Sensitivity enhancement in swept-source optical coherence tomography by parametric balanced detector and amplifier

Jiqiang Kang, Xiaoming Wei, Bowen Li, Xie Wang, Luoqin Yu, Sisi Tan, Chandra Jinata, and Kenneth K. Y. Wong

Department of Electrical and Electronic Engineering, The University of Hong Kong, Pokfulam Road, Hong Kong, China

*kywong@eee.hku.hk

Abstract: We proposed a sensitivity enhancement method of the interference-based signal detection approach and applied it on a swept-source optical coherence tomography (SS-OCT) system through all-fiber optical parametric amplifier (FOPA) and parametric balanced detector (BD). The parametric BD was realized by combining the signal and phase conjugated idler band that was newly-generated through FOPA, and specifically by superimposing these two bands at a photodetector. The sensitivity enhancement by FOPA and parametric BD in SS-OCT were demonstrated experimentally. The results show that SS-OCT with FOPA and SS-OCT with parametric BD can provide more than 9 dB and 12 dB sensitivity improvement, respectively, when compared with the conventional SS-OCT in a spectral bandwidth spanning over 76 nm. To further verify and elaborate their sensitivity enhancement, a bio-sample imaging experiment was conducted on loach eyes by conventional SS-OCT setup, SS-OCT with FOPA and parametric BD at different illumination power levels. All these results proved that using FOPA and parametric BD could improve the sensitivity significantly in SS-OCT systems.

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1. Introduction

Optical coherence tomography (OCT) is one of powerful imaging modalities for biological applications [1]. In conventional swept-source OCT (SS-OCT) systems, the illumination optical power is usually boosted by optical amplifiers, e.g. booster optical amplifier (BOA) [2,3] and Raman amplifier [4], right before the bio-samples for a good imaging sensitivity. Those optical amplifiers are commonly required to operate in a broadband and low-noise regime since OCT offers an axial resolution inversely proportional to the spectral width of the laser source [5]. However, in some cases, although the sample is illuminated by high optical power, the back reflected power is still feeble to the detection part because of the highly-scattered samples [6,7]. More importantly, a lower optical power illumination is required for the in-vivo imaging when the photodamage, phototoxicity and photobleaching issues are concerned, particularly in ophthalmology [8,9]. In these cases, one more amplification stage can be used to amplify the OCT’s interference signal to improve detection sensitivity. Unfortunately, the frequently-used BOA is featured with large noise figure, particularly for those extremely poor signal from bio-samples [10], while the Raman amplifier requires a strong shorter-wavelength pump. Thus, a broadband optical amplifier with superior noise performance for weak signal has a great demand in enhancing the sensitivity of those OCT systems.

Fiber optical parametric process is a widely-explored phenomenon which is based on the third-order nonlinearity of the fiber materials [11]. It has been utilized for many important aspects, such as fiber optical parametric amplifier (FOPA) [12], wavelength conversion [12], and fiber optical parametric oscillator (FOPO) [13], etc. Briefly, fiber optical parametric process can amplify a weak broadband signal and generate a phase conjugated coherent component, namely idler, with a suitable pump power and phase matching between the signal, pump, and idler. To be used as an amplifier, it has been demonstrated that FOPA can provide a bandwidth over 270 nm [14], a gain high up to 70 dB [15], and a 1.1 dB noise Fig [16, 17], at a moderate pump power and a piece of highly-nonlinear fiber (HNLF). Other than signal amplification, another important feature of fiber optical parametric process is its phase-conjugated idler. This idler, locates at different wavelength region, carries both the same intensity and information as the original signal. For these reasons, idler generated through fiber optical parametric process has been widely used at wavelength exchange aspects [18]. More importantly, signal and idler can be used together to realize a parametric receiver for further detection sensitivity improvement. Kuo et al., and Liang et al. demonstrated this technique on ON-OFF keying wavelength-division multiplexing (WDM) systems and the sensitivity was raised by more than 1.6 dB than using signal or idler band alone [19, 20]. Recently, Kumpera et al. demonstrated parametric coherent receiver by launching a weak idler, an intense pump, and the signal under test into a spool of HNLF to generate a parametric mixer where the input idler served as a local oscillator (LO). The LO interfered with the signal under test through fiber optical parametric process and finally the idler (or optionally the signal) was detected by a photodetector (PD) [21]. The parametric receiver that uses signal and idler together is intrinsically different from the parametric coherent receiver because no extra idler is launched in parametric receiver case. For clarity purpose, we refer to this kind of parametric receiver as parametric balanced detector (BD), and FOPA only represents an amplifier without utilizing the newly-generated idler in the fiber optical parametric process.

