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Spectral-domain optical coherence tomography with dual-balanced detection for auto-correlation artifacts reduction

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Abstract: We developed a spectral domain optical coherence tomography (SD-OCT) to reduce auto-correlation artifacts (AC) using dual-balanced detection (DBD). AC were composed of the interference signals between different sample tissue depths, and shown up as artifacts in OCT images. This system employed a free-space Michelson interferometer, at the refraction plane of whose beam splitter, the light reflected experienced a π/2 phase shift with respect to the light transmitted. Then two phase-opposed interferometric spectra sharing the same spectrometer optics were obtained simultaneously using two lines of a three-line CCD. This new design was of lower cost compared to the dual spectrometer design reported previously. DBD enabled this SD-OCT to achieve two-fold increase in the interested signal amplitude inherently, and obtain a SNR increase of ~2.9 dB experimentally. To demonstrate the feasibility and performance of this SD-OCT system with DBD, we conducted an imaging experiment using a glass plate to obtain the optimal spectral matching between dual-balanced spectrometer channels. As a result, this SD-OCT achieved AC reduction up to about 9 dB and direct current (DC) term suppression up to about 30 dB by cancelling the identical components between dual-balanced spectrometer channels. The efficacy of AC reduction and DC suppression was validated by imaging the polymer coating of a drug-eluting stent and fresh swine corneal tissue ex vivo. The quality of DBD optimized images was significantly improved with regard to the single-channel images.

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References and links
1. Introduction

Being high resolution, high speed, and non-invasive, optical coherence tomography (OCT) has been broadly applied in ophthalmology and endoscopy imaging of transparent and turbid tissue [1]. Since Fourier domain OCT (FD-OCT) was introduced [2-5], researchers have made significant progress in vivo retinal imaging [6, 7], 3D volumetric imaging [8], ultra-high resolution imaging [9-11], Doppler blood flow determination in the human retina [12, 13], and so on. In classical theory of FD-OCT, AC composed of the interference signals between different sample tissue depths, shown up as fundamental artifacts in OCT images, especially for tissues with layered structures. These artifacts can be effectively suppressed by use of dual-balanced detection (DBD) [14-18]. DBD can be conveniently implemented in swept-source OCT with the off-the-shelf dual-balanced photo-detectors [19, 20]. However, there is few established DBD solution for spectral-domain due to unavailability of dual-balanced linear camera and spectrometer optics.

In this study, SD-OCT with dual-balanced detection (DBD) mainly for AC reduction is developed. Compared with single and unbalanced detection (SD), DBD enables SD-OCT system to achieve an SNR increase of 3 dB inherently. Most importantly, this SD-OCT with DBD achieves a good performance of AC reduction and DC suppression [16, 17]. Compared to a dual spectrometer OCT system [16-18], two phase-opposed interferometric spectra are
captured simultaneously using two lines of a three-line CCD in this study, and share the identical spectrometer optics. This new design can be of lower cost and better performance compared to the dual spectrometer design.

2. Theory

In FD-OCT, the detected interference signal in $k$-space, including DC term, cross-correlation (CC) term, and AC term, can be expressed in simplified formulation.

$$I_D(k) = DC + \int_{-\infty}^{\infty} a(z) \cos(2knz) dz + AC \tag{1}$$

where $a(z)$ is the amplitude of the elementary waves backscattered from the sample tissue and reference mirror at depth of $z$. $n$ is the refractive index of the sample tissue. DC term is a pathlength-independent bias, and its amplitude is proportional to the sum of the power from reference and sample arms. CC term that encodes depth-resolved information is necessary component for OCT imaging. AC term is mainly composed of the interference signals between different sample tissue depths, and served as artifacts in OCT images.

