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Correction of saturation effects in endoscopic swept-source optical coherence tomography based on dual-channel detection

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Abstract. Saturation artifacts that are commonly observed in endoscopic swept-source optical coherence tomography (SSOCT) images cause image degradation and loss of image information. We present work on the correction of saturation effects in endoscopic SSOCT imaging. This method utilizes a broadband power divider with excellent pick-off flatness to divide the detected interference signal into the two channels of an analog-to-digital converter. Based on the precise calibration of the splitting ratio between the two channels, the maximum measurable signal power of the system was drastically increased by using the low level signal in one channel to correct the saturated signal in the other channel. The experimental results demonstrated that this technique can efficiently correct the saturation artifacts in endoscopic two- and three-dimensional SSOCT images in an accurate and cost-effective manner.
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Keywords: optical coherence tomography; endoscopic OCT; saturation artifacts.

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1 Introduction

Optical coherence tomography (OCT) is a high-resolution, noninvasive optical imaging technique that provides cross-sectional imaging of biological tissues in micrometer scale. Compared with time-domain OCT (TDOCT), Fourier-domain OCT (FDOCT) can achieve a much higher sensitivity and imaging speed[1] and has been widely used in various biomedical applications. Using the Fourier-domain technique based on a high speed wavelength swept source, swept-source OCT (SSOCT) is capable of an A-line rate of up to multi-MHz[2] with a simple setup, which makes SSOCT attractive in clinical applications especially in endoscopic imaging of internal organs such as gastrointestinal tracts and airways.[3,4]

In spite of all the advantages of SSOCT, there still exist some impediments that degrade the image quality especially in *in vivo* endoscopic studies. One of the challenges is the saturation effects due to strong signals from highly reflective areas such as surfaces of internal organs that are commonly lubricated by mucus, catheters, and guide wires that are assembled in endoscopic probes, metallic stent struts, microcalcifications, etc. Since the incident angle is hard to control in endoscopic OCT imaging, the power of the reflected light occasionally exceeds the input range of the used detector or analog-to-digital converter (ADC). Fourier transformation of the saturated signal results in a bright line on the tissue surface accompanied with a band of artifacts across tissue depth that degrades the image quality and leads to complete loss of information in the areas with strong artifacts. One solution is to increase the input range of the used detector or ADC especially the latter since the limited bit depth of the ADC is usually the bottleneck of the input range of the system. However, a high-performance ADC with a sampling rate sufficient for SSOCT imaging and a high bit depth is either not available in the market or costly. Huang and Kang reported a method to correct saturation artifacts by linear interpolation of the signals in adjacent A-lines.[5] However, the interpolation-based reconstruction can be used only in the correction of sparse saturation artifacts. An adaptive optimization technique based on automatic adjustment of the reference power was used to suppress saturation effects in SDOCT[6] at the cost of significantly slowing down the frame rate due to complex design and calculations. Wu et al. utilized a multiexposure spectrum recording method to reduce saturation artifacts in SDOCT.[7] However, the compensation effect was limited by the inaccurate estimation of the multiexposure signal levels since the ratio of the levels cannot be precisely calibrated. Therefore, a real-time and accurate technique to correct saturation effect suited for SSOCT systems, especially in endoscopic imaging, is still absent. In this paper, we demonstrated an alternative design based on a dual-channel detection technique to suppress the saturation artifacts. The detected signal was split into two channels with the ratio of the signal levels precisely calibrated. The high level signal was used to reconstruct OCT images and the low level signal was used to correct the saturated signal in the case that the high level signal exceeds the input range of the system. This technique allows for a simple and cost-effective suppression of saturation artifacts in endoscopic SSOCT without the need of decreasing the incident power.

