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Autocorrelation optical coherence tomography for mapping transverse particle-flow velocity

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Abstract

We present an autocorrelation method to quantitatively map transverse particle-flow velocity with a Fourier domain optical coherence tomography system. This method is derived from the intensity fluctuation of the backscattered light modulated by flowing particles. When passing through the probe beam, moving particles encode a transit time into the backscattered light. The slope of the normalized autocorrelation function of the backscattered light is proportional to the transverse velocity. The proposed method is experimentally verified using intralipid scattering flow phantom.

Keywords

optical coherence tomography; transverse velocity; autocorrelation

Quantitative and non-invasive mapping of blood flow velocity *in vivo* is important for diagnostic and therapeutic purposes. Optical coherence tomography (OCT), especially after its advent of Fourier domain OCT, is a promising tool for providing high speed and high sensitive 3D images of blood flow velocity within biological tissues [1–5]. Recently, a number of literatures describing optical coherence tomography of flow velocity have been presented [6–9]. Most of them are derived from the Doppler shift or Doppler broadening of bandwidth. In this letter, we demonstrate an autocorrelation method for quantitative mapping of transverse particle-flow velocity with a Fourier domain OCT system. As opposed to the Doppler-based methods that analyze the Doppler effect of a moving particle on the probing light, the present method utilizes the statistical nature of the intensity fluctuation of backscattered light modulated by flowing particles. Particles follow flow without flip, and their moving behavior is quickly disturbed because of Brownian motion. However, particle motion may be “frozen” during a short period of time. This is analogous to Taylor’s frozen turbulence hypothesis [10]. When a particle transits through a probe beam, it continuously back-scatters the incident light along its path. This results in a relatively strong backscattered light pulse with a width of $\tau_0 = w/v$, i.e., the transit time, where w and v are respectively the transverse size of the probe beam and the transverse velocity. Thus, the backscattered light received by a detector is encoded with information about the flowing particles and becomes strongly correlated within a time width identical to the transit time τ_0 .

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Suppose stochastic particles transversely pass through a probe beam, the detected temporal interference fringe at the depth of z is given by

$$I(z, t) = A(z, t) \cos(2knz - \phi_1(z, t) - \phi_2(z)), \quad (1)$$

where $A(z, t)$ denotes the magnitude modulation that mainly depends on the backscattering light by the particles, $\phi_1(z, t)$ is the Doppler related phase, $\phi_2(z)$ is the phase that depends on the path length difference between the reference and sample arms. When a particle transits a probe beam, it causes relatively strong backscattered light with a pulse width identical to the transit time $\tau_0 = w/v$. So, $A(z, t)$ may be approximated as a sequence of rectangle functions,

$$A(z, t) = \sum_{i=1}^N M_i(t_i, z, \tau_0) \text{REC}(t_i, z, \tau_0), \quad i=1, 2, 3 \dots N$$

$$\text{REC}(t_i, z, \tau_0) = \begin{cases} 1, & t_i \leq t \leq (t_i + \tau_0) \\ 0, & \text{otherwise} \end{cases} \quad (2)$$

where the Brownian motion is neglected, the subscription i denotes the contribution by the i th particle that moves through the probe beam, N is the total number of the particles that pass through the probe beam, t_i denotes the beginning time to pass through the probe beam for the i th particle, $\text{REC}(t_i, z, \tau_0)$ represents a unit rectangle function with a width of τ_0 , and $M_i(t_i, z, \tau_0)$ is the magnitude modulation for the i th particle passing through the probe beam. From Eq.(2) the normalized autocorrelation function of $A(z, t)$ can be written as,

$$\frac{R(z, \tau)}{R(z, 0)} = \begin{cases} 1 - \frac{\tau}{\tau_0}, & \tau \leq \tau_0 \\ 0, & \tau > \tau_0 \end{cases} \quad (3)$$

where $R(z, \tau)$ denotes the autocorrelation function of $A(z, t)$ with a time lag of τ . Eq.(3) is identical to the equation used for measurement of fluorophores flow with fluorescence correlation spectroscopy [11]. Thus, one can obtain the transverse velocity v from the above equations.

The experimental setup used in this work is similar to that used in our previous work [9]. Here we briefly describe its parameters. We used a superluminescent diode as the light source with a central wavelength of 1310nm and a bandwidth of 56nm that provided a $\sim 13\mu\text{m}$ axial resolution in air. In the sample arm, we used a lens with a focal length of 30mm to achieve a measured effective probe beam waist of $\sim 16\mu\text{m}$ (ω). The output light from the interferometer was coupled into a home-built spectrometer. The frame rate of the line scan camera was 47KHz. Each M-scan, i.e. repeated scan at the same location, comprised of 1000 A-lines. The system phase-noise floor was experimentally determined at ~ 5 mrad.