SNR is defined as $\text{SNR} = \frac{\langle S_{\text{oct}}^2 \rangle}{\sum N_{\text{ref}}^2}$, which can be denoted in dB as

$$\text{SNR}[\text{dB}] = 10 \cdot \log \left( \frac{\sum N_{\text{ref}}^2}{N_{\text{ref}}^2 + N_{\text{sh}}^2 + N_{\text{RIN}}^2} \right) \tag{2}$$

where $N_{\text{ref}}$ is the number of electrons in a single pixel generated by reference arm light, and $\sum N_{\text{sam}}$ is the total number of electrons over a single pixel array generated by sample arm light returning from a 100% mirror. $N_{\text{el}}^2$ is the CCD electrical noise, which mainly consists of readout noise $N_{\text{read}}^2$ and dark noise $N_{\text{dark}}^2$ of the CCD camera. $N_{\text{ref}}^2$ and $N_{\text{RIN}}^2$ are the numbers of electrons generated by shot noise and relative intensity noise, respectively.

The reference and sample power can be given by

$$\begin{align*}
N_{\text{ref}} &= \frac{\eta \tau \ h \nu}{P_{\text{ref}}} \\
N_{\text{sam}} &= \frac{\eta \tau \ h \nu}{P_{\text{sam}}}
\end{align*} \tag{3}$$

where $\eta$ is the spectrometer efficiency, including the grating diffraction efficiency and quantum efficiency, and camera lens efficiency. $\tau$ is the time consumption for interference signal detection. $h\nu$ represents the energy of a single photon. $P_{\text{ref}}$ and $P_{\text{sam}}$ represent the power from reference and sample arms, respectively. Under the condition that $P_{\text{sam}} \cong P_{\text{ref}}$ in most application, noise components in Eq. (2) can be written as

$$\begin{align*}
N_{\text{el}}^2 &= N_{\text{read}}^2 + N_{\text{dark}}^2 \\
N_{\text{sh}}^2 &= N_{\text{ref}} \\
N_{\text{RIN}}^2 &= (f / \Delta \nu) N_{\text{ref}}^2
\end{align*} \tag{4}$$

where $f$ is the reciprocal of double the exposure time of the CCD camera. $\Delta \nu$ denotes the FWHM spectral bandwidth of the reference arm light received by a single pixel.

When DBD configuration is employed in SD-OCT system, then two phase-opposed interferometric spectra can be obtained simultaneously in two dual-balanced channels. The desirable signal increases by a factor of 2, and DC term and AC term can be counterbalanced between dual-balanced channels to a great extent. Under the ideal DBD condition, the relative intensity noise $N_{\text{RIN}}^2$, arising from the fluctuation of light source, can be eliminated to a great
extent [17]. Compared with single detection configuration, the SNR with DBD configuration can be illustrated as

\[ SNR_{DBD} [dB] = 10 \cdot \log \left( \frac{2N_{ef} \cdot \sum N_{sam}}{N_{ef}^2 + N_{sh}^2} \right) \]  

(5)