Applying fiber optical parametric process in OCT system has led to a few attractive features. Leonhardt et al. explored the nonlinear optical frequency conversion to Fourier domain mode-locked (FDML) laser, which not only shifted the working wavelength, but also increased the wavelength sweep rate of FDML laser [22]. Zhu et al., Xu et al., and Yu et al. reported dual-band and tri-band SS-OCT systems, which expanded the working bandwidth of
SS-OCT systems, thus improved the resolution through generating an idler band by using the FDML laser’s output as the signal to the fiber optical parametric process [23–25]. Recently, Zhao et al. demonstrated the imaging enhancement and time gating effects on charge-coupled detector (CCD)-based spectral-domain optical coherence microscopy (SD-OCT) in turbid media by using optical parametric amplifier (OPA) with a type I BBO crystal [26]. In this study, we proposed and demonstrated a sensitivity enhancement method of the interference-based signal detection on a SS-OCT system through FOPA and parametric BD. Parametric BD was realized by superimposing the amplified signal and the newly-generated idler in time domain and was then detected by combining signal and idler at a PD. Since it is similar to the traditional BD that uses two differential inputs to improve the sensitivity [27–30], we named it for resemblance, though parametric BD is a single-ended receiver. Parametric BD can further raise the sensitivity over FOPA usage alone.

The two new schemes were denoted as SS-OCT with FOPA and SS-OCT with parametric BD to distinguish those from the conventional SS-OCT utilizing a traditional BD to detect the direct interference signal. Moreover, for simplicity purpose, both FOPA and parametric BD were referenced as optical parametric amplifier for life-science (OPALS) when they were used in bio-imaging [31]. To the best of our knowledge, this is the first time that OPALS is applied to improve detection sensitivity in SS-OCT. After comparing the sensitivity enhancement and bio-sample imaging results of OPALS-SS-OCT with the conventional SS-OCT, OPALS is evident to be suitable for SS-OCT systems’ advancement.

2. Experimental setup

The experimental setup (Fig. 1) consisted of three parts, the SS-OCT imaging, OPALS and signal processing.

![Fig. 1. Experimental setup. CIR: circulator; PC: polarization controller; DC: dispersion compensator; GM: galvo mirror; HNL-DSF: highly-nonlinear dispersion-shifted fiber; PD: photodetector; W: wavelength division multiplexing (WDM) coupler. X and Y represent two measurement points; A and B represent two filtering channels.](image)

The SS-OCT imaging part is comprised of a home-built FDML laser source with wavelength swept between 1575 and 1651 nm and its A-scan rate is about 45 kHz. The FDML laser’s output was amplified by a BOA (BOA XSS1004, COVEGA) to obtain 10.8 dBm (12 mW) output power. The power was splitted by a 90/10 coupler first, and the 10% was used for power spectrum monitoring and recalibration for the imaging part. The 90% power was divided by another 90/10 coupler again, and the 10% branch went through the reference arm which consisted of a collimator, iris, dispersion compensator (LSM03DC, Thorlabs) and mirror, while the 90% one incident to the sample arm that was consisted of a collimator, galvo mirror and scanning lens (LSM03, Thorlabs). The effective focal length of
the scanning lens is 36 mm and its aperture is 4 mm. Assuming that the beam profile is Gaussian, the lateral resolution of the SS-OCT can be calculated [32] as

$$\Delta x = \frac{4\lambda f}{\pi d}.$$  

(1)

