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Acoustic lens with variable focal length for photoacoustic microscopy

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A liquid acoustic lens with variable focal length is described for photoacoustic microscopy. This liquid lens takes advantage of the elastic and deformable lens interface to tune its focal length in a pneumatic manner. The curvature of the lens interface as well as the dependant focal length was characterized as a function of the infusion volume of the liquid. Experiments were carried out to demonstrate the zooming ability of this liquid acoustic lens. Targets embedded at different depths were photoacoustically imaged without performing mechanically axial scanning.

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Photoacoustic imaging is an emerging technique based on the combination of pulsed laser illumination and ultrasound detection, which has been demonstrated as a deep tissue penetration,\textsuperscript{1} functional,\textsuperscript{2} and high-resolution\textsuperscript{3} imaging modality for biomedical diagnosis. In terms of configuration, photoacoustic (PA) imaging can be generally categorized as photoacoustic tomography (PAT) and photoacoustic microscopy (PAM). Basically, PAT employs an illumination on the full field of view (FOV), and the photoacoustic signal from the irradiated region is collected by either a scanning transducer\textsuperscript{4,5} or a transducer array.\textsuperscript{6,7} On the other hand, either a weakly or tightly focused light beam provides a local illumination in an acoustic or optical resolution (AR\textsuperscript{8} or OR\textsuperscript{9}) PAM configuration, and a focused transducer is used to collect the signal point by point. The distinctions of configuration between PAT and PAM determine that the required laser power for PAM is much lower than that for PAT. To achieve a comparable imaging quality between PAT and PAM, a considerably small detector on tens of micrometer scale is required for PAT configuration, which is difficult to fabricate with good sensitivity.\textsuperscript{10} The difference between OR- and AR-PAM is how the light focusing is implemented. The OR-PAM, employing a tightly focused beam, offers comparably higher lateral resolution with smaller penetration owing to the light scattering, whereas AR-PAM, utilizing a weakly focused or collimated light beam, can provide deeper penetration with lower lateral resolution. Therefore, AR-PAM is more suitable for deep tissue imaging, such as imaging of sentinel lymph nodes, organs, and tumors, whereas OR-PAM is ideal for studying superficial blood vessels or capillaries.

For most AR-PAM implementations, axial scanning of the detector is required owing to the degrading resolution out of the acoustic focus. However, in an example of endoscopic imaging configuration, one issue comes up that little room could be spared for the axial scanning of the detector. Thus, AR-PAM cannot be performed optimally in this case. Here, we present an acoustic lens with variable focal length to eliminate the axial scanning of the detector, and to make it possible to miniaturize AR-PAM systems for important clinical applications such as endoscopic examination.

Acoustic lenses have been commonly used in photoacoustic imaging systems for either converging\textsuperscript{9,11} or diverging acoustic wave.\textsuperscript{12,13} However, these acoustic lenses are solid-based, which brings up the issues of high acoustic impedance and low transmission efficiency. In particular, these solid-based acoustic lenses have a fixed focal length once the lens is fabricated. Recently, the field of optofluidics emerges to enlighten the idea that solid-based optical components could be replaced by liquids or fluids with better performance and novel functionalities.\textsuperscript{14} Optical liquid or fluidic lenses have been demonstrated to have an adjustable focal length by either pneumatically control\textsuperscript{15} or flow rate management.\textsuperscript{16,17} Liquid-lensing mechanism can also be applied to acoustics. Liquid-filled acoustic lenses have been reported for underwater imaging or sonar systems in early literatures.\textsuperscript{18,19} However, the characteristics of underwater imaging systems are quite different from those in the field of medical imaging, and these differences make the techniques and results from one field not directly transferable to another.\textsuperscript{20} For example, owing to the requirement of detection range and scattering issue, the acoustic frequency for underwater imaging is low (typically, 100 kHz to 2 MHz) and the lens aperture ranges from 10 cm to 1 m, which may not be applicable and appropriate for the purpose of medical imaging. Recently, we have reported a liquid acoustic diverging lens for PAT.\textsuperscript{21} This lens can be miniaturized, cost-effective, and adaptive to most photoacoustic modalities with the proposed fabrication method. In this work, we will demonstrate a liquid acoustic converging lens with an adjustable focal length and apply it to photoacoustic microscopy.

