# Feasibility of photoacoustic tomography for ophthalmology

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For the eyeball composed of membrane and liquid, the contrast of ultrasound imaging is not high due to its small variance in acoustic impedance. As a new imaging modality, photoacoustic tomography combines the advantages of pure optical and ultrasonic imaging together and can provide high resolution, high contrast images. In this paper, the feasibility of photoacoustic tomography for ophthalmology is studied experimentally. A $Q$-switched Nd:YAG pulsed laser with 7-ns pulse width is used to generate photoacoustic signal of a porcine eyeball *in vitro*. The two-dimensional (2D) optical absorption image of the entire eyeball is reconstructed by time-domain spherical back projection algorithm. The imaging results agree well with the histological structure of the eyeball and show a high imaging contrast.

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Photoacoustic tomography (PAT) combines the merits of pure optical and ultrasound imaging together. It can give high contrast and high resolution images inside biological tissues[1−3]. The principle of PAT is that heterogeneous biological structures will generate acoustic waves under the irradiation of short laser pulses[4−6]. For the eyeball composed of membrane and liquid, the contrast of pure ultrasonic image is not high due to the small variance in acoustic impedance. The efficiency of photoacoustic (PA) signal generation primarily depends on the value of optical absorption coefficients inside tissue at a certain wavelength, so it is possible for PAT to perform high contrast imaging inside eyeball, overcoming the contrast disadvantage of pure ultrasound imaging. In this paper, the feasibility of PAT for ophthalmology is studied. Based on a circle measurement configuration, a $Q$-switched Nd:YAG pulsed laser with 7-ns pulse width was used to generate PA signal of a porcine eyeball *in vitro*. The imaging results agree well with the histological structure of the eyeball and achieved high imaging contrast.

When a laser pulse is much shorter than the thermal diffusion time of the biological tissue, the thermal diffusion can be neglected. The deposited optical energy will cause expansion and subsequent contraction of the optical absorption area inside tissue and then generate PA signal. To reconstruct PA image, the spatial optical absorption function $A(\vec{r})$ should be found. The acoustic pressure $p(\vec{r},t)$ at position $\vec{r}$ in tissue is given by[7]

$$\nabla^2 p(\vec{r},t) - \frac{1}{c^2}\frac{\partial^2}{\partial t^2}p(\vec{r},t) = -\frac{\beta}{C_p}\frac{\partial}{\partial t}H(\vec{r},t), \tag{1}$$

where $c = 1.5$ mm/$\mu$s is the ultrasound speed in biological tissue; $\beta$ is the isobaric volume expansion coefficient; $C_p$ is the specific heat; $H(\vec{r},t)$ is the heating function that indicates the thermal energy deposited by the incident laser per volume per time[8,9]

$$H(\vec{r},t) = A(\vec{r})I(t), \tag{2}$$

where $A(\vec{r})$ is the spatial optical absorption function and $I(t)$ is the temporal illumination function. Then, $p(\vec{r},t)$ can be expressed as[9,10]

$$p(\vec{r},t) = \frac{\beta}{4\pi C_p}\iiint \frac{\mathrm{d}^3 r'}{\left|\vec{r}-\vec{r'}\right|}A(r')I'(t'), \tag{3}$$

where $I'(t') = \mathrm{d}I(t')/\mathrm{d}t'$.

To reconstruct PA image, $A(\vec{r})$ should be found by time-domain spherical back projection algorithm. In two-dimensional (2D) circular configuration, $A(\vec{r})$ is given by[9]

$$A(\vec{r}) = -\frac{r_0^2}{2\pi\eta c^4}\int_{\varphi_0}\mathrm{d}\varphi_0 \frac{1}{t}\frac{\partial p(\vec{r}_0,t)}{\partial t}\bigg|_{t=|\vec{r}_0-\vec{r}|/c}, \tag{4}$$

where $\eta = \beta/C_p$, $\vec{r}_0$ is the scanning coordinate of transducer. Equation (4) indicates that $A(\vec{r})$ can be reconstructed by the integration of PA signal detected at $t = |\vec{r}_0 - \vec{r}|/c$ around all scanning positions.

