurs at 1520 nm with FWHM of 10.5 nm. Theoretically, the FWHM of the resonance can be determined using [53]

$$\Delta\lambda_{1/2} = \frac{\lambda}{Q} \approx 10 \text{ nm}. \quad (2)$$

The theoretical value agrees well with experimental results.

The frequency response of the etalon structure is characterized using a high-frequency piezoelectric transducer. First, a pulse-echo signal of the piezoelectric transducer reflected from a glass substrate is recorded. Then the glass substrate is replaced by the etalon, where the signal from the piezoelectric transducer is also recorded. The frequency response of the etalon is derived by dividing the spectrum of the etalon signal by the square root of the spectrum of the transducer pulse echo. Both spectra, together with the derived frequency response of the etalon, are shown in Fig. 5. Clearly, the etalon is suitable for ultrasound detection exceeding 50 MHz.

In summary, the working mechanism of the all-optical ultrasound transducer is defined by the unique optical properties of the gold nanostructure. Large optical absorption near the resonance wavelength of the structure enables the gold nanostructure and the PDMS layer to operate as an optoacoustic transmitter using a pulsed laser beam. Meanwhile, very high optical reflection at wavelengths far from resonance enables the gold nanostructure, the gold layer, and the interleaved PDMS layer to form an etalon for ultrasound detection, where a continuous wave laser is used as the optical probing beam.

III. Ultrasound Pulse-Echo Experiment

As a preliminary evaluation of the all-optical ultrasound transducer, a simple ultrasound pulse-echo experiment is performed using the setup shown in Fig. 6(a). The integrated optoacoustic transducer is mounted at the bottom of a water tank. The pulsed laser excitation source is a commercial high-energy solid state laser with tunable wavelength (Surelite, with OPO Plus, Continuum, Inc., Santa Clara, CA), which produces a 5 ns laser pulse when tuned to the wavelength of 780 nm, roughly the resonance wavelength of the gold nanostructure. The beam is coupled into a multimode fiber with core size of 50 μm , output through a collimator with focal length of 15 mm and a convex lens with focal length of 20 mm, and reflected from a dielectric mirror before it is focused onto a 70 μm spot on the gold nanostructure with energy of about 1 μJ /pulse (fluence of 26 mJ/cm^2). The dielectric mirror is designed to be highly reflective at wavelengths of 750 nm to 850 nm and highly transmittable at wavelengths of 1300 nm and higher. A 70 μm spot is the minimum diameter for our current setup.

The generated acoustic waves are launched into the water and reflected back from a glass slide aligned parallel to the etalon surface and placed about 1.5 mm away. To detect the pulse-echo ultrasound signal, a CW laser beam at 1517 nm with power of 4 mW is used. It travels through a polarizing beam splitter and a quarter-wave plate before being focused onto a concentric 20 μm spot through the dielectric mirror. A diagram of the transmit/receive elements is shown in Fig. 6(b). A smaller detection spot size is used to maximize the angular response of the equivalent array element for imaging applications. Reflected light, which has been modulated by the pulse-echo ultrasound waves, is collected using an amplified InGaAs photodiode. The signal is amplified by 30 dB before data capture.

Fig. 7 shows the pulse-echo signal with 1000 averages, and the upper right inset shows a single-shot acquisition of the pulse-echo signal. The signal to noise ratio (SNR) is measured to be more than 10 dB for this experiment in which only a very small fraction of the transmitted acoustic power is captured by the 20 μm diameter receive aperture. According to our previous work [30], the acoustic pressure is about 500 kPa at a distance of 10 mm away from the device surface with input optical power of 25 μJ . Because the acoustic pressure is proportional to the input optical power over distance [26], and assuming that the glass substrate is a perfect acoustic reflector, we conclude that the acoustic pressure echoed back at the device surface is about 67 kPa.

The noise equivalent pressure of the etalon in this setup has also been measured using a piezoelectric transducer. First, the piezoelectric transducer is calibrated using a fiber-optic hydrophone [54], which showed an acoustic pressure of 2.6 MPa at the focus of the transducer. Then the etalon replaces the hydrophone at the focus to detect the ultrasound signals generated by the piezoelectric transducer, which shows an SNR of 42.7 dB, yielding a noise equivalent pressure of about 20 kPa over a 50 MHz band-width for the etalon. Combining the estimated pulse-echo acoustic pressure of about 67 kPa and the etalon noise equivalent pressure of roughly 20 kPa, the expected SNR of a single-shot signal is estimated to be 10.5 dB, consistent with experimental results.