The tunable function of this liquid acoustic lens is realized by pneumatically controlling the infusion volume of the liquid. Specifically, a syringe pump (KDS 210, KD Scientific) is used to accurately infuse or withdraw liquid into or from the lens chamber with a maximum pumping flow rate 474.8 µl/s when using B-D 5 ml syringe. In this way, the lens interface, which is an elastic membrane, can be tuned with its curvature. Therefore, the focal length of this acoustic lens can be adjusted simply by changing the infusion volume of the liquid. The relationship between the curvature of the interface and the infusion volume of the liquid is characterized and
shown in Figure 1. Assuming that the lens interface has a spherical shape and the liquid is incompressible, the infusion volume can mathematically determine the radius of the interface curvature (solid line in Figure 1) via

$$V = \pi h^2 (r - h/3),$$  \hspace{1cm} (1)

where $V$ represents the infusion volume of the liquid, and $h$ and $r$ denote the thickness of the lens and radius of the lens interface, respectively. The experimental results show a good agreement with the mathematical prediction.

Setting this liquid acoustic lens on its focusing or diverging mode can be realized by purposely choosing the working liquid inside the lens chamber, which by principle is similar as manipulating optofluidic lens. In this work, we choose silicone oil as the working liquid, which has a relative refractive index of 1.4 and density of 0.986 g/cm$^3$. The higher refractive index (relative to water) as well as the convex shape of the lens will enable an ability of focusing acoustic beam. To characterize the focal length of this liquid acoustic lens, a transducer (30 MHz, V213, Olympus) was used in its transmission mode for providing ultrasound pulses. The liquid acoustic lens was attached to the transducer head to focus the acoustic wave. A hydrophone (HGL-0200, ONDA) was used to scan laterally and axially while detecting the ultrasound pulses from the transducer.21

![Graph showing curvature of the lens interface as a function of the infusion volume of liquid. The inset photographs illustrate the lens interface tuned by the infusion volume of the liquid.](Image)

FIG. 1. Curvature of the lens interface as a function of the infusion volume of liquid. The inset photographs illustrate the lens interface tuned by the infusion volume of the liquid.

Mathematically calculated according to the conjugate relationship be image and object

$$\frac{1}{S_o} + \frac{1}{S_i} = \frac{1}{f},$$  \hspace{1cm} (2)

while considering the natural focus as the object $S_o$ and focal position as the image $S_i$. The position of the natural focus can be calculated by

$$S_o = D^2 F/4c,$$  \hspace{1cm} (3)

where $D$ is the diameter of the transducer, $F$ is the working frequency, and $c$ is the velocity of sound. The theoretical focal length of the liquid acoustic lens can be derived as a function of infusion volume using Eq. (1)

$$f = (V/\pi h^2 + h/3)/(n - 1),$$  \hspace{1cm} (4)

where $n$ is the acoustic refractive index. From Eqs. (2) to (4), the focal position can be obtained

$$\frac{1}{S_i} = \frac{1}{(V/\pi h^2 + h/3)/(n - 1)} - \frac{4c}{D^2 F},$$  \hspace{1cm} (5)

which is represented as the solid line in Figure 2. The results from the experiment shows an agreement with the theoretical analysis. The purpose of this experiment is to give a validation that the liquid acoustic lens can effectively focus acoustic wave with variable focal length, and also provides a calibration of the relationship between the focal positions and the infusion volume for the later phantom experiment.

The advantage of this liquid acoustic lens with an adjustable focal length lies in imaging objects at different depths without axial scanning of the detector. We designed and

![Graph showing position of focal spot as a function of the infusion volume of liquid.](Image)

FIG. 2. (a) Pressure distribution mappings of the transducer with and without the liquid acoustic lens. The arrow shows the propagation direction of the acoustic beam; (b) focal length of the liquid acoustic lens as a function of the infusion volume of liquid.
carried out two groups of tests to demonstrate this advantage: one using human hair as micro-scale targets and the other using excised tumors (from a mouse) as macro-scale targets (Figure 3). The targets were buried in a background with an absorption coefficient of $\mu_a = 0.007 \text{mm}^{-1}$ and a reduced scattering coefficient $\mu_s = 0.1 \text{mm}^{-1}$. A 30 MHz transducer (V213, Olympus) was used to collect the signal from the hair targets, and a 10 MHz transducer (V315, Panametrics) was employed for the tumor targets. Since the 30 MHz and 10 MHz transducers have different outer diameters (11 mm and 25 mm), they were equipped with corresponding lenses whose apertures fit their outer diameters. The targets were illuminated and scanned by laser pulses with wavelength 532 nm and pulse duration 6 ns (NL 303HT, EKSPLA) coupled with a translational stage (Zaber Tech T-LSM200A). Upon the energy deposition, an acoustic wave was generated from the local thermo-expansion and reflected by a glass slide to the ultrasound detector. The liquid acoustic lens was attached to the transducer and mounted on a hard-resin holder. The liquid was injected into the lens chamber through needle and tubing by a syringe pump (KDS 210, KD Scientific) with a flow rate of 1 $\mu\text{l/s}$. During the data acquisition, the axial distance between the targets and the transducer was fixed, and the translational stage provided a lateral scanning under each condition of infusion volume of the liquid.