We used the circular measurement configuration shown in Fig. 1. The experimental setup is shown in Fig. 2. A 532-nm pulsed Nd:YAG laser of 10-Hz repetition rate (Quanta-Ray PIV, Spectra Physics) was employed as the pumping source. The pulse width was 7 ns and pulse energy was 80 mJ. The incident laser beam was 10 mm in diameter and was homogenized by a ground class

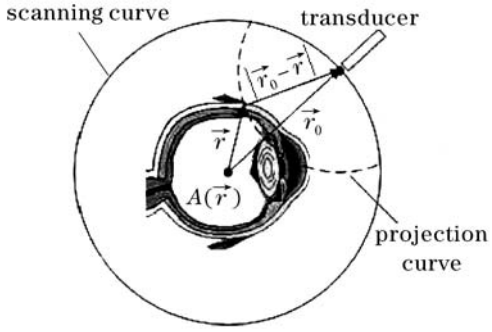


Fig. 1. PA signal detection scheme.

 

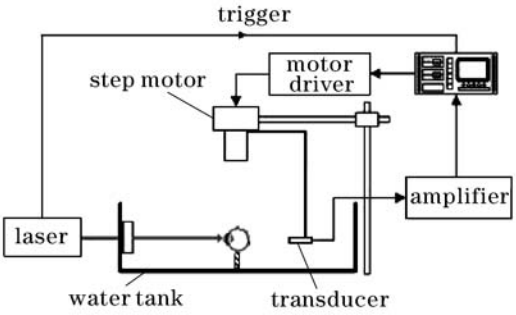


Fig. 2. Experimental setup.

before it projected into the pupil of the porcine eyeball. A wide-band polyvinylidene difluoride (PVDF) unfocused acoustic transducer (HPM1/1, Precision Acoustics) with the active detecting element of 1 mm in diameter was used to collect the PA signal. The frequency response range of the transducer was from 3 to 15 MHz. An oscilloscope (TDS5054, Tektronix) was triggered after laser irradiation by 10-$\mu$s delay time. The data sampling rate was 250 MHz. The transducer was driven by a step motor and rotated around the porcine eyeball in step of 3.6° with totally 158 scanning positions. The distance between the transducer and the eyeball was 52 mm. The eyeball was embedded into the surface of a cylindrical phantom made of the mixture of 100-mL water and 10-g agar.

The imaging result in Fig. 3(b) agrees well with the histological structure of the porcine eyeball *in vitro* and the imaging contrast is high, especially for cornea, anterior chamber, and crystalline lens. The cornea, sclera, ciliary body, and retina are primarily made up of

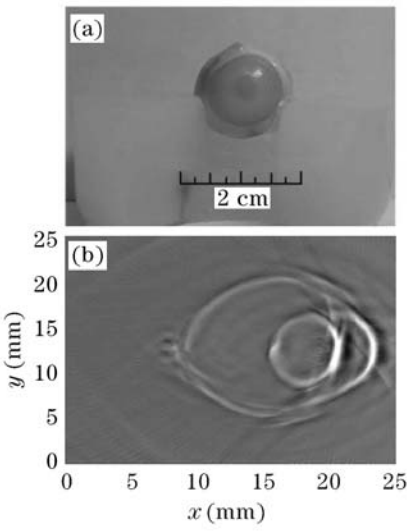


Fig. 3. (a) Photograph of the sample; (b) reconstructed PA image of the porcine eyeball *in vitro*.

connective tissue, pigmented epithelium, and melanin. The crystalline lens and vitreous body are filled with clear liquid. The variance of optical absorption coefficients inside eyeball is larger in comparison with that of ultrasound impedance. For example, the optical absorption coefficient of melanosomes isolated from the bovine retinal pigmented epithelium at 532 nm is 2370 $cm^{-1}$[11]. So, it is possible to get the higher imaging contrast by PA imaging. In pure clinical ultrasound imaging of eyeball, the resonance frequency of the acoustic transducer ranges from 50 to 100 MHz. As PA signal has a wide frequency band up to 100 MHz, PA imaging with the spatial resolution better than 100 $\mu$m is possible by using the transducer with the higher central resonance frequency[12]. So, it is possible to perform PA microscopy for ophthalmology. By applying the safety laser for human eye, PAT is a potential feasible clinical imaging method for ophthalmology.

In conclusion, the feasibility of PAT for ophthalmology was studied. The PA signal was collected by an unfocused PVDF needle acoustic transducer by circular scanning. The PA image was reconstructed by time-domain spherical back projection algorithm. The experimental results have clearly revealed the main histological structure inside eyeball and achieved preferable imaging contrast. The study demonstrated the potential feasibilities of PAT for ophthalmology.

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