3. Principle

3.1 SD-OCT system configuration

The optical configuration of SD-OCT system with dual-balanced detection was depicted in Fig. 1, which assumed a free-space Michelson interferometer structure. Light from a super luminescent diode (SLD) array (Superlum Broadlighters T-850-HP) with a center wavelength of 850 nm and a spectral full-width-half-maximum (FWHM) of 165 nm respectively was collimated by a lens L1 (AC050-010-B-ML, Thorlabs Inc.), and then split into a sample arm and a reference arm by a 50:50 non-polarizing cube beam splitter BS1 (BS011, 700-1100 nm, Thorlabs Inc.). The light beam was focused by an objective lens L5 (AC127-025-B-ML, Thorlabs Inc.). The numerical aperture of the objective lens was 0.052, resulting in confocal parameter of 1.3 mm and FWHM transverse resolution of 6.013 µm [21]. On the refraction plane of beam splitter BS1, the light reflected experienced a π/2 phase shift with respect to the light transmitted. The light backscattered or reflected from the sample interfered with the reference arm light in BS1. The second and third non-polarizing cube beam splitter BS2 and BS3 (BS029, 700-1100 nm, Thorlabs Inc.) were placed before and after the Michelson interferometer to balance the spectral power density of the light beam backscattered from the sample arm and reflected from the reference arm. As two light beams underwent a double pass through BS1, so two collimating lenses L3 and L4 (AC050-010-B-ML, Thorlabs Inc.) obtained two phase-opposed interferometric spectra concurrently, which were guided to the spectrometer through two single mode fibers of the 8 fiber v-groove array VGA (VGA-8-250, OZ OPTICS) whose v-grove spacing was 250 µm, and collimated by an achromatic lens L6.
The spectrometer was composed of a 1765 lines/mm diffraction grating (PING-Sample-020, Ibsen Photonics Inc.), a camera lens (Nikon, 85mm, f/1.8D), and a three-line CCD camera (ELIIXA 3V, e2V Inc.). Because the spacing between adjacent two lines of the CCD sensor was 10 µm, and the pixel size was 10 µm x 10 µm, so the vertical spacing between two channels was 20 µm. In order to obtain this tiny vertical spacing, we rotated VGA by 1.35 degrees. Two fiber ends representing CH1 and CH2 were shown in Fig. 1 and marked as ⊙ and ⊙. Then two dispersed spectra distinguished with a phase difference of π were projected onto the first and second line of the CCD camera. Among 4096 pixels in a single line, 2950 pixels were occupied to detect a total spectral range of 165 nm and each CCD sensor pixel corresponded to a spectrum band of ~0.06 nm. The detected signals were digitized at 12-bit resolution and transferred to personal computer via camera link cable and an image acquisition board (KBN-PCE-CL4-F, Bitflow). The timing of triggering signals generated by personal computer was well controlled, and used to synchronize the camera for image capture and the galvo scanner (GVSM002/M, Thorlabs Inc.), respectively. The galvanometer-mounted mirror was driven by a saw-tooth pulse to provide transverse scanning.

3.2 Sensitivity analysis

The light source operated at a center wavelength of 845 nm and a FWHM of 165 nm. The spectrometer efficiency, including the grating diffraction efficiency and quantum efficiency, was determined by the ratio of the power detected by two lines of CCD camera and the power at the entrance of the spectrometer, and estimated to be 0.45. The receiver noise (including readout noise, dark noise and quantization noise) was $N_{\text{rec}} = 60 e^-$, and the full well capacity (FWC) of CCD was 100 Ke. For the exposure time of 50 µs, this SD-OCT system operated at a line rate of 10 KHz. The maximum SNR [6, 17, 22] in a single pixel was achieved on the condition that $N_{\text{KN}}$ was equivalent to $N_{\text{el}}$ and shot noise dominated both the electrical noise and RIN. With a FWHM spectral bandwidth $\Delta \nu = 17.09$ GHz, we obtained $N_{\text{ref}} = 7.85 \times 10^7 \, e^-$, which corresponded to an average fill factor of 0.783 and a reference arm power of 0.56 nW. As a result, the shot noise was $N_{\text{sh}} = 280 \, e^-$ and dominated 70% of the total noise power. The sample power described by Eq. (3) was measured to be 1.98 mW, which corresponded to $N_{\text{sam}} = 1.04 \times 10^{10} \, e^-$. So the sensitivity was estimated to be 99.8 dB using Eq. (2) and 103.0 dB using Eq. (5) respectively. In order to characterize the sensitivity of our SD-OCT system with SD and DBD configuration, we placed a partially reflected mirror (~42.7 dB reflectivity) at the focal plane of the object lens and maintained the path-length difference between the sample and reference arm at 0.1 mm. The sensitivity was measured to be 97.7 dB with SD configuration, and calculated to be 100.6 dB with DBD configuration. The experimental results agreed well with the theoretical results. In practice, this scheme realizing DBD with two lines of a three-line CCD even achieved a significant SNR enhancement comparing to that employing two independent spectrometers, because the mismatch between every corresponding pixel in two independent spectrometers including gain and noise, may be greatly balanced with DBD configuration using one spectrometer.

