Intrasweep phase-sensitive optical coherence tomography for noncontact optical photoacoustic imaging

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We introduce a method to extract the photoacoustic (PA) signal from the phase time evolution of an optical coherence tomography (OCT) swept source spectral sweep. This all-optical detection is achieved in a noncontact fashion directly on the sample surface by using its specular reflection. High-speed measurement and referencing allow for close to shot noise limited phase-sensitive detection. It offers a simple way to perform OCT and PA imaging by sharing the same system components. © 2012 Optical Society of America

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Photoacoustic microscopy (PAM) is an emerging biomedical imaging technique that combines high resolution with high absorption sensitivity [1]. Targeting hemoglobin allows for comprehensive assessment of tissue microvasculature in vivo. A drawback is the missing contrast of the embedding low-absorptive tissue that would often be required for proper image interpretation. An alternative is optical coherence tomography (OCT) that shows high endogenous contrast of tissue scattering but lacks absorption sensitivity [2]. Their combination would therefore allow for highly sensitive complementary characterization of tissue structure and its metabolism. Functional OCT methods could in addition provide quantitative tissue perfusion, yielding a complete picture of tissue nutrition.

In fact, PAM with optical resolution combines well with OCT due to comparable resolution and penetration depth performances [3]. Furthermore, current high-speed OCT systems share similar electronic detection bandwidth requirements with PAM (~several hundreds of MHz to GHz). PAM measures the temporal photoacoustic (PA) signal that results from the depth absorption profile. Previous studies already demonstrated combined OCT with PAM imaging. They employ an acoustic transducer in contact with the sample through a coupling media, being sensitive to pressure [3–6]. A recent approach used a single light source for both OCT and PAM imaging; however, PA detection was still performed with a transducer [7]. So far, only one PA method employing an optically transparent polymer-film-based Fabry–Perot sensor [8] has been successfully combined all-optical with OCT imaging [9]. Still, tissue contact with the sensor that is highly sensitive to deformations due to acoustic pressure was needed. Also, OCT imaging and sensor interrogation required different wavelengths, increasing the system complexity. A noncontact technique for optical detection of the PA signal based on surface displacement sensing was demonstrated using low-coherence interferometry locked to a position of high sensitivity [10]. This could potentially be combined with time domain OCT imaging, which would, however, require axial scanning of the reference arm. In the present work we demonstrate, for first time to the best of our knowledge, a combined all-optical simultaneous OCT/PA imaging device sharing the same detection unit. We employ high-speed swept source (SS) technology that allows for both phase-sensitive measurement of the PA signal and OCT imaging. In SS OCT, the light source k spectrum is scanned over time with sweep rates of up to several megahertz. Hence, the necessary sampling rate of up to 1 GHz easily allows for sampling oscillations of typical PA signals. Furthermore, being an interferometric technique, SS OCT resolves the phase difference between reference and sample arm over time, giving access to rapid small path length changes, such as interface displacement due to an impinging PA wave. The spectral intensity I(t) of interference from a single reflective surface in the sample arm is given by

\[
I(t) \propto \cos(\varphi(t)) = \cos(2kt[\Delta z_0 + \Delta z_p(t)]).
\]  

where \(\Delta z_0\) is the static path length difference between reference and sample arm, \(\Delta z_p(t)\) is the PA pressure-induced surface displacement that allows us to reconstruct the distribution of absorbing sample constituents. Note that we only consider a single spectral scan or sweep. Eq. (1) is similar to frequency modulation (FM) known from telecommunication. It is equivalent to heterodyne detection where the carrier frequency is given by the path length difference \(\Delta z_0\) modulated by the vibration. It has the advantage of shifting the detection bandwidth away from the low frequencies and the 1/f noise in particular. Furthermore, we can assume that the amplitude of vibration remains much smaller than \(\Delta z_0\) and \(\Delta z_p\). According to Carson’s rule, this leads to narrow band operation with a bandwidth requirement that is twice the bandwidth of the acoustic signal [11]. We further take advantage of the large specular reflection occurring at
the sample-air interface to measure the spectral phase $\phi(t)$ evolution with high sensitivity.

