

Detection of Ultrasound Modulated Photons in Diffuse Media Using the Photorefractive Effect

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ABSTRACT

Ultrasound modulated optical tomography is a dual-wave sensing technique in which diffusive light in a turbid medium interacts with an imposed acoustic field. A phase-modulated photon field emanates from the interaction region and carries with it information about the opto-mechanical properties of the media. We present a technique for the detection of the ultrasound induced optical phase modulation using an adaptive, photorefractive crystal (PRC) based interferometry system. Experimental results are presented demonstrating the detection of ultrasound modulated signals in highly scattering media using pulsed ultrasound insonation.

A number of optical imaging techniques¹ have been investigated for biomedical imaging based on differences in the optical properties of tissue types. The main difficulty in implementing such methods is that light is highly scattered in biological tissue. A technique that holds promise in improving the resolution in the imaging of optical properties inside of highly scattering media makes use of a combination of light and ultrasound. In this technique, ultrasound beams are used to phase modulate or “tag” diffuse light, and the detection of this tagged light yields spatially resolved optical information. The ultrasound modulates the phase of the diffuse optical field through the displacement of scatters or ultrasound-induced changes in refractive index.² This hybrid technique retains the benefits of both the optical and ultrasonic techniques. It can reveal the optically relevant physiological information about biological tissue while maintaining ultrasonic resolution, due to the fact that the light-sound interaction region is defined by the spatial extent of the ultrasonic beam.

Marks *et al.*³ reported the modulation of light with focused, pulsed ultrasound in a homogeneous, scattering media in 1993. This was closely followed by the work of Wang *et al.*⁴ and Kempe *et al.*⁵ who demonstrated the utility of using the tagging of diffuse photons for imaging purposes. The former group developed *ultrasound modulated optical tomography* using a CW ultrasound transducer and a CW laser source. Wang and Ku⁶ later introduced a technique in which the ultrasonic source is chirped, thereby encoding spatial information on the transmitted laser light and allowing for 1-D scans along the ultrasonic axis to be produced from a single waveform.

In order to overcome signal-to-noise ratio (SNR) limitations of single detector techniques, Leveque *et al.*⁷ presented a new approach to detect the modulated signals in parallel using a CCD array. In

their technique, the speckle size is adjusted to approximately match the size of a single element of the CCD array and the modulation amplitudes measured at all of the pixels are summed. The effective light gathering capability of this system, characterized by the optical etendue, is increased by $\sim N$, where N is the number of coherence areas detected, corresponding to the number of pixels on the CCD. This results in an improvement in SNR of $\sim N^{1/2}$ over single detector schemes. The main limitation of the parallel detection scheme is that it is sensitive to the decorrelation of the speckle pattern during the measurement time.

We report a novel technique for the detection of the ultrasound-modulated optical signals using the two-wave mixing process in a photorefractive crystal (PRC). PRC-based interferometers have seen widespread use in nondestructive characterization applications.⁸⁻¹¹ In addition, photorefractive semiconductor heterostructures have recently been used in a hybrid imaging technique combining optical coherence domain reflectometry and ultrasound,¹² and the use of four-wave mixing in PRC's has been suggested for use as a frequency filter in the detection of ultrasound modulated optical signals.¹³ In our detection system, two-wave mixing is used to derive local oscillator (LO) that is wave front matched to the diffusely scattered light from the target media, referred to as the signal beam. The signal beam and the LO interfere at the optical detector where the phase modulation accumulated by the signal beam through interaction with the ultrasound field is converted to an intensity modulation. The wave front matched LO allows for the detection of phase modulation over the entire collected speckle field giving a large etendue. In addition, the PRC is adaptive and compensates for low frequency shifts in the speckle pattern, making it insensitive to speckle shifts on time scales longer than the response time of the crystal.

