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Wavelength-Filter Based Spectral Calibrated Wave number - Linearization in 1.3 mm Spectral Domain Optical Coherence

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Abstract— In this study, we demonstrate the enhanced spectral calibration method for 1.3 µm spectral-domain optical coherence tomography (SD-OCT). The calibration method using wavelength-filter simplifies the SD-OCT system, and also the axial resolution and the entire speed of the OCT system can be dramatically improved as well. An externally connected wavelength-filter is utilized to obtain the information of the wavenumber and the pixel position. During the calibration process the wavelength-filter is placed after a broadband source by connecting through an optical circulator. The filtered spectrum with a narrow line width of 0.5 nm is detected by using a line-scan camera. The method does not require a filter or a software recalibration algorithm for imaging as it simply resamples the OCT signal from the detector array without employing rescaling or interpolation methods. One of the main drawbacks of SD-OCT is the broadened point spread functions (PSFs) with increasing imaging depth can be compensated by increasing the wavenumber-linearization order. The sensitivity of our system was measured at 99.8 dB at an imaging depth of 2.1 mm compared with the uncompensated case.

Index Terms—SD-OCT, Wavelength-filter, wavenumber-linearization.

I. INTRODUCTION

Optical coherence tomography (OCT) is a real-time high-resolution imaging technology which has been developed and popularly used to obtain cross-sectional images of biological tissues. This ultrasound imaging method directs a narrow beam of pulsed light towards a tissue sample. The propagated light beam is backscattered from the tissue. By these backscattered light beams, it is possible to create a depth-resolved line profile of a tissue. A two dimensional (2-D) depth resolved image of the biological tissue can be obtained when the scanning direction is perpendicular to the original propagated light beam. An image with a high resolution of 1~15 µm can be obtained by this imaging method which is considered to be more close to the histology of the imaging object [1-2]. OCT can be classified into time domain (TD) and frequency domain (FD-OCT) according to the structure of the reference arm optics. The frequency domain OCT can be classified into swept-source (SS) and spectral domain OCT (SD-OCT). This classification is done according to the difference between the light receiving detector and the characteristics of the light source [3]. Time domain OCT (TD-OCT), uses a moving reference mirror for measuring the time it takes for light to be reflected. This relatively slow, mechanical process limits both the amount of data that can be captured as well as image quality. Mainly in such a system, the input light beam is split two through the beam splitter. One beam component is propagated to the reference mirror and the other is propagated to the sample arm respectively. The backscattered light beam is detected by a photo detector. A spectrometer- based spectral domain OCT is quite similar to the time domain OCT system. The displacing reference mirror and the photo detector are replaced by a static reference mirror and a spectrometer [4-6].Time domain OCT is a slow, mechanical process limits both the amount of data compared to the spectral (or Fourier) domain OCT (SD-OCT) as it uses a significantly faster, non-mechanical technology. The SD-OCT simultaneously measures multiple wavelengths of reflected light across a spectrum, hence the name spectral domain. The SD-OCT system is faster than TD-OCT as the mechanical A-scan has been eliminated by employing a wavelength-resolving detection scheme using a spectrometer [7]. Accurate spectral calibration is an important factor for obtaining a cross sectional image using SD-OCT. Broadening of the PSFs with increasing depth and the limited spectral resolution can be illustrated as the certain drawbacks of SD-OCT. There is a negative effect on the resolution of the system if the linear relationship between wavenumber and the pixel position is not maintained. The PSFs becomes distorted and broadened due to this factor. Several compensation methods for SD-OCT has been reported previously to overcome these drawbacks. One is a commonly used wavelength and spectral interferogram mapping method [8]. Another method is a simple pixel shift technique to double the spectral sampling rate. It is a widely used method in advanced digital camera to improve the spatial resolution [9]. It can be shown that the main drawback of the wavelength mapping method and the pixel shifting method is the insufficiency of the sampled wavelength or the pixel information. Another method is which analyzes factors influencing sensitivity drop-off in Spectral OCT by employing an optical frequency [10]. The main drawback of this technique is the requirement of an expensive additional hardware unit such as Fabry-Perot interferometer and to compensate the linearity it requires an additional signal processing as well. Another compensation method is placing a prism in the spectrometer to estimate the level of residual non equidistance which can be considered as a costly method and also it has a limitation on the degree of linearity as well [11-12].

