Image restoration method based on Hilbert transform for full-field optical coherence tomography

Jihoon Na,¹ Woo June Choi,¹ Eun Seo Choi,² Seon Young Ryu,¹ and Byeong Ha Lee¹,*
¹Department of Information and Communications, Gwangju Institute of Science and Technology, 1 Oryong-dong, Buk-gu, Gwangju, 500-712, Korea
²Department of Physics, Chosun University, 375 Seosuk-dong, Dong-gu, Gwangju, 501-759, Korea
*Corresponding author: leebh@gist.ac.kr

Received 4 September 2007; revised 5 November 2007; accepted 9 November 2007; posted 15 November 2007 (Doc. ID 87016); published 18 January 2008

A full-field optical coherence tomography (FF-OCT) system utilizing a simple but novel image restoration method suitable for a high-speed system is demonstrated. An en-face image is retrieved from only two phase-shifted interference fringe images through using the mathematical Hilbert transform. With a thermal light source, a high-resolution FF-OCT system having axial and transverse resolutions of 1 and 2.2 μm, respectively, was implemented. The feasibility of the proposed scheme is confirmed by presenting the obtained en-face images of biological samples such as a piece of garlic and a gold beetle. The proposed method is robust to the error in the amount of the phase shift and does not leave residual fringes. The use of just two interference images and the strong immunity to phase errors provide great advantages in the imaging speed and the system design flexibility of a high-speed high-resolution FF-OCT system. © 2008 Optical Society of America

OCIS codes: 110.3010, 110.6955, 120.3180.

1. Introduction

As a medical diagnostic imaging modality, optical coherence tomography (OCT) allows one to obtain noninvasive multidimensional images of biological tissues with a high axial resolution [1,2]. The high axial resolution of OCT enables one to visualize the microstructured anatomy of biological tissues in a diverse range of clinical applications. By combining with other optical sensing techniques, OCT can provide not only the detailed structural information of a biological sample but also its functional information, such as a flow speed or a birefringence variation [3,4]. To overcome the limitations of a slow imaging speed and a low signal-to-noise ratio of a conventional OCT system, new optical schemes such as Fourier-domain and spectral-domain OCT systems have been studied [5,6].

Full-field OCT (FF-OCT) has been widely studied as a technique allowing concurrent high axial and transverse resolutions [7–10]. By performing in-vivo retina imaging, the feasibility of FF-OCT in a practical ophthalmic diagnostic application was demonstrated [11,12]. An FF-OCT system is usually based on a Mirau or Linnik interferometer configuration [13,14]. A spatially and temporally low coherent or incoherent light source and a two-dimensional array detection device, such as a CCD camera, are utilized in general. Since the axial resolution of FF-OCT is inversely proportional to the spectral bandwidth of the light source, a tungsten–halogen lamp or a xenon arc lamp is frequently adopted for a high-resolution imaging system due to its broad bandwidth. As two-dimensional array detectors, silicon-based CCD cameras are commonly used in the visible range, and InGaAs cameras have been exploited for infrared light detection [15].

The most powerful advantage of a FF-OCT system over conventional OCT systems is that the FF-OCT system creates a transverse cross-sectional or en-face image at a depth of a biological sample without doing any lateral scan. By successively taking the en-face images at different depths, we can get the depth-resolved en-face images and generate three-dimensional structural images of the sample. How-
ever, to retrieve one en-face image at a depth, in general, it is necessary to capture several interference fringe images by using the two-dimensional array detection device. Since an appreciable time is required to capture the images and great attention is necessary to adjust the phase for each image, various image restoration methods for reducing the number of image captures and simplifying the optical configurations have been proposed [11,16–22].

The commonly used image restoration method is using the phase differences among the captured images. According to the way of attaining the necessary phase shift, it is categorized into two; one is the discrete (phase-stepping) [7,8] method and the other is the continuous (integrating-bucket) [9,16] method. In most cases, more than three interference fringe images are required to retrieve one en-face image [17]. A number of studies have been undertaken in an attempt to reduce the number of required interference fringe images, since the image acquisition time is one of the critical factors for realizing a real-time system.