Where $\lambda$ is the center wavelength of the light source, $f$ and $d$ are the effective focal length and aperture size of the scanning lens, respectively. Assuming that the spectrum is Gaussian shape, the axial resolution can be calculated by [32]

$$\Delta z = \frac{2\ln 2 \lambda^2}{\pi \Delta \lambda}.$$  

(2)

Where $\Delta \lambda$ is the full width at half maximum (FWHM) of the spectrum. The dispersion compensator (DC) was used to compensate the dispersion of the scanning lens. The loss from collimator to the sample surface was about 2.5 dB. The back reflected light by the mirror on the reference arm and by the sample interfered in a 2 × 2 50/50 coupler to obtain the interference signal of the SS-OCT. In conventional SS-OCT setup, two interference signal with $\pi$-phase shift were generated by the coupler and both of them were then launched into a two-channel BD to reduce common mode noise. On the other hand, in SS-OCT with FOPA or SS-OCT with parametric BD setup, only one of them was used in OPALS while the other signal was dropped, since an interference signal was sufficient for OPALS. It should be noted that OPALS will not affect the OCT resolution as it only amplifies the OCT’s interference signal without any influence on FDML source’s beam profile nor spectrum shape.

Meanwhile, a tunable continuous-wave (CW) laser source (TUNICS-OM 1560) working at 1554.7 nm wavelength, was a core part of OPALS. It was first phase modulated with a 4-tones radiofrequency oscillator to suppress stimulated Brillouin scattering (SBS) [33] and was then amplified by two stages erbium-doped fiber amplifiers (EDFAs) to obtain 2.3 W pump power. The detailed structure was not shown in Fig. 1 for the limited space. The pump power and the OCT’s interference signal which came from one port of the 2 × 2 50/50 coupler were combined with a C/L band WDM coupler and then these were launched into a 150 m highly-nonlinear dispersion-shifted fiber (HNL-DSF) with nonlinear coefficient of 30 W⁻¹km⁻¹ and zero-dispersion wavelength at 1554 nm. The pass band of C/L band WDM coupler was 1480 nm to 1563 nm for C band and 1570 nm to 1660 nm for L band. At the output port of OPALS, there were three optical wave bands, namely amplified signal band, idler band, and residual pump. The time domain superimposed signal and idler band, referred as combination band, can be obtained only if the residual pump was filtered. Therefore to acquire the combination band, two customized WDM notch filters, whose stop bands were 3 and 20 nm, were incorporated and centered at 1555 nm (W1 and W2 in Fig. 1). Combination band subsequently went through channel A, the parametric BD channel. Other than that, to attain only the amplified signal band, one more C/L band WDM coupler (W3 in Fig. 1) was utilized to extract it from the combination band (Channel B, the FOPA channel).

Finally, the output of channel A or B was detected by a PD (PDB110C-AC, Thorlabs) and it was then digitized with a 14-bit precision digitizer card. The sampling points in a single A-line are 2048. The sampling clock, 125 MHz, comes from the digitizer card’s on-board clock (Alazar Tech, ATS 460) and the system’s trigger signal comes from the fiber Fabry–Perot tunable filter (FPP-TF) driver board. All the calculations were performed with a graphical processing card (GTX460, NVIDIA) based on LabVIEW platform. The interpolation and Fast Fourier Transform (FFT) algorithms were packaged into a dynamic link library (.dll) file to speed up the processing and thus realized real-time imaging.
3. Results and discussion

3.1 Gain spectrum of OPALS

Since the poor interference signal of SS-OCT need to be amplified, whose bandwidth was the same with the FDML laser’s output, we could first verify the amplification and idler generation performance with the output of FDML laser. Therefore, the FDML laser’s output was launched into the HNL-DSF with the CW pump and then the output spectrum was measured by an optical spectrum analyzer (Agilent 86142B) after filtering the major part of the pump power at point X in Fig. 1. The spectrum of FDML laser and the gain spectrum of OPALS is shown in Fig. 2 as red and blue lines, respectively.