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In this paper we describe a rapid calibration method which rescales the spectral information by utilizing the total number of pixels of the array. To fulfill this requirement, a grating based wavelength-filter is designed and demonstrated on behalf of the rapid calibration method, improvement of the signal-to-noise ratio and for the enhancement of the PSF width by the increasing depth. Also this experiment was conducted for the SD-OCT at 1.3 µm which has not been actively studied due to the availability of the SD-OCT at 800 nm regions. Also the line scan camera which used for 1.3 µm has a limited number of pixels and high cost and broadband sources with a large FWHM was available. However, SLD with a bandwidth of more than 150 nm and also high-speed InGaAs cameras were commercialized.

II. EXPERIMENTAL SETUP AND METHODS

A. *1.3µm SD-OCT system and the experiment.*

Figure.1 shows the schematic diagram of the Michelson interferometer based SD-OCT system using the wavelength-filter. The broadband source which was used for the light emission is Denselight SLED broadband source (λc=1310nm ,$\Delta\lambda$=135 nm). 90:10 fiber coupler is used to split the broadband light into reference and sample arms. The sample arm consists of two galvanometer-based optical scanners for the transverse scanning. The scanners are driven by an analog input/output card (PCI-6353; National Instruments, USA). An optical spectrometer is used as the detector containing a collimator, a volume holographic diffraction grating(1145 lines/mm ,Wasatch Photonics, USA), and an achromatic doublet lens.12 bit line-scan CMOS camera with 1024 pixels(SU-1024LDM Compact; Goodrich USA) and a frame grabber(PCIe-1433; National Instruments, USA) is used to capture the interference fringe. In the SD-OCT system, broadband light beam is propagated to the fiber coupler and the beam is split into the reference mirror which performs the depth scan and to the sample arm which performs the lateral OCT scan using galvoscanner. The backscattered beams from the sample and the reference arms are interfered and propagated to the grating based spectrometer. When the beam passes through the grating, the spectral variation of the light beam is occurring and the varied spectrum is propagated to the line scan camera. The obtained raw data using the camera are sent to the computer using frame grabber and the data and signal processing is performed to build up an 2-D image using software programs.

B. *Wavelength-filter*

The wavelength-filter which is designed for the calibration process of the proposed method is separately connected to the SD-OCT system in the figure 1. This wavelength-filter filters the light beam by selecting a narrow light beam from the propagated broadband spectrum. Step motor based translating slit in the figure.1 scans the central wavelength of the filtered narrow light beam over the entire spectrum. The filtered spectrum passes to a spectrometer and to an optical spectrum analyzer OSA. Optical spectrum analyzer provides information about the certain wavelength. The wavelength-filter consists of a collimator, a reflection-type diffraction grating(600lines/mm),a focusing lens(f=75 mm),a movable optical slit (width=80 µm) and a reflective mirror.

C. *The calibration process of the wavelength-filter*

The calibration process is performed according to the following manner. The broadband light beam is propagated to wavelength-filter through the 1st and the 2nd terminals of the optical circulator. When the light beam enters the wavelength-filter, the beam is collimated and then diffracts according to the angle of the wavelength once it strikes the reflection grating. A focusing lens is placed right after the grating at the back focal distance of the lens. The light beam propagates through the lens and become collimates after passing through the lens. A slit and a mirror are placed at the focal plane of the lens. The slit and the mirror are positioned as close as the geometry permits. When the diffracted light beam reaches the slit, it is filtered and a small spectral band can pass through the slit. The wavelength of the transmitting light is dependent on the horizontal position of the slit. When the slit moves horizontally, a specific wavelength is scanned for the entire spectral range. While the wavelength selection is performed in the horizontal direction, the light passing through the slit has vertically focused on the mirror so that the coupling ratio of the power back to the interferometer is maximized. Then the light is reflected from the mirror and then it is collimated again after the focusing lens and converges in the horizontal view. The optical fiber of the collimator is connected to the optical circulator and the light exits from the 3rd terminal of the circulator and connects to the 90:10optical fiber coupler. The returned spectrum has a line width of (~0.5 nm) which propagates to the line-scanning camera of the spectrometer. Once the line-scanning camera obtains the light beam, data processing can be completed and the wavelength spectra can be coregistered along with the pixel positions of their respective center wavelengths which were obtained by using OSA. The wavenumber-linearization pattern can be generated using the start and the end points of the wavelength of the spectrum. To linearize the measured spectrum in the wavenumber-domain, a lookup table can be generated by finding the pixel positions from the measured spectrum that match the wavenumber linearized pixels. Therefor the image display speed is dramatically increased as an additional fitting algorithm is not required during image acquisition.