The spectrum of the pulse-echo signal (solid curve), averaged 1000 times, is shown in Fig. 8. The center frequency is 40 MHz, with -6 dB bandwidth of 57 MHz. The theoretical spectrum is also shown for comparison. It is derived by taking the spectrum of the time derivative of the laser pulse, multiplying by the etalon frequency response, then taking into account the attenuation in water (0.0022 dB/cm/MHz² [55]). Apparently, the 2 curves are in reasonably good agreement with each other. As noted above, the temporal profile of the generated ultrasound is proportional to the derivative of the input optical pulse. Therefore, the center frequency and bandwidth of the pulse-echo signal is mainly determined by the incident laser pulse, as well as the frequency response of the etalon.

The most straightforward approach to achieve higher center frequency and broader bandwidth is to use a shorter pulse. Also, the frequency response of the etalon is directly determined by the thickness of the polymer bulk [35] and can be greatly improved by further reducing the overall thickness of the structure. When a 2 ns laser pulse is used instead of a 5 ns one, combined with an etalon thinner than 1.5 μm , the bandwidth is expected to be enhanced to more than 100 MHz.

The surface acoustic pressure at the moment of generation can be calculated based on the value of the far field acoustic pressure using the following standard Rayleigh-Sommerfeld diffraction formulas [53], [56]:

$$P_{\text{far}}(\vec{r}', t) = \iint_{\Sigma} \frac{1}{2\pi c|\vec{r}' - \vec{r}|} \frac{dP_{\text{near}}(\vec{r}, t - \frac{|\vec{r}' - \vec{r}|}{c})}{dt} dS. \quad (3)$$

Here c is the acoustic velocity, $\vec{r}'$ is in the far field, $\vec{r}$ is at the surface of the optoacoustic transmission element, P_{far} is the far field acoustic pressure, and P_{near} is the surface acoustic pressure. Assuming that the element size is much smaller than the distance, which is normally the case, the above formula can be rewritten as

$$P_{\text{far}}(t) = \frac{S}{2\pi cr} \frac{dP_{\text{near}}(t - \frac{r}{c})}{dt}. \quad (4)$$

On the other hand, we have previously shown that [26]

$$P_{\text{far}}(t) = \frac{1}{4\pi} \frac{3B^2\alpha_L}{\rho^2 c^2 C} \frac{1}{r} \frac{dI(t - \frac{r}{c})}{dt}. \quad (5)$$

Here I is the total optical energy incident onto the film, ρ is the density, C is the specific heat capacity, B is the bulk modulus, and α_L is the linear coefficient of thermal expansion. Comparing the above 2 equations, it is not difficult to show that P_{near} is proportional to the total optical energy and thus should follow the formula of the incident laser pulse

$$P_{\text{near}}(t) = A_{\text{near}} \exp\left(-\frac{4\ln 2 \cdot t^2}{\tau^2}\right). \quad (6)$$

Here A_{near} is the amplitude of the surface acoustic pressure and τ is the laser pulse duration. Therefore, substituting (6) into (4), the far field pressure is

$$P_{\text{far}}(t) = -\frac{8\ln 2}{\tau^2} \frac{S}{2\pi cr} A_{\text{near}} \left[\left(t - \frac{r}{c} \right) \exp \left(-\frac{4\ln 2 \left(t - \frac{r}{c} \right)^2}{\tau^2} \right) \right]. \quad (7)$$

Thus, the amplitude of the far field acoustic field A_{far} is

$$\begin{aligned} A_{\text{far}} &= \frac{8\ln 2}{\tau^2} \frac{S}{2\pi cr} A_{\text{near}} \\ &\quad \cdot \max \left[\left(t - \frac{r}{c} \right) \exp \left(-\frac{4\ln 2 \left(t - \frac{r}{c} \right)^2}{\tau^2} \right) \right] \\ &= \sqrt{\frac{8\ln 2}{e}} \frac{S}{2\pi cr \tau} A_{\text{near}}. \end{aligned} \quad (8)$$

Consequently, A_{near} can be expressed as

$$A_{\text{near}} = \sqrt{\frac{e}{8\ln 2}} \frac{2\pi cr \tau}{S} A_{\text{far}}. \quad (9)$$

In our current experiment, the optoacoustic transmission element is about $70 \mu\text{m}$, the pulse duration is 5 ns, the distance is 3 mm, and the far field acoustic pressure is 67 kPa. Therefore, it is not difficult to calculate that the surface acoustic pressure is estimated to be 1.93 MPa. High surface pressures exceeding 1 MPa enable good SNR and high-quality imaging even though the transmitter emits a far-field, diverging radiation pattern where intensity falls off as the inverse square of distance.