4. Experiments

4.1 Optimal spectral matching

In order to match two spectra acquired from two lines of the CCD sharing the same spectrometer optics, we developed a spectrum matching algorithm [10, 15]. Firstly, we acquired ten sets of interference fringes and their corresponding background spectra from two channels by changing the reference delay when a partially reflecting mirror was placed at the focal plane of the objective lens. Then each of the interference fringes was subtracted by the respective background spectrum. Secondly, we manually selected parts of the spectrum whose
amplitude could not be neglected, detected the zero-crossing points, and achieved their indices. Thirdly, we selected one dense interference fringe as the template, and executed a limited search by moving the other interference fringes in the same channel ranging from -N/2 to N/2 (N is the number of zero crossing points) zero-crossing points. The square of the difference of two fringes in two channels respectively got its maximum value when the optimal spectral matching was achieved. Finally, the optimal spectral matching data was verified as follows.

The original interference fringes acquired from CH1 and CH2 were depicted in Fig. 2(a). Both spectra were subtracted by their background respectively. Then the optimal spectral matching algorithm was applied to two corresponding signals from two channels, and results were shown in Fig. 2(b). It was obvious that there was a π phase difference between signals in CH1 and CH2. The dual-balanced signal achieved by subtracting signals in CH2 from that in CH1 was shown in Fig. 2(c). The interested signal increased with two times and presented a symmetrical shape. Finally, the depth resolved signals after FFT were presented in Fig. 2(d). It was obvious that DBD resulted in a significant SNR increase of ~2.9 dB and DC suppression of ~30 dB comparing to single detection. So this algorithm achieved a good performance of matching two spectra acquired from two lines of the CCD.

![Graphs](image)

Fig. 2. (a) Original interference fringes from CH1 (black curve) and CH2 (red curve). (b) Original interference fringes from CH1 (black curve) and CH2 (red curve) subtracted by their background respectively. (c) Fringes achieved with DBD configuration. (d) The depth resolved signals after FFT from SD-OCT with SD (black curve) and DBD (red curve) configuration respectively.
4.2 Auto-correlation artifacts reduction verification

To demonstrate the performance of optimal spectral matching between two lines of the CCD, reduce AC and suppress DC, we chose a glass plate (Menzel-Glaser, BB024050A1) with a thickness of about 0.2 millimeter as the multilayer sample. A simple schematic of multilayer sample was illustrated in Fig. 3(a). The front surface and back surface indicated by FS and BS were corresponded to Fig. 3(b) with arrows, which included three cross-sectional images of the glass plate obtained using SD from CH1 and CH2 respectively, and using BDB. Comparing cross-sectional image captured using DBD to that captured using SD, the results showed that AC was nearly completely reduced, and DC term was suppressed to a great extent. The profiles of one single A-line indicated in red line in Fig. 3(b) were depicted in Fig. 3(c). Comparing A-line captured using DBD to that captured using SD, peaks that represented the front surface and back surface of the glass plate doubled, and DC term was largely suppressed up to 10.6 dB. Most importantly, the AC reduction in CH1 and CH2 were 9.39 dB and 9.11 dB, respectively. This experiment demonstrated that DBD achieved a good performance of AC reduction and DC suppression performance.

4.3 Auto-correlation artifacts reduction performance in imaging

To demonstrate the performance of AC reduction and DC suppression, we conducted two imaging experiments: the polymer coating of a drug-eluting stent and the swine cornea ex vivo. Drug-eluting stent (DES) was composed of a bare-mental stent coated with an anti-proliferative polymer, which gradually allowed drug elution into the coronary wall for weeks after stent implantation so as to inhibit cell hyperplasia that causes restenosis. By the time entire drug had been released in six to nine months, the main risk of restenosis had been minimized. However, the anti-proliferative polymer coating was clinically broken on occasion. For such a drug-eluting stent with layered structures, AC was the dominant artifact. AC reduction helped us obtain a better cross-sectional image of drug-eluting stent, and a better understanding of how drug-eluting stent works and how to redesign drug-eluting stent. Partial anti-proliferative polymer of an unemployed drug-eluting stent was broke when manually expanding and flating this stent to scan.

