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Contrast enhancement of spectral domain optical coherence tomography using spectrum correction

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ABSTRACT

We report a spectrum correction method to enhance the image contrast of spectral domain optical coherence tomography (SD-OCT). Our method treats SD-OCT signals as the product of harmonic signals backscattered from a sample comprising a series of discrete reflectors and a window corresponding to the light source spectrum. The method restores the magnitude of the main lobe of the axial point spread function (PSF) by estimating the magnitudes of the backscattered harmonic signals and strengthens OCT signals using these estimated values. Experimental results acquired from fresh rat corneas and fixed human aortic atherosclerosis tissues show that our method provides clearer microstructural information than the conventional methods by improving the contrast to noise ratios (CNRs) by 1.4779 dB and 3.2595 dB, respectively. This improved image quality is obtained without any hardware change, making our method a cost-effective alternative to compete with hardware advances.

1. Introduction

Optical coherence tomography (OCT) is a powerful non-invasive imaging tool that is widely used for three-dimensional imaging of biological tissue [1–3]. By measuring the magnitude and echo time delay of backscattered light using a low coherence light source, OCT can provide high-resolution images of the microstructure of biological tissue [4–7]. Therefore, it has become one of the most widely used imaging tools for disease diagnoses [6,8–10]. Improving image quality can be achieved by improving the axial resolution and contrast in both biomedical and industrial applications. Many works have been dedicated to improving OCT image quality by contrast enhancement methods using hardware or software-based methods. For example, better quality images have been obtained using exogenous contrast agents, such as gold nanoparticles, nanorods, and microspheres [11–15]. However, these methods confer higher hardware costs, are inconvenient to operate and have poor applicability. Software-based methods for OCT image contrast enhancement can be categorized into A-scan and B-scan based methods. Typically, A-scan based methods perform spectroscopic analysis to add new contrast such as particle size to images [16]. Nevertheless, previous A-scan based methods are computationally costly. The B-scan based methods, such as statistical modeling [17], histogram matching [18], multiple B-scan averaging [19], image fusion [20] and adaptive compensation [21,22], also improved OCT images of the retina, the optic nerve head, the lamina cribrosa, etc. However, B-scan based methods always can't deal with the contrast reduction caused by sidelobe artifacts well and clean the signal, especially for micro OCT (μOCT) with spatial resolution of 1–2 *um* [23–25].

OCT interference signals can be regarded as the product of harmonic signals backscattered from a sample comprising a series of discrete reflectors and a window corresponding to the light source spectrum. When using fast Fourier transform (FFT) to process SD-OCT spectral interference fringes, the FFT result is the convolution of the Fourier transforms of the light source spectrum and the backscattered harmonic signals. Windowing these backscattered harmonic signals using the light source spectrum broadens their main lobe and introduces sidelobe artifacts, which degrades OCT image resolution and contrast. Spectral shaping techniques have been thoroughly investigated to reduce sidelobe artifacts in axial PSF for non-Gaussian-shaped laser source spectra [26]. However, these methods might broaden the main lobe and detailed information of tissue microstructures maybe therefore lost.

In this paper, we report a spectrum correction method to enhance SD-

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OCT image contrast, from a new perspective of digital signal processing. By strengthening OCT signals through estimating the amplitudes of backscattered harmonic signals and restoring the amplitudes of the main lobe of the axial PSF with the estimated values, this method suppresses sidelobe artifacts and enhances SD-OCT image contrast. We preformed imaging experiments on fresh rat corneas and fixed human aortic atherosclerosis tissues to demonstrate the advantages of our method over the existing methods.

2. Methods

2.1. Fundamental methods

For a harmonic signal $B\cos(z_0 t)$, where t is a variable, z_0 and B are constants, the main lobe of its Fourier transform is $B\delta(z \pm z_0)$, which has a nonzero value B at $\pm z_0$ without any sidelobes. However, when it is multiplied by a window, windowing enlarges the full width at half maximum (FWHM) of its main lobe, reduces its main lobe amplitude and introduces sidelobe artifacts. In SD-OCT imaging, the interferometric signal can be expressed as $i_{OCT}(k)$, shown in Eq. (1),

$$i_{OCT}(k) = \frac{\rho}{2}\sum_{j=1}^{M}\left[s(k)\cdot\sqrt{R_R R_{Sj}}\cos\left[2k\left(z_R - z_{Sj}\right)\right]\right] = s(k)\cdot d(k) \qquad (1)$$

where R_R and R_{Sj} are the backscattering indices of the reference arm and the sample interfaces, k is the wavenumber, $s(k)$ is the spectrum of the light source, M is the number of interfaces in the sample between layers of different refractive indices, z_R and z_{Sj} are the axial positions of the reference reflector and $j-th$ interface in the sample, ρ is the responsivity of the detector, and $d(k) = \frac{\rho}{2}\sum_{j=1}^{M}\sqrt{R_R R_{Sj}}(\cos\left[2k\left(z_R - z_{Sj}\right)\right]$, which is the finite accumulation signal of the backscattered harmonic signals.

When OCT images are obtained using the Fourier transform $I_{OCT}(z)$ of $i_{OCT}(k)$, $I_{OCT}(z)$ can be expressed as Eq. (2),

$$I_{OCT}(z) = FFT[i_{OCT}(k)] = S(z)^{*}D(z) \qquad (2)$$

where z is the thickness position, $D(z) = \sum_{j=1}^{M}\frac{\rho}{4}\sqrt{R_R R_{Sj}}\delta[z \pm 2(z_R - z_{Sj})]$ is the Fourier transform of $d(k)$, $\delta(z \pm 2(z_R - z_{Sj}))$ is the Dirac delta function, and $S(z)$ is the Fourier transform of $s(k)$.