As a side note, the optoacoustic transduction efficiency η can be conveniently calculated using the surface pressure

$$\eta = \frac{E_{\text{acoustic}}}{E_{\text{in}}} = \frac{S \int \frac{P_{\text{near}}^2(t)}{Z} dt}{E_{\text{in}}} = \sqrt{\frac{\pi}{8\ln 2}} \frac{S \tau A_{\text{near}}^2}{Z E_{\text{in}}}. \quad (10)$$

Here E_{in} is the input optical energy, and Z is the acoustic impedance. In our experiment, the transduction efficiency is about 0.037%. Note that because A_{near} is proportional to E_{in} , and is inversely proportional to S , it is easy to conclude that the transduction efficiency is proportional to E_{in}/S . Therefore, the efficiency can be greatly enhanced by increasing input optical energy and reducing the element spot size.

Both the surface and the far-field acoustic pressures increase linearly with the optical energy absorbed by the gold nanostructure, thus the SNR can be further improved with higher input laser energy. The ultimate acoustic pressure available is determined by the thermal damage threshold of the structure, measured to be $25 \mu\text{J/pulse}$ delivered to a spot size of $25 \mu\text{m}$ (fluence of 5.1 J/cm^2) [30], that is, the fluence is a factor of about $200\times$ higher than that used in our experiment. If the transmission element remains at $70 \mu\text{m}$, then at this energy level, the far-field acoustic pressure could reach 1.6 MPa at 3 mm away from surface, while the surface pressure can be as high as 50 MPa. For high-frequency applications, an element size of $20 \mu\text{m}$ or less is highly desired, which limits the maximum energy input to $15 \mu\text{J/pulse}$, yielding a far-field acoustic pressure of 1 MPa at 3 mm away from the surface and a surface pressure of 350 MPa with enhanced transduction efficiency of 7%. Thus, even with reduced element

size of $20\ \mu\text{m}$, the generated acoustic pressure can be easily enhanced by 20 dB without damaging the device.

On the other hand, the noise equivalent pressure of the etalon can also be further improved. The sensitivity of the etalon can be derived by characterizing 2 separate aspects: how the reflected optical intensity changes with cavity length, and how cavity length changes with incident acoustic pressure. According to previous studies, the reflected optical intensity change is related to the cavity length change [37]:

$$\delta I_R = I_0 \frac{3\sqrt{3}}{2} \frac{Fn}{\lambda} \delta L = I_0 \frac{3\sqrt{3}}{4} Q \frac{\delta L}{L}. \quad (11)$$

Here δI_R is the change in reflected optical intensity, I_0 is the optical intensity of the etalon probing beam, F is the finesse of the etalon, n is the refractive index of the etalon bulk material, λ is the wavelength of the etalon probing beam, δL is the change in cavity length, L is the cavity length, and Q is the quality factor of the etalon.

In simplest form, the change in cavity length induced by incident acoustic pressure is

$$\delta L = \frac{L \cdot p}{E_Y}. \quad (12)$$

Here p is the acoustic pressure, and E_Y is the Young's modulus of the etalon bulk polymer material. Combining (12) and (11) yields

$$\delta I_R = \frac{3\sqrt{3}}{4} I_0 \cdot Q \cdot \frac{1}{E_Y} \cdot p. \quad (13)$$

Assuming that the optical detector sensitivity is I_s , the noise equivalent pressure p_{NEP} can be expressed as

$$p_{\text{NEP}} = \frac{4}{3\sqrt{3}} \frac{I_s \cdot E_Y}{I_0 \cdot Q}. \quad (14)$$

Therefore, the noise equivalent pressure of the etalon is proportional to the sensitivity of the photodetector and the Young's modulus of the etalon bulk material and is inversely proportional to the intensity of the etalon probing beam and the quality factor of the etalon. The most straightforward methods to reduce the noise equivalent pressure are to increase the probing beam intensity and increase photodetector sensitivity. Currently, the probing beam power is 4 mW, focused onto a spot size of $20\ \mu\text{m}$, yielding a fluence of $1.27\ \text{kW}/\text{cm}^2$. This is much lower than the damage threshold of gold films, which generally exceeds $1\ \text{MW}/\text{cm}^2$ [57], meaning that the probing optical intensity can be increased by at least 10 to 100 times without damaging the etalon. Of course, much higher optical energy should be avoided because part of the energy is able to transmit through the etalon and lead to an undesirable temperature rise at the boundary of the etalon and overlying samples. The quality factor can be increased by improving the optical reflection of the gold layer as well as the gold nanostructure. When the reflection is improved from 90% to 95%, the quality factor can be enhanced by a factor of 2. However, making thicker etalons to improve the quality factor is not desired because thinner etalons lead to a broader detection bandwidth. Assuming a 5 times higher probing laser