The heterodyne detection technique where the quadrature fringe images were obtained in a single acquisition by using a pair of CCD cameras and two liquid-crystal shutters was demonstrated [18]. Although this method could get two quadrature images directly, the synchronization between the movements of the reference arm and the shutters was challenging. The simultaneous use of a pair of CCD cameras with a quarter-wave plate was proposed to get ultrafast tomographic imaging [19,20], in which phase-opposed interference images were captured at the same time but with different polarization states. The methods based on sinusoidal modulation integrating-bucket techniques and two phase-opposed fringe images were proposed to achieve fast image reconstruction also [11,16,21]. With a single CCD, two CCD images were captured at the opposed phase states by adjusting the reference arm of the interferometer. However, the methods using two phase-opposed fringe images could not completely remove the interference fringes in the retrieved en-face image. Further, they need precise control to get the opposite phase. To avoid the problems caused by the incomplete cancellation of fringe signals, a four-step phase-shifting method with only two acquisitions was demonstrated by using a Wollaston prism and a single CCD camera [22]. However, the optical configuration of the system became a little bit complicated. Capturing and matching the two interference images in different polarization states and focused at different areas of a single CCD were also not simple.

In this paper, we propose the image restoration method that only needs two phase-shifted images but does not leave residual fringes in its retrieved en-face image, at least in principle. By using the mathematical Hilbert transform, the quadrature image necessary to remove the modulation signal could be obtained. The proposed method allows a large tolerance in the phase-shift error because of using the Hilbert transform, and enables a reduction in the image acquisition time because it needs only two image captures. By implementing an FF-OCT system via the use of a thermal light source, the feasibility of the proposed method is then verified. The spatial resolution and the sensitivity of the implemented FF-OCT system are subsequently analyzed, and the obtained high-resolution en-face images of biological samples, a piece of garlic, and a gold beetle (spilota plagicollis fairmaire), are presented.

2. Methods

A. Experimental Setup

The schematic of the proposed FF-OCT system is shown in Fig. 1. An illuminator (MILLE LUCE, M1000) having a 150 W customized quartz tungsten–halogen lamp was used as a broadband light source. It had a wide enough spectral bandwidth for a high-resolution system; however, the CCD camera as the detector array gave the limitation in the available or effective bandwidth of the source. The light from the illuminator was guided by a fiber bundle and split by a broadband beam splitter (BBS) to feed at both arms of the system; the reference arm and the sample arm. It was intended to collimate the input beam at both arms, and to focus the beams reflected from the sample and the reference mirror at the CCD plane.

In the reference arm, a silver coated mirror was used as the reference mirror, and a neutral density filter (NDF) was placed between the BBS and the reference mirror to adjust the interference contrast. The reference mirror was attached on a PZT (piezoelectric transducer) actuator in order to induce the phase shift to the interference fringes. In the sample arm, a glass plate (GP) was inserted to compensate the material dispersion caused by the NDF at the reference arm. A biological sample was placed on a high precision vertical motorized stage (0.1 μm in
At first, by taking the difference between two CCD point (generally expressed by the light intensity at each pixel) interference fringe pattern obtained by a CCD camera is determined by taking the square sum of two conjugate signals $S_1$ and $S_2$, and the new phase factor as

$$\Phi = \phi(x, y) - \left(\frac{\alpha}{2}\right). \quad (4.3)$$

Second, by taking the Hilbert transform [23] denoted by $H$, in Eq. (4.1) we can have the quadrature signal $S_2$ of the dc-free signal $S_1$ as

$$S_2 = H\{S_1\} = A'(x, y)\left[\cos\Phi(x, y)\right]. \quad (5)$$

Finally, the newly defined envelope signal is simply determined by taking the square sum of two conjugate signals $S_1$ and $S_2$,

$$A'(x, y) = \sqrt{S_1^2 + S_2^2}. \quad (6)$$

The interesting thing is that the new envelope signal $A'(x, y)$ is obtained regardless of the exact amount of the induced phase shift $\alpha$. As shown with Eq. (4.2), the new envelope signal has the same x-y dependency as the original envelop signal $A(x, y)$. The amount of phase shift $\alpha$ only affects the overall amplitude of the extracted signal. In other words, the amount of phase shift does not hamper the extraction of the envelope signal but affects the signal contrast only.

To confirm the effect of the induced phase shift on the signal contrast, by using the setup in Fig. 1, CCD image pairs of a flat mirror were taken at several phases. And then, by using the proposed process, the corresponding FF-OCT images of the mirror were extracted. As shown in Fig. 2, the maximum signal contrast occurred around at $\alpha = 180^\circ$, and then degraded smoothly with the deviation from it. The solid line of the figure was the one theoretically calculated by using Eq. (4.2). The two horizontal dotted lines of

![Fig. 2. Signal contrast of the proposed system measured in term of the applied phase shift $\alpha$. The solid line is the theoretical result. The signal has its maximum when the phase shift is $180^\circ$, and degrades with the error in the phase. Even with a phase error of $50^\circ$, the signal contrast is reduced by only 10% from its maximum. The measurements (data points) were made for every phase shift of $22.5^\circ$ from 0 to $180^\circ$.](image-url)
the figure show that even when the phase shift deviates from its most proper value of 180° by 50°, the signal contrast is reduced only by 10% from its maximum value. This measurement reveals that the proposed scheme has a large tolerance in the phase error. The discrepancy between the simulation and the experiment results was mainly due to the hysteresis of the PZT used for inducing the phase shift. For more accurate measurements, it is necessary to utilize an actuator operating in a closed loop control. Of course, since the system is a kind of an optical interferometer, we need to be careful in handling the vibration of the system also.

