Reflection-mode multifocal optical-resolution photoacoustic microscopy

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Abstract. Compared with single-focus optical-resolution photoacoustic microscopy (OR-PAM), multifocal OR-PAM utilizes both multifocal optical illumination and an ultrasonic array transducer, significantly increasing the imaging speed. A reflection-mode multifocal OR-PAM system based on a microlens array that provides multiple foci as well as an ultrasonic array transducer that receives the excited photoacoustic waves from all foci simultaneously is presented. Using a customized micromirror to reflect the incident laser beam to the microlens array, the multiple optical foci are aligned confocally with the focal zone of the ultrasonic array transducer. Experiments show the reflection-mode multifocal OR-PAM is capable of imaging microvessels in vivo, and it can image a $6 \times 5 \times 2.5 \, \text{mm}^3$ volume at 16 $\mu$m lateral resolution in ~2.5 min, which was limited by the signal multiplexing ratio and laser pulse repetition rate. © 2013 Society of Photo-Optical Instrumentation Engineers (SPIE) [DOI: 10.1117/1.JBO.18.3.030501]

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Photoacoustic microscopy (PAM) is a mode of photoacoustic tomography (PAT), which offers high sensitivity to optical contrast in biological tissues. Optical-resolution PAM (OR-PAM) is implemented by focusing the illumination through an objective lens; thus OR-PAM can achieve optical diffraction limited lateral resolution with maximum imaging depth. However, the original MFOR-PAM works in transmission mode, only suited for imaging thin biological samples.

For OR-PAM, high imaging speed is always desirable because it enables us to observe dynamic biological processes while mitigating motion artifacts. For most OR-PAM systems based on both a single optical focus and a single-element ultrasonic transducer in confocal configuration, mechanical raster scanning along two lateral directions provides limited volumetric imaging speeds. Although optical scanning (i.e., scanning using galvo mirrors) can improve the speed greatly, it is usually used with an unfocused or a single-axis focused ultrasonic transducer, which results in a limited field of view (FOV) or compromised signal to noise ratio (SNR). Moreover, the volumetric imaging speed for OR-PAM with either mechanical or optical scanning is eventually limited by the laser pulse repetition rate (PRR). Thus to achieve high imaging speed, lasers with high PRR (~100 kHz, or even faster) are employed. Currently, lasers with such high PRR usually lack the wavelength tunability, which is essential for imaging of some physiological functions, such as the oxygen saturation of hemoglobin ($\mathrm{SO}_2$). Moreover, high PRR translates to high average optical power, which may produce excessive tissue heating. Our lab previously reported a multifocal OR-PAM (MFOR-PAM) with ~20 times faster imaging speed (without considering the multiplexing effect) than a single-focus OR-PAM, so it relieves the demand for a high laser PRR. Besides the imaging speed improvement, the $\mathrm{SO}_2$ measurement capability makes MFOR-PAM potentially promising in clinical and preclinical applications. However, the original MFOR-PAM works in transmission mode, only suited for imaging thin biological samples. Therefore, it is necessary to develop reflection-mode MFOR-PAM.

We implemented the reflection-mode MFOR-PAM by placing a micromagnets array directly below the ultrasonic array transducer, with both immersed in water, as shown in Fig. 1. To get one-dimensional (1-D) multifocal optical illumination, the micromagnets array (250 $\mu$m pitch and 1.2 mm focal length in air, 11-1640-123-000, SUSS MicroOptics, Switzerland) was precisely cut by Biomedical-Optics LLC to 0.5 mm wide and 10 mm long. Due to the limited detection range of the ultrasonic array transducer, only 20 micromagnets were used to provide 20 focused optical illumination spots along the lateral axis. At a laser wavelength of 570 nm, Zemax (Radiant Zemax LLC) simulation shows that the water-immersed micromagnets can produce a diffraction-limited focus with a radius of ~7 $\mu$m at the waist and an effective focal length of 3.4 mm. The centers of the micromagnets array and the ultrasonic array transducer were aligned in both the elevational (x axis in Fig. 1) and lateral (y axis in Fig. 1) directions. The micromagnets array was carefully adjusted so that the optical foci of the micromagnets array and the cylindrical focus of the ultrasonic transducer overlap. To reflect the laser beam, which is incident from the elevational side, a customized microlens ($0.5 \, \text{mm wide} \times 0.5 \, \text{mm high} \times 7 \, \text{mm long}$, Biomedical-Optics LLC) was used. The microlens and the micromagnets array were aligned on their centers and glued together with optically transparent adhesive. The ultrasonic array transducer consisted of 48 elements with 100 $\mu$m pitch, and each element ($82 \, \mu$m $\times 2 \, \text{mm}$) had a central frequency of 30 MHz and a 70% receiving-only bandwidth. A tunable dye laser (Cobra, Sirah, Germany) pumped by a Q-switched Nd:YLF laser (INNOSLAB, EdgeWave, Germany) provided photoacoustic excitation at 1.35 kHz. Technical details about the ultrasound array system can be found in our previous publications.