intensity, which is a very conservative estimate, combined with a 2 times higher quality factor, we conclude that the noise equivalent pressure can be reduced by at least 20 dB. Because the generated acoustic pressure can also be improved by another 20 dB, the overall SNR of a single shot pulse-echo signal from a single 20 μm transmit/receive element can easily reach as high as 50 dB.

IV. Ultrasound Imaging

To further evaluate this all-optical ultrasound transducer, its ultrasound imaging capabilities must be demonstrated. An imaging object replaces the glass slide in the setup shown in Fig. 6 while everything else remains the same, and a 1-D synthetic aperture is formed by mechanically scanning the imaging object. First, a 70 μm diameter metal wire is used as the imaging target. A 2 mm long imaging aperture is scanned in 20 μm steps. At each position, the signal is averaged 100 times before recording. Fig. 9(a) shows a wavefield plot of the detected acoustic field. Band-pass filtering (25 MHz to 85 MHz) and demodulation are applied, and then beam forming according to a simple synthetic aperture focusing technique (SAFT) is performed to reconstruct the image, shown in Fig. 9(b).

Waves detected by the transducer are reflected from both front and back edges of the wire. Because metal is such a good acoustic reflector, most energy is reflected back from the front edge, while the signal from the back edge is quite small. As a result, 2 curves are observed in the wavefield plot, with the first much stronger than the second. These features are captured in the reconstructed image, where the front edge is much brighter than the back. For comparison, the metal wire is replaced by a 70 μm diameter human hair, whose acoustic reflection is much lower than metal. The wavefield plot of the detected acoustic field is shown in Fig. 9(c) and the reconstructed image in Fig. 9(d). Clearly, both front and back edges are well represented in the wavefield and reconstructed image.

To determine the imaging resolution of the current system, a 25 μm diameter metal wire is chosen as the imaging object, while the scanning distance remains 2 mm with 20 μm step separation. The reconstructed image is shown in Fig. 10. The -6 dB axial resolution is determined to be 19 μm , consistent with the bandwidth of the pulse. This also confirms that only the echo from the wire front edge contributes to the image because the axial resolution is smaller than the actual wire diameter. The -6 dB lateral resolution is 38 μm , representing about one acoustic wavelength at the 40 MHz center frequency.

The transmit element size relative to an acoustic wavelength significantly affects the divergence of the radiation pattern [28]. A small element, especially 20 μm or less, can emit considerable energy even at angles above 45 degrees, which is highly desired for a high-frequency array system. Due to limitations in our current illumination system, the optoacoustic transmission element size is 70 μm , several times the acoustic wavelength at high frequencies. This means that recorded echo signals have lower bandwidths when the object is moved toward the end of the scan, thus limiting image quality.

To reduce transmit element size below 20 μm , the most straightforward method is to couple a pulsed laser directly with a multimode fiber having a core size smaller than 20 μm or, preferably, a single-mode fiber, which allows more convenience and flexibility for a smaller focal spot.

V. Potential for Real-Time 3-D Imaging

At this stage, the imaging target is mechanically scanned to form an equivalent 1-D synthetic aperture. However, real-time imaging applications require simultaneous detection at all elements in an array system and even simultaneous excitation in some configurations. This can

be realized by splitting both the ultrasound generation and detection beams into an array of focused spots on the surface of the device. Three main questions must be answered for this approach: 1) how to deliver an array of laser beams; 2) whether the device thickness is uniform enough for multipoint ultrasound detection; and 3) how to detect an array of reflected CW laser beams simultaneously modulated by echo acoustic waves.

For laser beam delivery, we propose to use a graded index (GRIN) fiber bundle for simultaneous illumination and detection. Typically, a fiber bundle contains several thousand individual light guides; each has a diameter of 10 to 20 μm , and the spacing is also about 10 to 20 μm . The small size of each individual light guide provides convenience to achieve smaller element size, which is critical for limiting the divergence of the radiation pattern for high-frequency imaging. Laser power should also be increased corresponding to the total number of elements used, which can be done using Erbium doped fiber amplifiers (EDFA).

