Published in final edited form as:

Opt Lett. 2020 April 01; 45(7): 1675–1678. doi:10.1364/OL.382045.

Optically computed optical coherence tomography for volumetric imaging

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Abstract

We describe an innovative optically computed optical coherence tomography (OC-OCT) technology. The OC-OCT system performs depth resolved imaging by computing the Fourier transform of the interferometric spectra optically. The OC-OCT system modulates the interferometric spectra with Fourier basis function projected to a spatial light modulator and detects the modulated signal without spectral discrimination. The novel, to the best of our knowledge, optical computation strategy enables volumetric OCT imaging without performing mechanical scanning and without the need for Fourier transform in a computer.

Optical coherence tomography (OCT) is a high-resolution tomographic imaging modality and has found a wide range of biomedical applications, such as ophthalmology diagnosis, surgical guidance, and tissue characterization for cancer research [1]. One emerging application of OCT is high-speed imaging of dynamic scenes, such as quantitative blood flow imaging, dynamic optical coherence elastography, and large-scale neural recording [2-4]. In many applications of OCT, high-speed OCT imaging in a specific dimension such as in the *en face* plane is more desirable than slow acquisition of the entire 3D volume. However, OCT imaging in an arbitrary dimension still has limited imaging speed, although on average it takes an extremely short period of time to acquire an OCT pixel [5]. To obtain a depth resolved signal, time domain (TD) implementation of OCT performs mechanical scanning in the axial (z) dimension and acquires pixels of an A-scan one after another sequentially. Fourier domain OCT (FD-OCT) was developed afterwards reconstructs all the pixels within an A-scan simultaneously, by taking FD measurement and calculating the Fourier transform of the interferometric spectrum in a computer. FD-OCT eliminates mechanical scanning in axial direction and offers significant advantage in imaging speed and sensitivity compared to TD-OCT [6,7]. Nevertheless, the strategy for 3D data acquisition remains the same for TD and FD implementations of OCT, and has become a major challenge in high-speed imaging of dynamic events.

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Disclosures. The authors declare no conflicts of interest.

For volumetric imaging in a Cartesian coordinate system (x, y, z) , a conventional OCT system (both TD and FD techniques) performs fast axial (z) scanning, and performs slow scanning in the x and y directions to generate a B-scan with multiple A-scans and to generate a C-scan with multiple B-scans by steering the probing beam mechanically. The raster scanning strategy allows high-speed B-mode OCT imaging, but results in an extremely slow speed for 2D imaging in any non-B-scan plane. Consider a plane that is not parallel with the B-scan plane. The normal of the plane, $\mathbf{n}$, is not along y axis of the Cartesian coordinate system. To obtain a 2D image from this plane, the OCT system has to scan the entire 3D volume and select pixels within the plane through postprocessing. In addition to limiting the imaging speed, the current strategy for 3D OCT data acquisition uses mechanical scanners (galvanometers, MEMS scanners, scanning motors, etc.) to steer the light beams, resulting in bulky instrument footprint and complex system configuration. Moreover, the scanning of the 3D volume generates a huge amount of data. It is extremely challenging to acquire, transfer, process, and store the 3D image data. Optical computing that directly uses photons to carry out computation tasks may provide a more efficient way to address 3D spatial coordinate and manage massive data [8]. For example, in Ref. [9], Zhang *et al.* used arbitrary waveform generation to impose fast temporal modulation for optical computation in OCT imaging.

In this Letter, we describe an optically computed OCT (OC-OCT) technology that takes a highly innovative optical computation approach to perform Fourier transform and completely eliminates the need for mechanical scanning in 3D-OCT imaging. Unlike conventional OCT where data acquisition is performed before signal processing, OC-OCT performs signal processing optically before data acquisition. The key component of our OC-OCT system is a spatial light modulator (SLM). SLM has been used for optical pulse shaping, structured illumination, optical computation, and other applications of optical imaging [10,11]. In OC-OCT, we innovatively exploit the capability of SLM in precisely manipulating a light wave to generate output with desired amplitude and phase. Optical computation of Fourier transform is achieved by modulating the interferometric spectra with a programmable SLM and then performing spectrally nondiscriminative detection. The optical computation strategy implemented for OCT imaging in this study is unique and has not been demonstrated before. To the best of our knowledge, we report the first demonstration of volumetric OCT imaging through optical computation.

OC-OCT is a FD technique that achieves depth resolution by optically Fourier transforming a spectral interferogram. Consider an A-scan $\mathbf{S}$ ($\mathbf{S} \in \mathbb{C}^N$ and $\mathbf{S} = [s_1, s_2, s_3, \dots, s_N]^T$). With sample light originating from a specific transverse coordinate, the interferometer generates a spectral interferogram $\mathbf{M}$ after a disperser. $\mathbf{M}$ is a 1D vector ($\mathbf{M} \in \mathbb{R}^N$ and $\mathbf{M} = [m_1, m_2, m_3, \dots, m_N]^T$) and is mathematically related to the spatial domain A-scan through Fourier transform $\mathbf{S} = \mathbf{F}\mathbf{M}$, which is more explicitly shown in Eq. (1) (s_n represents spatial domain OCT signal at the n th discrete depth in an A-scan, m_k represents spectral signal at the k th wavenumber, $\mathbf{F} \in \mathbb{C}^{N \times N}$ is the Fourier transform matrix, and $F_{nk} = e^{j2\pi nk/N}$).