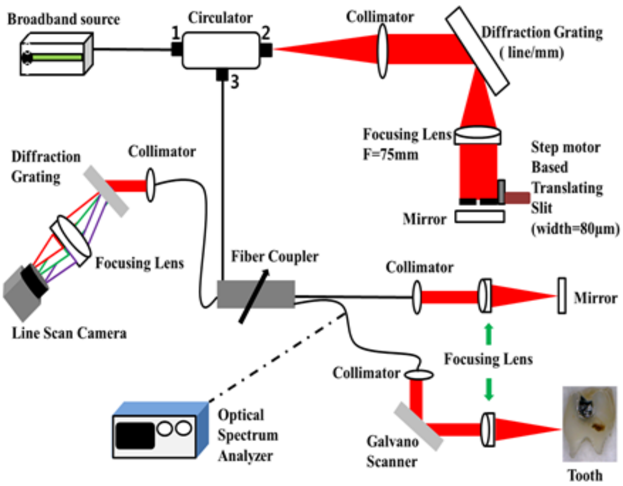

Figure 1. Schematic diagram of the wavelength-filter based SD-OCT.

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III. EXPERIMENTAL REULTS

Figure.2 .A shows the relationship between the measured wavenumber and the pixel position of the wavenumber distribution on the line-scan camera . Figure.2 .B shows the magnified plot of the above figure. The red color curve shows the practically measured wavenumber–pixel-position information using a wavelength-filter and the black dotted curve shows the spectral calibrated wavenumber linearized curve fitting using the start and end wavenumbers. The red color curve shows that experimentally the wave number spacing between the pixel positions is not linear owing to the diffraction grating. The wavenumber-linear pattern was designed by selecting the closest value between the measured wave numbers and the spectral calibrated wavenumber. The calibration order is the number of pixels with known wavenumbers to be used in calibration of the wavenumber-linearization. Therefore the image display speed can be enhanced without using any software fitting algorithm as the proposed calibration method utilizes all the pixels with known wave numbers.

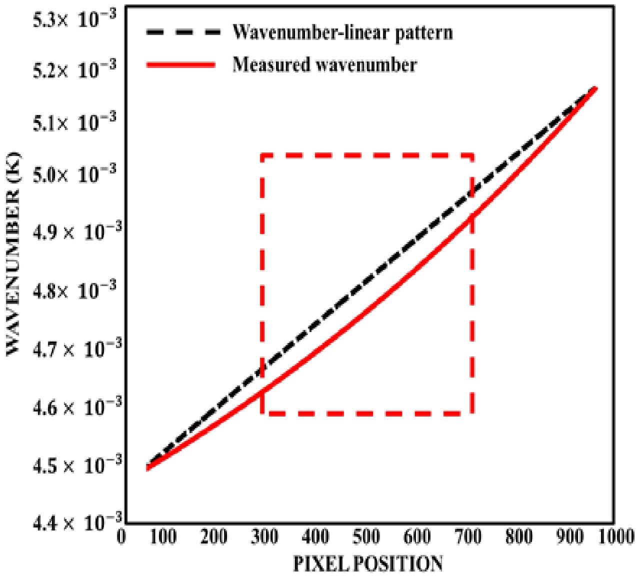

Figure.2.A Wavenumber graph compared with the spectral calibrated wavenumber-linear patte rn.

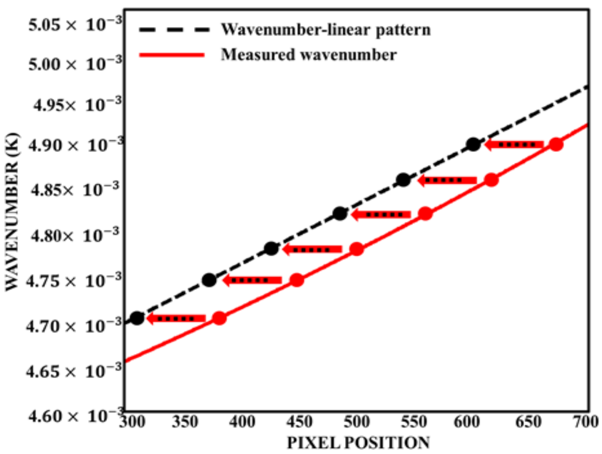

Figure.2.B Magnified plot of the dotted box

Figure.3 shows the results of the measured depth dependent sensitivity with and without spectral calibration. The blue color dotted curves represent the PSF without spectral calibration, and the other colored solid curves represent the PSF after spectral calibration respectively. The width of the PSF is increased compared to the depth before the calibration. When it is calibrated, the width of the PSF is constant with the increase of the depth. This falloff is consistent with or without the calibration. The figure clearly shows a 50 dB drop within a depth of 2.1 mm in both cases as this falloff can be considered as a consistent falloff for both cases.