The uniformity of the device is important especially for ultrasound detection, because the thickness of the structure determines the resonance wavelength of the etalon. This correspondingly determines the optimal wavelength for the detection laser beam, which must be equal to or close to the actually used wavelength at all element spots. Previously, we have shown that in an etalon with SU-8 as bulk material, the optimal wavelength is uniform over regions of several millimeters [38], which is enough to build a high-frequency array of thousands of elements. We expect to achieve similar uniformity with PDMS etalons after further refinement of our fabrication methods. Meanwhile, In-GaAs photodiode arrays are commercially available, which can be integrated into our system with appropriate modifications. Therefore, expanding a single-element optoacoustic transducer into an array system can be practically done by splitting the laser beams and focusing them onto an array of focused spots.

Another practical consideration for building an array system is to achieve SNR higher than 10 dB in our current single-element system. This requires higher generated acoustic pressure and lower etalon noise equivalent pressure. As described in previous sections, the most efficient way to generate higher acoustic pressure is to increase input pulse laser energy, and the best methods to reduce etalon noise equivalent are to increase probing laser beam intensity, improve etalon quality factor, as well as use photodetectors with higher sensitivity. The acoustic pressure has the potential to be improved by 20 dB simply by increasing the input energy from 1 $\mu\text{J}/\text{pulse}$ to above 10 $\mu\text{J}/\text{pulse}$, and it is not difficult to reduce the noise equivalent pressure by at least another 20 dB by increasing the probing laser power from 4 mW to 20 mW and the etalon quality factor from 150 to 300. Thus, the SNR of a single shot pulse-echo signal from a single element can easily exceed 50 dB without damaging the device, a value more than sufficient for a 2-D array element.

VI. Conclusions

In summary, we have designed, fabricated, and tested the first all-optical ultrasound transducer for high-frequency array applications. The pulse-echo signal from a 70 μm diameter transmit element combined with a 20 μm diameter receive element probing a near perfect reflector positioned 1.5 mm from the transducer surface is more than 10 dB, which can be further improved with larger input optical power and improved etalon sensitivity. The center frequency is 40 MHz with -6 dB bandwidth of 57 MHz, with room for improvements if shorter laser pulses and thinner etalon structures are used. High-resolution ultrasound imaging capabilities have been demonstrated using a 1-D synthetic aperture formed by mechanically scanning the imaging target, where the -6 dB lateral resolution is 38 μm . Planned next steps include optimizing the gold nanostructure for maximal optical absorption at the resonance wavelength, modifying the experimental system to achieve higher SNR, and expanding a single transducer

element to an array system for real-time imaging. We believe that these all-optical ultrasound transducers are suitable for 2-D high-frequency arrays providing real-time 3-D high-resolution imaging capability.

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Biography

Yang Hou received a B.S. degree in physics from Peking University, Beijing, China, in 2004, and an M.S.E. degree in electrical engineering from the University of Michigan, Ann Arbor, MI, in 2006. Currently he works in the Biomedical Ultrasonics Lab at the University of Michigan, where he's completing his doctoral dissertation on broadband all-optical ultrasound transducers for high-resolution imaging. His research interests include high-frequency ultrasound transducers and imaging, optoacoustic devices, and photoacoustic imaging.

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Fig. 1.

(a) SEM picture of the top view of the 2-D gold nanostructure, (b) sketch of the side view of the gold nanostructure, and (c) sketch of the side view of the all-optical transducer.

Fig. 2.
Optical transmission vs. wavelength (400 nm-1650 nm) for the gold nanostructure.

Fig. 3. Optical reflection vs. wavelength far from the resonance wavelength (1460 nm-1560 nm) for the gold nanostructure.

Fig. 4.
Optical resonance of the etalon structure.

Fig. 5. Spectrum of the etalon signal, square root of the spectrum of pulse-echo signal of the piezoelectric transducer, and the derived etalon frequency response.

Fig. 6. (a) Setup of an ultrasound pulse-echo experiment and (b) diagram of the transmit/receive elements.

Fig. 7. Pulse-echo signal; insert is a single shot recording and the primary signal represents an average of 1,000 recordings.

Fig. 8. Spectrum of pulse-echo signal (solid curve) and simulation curve (dashed curve).

Fig. 9.

(a) Wavefield and (b) reconstructed image of a 70 μm diameter metal wire; (c) wavefield and (d) reconstructed image of a 70 μm diameter hair.

Fig. 10.
The reconstructed image of a 25 μm diameter metal wire.