The experimental setup is given in Figure 1. An 80mW frequency doubled Nd:YAG laser source is sent to a variable beamsplitter where it is split into signal and reference beams. The reference beam is directed around the test tank and sent directly to the PRC. The signal beam is sent to the submerged tissue-mimicking phantom. The scattered and ultrasonically modulated light is collected after the tank and is directed into the PRC where it interferes with the reference beam. Our PRC-detector employs a BSO crystal with dimensions of $5\text{mm} \times 5\text{mm} \times 7\text{mm}$. A 4 kHz AC field of 10-kV/cm peak-to-peak is applied to the crystal to enhance the two-wave mixing gain. Under our experimental conditions, the response time of the PRC is approximately 150ms. After the signal beam passes through the PRC, the signal beam and diffracted reference beam (LO that is wave front matched to the signal beam) are sent to an avalanche photodiode (APD). The sound source is a spherically focused piezoelectric transducer (Sonic Concepts), with a 6.32-cm focal distance and a center frequency of 1.1 MHz. The ultrasonic axis is perpendicular to the phantom illumination direction. The tissue phantoms consist of a transparent polyacrylamide gel, with 0.4- μm diameter polystyrene microspheres added to control the optical scattering coefficient. The dimensions are $4.3\text{cm} \times 4.3\text{cm} \times 2.9\text{cm}$ and the phantoms are oriented such that the laser passes through the 2.9-cm thickness.

The operation of PRC interferometers has been detailed in the literature. For simplicity, the case of a planar signal and reference beams will be treated here, although it is noted that this can be easily extended to the case of a speckled signal beam. The signal and reference beams interfere in the PRC creating an index grating in the material, and the signal beam experiences a photorefractive amplitude gain (γ) through the two-wave mixing process. It is assumed that the ultrasound causes a sinusoidal phase modulation in the signal beam, and that the overall length of the ultrasound tone

burst is short with respect to the response time of the crystal, such that the photorefractive grating remains static during the measurement period. At the exit of the crystal the intensity I_{SE} is given by:¹⁰

$$I_{SE} = \exp(-\alpha L) I_{SO} \left\{ \left| e^{\gamma L} - 1 \right|^2 + 1 + 2 \operatorname{Re}[(e^{\gamma L} - 1) * \exp(i\phi_a \sin(\omega_a t + \chi_r))] \right\}, \quad (1)$$

where I_{SO} is the intensity of the signal beam at the front surface of the crystal, α is the crystal absorption coefficient, L is the crystal length, ϕ_a and ω_a are the amplitude and the angular frequency of the ultrasound induced phase modulation, respectively, and χ_r is a constant that depends on the optical path. The gain coefficient is given by $\gamma = \gamma' + i\gamma''$, where γ' is the real part of the gain and γ'' is the imaginary part. Rewriting Eqn. (1) in terms of DC and AC components, and retaining only the lowest order terms in the Bessel function expansion, we obtain:

$$I^{DC}_{SE} = \exp(-\alpha L) I_{SO} \left\{ \left| e^{\gamma L} - 1 \right|^2 + 1 + 2[e^{\gamma' L} \cos(\gamma'' L) - 1] J_0(\phi_a) \right\}, \quad (2)$$

$$I^{AC}_{SE} = 4 \exp(-\alpha L) I_{SO} e^{\gamma' L} \sin(\gamma'' L) J_1(\phi_a) \sin(\omega_a t + \chi_r). \quad (3)$$

Note that these equations give the optical signal at the exit of the crystal when the signal beam incident on the PRC has a fixed, time dependent phase modulation corresponding to a single optical path. In the case of signal detection from highly scattering media, light travels over multiple paths and the AC component of the signal observed at a single detector is the summation of the signals from each of the paths. As the modulation induced by the ultrasound is not uniform spatially (χ_r is path dependent), it is not expected that the AC components of the signals will add coherently at the detector. The DC component, however, depends only on the amplitude of the phase modulation and

thus allows for the measurement of the magnitude of the mean phase shift induced by the ultrasound on the multiply scattered optical field. In the linear detection of small amplitude signals, photorefractive interferometers are typically designed such that the diffracted reference beam and transmitted signal beam are placed in quadrature. However, in the present experiment the AC component of the signal (Eqn. 3) is very weak when compared to the DC shift (Eqn. 2) and it is the latter component that we propose to use for sensing and imaging applications. In order to maximize this signal, the photorefractive interferometer configuration shown in Fig. 1 was chosen, where the diffracted reference beam and transmitted signal beam are in phase giving pure (real) photorefractive gain.