In this reflection-mode MFOR-PAM, the PA wave was partially reflected by the micromagnets array while propagating to the transducer. Figure 1(c) shows the dimensions that were used to estimate the fractional PA wave loss, which is $w/\omega_2 = w_1/\omega_1$. Here we neglect the acoustic diffraction effect because the acoustic wavelength is much smaller than the width of the
shows a snapshot from the 3-D animation (Video 1). Therefore, hereafter we use this depth is shallower because of the PA wave. However, this shows that the black needle can still be observed 0.7 mm beneath the skin surface of the mouse leg. Compared to the maximum imaging depths of other OR-PAM systems (up to 1.2 mm), 2,6 this depth is shallower because the microlens array reflects ~60% of the PA wave. However, this maximum imaging depth is still sufficient for observations of many shallow structures.

Figure 3(a) shows an in vivo photoacoustic image of a mouse ear microvasculature acquired by the reflection-mode MFOR-PAM system. The MAP image was processed by the 2 × 2 Gaussian low pass filter. The laser wavelength was set to the isosbestic wavelength of 570 nm, where hemoglobin has high absorption. Figure 3(b) shows a snapshot from the 3-D animation (Video 1). Microvessels were clearly imaged in both the MAP image and the 3-D animation. The imaging time to acquire the data set of Fig. 3 (600 × 500 × 200 voxels) was ~2.5 min. The incident laser energy on each microlens was limited to ~62 nJ/pulse.
and the laser fluence on the ear surface was estimated to be 16 mJ/cm², lower than the ANSI limit 20 mJ/cm². With the flexibility of reflection mode, the in vivo PA image of a nude mouse brain was also acquired. Figure 4(a) shows the MAP image of a living adult mouse brain. The MAP image was processed by the 2 × 2 Gaussian low pass filter. The skin of the mouse brain was removed, but the skull was kept intact, as shown in Fig. 4(b). From Fig. 4(a), the microvessels can be clearly observed, even though the intact skull attenuated both the incident light and the PA wave. Figure 4(c) is a snapshot from the 3-D animation, showing the mouse brain from various viewing angles (Video 2). The 3-D animation shows the curved shape of the mouse brain.

In summary, we have developed a reflection-mode MFOR-PAM system using a microlens array and a linear ultrasonic array transducer. The reflection-mode MFOR-PAM was capable of acquiring volumetric PA images in vivo, with a lateral FWHM resolution of 16 μm, axial resolution of ~25 μm, and in vivo penetration depth up to 0.7 mm. Due to the combination of multiple optical foci and the ultrasonic array transducer, the lateral scanning range was reduced to only one pitch of the microlens array; thus the imaging speed was much faster than single-focus OR-PAM with mechanical scanning. Potentially the imaging speed can be further improved by eliminating the 6:1 multiplexer, and using a higher PRR laser. Optical scanning within one pitch in the lateral direction can further improve the imaging speed by several times. A more densely packed microlens array with a smaller pitch might improve the imaging speed, but this usually means a shorter working distance if we want to maintain the NA or lateral resolution. Other multifocal designs might be developed to provide both a high NA and long working distance.

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References

Fig. 4 In vivo trans-cranial photoacoustic images of a nude mouse brain with the skin removed. (a) MAP image of the brain. (b) Photograph of the mouse brain with skin removed. The dashed box is the imaging area presented in (a). (c) Snapshot from the 3-D animation showing the structure in (a). (Video 2, MPEG, 4 MB) [URL: http://dx.doi.org/10.1117/1.JBO.18.3.030501.2]