Homodyne Time-Domain Acousto-Optic Imaging for Low-Gain Photodetectors

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Acousto-optics imaging (AOI) is a hybrid imaging modality that is capable of mapping the light fluence rate in deep tissue by local ultrasound modulation of the diffused photons. Since the intensity of the modulated photons is relatively low, AOI systems often rely on high-gain photodetectors, e.g. photomultiplier tubes (PMTs), which limit scalability due to size and cost and may significantly increase the relative shot noise in the detected signal due to low quantum yields or gain noise. In this Letter, we have developed a homodyne AOI scheme in which the modulated photons are amplified by interference with a reference beam, enabling their detection with low-gain photodetectors. We experimentally demonstrate our approach with a silicon photodiode, achieving over a 4-fold improvement in SNR in comparison to a PMT-based setup.

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Deep-tissue optical imaging is generally performed by illuminating the tissue over a large area and using the reemitted diffused light to form an image representative of the optical properties of the tissue [1]. By using tomographic illumination and detection patterns, combined with optimization-based inversion algorithms, depth-resolved imaging may be performed, as been demonstrate in the field of diffuse optical tomography (DOT) [2, 3]. However, purely optical techniques such as DOT are inherently limited in their spatial resolution due light diffusion, mathematically characterized by an ill-conditioned inverse problem.

Acousto-optics imaging (AOI) [4, 5] can improve the resolution of purely optical techniques of deep-tissue imaging by using ultrasound (US). In AOI, the tissue is both illuminated with a coherent laser and insonified with an ultrasound transducer, leading to a pressure-induced refractive-index modulation and vibrations of the optical scatterers in the insonified regions [6]. As a results, light that travels through the insonified regions experiences a phase modulation with the same frequency as that of the ultrasound wave. Because of the high coherence of the laser, the reemitted light exhibits a speckle pattern in which the intensity of each speckle grain is temporally modulated with the ultrasound frequency. The modulation depth of the speckle pattern may be measured using various methods [7], enabling a localized detection of light in the tissue. AOI may be performed with a focused US beam that scans the imaged region [8] or a set of plane US waves [9], enabling the visualization of the light fluence rate within the tissue with acoustic resolution.

To optimize the signal-to-noise ratio (SNR) of the measurement, parallel detection of uncorrelated speckle grains is required, which is often performed by digital cameras. However, the low time resolution of cameras introduces two difficulties that limit their use \textit{in vivo}. First, it complicates the distinction between the effect of US modulation and speckle decorrelation, as both these phenomena are faster than the camera frame-rate [10]. Second, it limits the use of US bursts, for which the acoustic time of flight may be used for depth sectioning.

In order to overcome speckle decorrelation and enable the use of acoustic bursts, time-domain AOI (TD-AOI) may be used, in which the optical detection is performed with detectors that operate at a higher bandwidth than that of the US transducer [11–13]. The immunity of TD-AOI to speckle decorrelation has facilitated \textit{in vivo} demonstrations in small animals [14] as well as clinical testing for the diagnosis of osteoporosis [15]. An additional advantage of TD-AOI is its ability to locally quantify blood flow at different depths from the spectral broadening of the AOI signal [16, 17] — a capability that has been demonstrated for monitoring cerebral blood flow [18].

Because of the low intensity of the US-modulated light, TD-AOI is conventionally performed with photodetectors with a high internal gain, e.g. photomultiplier tubes (PMTs) that use the avalanche effect to multiply the current created by a single photon [13, 17, 19]. However, despite the high gain, this approach may lead to a lower SNR in the AOI measurement. First, the avalanche is an inherently stochastic process in which the gain varies randomly, thus increasing the relative shot noise in the signal. Second, in the case of PMT, the relative shot noise is further increased by the low quantum yield of the detector. Third, the cost and complexity of high-gain detectors limits their scale-up to multi-element arrays, required for high-SNR operation.

In this work, we have developed a new approach for TD-AOI in which the reemitted light is not detected directly, but is rather interfered with a reference beam in a homodyne configuration. The interference leads to an optical amplification of the US-modulated light, enabling its detection with low-gain photodetectors. While homodyne detection often requires con-
trolling the exact phase difference between the signal and reference arms [20], no such control is required in our scheme due to the stochastic nature of the AOI signal. We have experimentally tested our approach with a silicon photodiode (PD), demonstrating over a 4-fold increase in SNR in comparison to a PMT, in agreement with the theoretical prediction.

Assuming that the detection is performed over time scales faster than the speckle decorrelation time, the field of a single speckle may be represented by

$$E_i = E_{bkg,i}e^{i\omega t} + E_{AO,i}e^{i\omega t}e^{i(2\pi f_{US}t + \phi)} \tag{1}$$

where $\omega$ is the angular frequency of the light, $f_{US}$ is the US frequency and $\phi$ is a random phase distributed: $\phi \sim \mathcal{U}[0, 2\pi]$ that represents the phase difference between the modulated and unmodulated parts of the field. $E_{bkg,i}$ and $E_{AO,i}$ are the amplitudes of the unmodulated and modulated fields, respectively, where it is assumed that $|E_{bkg,i}| \gg |E_{AO,i}|$.