3. Results and Discussion

A. Image Resolution and Detection Sensitivity

As in the case of a conventional OCT, when the spectral shape of the light source is assumed as Gaussian, the theoretical axial resolution of FF-OCT is given as

$$\Delta z = \frac{2 \ln 2}{\pi \left( \frac{\lambda_s^2}{\Delta \lambda} \right)}$$

where $\lambda_s$ is the center wavelength of the light source and $\Delta \lambda$ is its full width at half maximum (FWHM). Although the white-light source used for the experiment had an ultrabroad bandwidth, the silicon-based CCD camera could not cover the whole spectral range of the source due to its limited spectral responsivity. Thus, the axial resolution of the implemented system was determined by the effective spectral bandwidth of the CCD camera, not by the light source. To
confirm the effect of the limited spectral responsivity of the CCD camera, the spectrum of the white-light source was measured with the spectrometer (HR4000, Ocean Optics) consisted of a similar silicon-based line CCD. Figure 3(a) shows that the measured spectrum had an overall Gaussian-like spectral shape. The FWHM and the center wavelength of the spectrum were about 164 nm and 0.6 µm, respectively. From this spectral measurement, and by using Eq. (7), the axial resolution of the system was estimated as 0.97 µm.

The actual axial resolution of the implemented FF-OCT system was measured by imaging the surface of a flat mirror. The mirror in the sample stage was moved along the axial direction (Z-scan) of the system in steps of 0.1 µm by using a high precision linear motorized stage. Figure 3(b) shows that the axial resolution was measured about 1 µm, which was very well matched with the calculated value of 0.97 µm.

To measure the transverse or lateral resolution of the system, the en-face image of a standard U.S. Air Force 1951 resolution test pattern was taken. As shown in Fig. 4, we could distinguish all lines within the group code number 7 of the test pattern. In that group, since the smallest line pair was 228 lp/mm, by taking the half-width between two line pairs, we can say that the transverse resolution of the system was better than 2.2 µm. As is well known, the transverse resolution of an imaging optics is determined by the numerical aperture (NA) of the used microscope objective (MO). In our experiment, a water-immersion MO having 10× magnification and 0.3 NA was used.

The minimum detectable reflectivity, which determines the imaging quality and the measuring depth of OCT, is approximately given by [22,23]

$$R_{\text{min}} = \frac{(R_{\text{ref}} + R_{\text{inc}})^2}{2N\zeta_{\text{sat}}R_{\text{ref}}}$$  (8)

where $R_{\text{ref}}$ is the reflectivity of the reference mirror, $R_{\text{inc}}$ is the reflectivity of the incoherent light to the reflection, $N$ is the number of data sets measured for average, and $\zeta_{\text{sat}}$ is the full-well capacity (FWC) of the CCD camera. Due to the small cell size of the CCD camera (7.4 µm × 7.4 µm), the FWC level was un-

![Fig. 6. OCT images of epidermal cells of a garlic. The microscopic images (left column; (a)–(c)) and the retrieved en-face images (right column; (b)–(d)) of the sample. The images were taken at two different depth positions; at the top surface (upper row; (a) and (b)) and at a plane 40 µm below the top surface (lower row; (c) and (d)). The image size of each figure is 512 µm × 452 µm.](image-url)
Fortunately, as low as $\zeta_{\text{sat}} = 4 \times 10^4$, so that the theoretical limitation of the sensitivity was only about 60 dB when no average was taken. Therefore, to improve the sensitivity, multiple measurements (200 times) were made and their average was taken. Figure 5 shows the measured system sensitivity. For the measurements, a glass plate whose top surface had a 2% reflectivity was placed under the MO in the sample arm, and the optical power of the beam coming from the reference arm was adjusted with the NDF until it saturated the pixel depth of the CCD camera. While moving the sample along the axial direction in steps of 0.5 μm, successive en-face images of the sample were taken. The figure shows that the sensitivity of the system was about 83 dB and the dynamic range was about 70 dB.

B. En-Face and Volumetric Imaging Performance

The feasibility of the proposed FF-OCT system, which was characterized mainly by using only two interference fringe images per each en-face image, was examined by imaging biological tissues ex-vivo.