An intuitive schematic of the drug-eluting stent was manifested as Fig. 4(a). The blue plane in indicated the scanning direction. Cross-sectional images of the stent captured from CH1 and CH2 using SD and BDB respectively were shown in Fig. 4(b). A-line profiles indicated in
red in Fig. 4(b) were depicted in Fig. 4(c). Peaks represented the surfaces of the metal and coating had doubled, and the DC term had also been eliminated to a great extent. Auto-correlation reduction in CH1 and CH2 were 5.51 dB and 6.57 dB respectively. So the performance of AC artifacts reduction had been demonstrated when imaging a stent clinically.

In order to validate this advantage, the other experiment was conducted on a swine cornea. The results shown in Figs. 5(a)-5(c) demonstrated the reduction performance of AC artifacts between epithelium layer and endothelium layer in a swine cornea. Figs. 5(a)-5(c) compared cross-sectional images acquired using SD and DBD, respectively. Figs. 5(a) and 5(b) showed the images processed by SD and background subtraction. Partial DC term and mass AC artifacts existed. Fig. 5(c) shows the image processed by DBD, where DC term and AC artifacts were greatly suppressed. This experiment demonstrated that DBD achieved a good performance of AC reduction and DC suppression performance when imaging a cornea, but there still existed partial AC due to imperfect spectral matching.

Fig. 4. (a) An intuitive schematic of the drug-eluting stent. The blue plane indicates the scanning direction. (b) Cross-sectional images of the stent obtained using SD from CH1 and CH2 respectively, and using BDB. DC: direct current term; AC: auto-correlation term; PC: polymer coating; MA: mental alloy; MI: mirror image; PD: petri dish. (c) A-line profiles indicated in red line in (b).

Fig. 5. Cross-sectional images of the central area of a swine cornea captured using SD from CH1 (a) and CH2 (b) respectively, and using BDB (c). EP: epithelium layer, BL: Bowman’s
layer, ST: Stroma, DM: Descemet’s membrane, ED: Endothelium layer. Three images consist of 674 axial × 472 transverse pixels covering 0.32 mm × 1.0 mm.

5. Discussion and conclusion

In summary, auto-correlation artifacts reduction performance were demonstrated experimentally using dual-balanced detection in SD-OCT. Firstly, dual-balanced detection provides a practical approach to suppress direct-current background noise and reduce the auto-correlation artifacts, which are shown up as fundamental artifacts in OCT image, especially for tissues with layered structures of highly scattering/reflection. Due to the relative large spacing between adjacent channels of VGA, noticeable mismatch still exists between two dual-balanced channels even sharing the same spectrometer optics, resulting in about 10% residual auto-correlation artifacts in images using dual-balanced detection. This issue can be solved by use of VGA of smaller spacing, for instance, 80 μm. Secondly, dual-balanced detection enables SD-OCT to achieve a SNR increase of 3dB inherently. As a result, a shorter exposure time and higher scanning rate were achieved. However, the sensitivity of the proposed SD-OCT system with dual-balanced detection is suboptimal in that about 20% of focused light was not detected by the camera sensor because the beam spot size on CCD is larger than the pixel size. This problem can be solved by using a grating with larger area in the spectrometer. In addition, the noticeable chromatic focal shift is another reason for suboptimal sensitivity. Thirdly, dual-balanced detection employing two lines of a three-line CCD can be of lower cost compared with two independent spectrometers design. Experiments on a stent and a swine cornea have demonstrated the feasibility of dual-balanced detection in SD-OCT.

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