A schematic of the experimental setup is shown in Fig. 1. The light source is a Fourier-domain mode-locked (FDML) laser operating at 110 kHz sweep rate, and centered at $\lambda_0 = 1310$ nm with a 120 nm full bandwidth giving $\sim 12 \mu m$ axial resolution in air. FDML lasers are particularly apt for phase-sensitive measurement as they provide high phase stability due to pseudostationary operation [12]. The interferometer is a fiber-based Michelson using a circulator (CIRC) and a 50/50 fiber coupler FC. In the sample arm, the light, after collimation through L1, is focused onto a $\sim 30 \mu m$ spot size in the sample with an achromatic lens L2 ($f = 30 \text{ mm}$). The interference signal is measured with a dual balanced detector (DBD, Thorlabs, PDB150C). The signal is digitalized at 250 MSamples/s (DAQ, Alazartech, ATS9870). The PA excitation is performed with a Q-switched pulsed laser operating at 532 nm (Coherent, Corona). A dichroic mirror D is used to have its beam collinear to the OCT beam. The Q-switched laser beam diameter before L2 was $\sim 8 \text{ mm}$. A function generator GEN is used to drive the FDML with a sine function. The sync channel to that signal serves as trigger to the DAQ. The function generator controls the repetition rate of the pulsed laser that was set close to its maximum value at 5 kHz. Every channel was phase locked to ensure that the pulse occurs at the same time compared to the sweep. The pulse width is $\sim 250 \text{ ns}$, which limits the axial resolution to $\sim 375 \mu m$ by considering a speed of sound in tissue of $c \sim 1500 \text{ m/s}$. The sample is placed on a linear stage LS in order to measure a full tomogram.

We take advantage of the fast sweep rate to measure several spectra per pulse, i.e., one M-Scan per pulse. The pulse trigger is phase shifted relative to the start of the M-Scan so that the first sweep can be used as reference during which no vibration occurs. For millimeter PA imaging and the given speed of sound in tissue $c$, it is sufficient to use a single microsecond-long sweep for phase analysis.

Figure 2 presents the processing steps required to extract the vibration signal from the interference intensity dataset of a single sample interface. In the following, for sake of simplicity, we suppose a linear sweep in $k$. This assumption is also a good approximation for the central part of a sinusoidal sweep [see Fig. 5(a)]. After Fourier transform of the recorded spectrum, the obtained depth profile $I(x)$ is multiplied with a rectangular filter of width $B$ centered at the interface location. The filter selects the depth of interest and reduces the phase noise. The choice of $B$ is therefore a trade-off between phase sensitivity and temporal-axial resolution of the acoustic signal. The filtered spectral phase $\phi(t)$ is assessed by unwrapping the argument of the filtered signal after inverse Fourier transform (IFFT). These steps are repeated for each sweep in the M-Scan. By taking the difference of the extracted spectral or temporal phase to that of a reference sweep, PA signal independent phase terms can be rejected in $\Delta \phi(t)$. The resulting time-dependent displacement $\Delta z_p(t) = \Delta \phi(t)/2k(t)$ allows then for PA reconstruction. In order to improve the SNR and the phase sensitivity further, we average the phase difference over several M-Scans or pulses per spatial position.

For a proof of principle, a phantom was made consisting of a black tape embedded in agar gel. The tape was tilted relative to the phantom surface, thus leading to increasing acoustic delays over the lateral scanning axis $x$. 200 M-Scans were acquired at each lateral position. Figure 3(a) shows the referenced spectral phase traces for the central part of the sweep with and without optical pulse excitation. The PA signal can be well resolved. The full width at half-maximum (FWHM) is $\sim 200 \text{ ns}$. The power of the OCT and the pulsed laser was $6$ and $150 \text{ mW}$, respectively. The numerical filter was centered at a carrier frequency of $15 \text{ MHz}$ ($\Delta z_p = 1 \text{ mm}$) with a bandwidth $B$ of $13 \text{ MHz}$. This value does not reduce the PA axial resolution since it is already limited by the laser pulse duration. Typical SS systems have a SNR decay with increasing delay due to finite instantaneous spectral linewidth, in our case $\sim 1.2 \text{ dB/mm}$. Hence, the interface delay has to be chosen so as to optimize SNR for a given

![Fig. 1. Experimental setup for OCT imaging and phase-sensitive measurement of PA signals.](image1)