Fig. 2(a) shows the measured focal pressure generated by the sound source driven by a 20-cycle pulse at 1 MHz center frequency. The peak pressure is approximately 0.4 MPa. The corresponding ultrasound modulated optical signals are shown in Fig. 2(b) for a phantom with a reduced scattering coefficient of $\mu'_s = 6 \text{ cm}^{-1}$. The signals are temporally averaged 1000 times to improve the SNR and the detection bandwidth is 25 MHz. The top trace in Fig.2 (b) shows the signal detected in the absence of a reference beam to the PRC. In this case, the PRC has no effect except for absorbing a small amount of the signal beam. The lower trace shows the signal when the reference beam is present and two-wave mixing takes place. In this case a signal is observed that follows the envelope of the pulse train. One can readily see that the PRC interferometer, facilitated by the addition of a high voltage AC bias field, dramatically enhances the signal. The fact that the system has sufficient sensitivity to use pulsed ultrasound provides two important advantages over the systems which use CW ultrasound: the use of short ultrasound pulse trains allows for spatial resolution along the

ultrasound axis and pulsed ultrasound minimizes deleterious thermal bio-effects that can result from high-intensity ultrasound exposure.

Line scans were also performed to demonstrate the ability of this technique to detect a buried optical inhomogeneity. A $4.3\text{cm} \times 4.3\text{cm} \times 2.9\text{cm}$ phantom with a reduced scattering coefficient of 6cm^{-1} was fabricated that contained a $1.0\text{cm} \times 1.0\text{cm} \times 0.8\text{cm}$ buried optical absorber, where the 0.8 cm dimension was along the light propagation direction. The absorber was made of the same acrylamide gel as the surrounding media, but with a small amount of India ink added such that the absorption coefficient is approximately 4.6cm^{-1} . The phantom was scanned in 1mm steps perpendicular to the light propagation direction (i.e. out of plane in Figure 1). At each point, the ultrasound-modulated signal was averaged 1000 times and digitally low pass filtered at 300kHz. The depth of modulation of the envelope signal was then measured. Figure 3 shows the results from averaging two such scans. The position of the absorber is approximately in the center of the scanned region and a reduced modulation depth in this region is clearly evident in the figure. The signals are normalized by the total amount of light collected from the sample, eliminating any shadowing effects that result in changes in signal beam level as the sample is scanned.

The envelope signal was clearly observed when using PRC-based interferometry and only a limited number of ultrasound cycles. The use of short ultrasound pulse trains should allow for high axial resolution. While other researchers have shown that axial resolution is possible using CW chirped ultrasound insonation⁶, it is clearly desirable to minimize the ultrasound exposure by using pulses. This technique also has advantages in terms of ease of implementation over that proposed recently by Lev and Sfez¹⁴, where CW signals from each depth within the phantom are built up using a

reshaping algorithm with phase conserving pulse-to-pulse transmission. The detection system that has been developed is somewhat insensitive to speckle decorrelation as the crystal is adaptive and provides a LO that is wavefront matched to the signal beam, provided that the crystal response time is short with respect to the speckle decorrelation time. Finally, the PRC based system allows for a high optical etendue, giving an additional advantage over source synchronized lock-in techniques which use a CCD camera as a detector array and are thus limited in the number of speckles detected by the number of pixels on the CCD. The PRC based system requires only a single detector and relatively simple detection electronics. Future work is required to quantify the SNR and etendue of the system for direct comparison to CCD array detection approaches.

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Figure Captions

Figure 1. Experimental setup for PRC based detection of ultrasound modulated optical signals: VBS- variable beamsplitter, RB- reference beam, SB- signal beam, BPF- optical bandpass filter, PRC- photorefractive crystal, and APD- avalanche photodiode.

Figure 2. Experimental results showing a) ultrasound focal pressure and b) optical signal observed without a reference beam on the PRC (top) and with a reference beam (bottom).

Figure 3. The modulation depth of the envelope signal observed as the transducer is scanned across a 10 mm diameter optical inhomogeneity buried in a tissue phantom.