In conventional TD-AOI, the measured optical power consists of spatial integration over $N$ speckles, which are statistically independent and identically distributed, leading to the following expression:

$$P^N \sim N \sum_{i=1}^{N} |E_{bkg,i}e^{i\omega t} + E_{AO,i}e^{i\omega t}e^{i(2\pi f_{US}t + \phi)}|^2 \tag{2}$$

$$= N \sum_{i=1}^{N} I_{bkg,i} + 2 \sum_{i=1}^{N} \sqrt{I_{bkg,i}I_{AO,i}\cos (2\pi f_{US}t + \phi)}.$$  

where $I_{bkg,i} = |E_{bkg,i}|^2$ and $I_{AO,i} = |E_{AO,i}|^2$. Since the phase $\phi$ is uniformly distributed over $2\pi$, one obtains that $\mathbb{E}[P^N] = N\mathbb{E}[I_{bkg,i}]$ and $\text{Var}[P^N] = N\mathbb{V}[I_{bkg,i}]$. Accordingly, the signal in AOI is often calculated as the standard deviation of $P^N$ at the frequency $f_{US}$ and is proportional to $\sqrt{N\mathbb{E}[I_{bkg}I_{AO}]}$, where $I_{bkg} = \mathbb{E}[I_{bkg,i}]$ and $I_{AO} = \mathbb{E}[I_{AO,i}]$.

In the proposed homodyne scheme for TD-AOI, the field in Eq. 1 is interfered with a reference beam, leading to the following expression for a single speckle grain:

$$E'_{i} = E_{ref}e^{i\omega t} + E_{bkg,i}e^{i\omega t} + E_{AO,i}e^{i\omega t}e^{i(2\pi f_{US}t + \phi)} \tag{3}$$

Assuming $I_{ref} \gg I_{bkg,i}$, the following expression for the power of $N$ grains is obtained by neglecting $I_{bkg}$:

$$P^N_{hom} \propto N I_{ref} + 2 I_{ref} N \sum_{i=1}^{N} \sqrt{I_{bkg,i}I_{AO,i}\cos (2\pi f_{US}t + \phi)} \tag{4}$$

Accordingly, the magnitude of the AOI signal in Eq. 4, described by the standard deviation of the expression, is given by $\sqrt{N I_{ref} I_{AO}}$, and the average power is given by $NI_{ref}$.

As can be seen from the above analysis, the use of homodyne detection enables one to amplify the AOI signal since $I_{ref} \gg I_{bkg}$, where the goal is to achieve sufficient amplification such that shot noise becomes the dominant noise factor even when low-gain photodetectors are used. In the shot-noise-limited case, the noise is proportional to the square root of the average power, leading to an SNR that is proportional only to $\sqrt{I_{AO}}$ and is independent of all the other parameters in Eq. 2 and 4. Thus, when examining only inherent noise in the optical signal, homodyne TD-AOI attains the same SNR as conventional TD-AOI. Accordingly, the advantage of the homodyne approach is that it enables the use of low-gain photodetectors, with potentially lower cost and noise factors, without the cost of increasing the inherent SNR of the optical signal before detection.

Assuming shot-noise-limited detection, the SNR at the output of the photodetector is given by [21]:

$$\text{SNR}_{det} = \sqrt{\frac{\eta P}{2\hbar c f_{US}}} \tag{5}$$

where $\eta$ is the quantum efficiency, $P$ is the optical power, $F$ is the noise factor, $h$ is Planck’s constant, and $f_{US}$ is the measurement bandwidth. The noise factor $F$ is a result of gain fluctuations that occur when the gain is achieved via an avalanche process, and is typically smaller than 1.5 [21]. In the case of a PD, whose gain is provided by a trans-impedance amplifier, the gain may be considered constant, i.e. $F = 1$. Accordingly, when shot-noise-limited detection is assumed, the gain in SNR achieved by a PD over a PMT may be expressed by

$$G_{SNR} = \frac{\eta_{PD} F_{PMT}}{\eta_{PMT}} \tag{6}$$

Using the values $\eta_{PD} = 76\%$, $\eta_{PMT} = 5\%$, and $F_{PMT} = 1.4$, given for the components used, a theoretical SNR gain of $G_{SNR} = 4.6$ is obtained.

Fig. 1 shows the experimental setup used in this work to test the performance of homodyne TD-AOI. The system’s optics is schematically divided into 2 sub-system: (1) splitting module and (2) merging module. In the splitting module, a linearly polarized CW laser (DL Pro 780, Toptica) with a linewidth of 50 kHz and wavelength of 780 nm is split into 2 branches by two half-wave plates ($\lambda/2$) and a polarization beam splitter (PBS). The first $\lambda/2$ plate angle is set such that most of the optical power is reflected by the PBS into the phantom via a multi-mode fiber with a 62.5µm core diameter, 2 m length that delivered 200 mW to the phantom boundary. The light transmitted through the PBS is used for the reference beam. To optimize the reference beam power, it passes through a second motorized $\lambda/2$ plate and a polarizer. Eventually, the reference beam is coupled to a polarization-maintaining (PM) fiber with a length of 4 m.

