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Real-time phase-resolved functional optical coherence tomography by use of optical Hilbert transformation

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We have developed a novel real-time phase-resolved functional optical coherence tomography system that uses optical Hilbert transformation. When we use a resonant scanner in the reference arm of the interferometer, with an axial scanning speed of 4 kHz, the frame rate of both structural and Doppler blood-flow imaging with a size of 100 by 100 pixels is 10 Hz. The system has high sensitivity and a larger dynamic range for measuring the Doppler frequency shift that is due to moving red blood cells. Real-time images of *in vivo* blood flow in human skin obtained with this interferometer are presented. © 2002 Optical Society of America

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Optical coherence tomography¹ (OCT) is a recently developed imaging modality that uses coherent gating to obtain micrometer-scale, cross-sectional images of biological tissues. OCT is analogous to ultrasound imaging. However, OCT uses near-infrared optical waves instead of sound waves. Consequently, the spatial resolution of OCT is more than an order of magnitude better than that of the ultrasound. A number of extensions of OCT capabilities for functional imaging of tissue physiology have been developed. Optical Doppler tomography, for example, combines the Doppler principle with OCT to yield high-resolution tomographic images of tissue structure and blood flow simultaneously.²⁻⁶ Spectroscopic OCT combines spectroscopic analysis with OCT to yield depth-resolved tissue absorption spectra.⁷ Light is also a transverse wave, and one can use the polarization state of light to acquire information on tissue physiology. Polarization-sensitive optical coherence tomography (PS-OCT) combines polarization-sensitive detection with OCT to determine tissue birefringence.⁸

Similar to Doppler ultrasound, optical Doppler tomography, or Doppler OCT, has been developed to image the microvasculature in patients when blood-flow monitoring is important.^{2-4,9} However, one must overcome several technical challenges to obtain images at high speed with high velocity sensitivity. Here, we describe an approach that utilizes the polarization properties of light to obtain phase information on the Doppler frequency shift caused by moving particles inside the sample under study. This technique has the following advantages: (1) A stable high-frequency phase modulator is not required and (2) no Fourier or Hilbert transformation by computer is necessary. The frame rate can be as high as 10 Hz for both structural and Doppler blood-flow imaging, with a size of 100 by 100 pixels.

Optical Doppler tomography is based on the principle that moving particles, such as red blood cells inside biological tissues, cause a Doppler frequency shift in the OCT signal. Several techniques have been developed to detect the Doppler frequency shift. The most straightforward method is spectral analysis using a fast Fourier transformation algorithm.²⁻⁴ However, this method depends on the fast Fourier transformation window time, which limits

axial scanning speed and spatial resolution when one is measuring slow-moving blood flow in small vessels, which requires high velocity sensitivity. A phase-resolved technique has been developed^{6,9,10} that can decouple the Doppler sensitivity and the spatial resolution while maintaining high axial scanning speed. Phase-resolved functional optical coherence tomography (F-OCT) is based on a Hilbert transformation that converts the OCT signal from the real to the complex domain to yield phase information at each pixel. In addition to structural imaging, phase-resolved F-OCT has been used to image and quantify physiological parameters, including blood flow, birefringence, dispersion, and spatial phase variation.^{6,8,10-12} For blood-flow imaging, the phase difference between sequential axial scans at each pixel can be used to determine the Doppler shift. Because the time interval between sequential scans is much longer than the pixel window time, the sensitivity of the phase-resolved technique is much higher than that of the fast Fourier transformation method.⁶

The limitation of phase-resolved F-OCT is an aliasing phenomenon caused by 2π ambiguity in the arctangent function, which limits the maximum determinable Doppler shift to half that of the axial scanning rate.¹³ Many F-OCT systems utilize a rapid-scanning optical delay line (RSOD) based on a galvanometer to provide a maximum axial scanning rate of 1 kHz. Consequently, the maximum determinable Doppler shift is only 500 Hz in such systems. Although a phase-tracing technique can extend this limited range,¹³ such an approach is helpful only in special cases in which the spatial variation of the flow velocity is small and smooth. Another solution to this problem is to replace the galvanometer in the RSOD with a faster device such as a resonant scanner, to provide a scanning rate of 4–8 kHz. However, faster scanning rates lead to other problems. First, the photoamplifier bandwidth must be sufficiently wide because, due to the fact that the modulation frequency is proportional to the scanning rate, it is more difficult to design a circuit with low noise and wide bandwidth. Second, a higher modulation frequency requires a higher data acquisition rate, which complicates the system design. Third, a higher data acquisition rate requires faster processing speed for

real-time imaging. Current processors in personal computers are not fast enough to process data in real time. Therefore, a custom-designed digital signal processor or hardware demodulation may be necessary to accelerate processing speed. In this Letter we describe a novel technique for overcoming the problems mentioned above by incorporating an optical Hilbert transformation into our phase-resolved F-OCT system.