2 Methods

The schematic diagram of the SSOCT system is shown in Fig. 1. A swept source at 1310 nm with a bandwidth of 87 nm, a sweep frequency of 100 kHz, and an output power of 20 mW (Santec Corporation, Japan) was used as the light source. The input light was split by a 1×2 coupler into the sample and reference arms, respectively. In the sample arm, a 1.3-mm proximal scanning endoscopic probe was employed for three-dimensional (3-D) imaging. The helical scanning probe was driven by a rotary motor with a rotational rate of 50 rounds/s and a stepper motor translational stage with a pulling-back speed of 1 mm/s, respectively. By using a phase-resolved algorithm to computationally compensate the dispersion generated by the endoscope optics, the SSOCT system is capable of an axial resolution of

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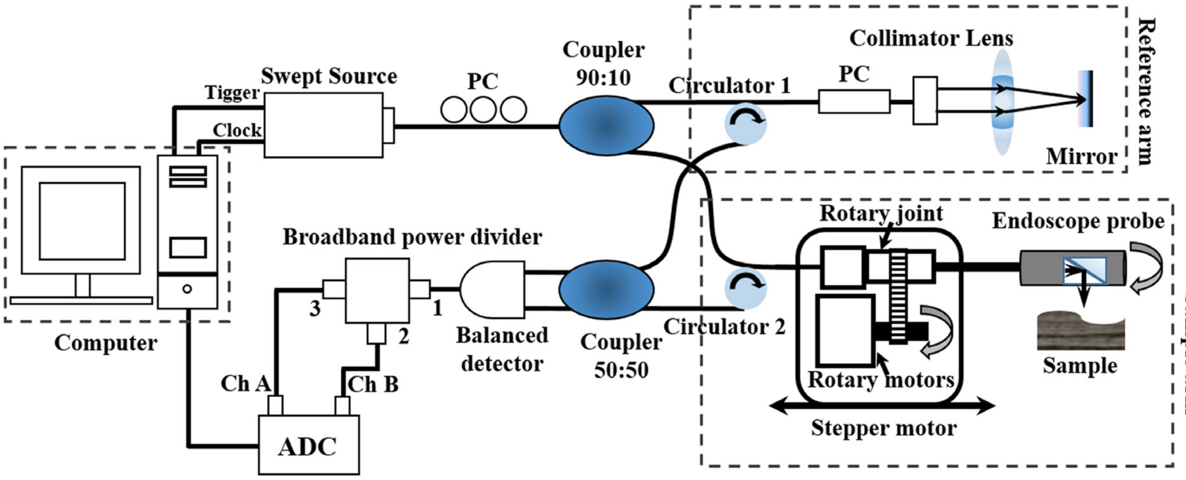

Fig. 1 Schematic of the SSOCT system. PC, polarization controller; ADC, analog-to-digital converter.

8 μm in tissue and a lateral resolution of 20 μm, respectively. The total reference power was set to be 25 μW for optimization of the system sensitivity. In the detection arm, a balanced detector (Newport, 1817) was carefully selected in regard to the detection bandwidth, gain, and noise spectrum so that the noise level of the detector is comparable to the quantization noise of the used ADC. In order to compensate saturation effects, the interference signal was divided into two paths by a broadband (DC-20 GHz) power divider (power ratio $\sim$1∶25) and then digitized by a 12-bit two channel ADC (Alazartech, ATS9360). In each channel, 1024 samples were acquired using the k-clock from the laser source as an external clock signal. By utilizing a high-performance oscilloscope, the splitting ratio of the power divider was accurately calibrated showing excellent pick-off flatness especially in low frequency range. The signal collected by the high-level channel (ChA) was used for OCT imaging. Our OCT algorithm is processed on a graphical processing unit (GPU) featuring a multithreaded real-time data acquisition, image processing, and display at the rate of 50 frames/s with 2000 A-lines in each frame. To detect saturation in ChA, a threshold of the low level signal in ChB was set to be equal to the input range divided by the splitting ratio of the power divider. Hence, the saturated signal in ChA over the maximum input range can be reconstructed with the signal spontaneously detected in ChB. By multiplying the splitting ratio with the signal in ChB, the saturated signal due to strong reflection is compensated as shown in Fig. 2(a). The corresponding artifact peaks in depth-domain after Fourier transformation were significantly suppressed [Fig. 2(b)]. This design using a power divider with a ratio of 1:25 can