As biological samples, a piece of garlic and a gold beetle (*spilota plagicollis fairmaire*) were used. The optical power incident on the sample was about 510 μW.

Figure 6 shows the optical microscope (OM) images (left column) and the extracted en-face OCT images (right column) of a piece of garlic, which were taken at two different depths; the top surface of the sample (upper row) and the surface 40 μm below it (lower row). The OM images (512 μm × 452 μm in size) [Figs. 6(a) and 6(c)] could be simply taken with the implemented FF-OCT setup by intentionally blocking the reference arm. With the OM images, it was not easy to distinguish the structure information of the sample along its depth. On the other hand, with the en-face images in the right column [Figs. 6(b) and 6(d)], we could get the depth-resolving structure information of the sample.

A volumetric OCT image of a gold beetle was also taken by using the implemented FF-OCT system. While moving the sample in steps of 0.2 μm, 570 en-face OCT images (433 μm × 325 μm in size) were taken. Figure 7 shows the reconstructed three-dimensional image of the beetle.

![Fig. 7. Volumetric OCT image of a gold beetle reconstructed with 570 en-face OCT images. Three different layers, the hard forewing, the flight wing, and the abdomen, can be clearly observed in the reconstructed three-dimensional image.](image-url)
acquired and combined into a single volumetric OCT image by using a volume rendering software (AMIRA 4.1, Mercury Computer Systems). As Fig. 7 shows, three layers of the sample can be clearly distinguished: the hard forewing, the flight wing, and the abdomen of the gold beetle. The two layers of the flight wing in the middle of the image are thought to result from folding of the wing. The cross-sectional views of the sample processed by the same rendering software are presented in Fig. 8; the volumetric OCT image [Fig. 8(a)] was sliced along the x-y plane [Fig. 8(b)], the x-z plane [Fig. 8(c)], and the y-z plane [Fig. 8(d)].

Fig. 8. (a) Different view of Fig. 7 presenting the image slicing direction, (b) x-y image (433 μm × 325 μm size), (c) x-z image (433 μm × 150 μm size), and (d) y-z image (325 μm × 150 μm size).

By using the mathematical Hilbert transform, we could extract one \textit{en-face} image from only two interference fringe images. The amount of the phase difference between the captured images was not critical. Moreover, the extracted image was free from residual interference fringes at least in principle, which opens the application to other fields; for example, the inspection of micro-optics. However, the mathematical Hilbert transform needed an appreciable calculation time depending on the performance of the personal computer. The acquisition time for taking one \textit{en-face} image with the implemented system was about 1 s. Most of the time was used to control the sample translation stage and the PZT actuator. At each position of the sample along its depth, we should capture two images at different phases. Therefore, the sample stage was moved in the go-and-stop mode, which limited the system rate. Further efforts should be made in customizing the sample stage and the PZT control in order to get a high system rate. The pure elapsed time for the mathematical Hilbert transform was in the order of taking Fourier transform. By introducing an embedded hardware, it is expected that the processing time can be reduced significantly.

As was emphasized several times, owing to introducing the mathematical Hilbert transform, the amount of the induced phase shift becomes not critical in extracting the \textit{en-face} image. The big tolerance in the phase control, in addition to using only two interference fringe images, is believed to allow the implementation of a high-speed FF-OCT system.

4. Conclusion
We have proposed an image restoration method for an FF-OCT system and demonstrated its performance by presenting some \textit{ex vivo} FF-OCT images of biological samples taken with an implemented sy-
The proposed image restoration method was characterized by using only two interference fringe images, an original and a phase-shifted, to retrieve one en-face image. To reduce the number of required fringe images while removing residual fringe patterns, a numerical Hilbert transform was utilized. It was found that the performance of the image restoration was not easily degraded by the error in the degree of the phase shift, though the maximum image contrast was obtained with a phase shift of 180°.

The high-resolution FF-OCT system implemented with a thermal light source and the proposed image restoration method had an axial and a transverse resolution of 1 and 2.2 μm, respectively. The system sensitivity was measured as 83 dB, based on the average of 200 accumulated images. The en-face images of a piece of garlic and a volumetric OCT image of a gold beetle showed that the proposed scheme worked well. The images clearly distinguished the micro-structured layers of the samples. Since the proposed image restoration method needs only two interference fringe images, even it does not leave residual fringes, the image acquisition time and the memory requirement of the system can be reduced. Further, the robustness in the phase error is believed to allow great flexibility in the system design, both in software and hardware.

This work was supported in part by the BK21 project, the Ministry of Commerce, Industry and Energy (MOCIE) of Korea through the Industrial Technology Infrastructure Building Program, the GIST Top Brand Project (Photonics 2020), and by MOCIE of the Advanced Technology Center (ATC) project.

References