The design of a phase-resolved F-OCT that incorporates an optical Hilbert transformation is illustrated in Fig. 1. The interferometer is constructed on a custom-designed optical table with five separate fiber ports connected to the light source, reference arm, sampling arm, and two photodetectors. The light source has a bandwidth of 80 nm centered at 1310 nm, which results in an axial resolution of 9 μm . The source light is linearly polarized (45°) by a polarizer (P1) after it is coupled into port A and is then split into the reference and the sampling arms of the interferometer. There is an identical 45° polarizer (P2) in the sampling arm so that light coming back from port B can maintain a 45° linear polarization at the beam splitter. We insert a quarter-wave plate and a linear polarizer (P3) into the reference arm after port C to change the polarization state of the light from 45° linear polarization to circular polarization at the beam splitter. The beam splitter then splits the interference signal into two channels, received by photodetectors 1 and 2.

The two interference fringe signals detected from photodetectors 1 and 2 are proportional to

$$\begin{aligned} S_1(t) &= A(t)\cos(\omega t + \phi), \\ S_2(t) &= A(t)\sin(\omega t + \phi), \end{aligned} \quad (1)$$

where $A(t)$ is the amplitude of the OCT signal determined by the convolution of the backscattering light intensity from the sample with the power spectrum of the partial coherent source, ω is the phase-modulation frequency, and ϕ is the phase difference between the reference and the sampling arms. The two interference signals have a 90° phase difference because of interference between circular polarization and 45° linear polarization. The two signals are treated as real and imaginary parts of the complex signal, which allows us to determine both amplitude and phase information.

A RSOD based on a 4-kHz resonant scanner is used in the reference arm for axial scanning. The resonant scanner is aligned at the center position so that no phase modulation will be introduced during scanning. The advantage of a phase-resolved F-OCT that incorporates optical Hilbert transformation is that a modulation device is not required in the reference arm, as noted in Eqs. (1). In conventional F-OCT, phase modulation is required for generation of a carrier frequency that is determined by the axial resolution and the scanning frequency. The phase-modulation frequency is usually chosen so that there is a minimum of two fringes per coherence length scanned. Because no phase modulation is required in the current F-OCT system, the detector bandwidth, which is determined

by the phase-modulation frequency, can be reduced by at least a factor of 2 (2 MHz in our system). Moreover, the data acquisition rate can also be significantly reduced (5 MHz in our system) because we do not need to capture the full fringe generated by the phase modulation. Finally, the computer does not need to perform the Hilbert transformation, which was the most time-consuming process in our previously described phase-resolved F-OCT system.^{6,10} As a result, the computer can process and display the Doppler images in real time. The frame rate can be as high as 10 Hz for both structural and Doppler imaging, with a size of 100 by 100 pixels, which is very useful when one is monitoring blood-flow changes in a specific vessel.

A 1 × 2 50/50 coupler is inserted into the reference arm between the RSOD and port C for balanced detection by use of differential amplification by photodetectors 1 and 2. Balanced detection removes the low-frequency background noise caused by different reflections of the RSOD during axial scanning and reduces the intensity noise. The system's dynamic range is slightly more than 80 dB, which is smaller than that of the previous system, which used a nonresonant galvanometer. The reduction in the dynamic range is due mostly to the higher scanning rate used in our current system.

Figure 2 shows an *in vitro* example of particle flow imaging with phase-resolved F-OCT with optical Hilbert transformation. The model is a glass tube (inner diameter, 500 μm) with 1% Intralipid flowing through the tube at a velocity of 2 mm/s. Figure 2A shows the structural image, which is distorted by the refractive-index difference between glass and air. Sine oscillations of the resonant scanner will also produce nonlinear changes during axial scanning, which further contributes to image distortion. Figure 2B shows a Doppler shift image representing flow