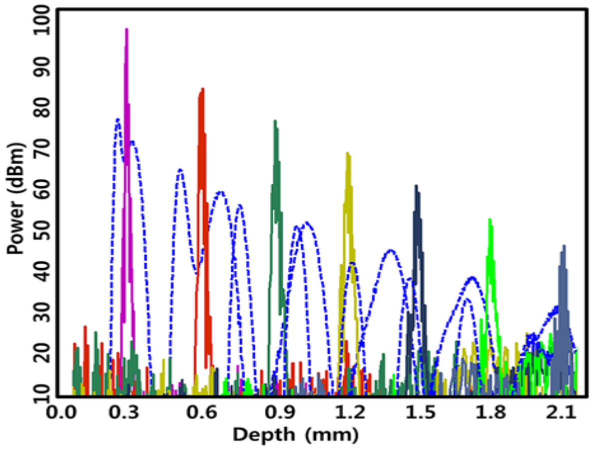

Figure 3. PSF with respect to the depth before (blue dotted curves) and after (color solid curves) the spectral calibration.

Figure.4 shows the labview interface of the SD-OCT system which was used for real-time imaging. Part (a) is the system control panel which controls the galvanometer to according to the variable input data. According to the input details, the galvanometer rotates to acquire 1-D, 2-D or 3-D images. Part (b) is the raw data spectrum which shows the spectrograph of the light source. This observes signal errors of the system under the influence of the external influence and rearranges waveform. Part (c) shows the background profile of the system before the Fourier transformation. Part (d) shows the observed output signal in log units acquired from the Fourier transformed 1-D data. Part (e) shows the real-time display which provides a clear view of a 2-D image of the object.

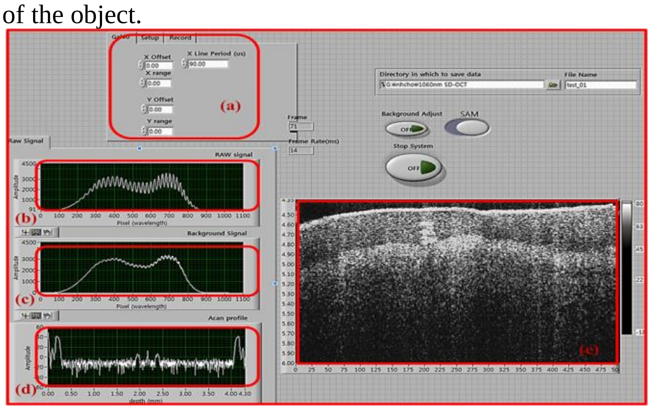

Figure.4 SD-OCT based on Labview interface. . (a) System control panel. (b) Raw data spectrum. (C) Background spectrum. (d) FFT spectrum. (e) OCT image display .

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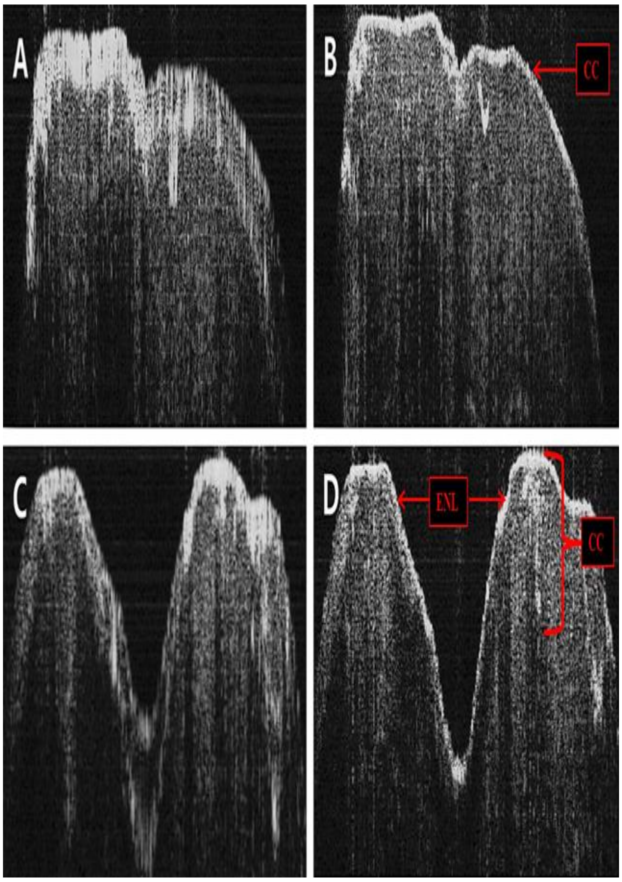