As shown in Fig. 2, the output of FDML laser had a full bandwidth of 76 nm with FWHM at 62 nm. The gain spectrum of OPALS covered FDML laser’s full bandwidth with the maximum gain up to 18.3 dB and a 66 nm bandwidth idler was generated simultaneously. In general with FDML OCT at 1.5 μm window, the bandwidth can be over 100 nm centered at about 1550 nm [34]. Here, the 76 nm bandwidth starting at 1575 nm was selected for the following reasons: First, the passband of the C/L band WDM coupler in OPALS starting at 1575 nm limits the shorter wavelength. Second, the BOA (COVEGA BOA1080) in the FDML laser cavity cannot provide efficient gain to the wavelength beyond 1650 nm that limits the longer wavelength.

The combined signal and idler bands is required to obtain a parametric BD. It is straightforward to achieve the target as long as the pump after the HNL-DSF is filtered. Therefore, a reliable filter is crucial for the setup. In our setup, although two customized notch filters were used, the pump was not completely suppressed after these filters (see residual pump in Fig. 2) because the CW pump power launched into these filters was still high, around 1.1 W. However, this narrow band residual pump had no influence on the imaging part because it was generated in the OPALS part, and its negative effects could be removed by averaging on the final images.
3.2 Resolution of SS-OCT

Lateral and axial resolution are critical parameters to OCT systems. In our setup, from Eq. (1) and Eq. (2), the calculated lateral resolution is 17 μm which corresponds to about 12.1 μm in tissue, and the calculated axial resolution is 18.7 μm with the FWHM (62 nm) centered at 1620 nm in air which corresponds to about 13.4 μm in tissue. To confirm both of them, we measured the lateral and axial resolutions which were shown in Fig. 3(a)-3(b).

The lateral resolution was measured by placing a United States Air Force (USAF) 1951 resolution target at the focal plane of the scanning lens. As imaging with conventional SS-OCT, a series of two-dimensional (2D) axial images can be obtained by scanning the resolution target. After synthesizing a three-dimensional images with those 2D axial images, we can get the resolution target’s surface image which can be used to determine the lateral resolution. In Fig. 3(a), the image of the 5th group of the target is shown, and the lines in 6th element (labelled with green dash box) can be separated clearly. The distance between two adjacent lines is 17.54-μm. This result is in line with the calculated value.

Since the axial resolution is determined by the actual axial point spread function (PSF), a mirror served as a single layer sample can be placed at the focal plane of the scanning lens to measure the PSF. Furthermore, by doing linear interpolation and FFT to the interferometric signal, the PSF data can be acquired. After doing Gaussian fit to the PSF, the measured axial resolution is 35 μm in air which corresponds to about 25 μm in tissue, as shown in Fig. 3(b). The measured value was degraded from the theoretical value mainly because that the source spectrum was not in an ideal Gaussian shape in addition to the uncompensated dispersion mismatch in the system.

3.3 Sensitivity and its roll-off

Sensitivity and its roll-off are another important parameters for SS-OCT systems which exhibit the instantaneous linewidth of the swept-source. By measuring the PSF at different mismatch length positions between the reference and sample arm with the post-amplified swept-source power, one can obtain the roll-off curve. In order to avoid saturation to the PD, the back reflected power should be first attenuated in some degree by tuning the iris and the specific loss value should be recorded which was referred to calibrated loss here. By adding the calibrated loss and the the signal-to-noise ratio (SNR) of the PSF curve at a given mismatch length position, the sensitivity at the given position can be obtained. A detailed measurement method has been demonstrated elsewhere [2]. In this part, we not only measured...
the roll-off of conventional SS-OCT but also the roll-off of SS-OCT with OPALS for depths of 0-6 mm in 0.5 mm step while the sensitivity of them was measured at about 1 mm mismatch position, which are shown in Fig. 4.