In Eq. (1), $i_{OCT}(k)$ can be regarded as the windowed result of $d(k)$ using $s(k)$ (here, we regard $s(k)$ and $d(k)$ as the window and the signal, respectively), and thus $I_{OCT}(z)$ is the convolution between $S(z)$ and $D(z)$, as shown in Eq. (2). The main lobe of the Fourier transform of each harmonic signal in $d(k)$ is shown in Fig. 1(c), and its amplitude is lower than A because the convolution between the Fourier transform of the harmonic signal and the sampling window has a smoothing effect and

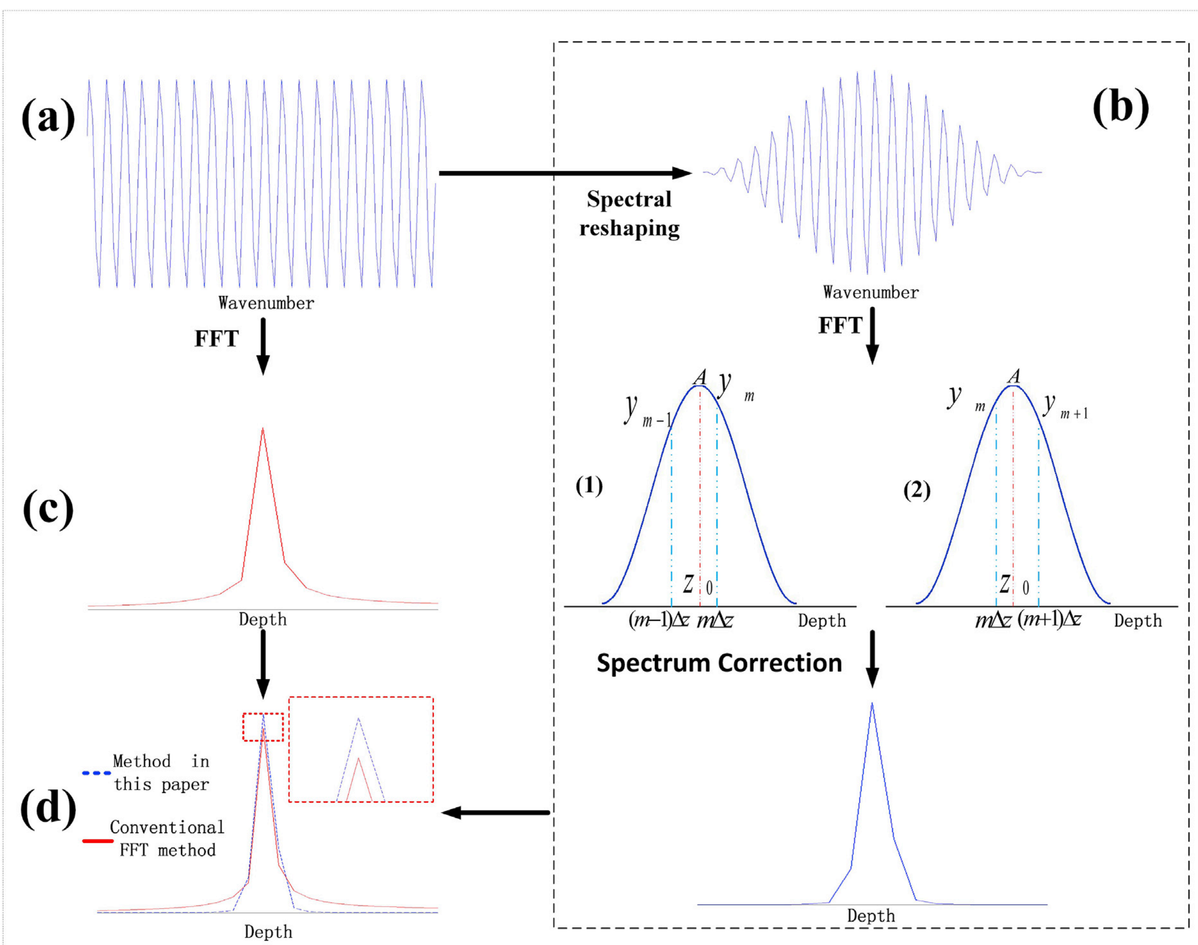

Fig. 1. Principles of the conventional FFT method and the method proposed in this paper, and the difference in the results between them. (a) Interferogram signal sampled from an OCT signal by a line scan camera (after removing constant terms from the signal). (b) The method in this paper using the spectrum correction method retrieves the depth profiles from the interferogram signal, including spectral reshaping, takes the Fourier transform, and finally performs spectrum correction to obtain the final data, where A is the theoretical value, and y_m, y_{m-1} and y_{m+1} are the values calculated from the interferogram signal. (1) the maxy_m is in front of the sub maxy_{m-1} in a main lobe, (2) the maxy_m is behind the sub maxy_{m+1} in a main lobe. (c) Conventional FFT method using Fourier transform to retrieve depth profiles from the interferograms. (d) Difference in the results between the conventional FFT method and the method proposed in this paper.