Figure 5. Real-time display OCT images of a human posterior tooth surface (front view A and B), (side view C and D) before and after spectral calibration.

Figure.5 shows the OCT images of a clinical crown (CC) part of a human posterior tooth sample. The A and B images show the front view of the tooth and the C and D images show the side view of the tooth. A and C images were obtained before the calibration process. The B and D images were acquired after the spectral calibration experiment. It can be seen that the internal as well as the external structure of the tooth crown part can be seen clearly after the wavenumber-linearization. Hence B and D images provide a clear and sharp view of the enamel (ENL) and the dentine structures compared to the non-calibrated image. Whoever A and C image has more blurred borders and unclear particles compared B and D images. Comparing to other wavelengths, 1.3μm has a high axial resolution and increased SNR value. Therefore deeper structures can be seen more sharply than the conventional methods.

Figure.6. Shows the top and the side views of the 3-dimensional posterior tooth which was obtained after the calibration. According to the two images, it can be identified that the top and the bottom structures of the images have a sharp view and provides a clear view of the layers after the spectral calibration as the deeper structures has more visibility due to the increase of the SNR.

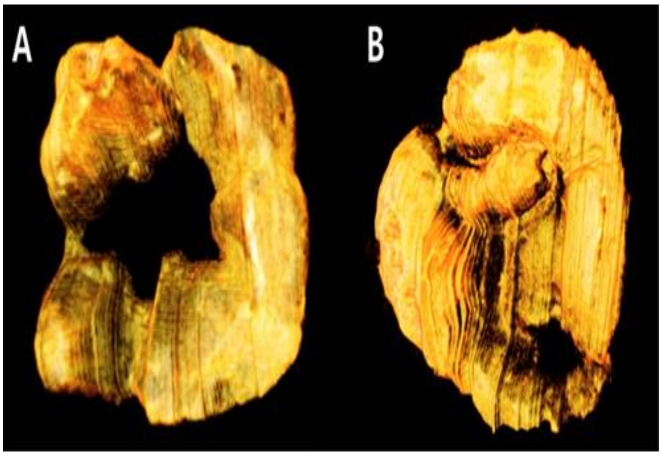

Figure 6. 3-dimensional real-time display OCT images of human posterior tooth surface. (A) side view, (B) top view.

IV. CONCLUSION

A new wavelength-filter based spectral calibration method for 1.3 μm SD-OCT system has been developed. This method does not require any hardware implementation of the SD-OCT system to measure the signal over an increased imaging depth range. This method can directly be used to acquire data as there is no requirement of rescaling wavelength-to-frequency. A wavelength-filter is used to acquire the spectral-pixel information. For the calibration process a lookup table was generated from the acquired data. The wavenumber-linearization was achieved by simply resampling the acquired camera output according to the lookup table. This method does not need any software fitting algorithm in real-time acquisition or display data and also the acquisition and the display rate is increased. Therefore the system provides more position information, surpassing the optimum without compromising the imaging speed. This calibration process is a one-time calibration process which provides an image with a better quality comparing to a un-calibrated image. SD-OCT has become a popular bio-imaging application in the medical field due to its distinct advantages. Specially in the surgical sector, surgery can be done by viewing the real-time images of the OCT. Therefore the proposed method can be used to obtain more information about the biological tissue due to its high sensitivity.Hence, in surgical applications,it is a necessity to obtain a clear and sharp view of biological structures for a successful surgery and this proposed method helps to overcome the drawbacks of the blurred images and unnecessary hardware or software utilizations.

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