Fig. 4. Sensitivity and its roll-off: (a) roll-off of conventional SS-OCT; (b) roll-off of SS-OCT with FOPA; (c) roll-off of SS-OCT with parametric BD; (d) Sensitivity comparison of (a), (b) and (c) at about 1 mm mismatch position.

In conventional SS-OCT case, the calibrated loss was 34.3 dB and the power emitted out from collimator 2 was 9.8 dBm (9.6 mW) and the illumination power on the sample (mirror here) was 6.3 dBm (4.3 mW) owing to 2.5 dB loss in free space. The power on reference arm was 1/9 of the sample arm. The optimal sensitivity of the system can be obtained by setting the back-reflected power of the reference arm (BPR) larger than the back-reflected power of the sample arm (BPS) to avoid the input of the PD too small, otherwise the excess noise would became large [2]. In our setup, the optimal sensitivity was 88.6 dB at about 1 mm mismatch position. The measured sensitivity was about 14 dB lower than the theoretical value (~103 dB) calculated by shot-noise limited detection [35, 36]. This discrepancy major was due to the insertion loss of the system, including fiber component loss (broadband WDM couplers, circulators, and etc.) and free-space coupling loss. However, although the sensitivity is optimal by setting BPR larger than BPS to some degree, it will sacrifice the modulation depth (or contrast) of the interference fringe since the maximum modulation depth to the interference fringe only happens when the power of BPR is equal to BPS [37]. In order to maintain the maximum modulation depth, and thus achieve the best contrast to the final images, we set the BPR equal to BPS although it sacrificed 16 dB sensitivity, and the sensitivity of conventional SS-OCT was 72.6 dB whose sensitivity roll-off is shown in Fig. 4(a).

In SS-OCT with FOPA and SS-OCT with parametric BD cases, the calibrated loss were 51.3 dB and 56.7 dB, respectively. The difference was caused by that different WDM filters were used in different channels (A & B) and another two extra attenuators, whose measured attenuation coefficient were 9.9 dB and 14.9 dB respectively, were used in channel A and B to avoid damaging the PD for the high gain in OPALS. By launching the interference signal with
BPR equal to BPS to the OPALS part, we measured the roll-off and sensitivity with the same procedures in conventional SS-OCT. The roll-off of SS-OCT with FOPA and SS-OCT with parametric BD were shown in Fig. 4(b) and Fig. 4(c) respectively and their sensitivity are 82.4 dB and 84.8 dB at about 1 mm mismatch position. The sensitivity of these three cases are compared as bar chart in Fig. 4(d).

From Figs. 4(a)-4(c), the SNR did not change significantly at depth from 0 to 1.5 mm. But from 1.5 to 6 mm, the descending slope were different in different figures. Figure 4(c), namely the SS-OCT with parametric BD, has the best performance whose slope is much flatter than Fig. 4(a) and 4(b) with slope of 2.59 dB/mm. Figure 4(a), the conventional SS-OCT, and Fig. 4(b), the SS-OCT with FOPA have the same slope about 3 dB/mm. This is due to the fact that the lower the slope, the better imaging depth it obtains. Furthermore Fig. 4(d) displays sensitivity improvement by 9.8 dB with FOPA and it was even exceeded by 2.4 dB more when parametric BD was used, compared to the conventional SS-OCT setup. All the results prove the sensitivity enhancement by using OPALS in SS-OCT system.