Figure 1

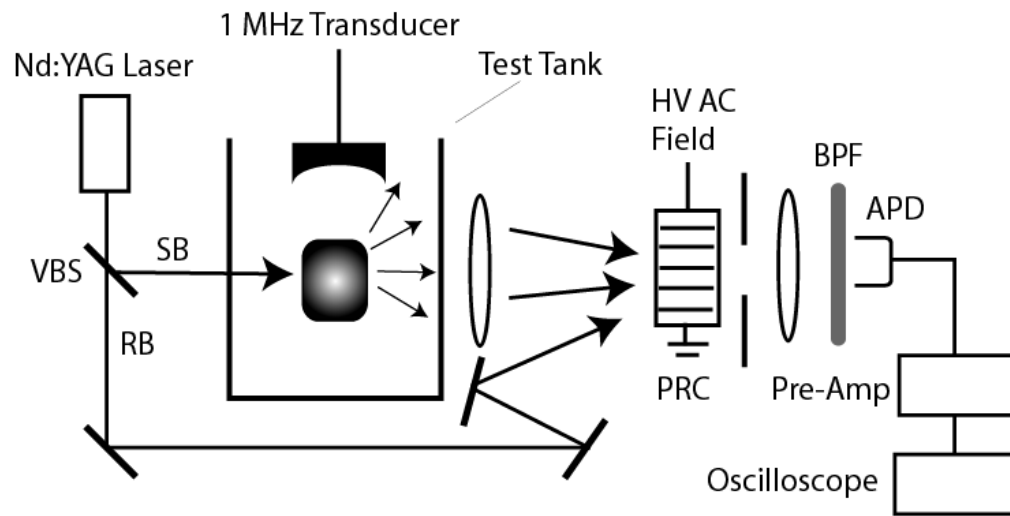


Figure 2

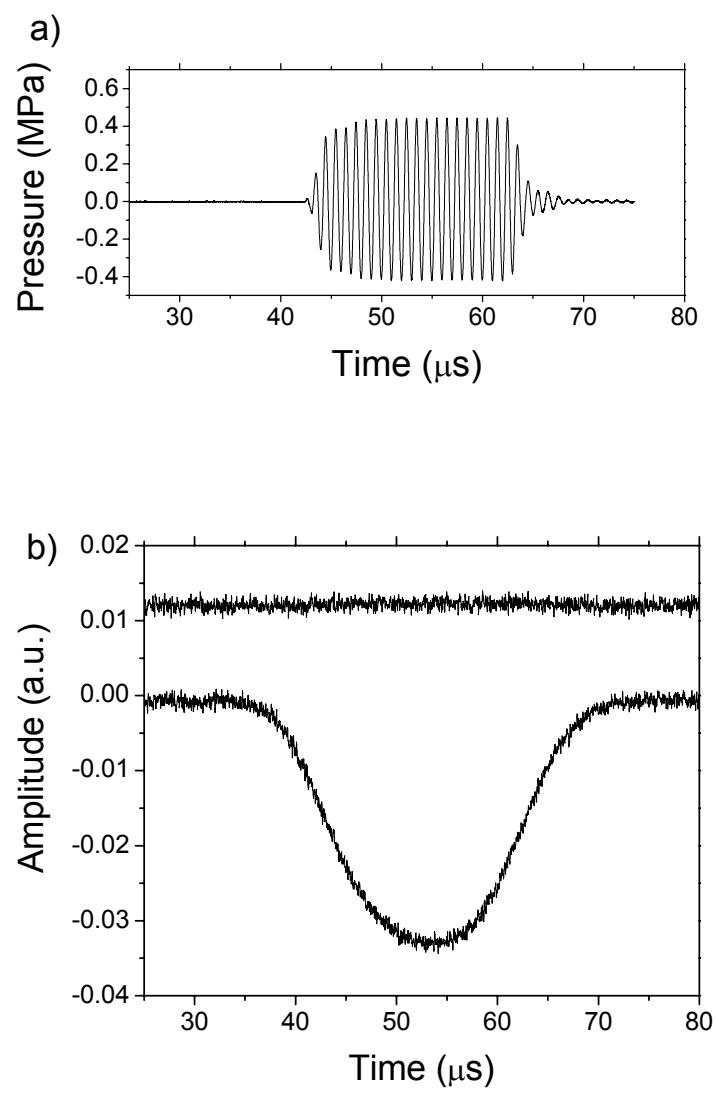


Figure 3