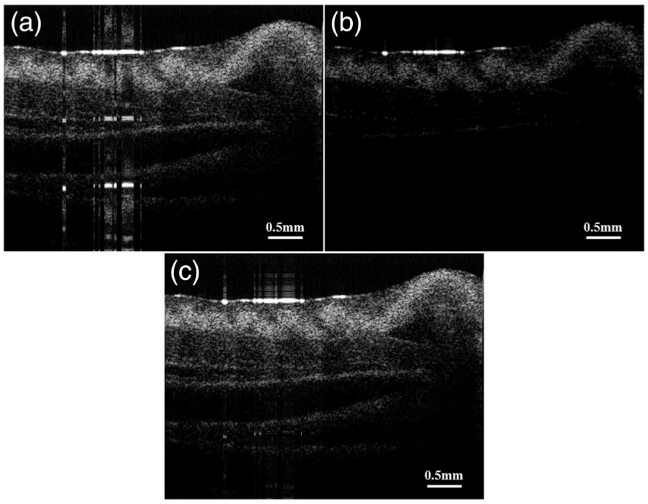

Fig. 3 OCT images of porcine airway captured by the (a) high and (b) low level channels before correction; (c) OCT image after correction.

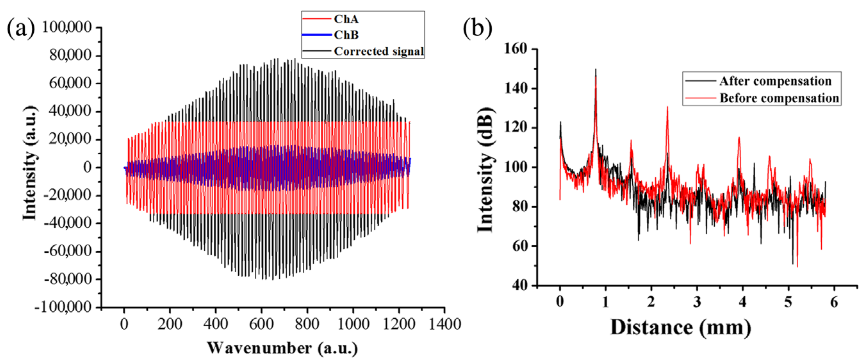

Fig. 2 (a) Interference signals recorded with ChA (red) and ChB (blue). Black line denotes corrected signals in ChA after compensation with signals in ChB; (b) Fourier transforms of signals before and after compensation.

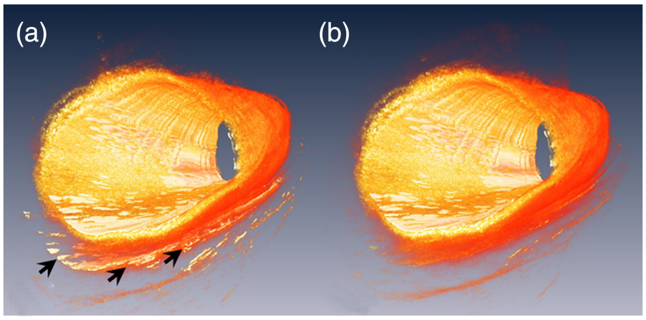

Fig. 4 Endoscopic 3-D OCT images of porcine airway (a) before and (b) after correction, the arrows denote the saturation artifacts.

theoretically increase the input range by 14 dB, which is sufficient to correct saturation effects in most endoscopic imaging.

3 Results

To evaluate the approach's capability of imaging tissues with high reflectivity, we have performed imaging of a section of porcine upper airway using our system. The bright lines in the middle of Fig. 3(a) due to the Fourier transformation of the saturated signal were removed as shown in Fig. 3(c). The structures hidden inside the bright lines were revived through significant suppression of the saturation artifacts.

Construction of 3-D data sets from 500 B-scans utilized a commercial software package. As shown in Fig. 4, the artifacts were removed with this technique resulting in a clean 3-D reconstruction of endoscopic OCT imaging.

4 Discussion and Conclusions

In clinical applications such as dermatological imaging, the reflectivity of dry tissue is less than 10^{-4}. However, the reflectivities of mucous fluid, catheters, and stent struts usually exceed 10^{-2} or more in endoscopic applications, resulting in saturation artifacts in OCT images. Increasing the number of bits of an ADC could reduce saturation effects, however, requiring complex and costly design. In this paper, we have presented a simple and cost-effective method to solve this problem. The results reported here demonstrate that this technique effectively suppresses saturation artifacts especially in endoscopic OCT imaging.

Disclosures

The authors have no relevant financial interests in this article and no potential conflicts of interest to disclose.

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