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Two dimensional photoacoustic imaging using microfiber interferometric acoustic transducers☆

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ABSTRACT

Photoacoustic imaging transducer with a pair of wavelength-matched Bragg gratings (forming a Fabry–Perot cavity) inscribed on a short section of microfiber has been developed. A tunable laser with wavelength that matched to one of selected fringe slopes was used to transmit the acoustic induced wavelength. Interferometric fringes with high finesse in transmission significantly enhanced the sensitivity of the transducer even under very small acoustic perturbations. The performance of this novel transducer was evaluated through the imaging studies of human hairs (~98 μm in diameter). The spatial resolution is 300 μm. We have demonstrated that the novel transducer developed in this study is a versatile tool for photoacoustic imaging study.

1. Introduction

Photoacoustic imaging becomes a powerful tool offering good spatial resolution for biomedical fields. Under a pulse laser irradiation, biological tissue absorbs the light energy and generates heat, which leads to an acoustic pressure distribution in the tissue followed by acoustic wave propagation to the surface of the tissue. Acoustic transducers can be used to detect the acoustic waves at the tissue surface for the reconstruction of photoacoustic images. Combining the advantages of high optical contrast in biological tissues and excellent spatial resolution of ultrasound imaging techniques, photoacoustic imaging, since it was firstly proposed in the end of the 20th century [1], has been regarded as a promising method in preclinical and clinical research [2,3]. Typically the frequency spectrum of photoacoustic signals covers a wide range from 100 kHz to 10 MHz. For imaging of absorption regions with different sizes, a broad band transducer is essential for the detection of photoacoustic signals. Hence, the development of wide band ultrasonic transducers could significantly enhance the sensitivity of photoacoustic imaging and expand the scope of its applications [4].

The detection in photoacoustic imaging is traditionally based on piezoelectric transducers, which are highly sensitive but have limited bandwidth owing to their resonant characteristics [5]. Although piezoelectric transducers based on the thin polymer films and appropriate matching materials can respond to a wide bandwidth, the sensitivity will be reduced considerably as the diameter and the electrical capacitance decrease [6]. Another drawback of piezoelectric transducers, which is related to their electrical nature, is that it is vulnerable to electromagnetic noise interference. In marked contrast to piezoelectric transducers, optical fiber interferometric transducers are not subjected to external electromagnetic disturbances and thermal crosstalk produced by the direct laser pulse illumination [7], which can be advantageous in photoacoustic imaging. Recently, Fiber Bragg grating Fabry–Perot (FBG-FP) transducer has been proposed as integrating line detectors for photoacoustic imaging [8]. It has been shown that high resolution two dimensional photoacoustic imaging can be obtained by using FBG-FP interferometric transducer [9]. Interferometers based photoacoustic detection has been reported by several groups. Paul et al. monitor acoustic pressure induced displacements of resonant Fabry–Perot polymer film cavity, which is the most commonly used optical method [10]. Daniel et al. are based on acoustic pressure induced refractive index variation using single-mode silica optical fiber (SMSOF) interferometric sensors [11]. A single-mode polymer (SMPOF) optical fiber has offered

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the possibility of developing a new class of high-sensitivity interferometric transducer, which increases in ultrasonic sensitivity than a single-mode silica optical fiber [12]. However, these SMPOF are still under development and present high loss and difficulties to coupling light into. On the other side, Hubert et al. obtain acoustic pressure induced refractive index variation with perfluorinated polymer optical fibers (PFPOF). The contrast of the 2D reconstructed image is higher than using SMSOF. The PFPOF with lower losses is easier to handle, but there is degradation in the visibility of the interference reducing the total sensitivity due to their big core diameter nature [8].

In this Letter, we present our results on the detection of photoacoustic signals using the microfiber FBG-FP interferometric transducer and the reconstruction of photoacoustic images. Two dimensional images of a loop hair can be reconstructed with high resolution. The influence of the measured position numbers are analyzed and discussed.

2. Principle

First, the principle of the operation of the microfiber FBG-FP interferometric acoustic transducers is based on the effective refractive index induced by the acoustic wave. The phase of the propagating light can be modulated by the effective refractive index which is given by:

$$\Delta\phi = k\,(n\Delta l + l\Delta n) \tag{1}$$

where k is the wave number, l is the length of the sensing segment, and n is the effective refractive index of the optical fiber. The strain induced, localized on a fiber segment of length l, can be under axially constrained, suggesting that the axial elongation $\triangle l$ is zero. Therefore, the phase shift is mainly governed by the factor of the effective refractive index change $\triangle n$. The phase shift can be magnified using a longer cavity length segment exposed to the acoustic wave. Then, the phase of the light can be demodulated by a variation of the optical intensity of microfiber FBG-FP interferometer. So, this optical signal is registered by a photodiode with transimpedance amplifier that converts the optical intensity in a voltage signal. The total sensitivity of the detector, i.e. the relation between the output voltage and the acoustic pressure, can be expressed as:

$$\frac{\Delta W}{\Delta P} = \frac{\Delta W}{\Delta I}\frac{\Delta I}{\Delta\phi}\frac{\Delta\phi}{\Delta P} \tag{2}$$

where W is the output of the voltage, P is the acoustic pressure amplitude, I is light intensity at interferometer output and ϕ is the phase of the light.

