Self-spectral calibration for spectral domain optical coherence tomography

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Abstract. A different real-time self-wavelength calibration method for spectral domain optical coherence tomography is presented in which interference spectra measured from two arbitrary points on the tissue surface are used for calibration. The method takes advantages of two favorable conditions of optical coherence tomography (OCT) signal. First, the signal back-scattered from the tissue surface is generally much stronger than that from positions in the tissue interior, so the spectral component of the surface interference could be extracted from the measured spectrum. Second, the tissue surface is not a plane and a phase difference exists between the light reflected from two different points on the surface. Compared with the zero-crossing automatic method, the introduced method has the advantage of removing the error due to dispersion mismatch or the common phase error. The method is tested experimentally to demonstrate the improved signal-to-noise ratio, higher axial resolution, and slower sensitivity degradation with depth when compared to the use of the zero-crossing method and applied to two-dimensional cross-sectional images of human finger skin.

Subject terms: optical coherence tomography; spectral domain; self-spectral calibration; interferometer.

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1 Introduction
Spectral domain optical coherence tomography (SDOCT) is a noninvasive, noncontact imaging modality,1,4 in which the depth profile of the tissue is retrieved by performing inverse Fourier transform of the interference signal generated by the light back-scattered from the interior of the sample. The light reflects from the reference mirror in a Michelson interferometer configuration. SDOCT has the advantages of wider dynamic range and higher signal-to-noise ratio (SNR) than the time-domain implementation.5 One limitation of SDOCT is that the spectrum is sampled uniformly in wavelength and unevenly sampled in wavenumber space (k-space) which leads to variations of the axial resolution and the image quality along with the axial or A-scan direction. Hence, accurate calibration of the spectrum is necessary to obtain high quality SDOCT cross-sectional images.6

Several methods have been proposed for spectrum calibration. One calibration method is based on a specially designed spectrometer that linearly samples the wavenumber.7–9 This method reduces post-processing needs but complicates system design. The other direct calibration methods proposed include a characteristic wavelength method10 and a spectral interferogram mapping method,11,12 which are based on measuring wave numbers at these zero-crossing points, the wave numbers at these zero-crossing points are known up to a scaling factor and an offset. Such methods take advantage of the linear relationship between phase or phase difference of interferograms and the wavenumber on CCD. However, they need separate measurement steps which may cause errors in calibration data because of environmental temperature fluctuation effects on the system or poor stability of handling practices. For the need of real-time imaging, a real-time or adaptive calibration routine is necessary. Mujat et al proposed a self-calibration method in which the intensity spectrum of the light source is modulated by an external sinusoidal signal in k-space with a thin glass slide inserted at the exit of the light source.13 Liu et al proposed an automatic spectrometer calibration method in which an accurate estimation of wavenumber assignment on CCD was obtained by adopting zero-crossing detection technique.15 Since the differences in wavenumbers between any two adjacent zero-crossing points are identical, by finding out all the zero-crossing points, the wave numbers at these zero-crossing points are known up to a scaling factor and an offset. Then spectrometer calibration can be realized through polynomial fitting process. The algorithm used in Ref. 16 is effective and simple and can achieve calibration with high precision. But this method is inherently unreliable in the cases where there are some large phase errors that arise from a large polarization mode dispersion mismatch between the reference arm and the sample arm, or when the phases of the calibration spectrum are contaminated by noises.

The automatic spectral calibration method presented here is based on two spectra that are from two different transverse positions of the tissue under observation. The use of the

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The mutual interference term of the output electric current is composed by the three interfaces within the tissue at the depth \(z_i\). Hence, in spite of the complexity of the tissue, amplitude reflectance \(R_i\) near the surface is much greater than that of other depths. When the digitized interference spectrum current is transferred to the computer and filtered by a band-pass digital filter with a centre frequency equal to the optical path difference (OPD) between the tissue surface and the reference mirror, approximately only the light reflected from the tissue surface is then obtained.

Before performing system spectrum calibration, direct inverse Fourier transforms were performed to all the interference spectra obtained at different transverse positions to form a two-dimensional (2-D) raw image. From the raw image, two of the interference spectra with different OPDs at the tissue surface in the spatial domain were chosen for calibration. The phases of the spectral signals were then extracted by performing Hilbert transform and further unwrapped.

From Eq. (1), it can be seen that the phase of the spectrum can be expressed as

\[ \phi_i(\lambda) = \frac{4\pi}{\lambda} z_i + \beta(\lambda). \]  

(2)

Hence the phase difference \(\Delta \phi(n)\) can be expressed as

\[ \Delta \phi(n) = \phi_B - \phi_A = \frac{4\pi}{\lambda_n} \Delta z, \]  

(3)

where \(\Delta z\) is the difference between OPDs from two points at the tissue surface, \(\phi_A\) and \(\phi_B\) are the phase distributions at the positions “A” and “B”, respectively, and \(n\) is the CCD pixel index. Obviously, the phase error term \(\beta(\lambda)\) has been canceled due to subtraction.