For the hair imaging, three hairs were aligned in parallel and buried at different depths within the background phantom with 2.4 mm distance between each other. The photoacoustic signal was first collected without using the liquid acoustic lens and processed with maximum amplitude projection (MAP) method. The images of the hairs (left column in Figure 4) show a broadened profile regarding the thickness of the hair. This lack of image fidelity is owing to the finite size of the effective diameter of the transducer. Considering the transducer has an effective diameter of several millimeters, the hairs were imaged to have approximately the same size regarding the thickness. After the transducer was equipped with the liquid acoustic lens, the PAM system managed to obtain sharper images of the hairs (with blue outlines in Figure 4) when the focus of the lens was axially adjusted to coincide with the position of the hair. During the test, the focal length was continuously shortened by increasing the infusion volume of liquid. Thus, hair 3, which was the farthest from the transducer, was first focused and imaged under a condition of low infusion volume of 10 $\mu\text{l}$. With the infusion volume going higher and focal length getting shorter, hair 2 and hair 1 came to their focus afterwards and were sharply imaged under the infusion conditions of 30 $\mu\text{l}$ and 100 $\mu\text{l}$, respectively.

We also looked into the comparison of photoacoustic signals between using and without the liquid acoustic lens, which is shown in Figure 5. The red line denotes the PA signal collected without using the liquid acoustic lens; the black line represents the PA signal collected using the lens with a liquid infusion volume of 10 $\mu\text{l}$. The collected PA signals from hair 2 and 3 when using the lens are stronger than those collected without the lens, while the difference of the PA signals of hair 1 goes to the opposite way. When employing the liquid acoustic lens with a 10 $\mu\text{l}$ infusion volume, the focus...
was approximately located near the regions of hair 2 and 3, so a larger volume of the spherically radiating PA signals can be covered and collected by the transducer. On the other hand, the bare transducer can only take in the PA waves with Poynting vectors that fit the limited directivity of the transducer. With regard to hair 1, its position was relatively farther from the focus of the lens, so the spherical PA waves can hardly be converged to enter the transducer. Meanwhile, part of the PA signal could be reflected by the lens membrane, so the overall detected signal from hair 1 when using the lens is less than that using the bare transducer.

For the tumor imaging, two tumor samples with an approximately meniscus shape (see the photograph in Figure 6 were buried in the background with a distance of 1.5 cm between each other. The superficially buried tumor 1 was located 0.5 cm and 5 cm away from the surface and transducer, respectively. In this test, a 10 MHz transducer was used to collect the photoacoustic signal. A liquid acoustic lens was attached to the transducer with its focal length varied by adjusting the infusion volume of the liquid. During the test, the infusion volume was increased and under each condition of infusion volume, lateral scanning was performed for the data acquisition. When the infusion volume went up to 50 µl, tumor 2, which was further away from the transducer, came into focus. Under this condition, tumor 2 was imaged to have a meniscus shape that was approximately the same as its real shape, whilst tumor 1 was out of focus to have a blurred image (Figure 6(a)). As the infusion volume was increased to 250 µl leading to a shorter focal length, tumor 1, which was closer to the transducer, came into focus and was shapely imaged. Figure 6(b) shows the photograph of the tumor samples for reference. Using the liquid acoustic lens, two targets with an axial distance of 1.5 cm can be imaged without the implementation of axial scanning but alternatively by adjusting the focal length of the acoustic lens.

In summary, we have presented a liquid acoustic converging lens with an adjustable focal length for photoacoustic microscopy. This lens is cost-effective to fabricate and easy to be attached to or detached from a commercial transducer.21 Multi-function, focusing or diverging acoustic wave, can be realized for a specific application by simply

(a) Photoacoustic microscopy using liquid acoustic lens

(b) Photograph of the tumor samples

FIG. 5. Comparison of photoacoustic signals from the three hairs. The red line denotes the PA signal collected without using liquid acoustic lens; the black line represents the PA signal collected using the lens with a liquid infusion volume of 10 µl.

FIG. 6. (a) Images of the tumors captured by a 10 MHz flat-head transducer with liquid acoustic lens attached under different conditions of infusion volumes; (b) photograph of the tumor samples.
replacing the liquid inside the lens chamber. We expect this liquid acoustic lens to be well suited for various PAM applications where axial scanning of the detector is prohibited.

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10 P. Beard, Interface Focus 1, 602 (2011).