Second, when the biological tissue is irradiated by a short pulse laser, the wave equation of the acoustic pressure $P\,(r,\,t)$ of the photoacoustic signal can be expressed as:

$$\nabla^2 p(r,t) - \frac{1}{c^2}\frac{\partial^2}{\partial t^2}p(r,t) = -\frac{\beta}{C_p}\frac{\partial}{\partial t}H(r,t) \tag{3}$$

where β is isobaric expansion coefficient and C_P is specific heat coefficient. The solutions based on Green functions are [13]:

$$p(r,t) = \frac{\beta}{4\pi C_p}\iiint_{t'=t-(|r-r'|/c)}\frac{d^3r'}{|r-r'|}\cdot\frac{\partial H(r',t')}{\partial t'} \tag{4}$$

where $H(r,t)$ is called thermal function, which represents the laser energy absorbed by biological tissue in unit time and volume. It can be expressed as the product of the space optical absorption distribution function and the time domain function of laser pulse intensity:

$$H(r,t) = A(r)I(t). \tag{5}$$

At this time, the acoustic pressure $P\,(r,\,t)$ measured by the transducer at the r_0 can be expressed as:

$$p(r_0,t) = \eta\iiint d^3r\cdot A(r)\frac{\delta'(t-|r_0-r|/c)}{4\pi|r_0-r|} \tag{6}$$

where $\eta = \beta/C_p$, $I\,(t) = \delta\,(t)$.

On the other hand, the dependence of phase change on acoustic signal can be expressed as:

$$\frac{d\varphi}{dp} = \varphi(\frac{1}{L_{eff}}\cdot\frac{dL_{eff}}{dp} + \frac{1}{n_{eff}}\cdot\frac{dn_{eff}}{dp}). \tag{7}$$

The first term in the bracket denotes the induced elongation of the cavity and the second term represents the induced effective-index change. As described in [14], the fiber is under axially constrained, suggesting that the axial elongation is zero. Therefore, only the index change contributes to the phase change.

We have measured the photoacoustic signal in the form of an ultrasonic pressure wave by using our microfiber FBG-FP interferometric transducer (see next section). Sensitivity is a key parameter in case of photoacoustic imaging. The acoustic sensitivity of the microfiber FBG-FP interferometric transducer is as high as 1.845 mV/kPa, which is 10 times higher compared with the single mode fiber FBG-FP interferometric transducer [15]. Fabry–Perot cavity is formed by cascading two wavelength matched FBGs, which are inscribed on the microfiber with a diameter of 5.2 μm. The transducer with a cavity length of 5 mm has the power of 10.8 dB higher than those of the single mode fiber FBG-FP interferometric transducer. The noise voltage is 0.1 mV, the photoacoustic signal voltage is 31.1 mV. Therefore, the SNR is 49.86 dB.

3. Experimental method

The microfiber which was tapered from a standard 62.5/125 multimode fiber (purchased from Corning Ltd) was further down to 5.2 μm by using a scanning flame as shown in Fig. 1(a) [16]. The multimode fiber was used because it had a larger Ge-doped photosensitive region, which enabled high efficient grating inscription without hydrogen loading or other photosensitization treatment [17]. The FBG-FP cavity was formed by inscribing two wavelengths-matched Bragg gratings with a 193 nm ArF excimer laser and a phase mask with a pitch of 1070 nm. The single-pulse energy and repetition rate were set to 3 mJ and 200 Hz, respectively. The length of the gratings was 3 mm and the blank space between the two gratings was 5 mm. After illumination for duration of 180 s for each grating, an FBG-FP cavity was then fabricated. Transmission spectrum of the FBG-FP cavity obtained under water by using an optical spectrum analyzer with a resolution of 0.02 nm is shown in Fig. 1(c). The maximum transmission depth was found to be 27.7 dB. The reflectivity of two FBGs was 91% and the finesse of microfiber FBG-FP interferometer was thirty. A tunable laser (1370–1660 nm, FP000012, Anritsu) was coupled to the microfiber FBG-FP interferometer. The output wavelength was adjusted to 1536.63 nm, at the sharpest slope of the interferometric fringes. In the present study, a human hair was used to perform the imaging study. Because of melanin, the hair was good absorber at 532 nm excitation laser. As shown in Fig. 1(b), the diameter of the hair was about 98 μm.

The experimental setup of our microfiber FBG-FP interferometric acoustic transducers is shown in Fig. 2. The light from a tunable laser, which operated at the wavelength of 1536.63 nm, was coupled into the interferometer. The measurement segment between two the FBGs was situated within a tank filled with water. Meanwhile, a 532 nm Nd: YAG pulse laser with a pulse laser of 15.8 mJ/cm^2 (average power of 1.6 kW, pulse duration of 10 ns and repetition rate of 20 Hz) was used for excitation. The light from the Nd: YAG laser was equally split into two light beams using a 50/50 optical beam splitter. The two beams were reflected by the mirrors to aim at the sample from two sides with an angle of about 90°. The two beams were merged to a diameter of approximately 18 mm to enable the excitation of the whole hair. The pulse energy was adjusted to keep the radiant exposure below the allowed maximum value of 20 mJ/cm^2 for biological tissue. An excitation wavelength of 532 nm was utilized to show the applicability for medical purposes [18].

Before mounting on the sample holder, the hair was tangled to form a loop with a diameter of approximately 2 mm as shown in Fig. 6(a).