To facilitate the analysis, we introduce a phase difference function \(\Delta_i\) by the formula

\[ \Delta_i = \phi_i(\lambda_M) - \phi_i(\lambda_N), \]  

(4)

where \(M\) and \(N\) denote the two different pixel indexes of the CCD. The difference of OPDs between positions “A” and “B” can be rewritten as

\[ \Delta z = (T - 1) \cdot z_A, \]  

(5)

where \(T = \Delta_B/\Delta_A = z_B/z_A\). \(z_A\) and \(z_B\) are the OPDs of spectra at the positions “A” and “B”, respectively. The values of \(M, N, N < \) CCD pixels number) for “A” and “B” are then substituted into Eq. (3) to obtain each phase differences \(\Delta_A\) and \(\Delta_B\) and \(T\). The value of \(\Delta z\) is not determined because \(z_A\) is unknown. Since \(z_A\) cannot be measured directly by the position of the maximum of the signal in spatial domain because of the position error induced by the direct inverse Fourier transform of the spectrum which is nonuniformly sampled in \(k\)-space. An initial value of \(z_A\) is chosen according to the position of signal maximum in spatial domain by direct inverse Fourier transform without calibration. Since a more accurate spectral calibration means a better OCT image sharpness, we may decide the precise value of the \(z_A\) by an iterative algorithm by maximizing the sharpness of the reconstructed OCT A-scan.

When \(\Delta z\) is determined, the whole wavelength distribution on the CCD is determined up to \(\lambda_0\) at the first pixel index on the CCD.
3 Experiment System and Results

In our experiment, an SLD with a centre wavelength of 834 nm and a bandwidth of 40 nm was used as the light source, corresponding to the theoretical coherence length 7.7 μm in air. A 2 × 2 50/50 fiber coupler splits the incident light beam into two beams. The detection arm consists of an achromat collimating lens with a 60-mm focal length, a 1200-line/mm holographic grating, and an achromatic lens of 150-mm focal length for focusing the diffracted beam into the line-scan CCD camera that has 1024 pixels with a 14-μm pixel size, and a 12-bit digital resolution (e2v AViiVA SM2CL). The spectral resolution of the system that is determined by the CCD pitch length and the resolution of optical grating is 0.0674 nm, which gives a 2.59 mm maximum imaging depth in air. The spectra range covered by CCD is 69 nm.

The spectral data from CCD were converted and digitalized into digital signals and transferred to the computer by the image acquisition system. Then a one-dimensional (1-D) reflection distribution of the tissue along with the depth direction by inverse Fourier transform of the spectral data on the computer was obtained. Furthermore, by scanning the light beam across the tissue surface, the spectral data at different transverse positions were acquired and a 2-D tomographic image of the tissue was then generated by performing the inverse Fourier transform.

A raw cross-sectional image of a volunteer’s finger skin was obtained with our experiment system (see Fig. 2). The spectra chosen for calibration are marked by “A” and “B” in Fig. 2 and are depicted in Fig. 3(a) and 3(b), respectively. The corresponding filtered interference spectra are shown in Fig. 3(c) and 3(d). Figure 3(e) and 3(f) show the amplitude spectra at positions “A” and “B,” respectively, obtained with the sample replaced by a mirror. When Fig. 3(c) is compared with Fig. 3(e) or Fig. 3(d) is compared with Fig. 3(f), it can be seen that the fringe period of the filtered spectra at “A” and “B” are nearly equal to that obtained with the mirror, as sampled with the same OPD, respectively. It should be noted that the positions of “A” and “B” indicated in Fig. 2 can be any of the two positions in the image.

The unwrapped phase distributions of the filtered spectra and the wavelength distribution can be determined by the

![Fig. 2 Raw image of human skin at fingernail without spectral calibration. “A” and “B” denote the positions where the spectra were used for calibration.](image)

![Fig. 3 Measured interference spectra corresponding to “A” (a) and “B” position (b); the corresponding filtered interference spectra for “A” (c) and “B” (d), respectively; and the spectra obtained by using mirror as the sample at positions “A” (e) and “B” (f).](image)
procedure described in Sec. 2. The results are shown in Fig. 4. With these calculated wavelength distributions on CCD pixels, an interpolation and sequent inverse Fourier transform of the spectra of the volunteer's skin of a fingernail were carried out and a 2-D cross-sectional image with good quality was obtained, as shown in Fig. 5(b). For comparison, the image reconstructed without calibration is also given in Fig. 5(a). The zoomed-in views of two local areas in the two images marked by the rectangles were also overlying on the images. From the enlarged local images in Fig. 5(b), one can see that the human sweat glands, the skin surface, and the bottom layer of the fingernail can be clearly visualized after calibration.