### 3.4 Bio-sample imaging

To further demonstrate the sensitivity enhancement of OPALS, a bio-sample imaging experiment was conducted on loach eyes at different illumination power levels shined on the sample. Three illumination power levels were set, namely level one, two and three with 6.3 dBm (4.3 mW), 2.97 dBm (2 mW) and −0.43 dBm (0.9 mW), respectively. At each power level, the back reflected power of the loach eye was measured first by power meter right before the PD (point Y in Fig. 1) and then the iris in reference arm was adjusted to make BPR equal to BPS by measuring the total interference power. The power values of each illumination level were listed and compared in Fig. 5 with bar chart with bio-sample photos attached as well. Moreover, although optimal sensitivity cases (BPR larger than BPS) sacrificed image contrast, it is kept here as a comparison group, and BPR was larger than BPS by about 15 dB in this case.

![Power values for different power levels](image)

Fig. 5. Different power value. One: power level one; Two: power level two; three: power level three.

The bio-sample imaging was conducted by conventional SS-OCT, SS-OCT with FOPA and SS-OCT with parametric BD. Each of them was conducted under illumination power...
level one, two and three. The loach eye images were shown in Fig. 6. Each column represents different power levels. And the first row was the images of conventional SS-OCT when BPS was equal to BPR, the second row is SS-OCT with FOPA, the third row is SS-OCT with parametric BD and the fourth row is conventional SS-OCT when BPR was larger than BPS by about 15 dB. Each image was averaged 20 times to reduce the speckle.

According to the conventional SS-OCT (where BPR = BPS) images, represented by the first row, only dark images and some relative high reflectivity could be observed, e.g. cornea and iris, in power level one. However, by increasing the BPR by about 15 dB, the better sensitivity was achieved. Consequently, another eye structure, the lens, emerged (the fourth row). Nevertheless, the respective iris structure was blurred, indicating low image contrast.

SS-OCT with FOPA (where BPR = BPS) images are shown in the second row. In this case, retinal structure was slightly revealed, exhibiting sensitivity improvement. By comparing it to the fourth row at power level one, the sensitivity was improved and the image contrast was well-maintained through the interference signal amplification rather than BPR addition.

The third row represents the images from SS-OCT with parametric BD, when both the signal and idler bands were also utilized and the interference signal (where BPR = BPS) was amplified by OPALS. These images are much clearer than the SS-OCT with FOPA images. For example, in the second column, the retina which was firstly slightly visible in the second row became evident in the third row. This result shows remarkable sensitivity improvement is achieved by adapting parametric BD, better than incorporating FOPA alone. Moreover, even at a lower illumination level, the retina can still be distinguished by the SS-OCT with parametric BD, strongly implying that it could be a promising solution for the in-vivo low illumination power imaging cases when the photodamage, phototoxicity, and photobleaching problems are engaged.

It should also be noted that minimum illumination power is still required to reveal deep structure of the bio-sample, even when the system has been advanced by OPALS. Retina, located in deeper position of the highly-scattered sample, was barely visible when the
illumination power was relatively low (the second and third columns at SS-OCT with parametric BD case).

4. Conclusions and future work

This work demonstrated, for the first time, a sensitivity enhancement method of the interference-based signal detection approach through all-fiber based FOPA and parametric BD and it was subsequently applied to a SS-OCT system. As the sensitivity enhancement had been verified, this method could potentially be a powerful solution to low-illumination in-vivo imaging cases.

The working bandwidth of SS-OCT with OPALS was limited at 76 nm for the band limit of the C/L band WDM coupler and the SOA in FDML laser cavity, which further limited the axial resolution to this sensitivity enhancement scheme. However, it is possible to expand the working bandwidth of SS-OCT with OPALS by both using a customized C/L band WDM coupler which has suitable pass band and selecting a suitable SOA which has a longer amplification wavelength.

This OPALS system can also be further extended to other OCT wavelength regions other than the 1.5-μm window used in this study, e.g. at 1 μm. The 1 μm OCT has several advantages. For instance, the shorter wavelength has better axial resolution than the longer ones (Eq. (2)) and the dispersion of the vitreous component is minimal in the 1 μm window [38–41].

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