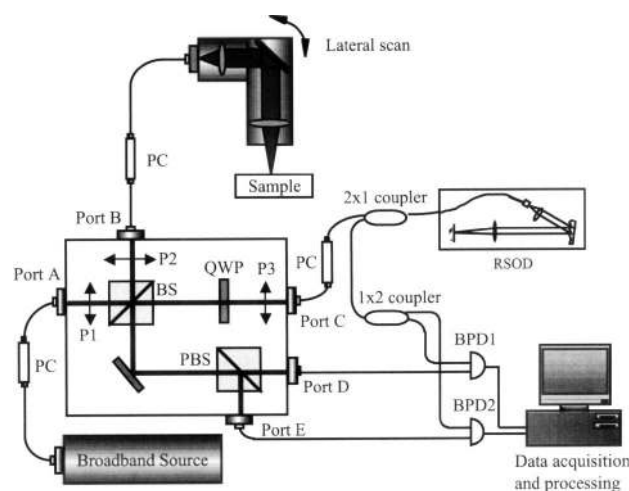


Fig. 1. Design of a phase-resolved F-OCT system that uses polarization to perform optical Hilbert transformation. Ports A–E, pigtailed fiber ports; P1–P3, linear polarizers; QWP, quarter-wave plate; BS, nonpolarized beam splitter; PBS, polarization beam splitter; PCs, polarization controllers; BPD1, BPD2, balanced photodetectors with preamplifiers.

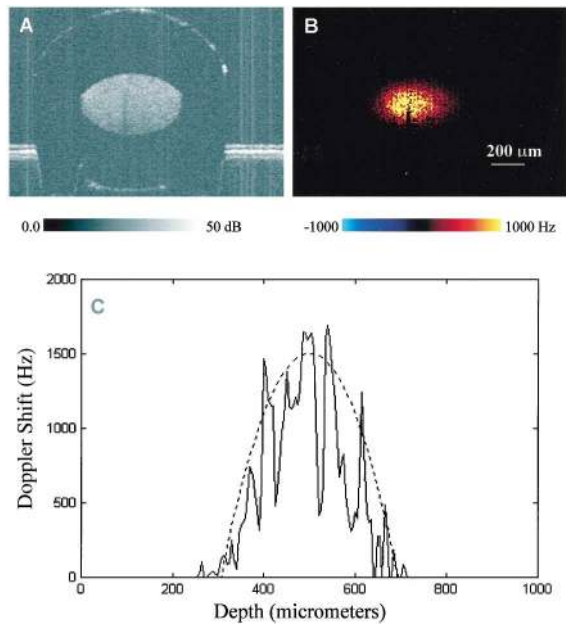


Fig. 2. *In vitro* particle flow imaging in a glass tube: A, structural image; B, Doppler shift image; C, Doppler shift profile along the center of the tube.

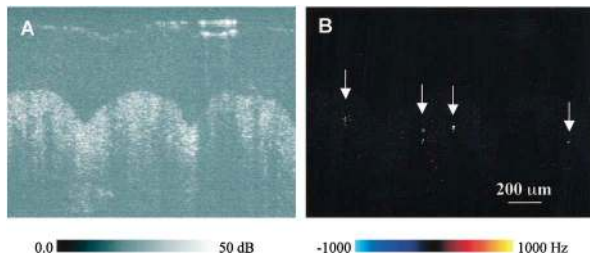


Fig. 3. *In vivo* results from human skin: A, structural image; B, Doppler shift image.

velocity. The angle between the probe beam and the flow direction is 74° . Both the maximum and the average Doppler shifts agree very well with theoretical calculations. Figure 2C shows the Doppler shift along the center of the tube, which, as predicted by theory, approximates a parabolic distribution.

Results of *in vivo* experiments with human skin are presented in Fig. 3. Images were taken from the finger of a human volunteer. Figure 3A and 3B show structural and Doppler shift images, respectively. Several small vessels can be identified at the epidermal–dermal junction, as indicated by the arrows in the Doppler image. The detected Doppler shift in these vessels is 500–1000 Hz (corresponding to a flow velocity of several hundred micrometers per second), which is consistent with the known capillary flow velocity. When the Doppler image was dynamically displayed on the computer monitor, the vessels alternately disappeared and reappeared in the Doppler images, because the motion of red blood cells is not constant in small vessels. However, the vessels always reappeared at identical positions in the

images, indicating that blood flow in these vessels was, in fact, detected.

In conclusion, we have developed a novel real-time phase-resolved F-OCT system that uses optical Hilbert transformation. This technique does not require phase modulation in the detection arm. Consequently, the technique reduces the requirements for the detection bandwidth of the photodetector and simplifies data acquisition and signal processing during real-time imaging. Our *in vivo* results demonstrate that this technique might be useful in the clinical management of patients when blood-flow monitoring is essential.

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