4 Performance Analysis

4.1 System SNR and Axial Resolution Improvement

To quantitatively evaluate the performance of the presented calibration method, the SNRs and axial resolutions of the system with and without calibration were calculated. In the following analysis, the SNR is defined as

$$\text{SNR} = 20 \log \frac{I_{\text{max}}}{\sigma},$$

where $I_{\text{max}}$ is the maximum of the signal intensity in spatial domain, $\sigma$ is the background noise root-mean-square error. The axial resolution is defined as the full width at 3 dB less than the maximum of the point spread function (PSF). The interference spectra at several different OPDs when the sample was replaced by a mirror were measured. Inverse Fourier transforming of these interference spectra yield several 1-D signal distributions along with the depth direction that represent variations of the system PSF with depth. The SNRs and axial resolutions for the different OPDs were calculated by use of the obtained PSFs.

The PSFs of the system at different depths without calibration and with calibration are illustrated in Fig. 6(a) and 6(b). Then, the SNRs and axial resolutions at different depths...
were calculated according to measured PSFs and a third-order polynomial fitting were applied to the calculated data, respectively [see Fig. 6(c) and 6(d)]. One can see that the SNR and the axial resolution of the system without calibration decrease greatly with depth. For comparison, the quadratic fitting coefficient and first-order fitting coefficient of the SNR fitting curve are $-0.0003$ and $-0.037$ for the uncalibrated system and $-0.0001$ and $-0.003$ for the calibrated system with the proposed self-calibration method, respectively. The cubic fitting coefficients are both less than $10^{-7}$. These values indicate that the degrading speed of SNR has been halved after calibrating with our method.

The axial resolution of the uncalibrated system is about 50 μm at the depth of 1 mm, which is about five times larger than that at zero OPD and further deteriorates with depth [see Fig. 6(d), dotted line]. As a comparison, the averaged axial resolution of the calibrated system over the 2.5 mm range is close to the theoretical axial resolution and is almost unchangeable with the depth [see Fig. 6(d), dash-dotted line].

4.2 Comparison with Zero-Crossing Method
One advantage of our method over the zero-crossing method is that it has the capability of compensating the large phase errors. To clearly demonstrate this fact, we analyzed the same unwrapped phase distributions denoted by “A” and “B” in Fig. 2 and calibrated the system with zero-crossing method without the complex optimization for fitting coefficients.

To this end, first, the positions of the zero-crossing points are counted and plotted in Fig. 7(b). One can see from the Fig. 7(b) that a few zero-crossing points deviate from its true positions significantly. This may arise from some phase errors, as shown in Fig. 7(a) (black arrows). This fact shows that the phase errors in determining zero-crossing positions can affect the spectrum calibration precision further.

Using the zero-crossing calibration method with the spectra at “A” and “B,” the 2-D cross-sectional images of the same finger skin were generated, respectively [see Fig. 8(a) and 8(b)]. Compared with the calibrated image in Fig. 5(b), it can be seen that although the image quality could be improved by the calibrating system with the zero-crossing method, the method presented in this paper is more effective with clearer boundaries.

The calculated PSFs corresponding to different OPDs of the calibrated system with the zero-crossing calibrating method with the spectra at “A” and “B” [see in Fig. 8(c) and 8(d)] further validate the above conclusions. The SNRs and axial resolutions for different OPDs and their fitting curves were calculated and drawn in Fig. 8(e) and 8(f) together with the results obtained by our method. Both the quadratic fitting and first-order fitting coefficients of the SNR curve obtained with the calibrated system by the spectrum at “A” are more than three times than that of the corresponding coefficients with our self-calibration and the corresponding coefficients of “B” are 1.8 times and one-tenth, respectively. The results show that our method has a stronger capability of suppressing the decline speed in SNR with depth than the zero-crossing method.

Fig. 6 (a) PSFs at different depths without calibration; (b) PSFs at different depths with calibration based on our method; (c) the fitting curve of SNR variations; and (d) the fitting curve of resolution versus depth (dot line and dash-dot line represent the results without and with calibration, respectively).
The average 3 dB width of the PSF for the zero-crossing method with the spectra at "A" and "B" at the depth of 2 mm are almost five times and three times wider than that obtained with our self-calibration method, respectively. Hence, our method has an advantage of improving the axial resolution of the system due to the fact that the phase errors in the interference spectrum signal are removed by subtraction of the two complex spectrum phases, so higher calibration precision can be achieved.

In addition, by comparison between the two calibration results obtained with spectrum at "A" and "B" with zero-crossing method, it shows that the spectrum "B" is more appropriate for calibration than the one at "A." Based on this analysis, we can see that the conditions are stricter for the
zero-crossing method to choose the appropriate spectrum for calibration.

5 Conclusions
An automatic calibration method which allows calibration for SD-OCT system during scanning operation without external calibration light source and external measurement step is presented in this paper. The method allows choosing calibration spectrum from all the measurement spectra more randomly and can remove most of the phase error between the reference arm and the sample arm. There is also no need of a next fitting coefficient correction. Experimental results show that the axial resolution and depth sensitivity of the 2-D reconstructed images from spectral domain after calibration have been improved significantly.

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