The reemitted light from the phantom, is collected by a second MM fiber with 600µm core diameter, 0.39 NA, and 2 m length and delivered to the merging module. The distance between the two fibers on the phantom boundary was 5 mm on X axis. Since the reemitted light is unpolarized, a linear polarizer is used to achieve a polarization that matches that of the reference beam. Both beams are merged into a single interfered beam by a 50:50 BS and then coupled into a 200µm core diameter fiber. The merged light is delivered to a silicon PD (FDS-02, Thorlabs) with a quantum efficiency of 76%. Both the PD and the PMT were connected to identical custom-made electronics, including a trans-impedance amplifier (Texas Instruments LMH-6626), a 2.5 MHz passive low pass filter, and a voltage buffer. A 14-bit digitizer (ATS-9416, AlazarTech) was used to sample the voltage signal at a frequency of 20 MHz.

The technique is tested on a tissue-mimicking phantom made of silicone mixed with 193nm TiO$_2$ particles forming a reduced scattering coefficient equals to $\mu'_s = 15cm^{-1}$ and speed of sound of 990 m/s [19]. The ultrasound modulation is generated by a focused piezoelectric transducer (Panametrics, A392S) with diameter of 38.1 mm, focal length of 9.4cm, a Rayleigh length of 5.56 cm, and a cross-section FWHM of 4.25 mm. The transducer is driven by an arbitrary function generator (Tabor, 8026), amplified to a 48 V peak-to-peak amplitude (±1.8-50EU26, SVPA),
leading to a peak pressure of approximately 250 kPa in the acoustic focus. The transducer may be used to deliver a single acoustic pulse into the phantom, which modulates different depths at different times, enabling depth-resolved mapping of the AOI signal without scanning by using the time-of-flight principle [22]. In our implementation, we used a coded sequence of pulses, rather than a single pulse, to maximize the acoustic modulation depth [19]. Our sequence was based on a code with 251 elements, where each pulse had a single period with a frequency of $f_{US} = 1.25$ MHz, corresponding to an axial resolution of 1.1 mm, a total imaging depth of 27.6 cm and sequence repetition rate of 5 kHz. In each measurement the signal was measured continuously and averaged over 10^4 repetitions (a total duration of 2.008 s).

To obtain optimal reference power, a preliminary calibration experiment was performed. The transducer was positioned in a fixed coordinates on the $X - Y$ plane and 100 nW of the remitted light was coupled into the sensor MM fiber (Fig. 1). In the first part of the experiment, the reference beam was blocked and the remitted light was directed towards a PMT (R5900U-20-L16, Hamamatsu) with a gain of $2.7 \times 10^5$. The PMT signal was used to calculate the baseline SNR of the measurement for conventional TD-AOI. The measurement was then repeated with the reference beam unblocked, i.e. in a homodyne configuration, for different reference powers. For each reference beam intensity, the SNR values were averaged over 30 measurements to minimize the variations between measurements.

Fig. 2 presents the SNR gain of homodyne TD-AOI as a function of $\gamma = P_{ref}/P_{re}$, where $P_{ref}$ and $P_{re}$ are the intensities of the reference beam and the remitted light respectively. The figure shows that for low values of $\gamma$, the SNR gain is lower than 1, i.e. the homodyne system achieves a lower SNR than conventional TD-AOI. The reason for this result is that at low levels of $\gamma$, the detection with PDs is not shot-noise limited, but rather dominated by additive noise from the detector. Accordingly, the SNR gain of homodyne-AOI increases approximately linearly with $\gamma$ until a maximum value of 4.17, achieved for $\gamma = 1.5 \times 10^4$, corresponding to reference power of 1.4 mW. For higher reference powers, the SNR decreased due to saturation of the PD.

Once the reference beam intensity was set to maximal SNR gain, the 1D profile of the fluence rate inside the phantom was mapped in two directions. In the $z$ dimension (Fig. 3a) the mapping was performed using the time-of-flight principle, whereas in the $y$ dimension (Fig. 3b) the transducer was mechanically scanned with a step size of 0.5 mm. The measurement was conducted for both the homodyne and conventional TD-AOI setups, and the results are presented in Fig. 3a and 3b for the $z$ and $y$ scans, respectively. In both cases, the same spatial profiles were obtained for the AOI signals both the technique. The asymmetry depicted in the fluence rate shown in Fig 3a, which is consistent for both the PMT and the PD measurements, caused by geometric imperfections of tilted fibers with relation to the US axis. In terms of sensitivity, the noise level in the PMT-based measurement was higher by approximately 10 dB than the one
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Data availability. The data files generated during and/or analyzed during the current study are available from the corresponding author on a reasonable request.
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