In order to test the proposed method, we measured the transverse velocity of 2% intralipid scattering flow in a plastic tube (inner diameter 1mm and out diameter 1.5mm). The tube was adjusted to be approximately perpendicular to the sample beam. In doing so, we first used phase-resolved Doppler OCT [4,5] to measure the axial components (vertical) of the flow velocity; and then we adjusted the tube orientation until the measured phase values fell within the phase noise floor of system (which is 5 mrad). Thus, the vertical velocity would have a negligible effect on the measurements of transverse velocity, because the maximum

possible vertical velocity was less than ~ 25.5 microns/s in this case (assuming that the refractive index of the fluid phantom is 1.35).

Fig. 1(a), (b) and (c) show the magnitudes of the detected light backscattered by flowing intralipid scattering solution with different transverse velocities controlled by a precision syringe pump: (a) 3.20mm/s, (b) 1.18mm/s and (c) 0.64mm/s. These signals were acquired from a single position near the center of the tube with a sampling frequency of 47KHz. From Fig. 1(a), (b) and (c), it can be clearly seen that the magnitudes of the backscattered light are modulated by the flowing particles and the modulation durations vary inversely with the transverse velocities. The normalized autocorrelation functions of the three signals are shown in Fig. 1(d) as the solid line (0.64mm/s), dash line (1.18mm/s) and dot line (3.20mm/s). The slopes of the normalized autocorrelation functions (calculated using the zero time lag and the first zero of the autocorrelation function) are approximately proportional to the transverse velocities. These experimental results are in agreement with the theoretical analysis.

The experimental results for quantitatively validating the proposed method are shown in Fig. 2. The dot line denotes the measured transverse velocities at a single position near the center of the tube with a sampling frequency of 47KHz, and the circle line represents the calibrated transverse velocities. For the case of relatively faster flow, the measured results are in excellent agreement with the calibrated velocities determined from the precision pump. However, relatively large deviations are observed for the case of relatively slower flow. This is because of the limitation of the dynamic range of the measurable velocity and the influence of Brownian motion (The detailed discussion is given below). Two profiles of the transverse velocities along a depth-scan passing through the center of the plastic tube are shown as the dots in Fig. (3)(a) and (b), where the maximum transverse velocities were respectively (a) 3.2mm/s and (b) 2.55mm/s as controlled by the syringe pump. The solid lines are the parabolic fits of the measured results. The flow velocities decrease approximately parabolically from the center to the wall, which is expected.

Unlike the phase-resolved Doppler OCT that has a limited dynamic range of velocity caused by 2π ambiguity in the arctangent function [4], the present method has no limitations related with phase wrapping. The maximum and minimum measurable transverse velocities depend on the integration time of the camera (the reciprocal of the frame rate) and the total time for an M-scan (the maximum probe beam dwell time at the moving particle, 0.0213 sec in this study), which limit the transit time τ_0 that can be measured. For the scanning parameters used for this study demonstrated above, the theoretical dynamic range is from ~ 0.7 mm/s to several ten mm/s. Fig. (3)(a) and (b) show that the minimum measurable velocity is ~ 0.64 mm/s, which is close to the theoretical prediction. For the case of slow particle flow, the random Brownian motion and the non-stabilization of the experimental setup would also influence the backscattered light and it causes nonlinear attenuation of the normalized autocorrelation function (the solid line in Fig. (1)(d)). This would result in the large deviations of the calculated slopes of the normalized autocorrelation. To mitigate these limitations, one may increase the record time for M-scan so that the system is sensitive to the slower velocities.

In conclusion, we demonstrate an autocorrelation method capable of mapping transverse particle-flow velocity with a Fourier domain OCT system. As opposed to the current Doppler-based methods, this method uses the stochastic nature of the intensity fluctuation of the backscattered light modulated by moving particles. The method is experimentally verified with intralipid scattering flow. By combining the Doppler-based methods with the method proposed in this paper, one may image the omnidirectional velocities of the moving particles.

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Fig. 1. The magnitudes of the detected light modulated by flowing intralipid scattering particles with a transverse velocity of (a) 3.20mm/s, (b) 1.18mm/s and (c) 0.64mm/s, and (d) the normalized autocorrelation functions.

Fig. 2.

The measured transverse velocities (dot line) and calibrated velocities (circle line).

Fig. 3.

The dots denote the profiles of the transverse velocities along a depth-scan passing through the center of the plastic tube, the solid lines are the parabolic fits of the measured results, with the maximum transverse velocities controlled by the syringe pump at (a) 3.2mm/s and (b) 2.55mm/s, respectively.