![Fig. 2. Processing steps to extract the vibration signal from the interference signal.](image2)

![Fig. 3. (a) Part of sweep spectrum and fringe intensity. (b) Phase difference over part of the sweep duration with and without optical pulse excitation.](image3)
To further determine the system sensitivity, we measured the phase difference standard deviation without PA signal [Fig. 3(b)]. The analysis was limited to the range over which the spectrum envelope is larger than \(1/e\) of its maximum. The resulting time span of \(5 \mu s\) corresponds to a PA ranging depth of 7.5 mm. The value of \(\sim 61\) pm is obtained over an average of 200 measurements. The differential phase standard deviation \(\sigma_{\Delta \phi}\) and associated displacement sensitivity (DS) \(\sigma_{p}\) depend in the shot noise limit on the SNR as

\[
\sigma_{\Delta \phi} = \frac{\lambda_0}{4\pi} \Delta \phi_p \approx \frac{\lambda_0}{4\pi} \sqrt{\frac{B}{W \text{SNR}}},
\]

\[
\sigma_{p} = \frac{\lambda_0}{4\pi} \Delta \phi_p \approx \frac{\lambda_0}{4\pi} \sqrt{\frac{B M/2}{W \text{SNR}}}
\]

with \(W\) being the electronic detection bandwidth, \(N\) the number of averaged differential phases \(\Delta \phi(t)\), and \(M\) the number of pixels over which the FFT is calculated, in our case \(M = 2000\) [2,12,13]. The measurement of the spectral SNR SNR\(_r\) is approximated by the more accessible measurement of the SNR after FFT SNR\(_f\). From Eq. (2), the smaller intensity at the beginning and end of the sweep causes a higher phase noise floor, as visible from Fig. 3(b). The signal at the interface had a SNR\(_f\) of \(\sim 63\) dB. With \(W = 50\) MHz, the theoretical DS in the shot noise limit is \(\sim 84\) pm. The experimental value is smaller since it is calculated only for a limited spectral range as outlined above. By considering an ultrasonic frequency of 6 MHz and according to [13], the experimental DS corresponds to a pressure sensitivity of \(\sim 1700\) Pa. The SNR of the PA signal, defined as the peak value divided by the noise rms in Fig. 3(b), is \(\sim 34\).

We then measured 40 different lateral positions over 8 mm (Fig. 4). First the focus was set on the phantom surface to maintain high SNR for PA detection [Fig. 4(b)]. The simple sample geometry and small detection spot size allow straightforward PA reconstruction in which the time coordinate of \(\Delta \phi\) measured at a spatial location is converted to a PA depth profile by knowledge of \(c\) in a similar fashion that the group refractive index, in our case of 1.33, is used to convert time delays to path length differences in OCT. The procedure is repeated for every lateral position to obtain a PA tomogram. The increasing noise floor caused by low interference amplitude at the borders of the differential phase plot is visible at the top of the picture. For OCT imaging we axially displaced the sample to position the black tape signal within the OCT measurement range [Fig. 4(a)]. Figure 4(c) shows the superposition of both signals using the red (PA) and green (OCT) channels of an RGB representation.

Sufficient SNR and dynamic range is required to obtain high DS for PA sensing. This might be a drawback for measuring PA signals at rough sample interfaces. Still, the topical application of an oil film could significantly improve the SNR [10]. The DS can be further optimized by switching to shorter wavelengths employing 1060 nm laser technology. A further improvement concerns the pulse width of the excitation laser. Current PA systems utilize pulse widths <10 ns. The axial resolution improvement comes at the cost of sensitivity since the filter width

\[B\]

would be larger. Nevertheless, it is expected that the acoustic generation efficiency is increased with shorter pulse widths. Finally, an even higher repetition rate would allow either employing increased averaging or decreasing the total acquisition time that remains critical for in vivo imaging. Optimally, the pulse repetition rate is half the SS OCT sweep rate allowing for one sweep as reference and the next scan recording the PA signal. Altogether, an improvement in DS by a factor of 10 at least is feasible.

In conclusion, we introduced a method to extract the PA signal generated by an absorbing structure from a SS OCT signal. It was achieved in a noncontact fashion directly on the sample surface by using its specular reflection. It offers a simple way to perform OCT and PA imaging by sharing the same system components.

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