$$\begin{bmatrix} s_1 \\ s_2 \\ \dots \\ s_N \end{bmatrix} = \begin{bmatrix} F_{11} & F_{12} & \dots & F_{1N} \\ F_{21} & F_{22} & \dots & F_{2N} \\ \dots & \dots & \dots & \dots \\ F_{N1} & F_{N2} & \dots & F_{NN} \end{bmatrix} \begin{bmatrix} m_1 \\ m_2 \\ \dots \\ m_N \end{bmatrix}. \quad (1)$$

As illustrated in Fig. 1(a), a conventional FD-OCT system measures the entire interferometric spectrum $\mathbf{M}$ that has N discrete Fourier bins, streams the data into a computer, performs Fourier transform in the computer, and reconstructs the entire A-scan. Figure 1(a) also implies N Fourier bins have to be acquired to fully reconstruct the A-scan $\mathbf{S}$, even if a small subset of pixels is of interest in the A-scan. OC-OCT takes a completely different approach to resolve a pixel in 3D space.

According to Eq. (1), s_n the OCT signal at the n th discrete depth in an A-scan that can be expressed in Eq. (2) shows s_n is the inner product between vector f_n (the transpose of n th row of the Fourier matrix F) and vector $\mathbf{M}$. In Eq. (2), $\bullet$ indicates vector inner product,

$$\begin{aligned} s_n &= F_{n1}m_1 + F_{n2}m_2 + \dots + F_{nN}m_N \\ &= [F_{n1} \ F_{n2} \ \dots \ F_{nN}] \begin{bmatrix} m_1 \\ m_2 \\ \dots \\ m_N \end{bmatrix} \\ &= f_n \bullet \mathbf{M}. \end{aligned} \quad (2)$$

Equation (2) provides an alternative approach to address a spatial location in OCT imaging. As illustrated in Fig. 1(b), the OC-OCT system calculates $f_n \bullet \mathbf{M}$ optically and directly obtains s_n at the point of data acquisition. Optical computation of Eq. (2) is further illustrated in Fig. 1(c). The chosen Fourier basis function (f_n) is projected to the SLM along the dimension of spectral dispersion. The spectrum modulated by the SLM is essentially the element-wise product of f_n and $\mathbf{M}$ ($f_n \circ \mathbf{M}$).

The detector then performs spectrally nondiscriminative detection, generating s_n the inner product between vector f_n and vector $\mathbf{M}$.

The configuration of the OC-OCT system that allows depth resolved imaging with an extended field of view (FOV) is illustrated in Fig. 2. The OC-OCT system uses a broadband source to illuminate the Michaelson interferometer with an extended field of view. A 2D reflective SLM is used for light modulation, and a 2D camera is used for signal detection.

The imaging principle of OC-OCT is explained as the follows. First, the OC-OCT configuration in Fig. 2 establishes a one-to-one mapping between the transverse spatial coordinate at the sample plane and at the detector plane, illustrated as solid and dashed light beam profiles in Fig. 2. This is similar to conventional light microscopy. Optical signals originating from the transverse coordinate (x_0, y_0) at the sample is mapped to the same y coordinate ($y = y_0$) at the detector, because the light beam is not altered along y dimension

by the grating or the SLM. On the other hand, the spectrum originating from a different x coordinate arrives at the SLM plane with a global shift proportional to the x coordinate after the diffraction grating. Reflected by the SLM and diffracted again by the grating, the light rays originating from (x_0, y_0) at the sample are collimated and eventually focused to (x_0, y_0) at the detector plane for spectrally nondiscriminative detection, for a magnification of 1 from the sample plane to detector plane without loss of generality. On the other hand, depth resolution is achieved through optical computation. The diffraction grating disperses the output of the interferometer along the x direction, and the SLM projects a Fourier basis function (f_n) to its row at a specific y coordinate ($y = y_0$), as illustrated in the upper right inset of Fig. 2. The spectral interferogram originating from a different x coordinate at the sample is modulated by a laterally (in x dimension) shifted version of f_n , which does not affect the results of optical computation of OCT signal magnitude. Spectrally nondiscriminative detection of the modulated interferometric spectrum generates a depth resolved OCT signal from the n th discrete depth, for pixels corresponding to different x coordinates. Notably, when all the rows of the SLM project the same pattern for spectral modulation, the OC-OCT system generates an *en face* imaging from a specific depth. If different rows of the SLM project different Fourier basis functions, signals can be simultaneously obtained from different depths. Therefore, OC-OCT allows snap-shot imaging from an oblique plane.

We established the OC-OCT system shown in Fig. 2. We used a mounted LED (Thorlabs) at 470 nm with 25 nm bandwidth (1 mm by 1 mm emitter size) as the broadband source. The interferometric spectrum was dispersed by a 600/mm grating, modulated by a 2D SLM (Holoeye LC-R 720), and detected by a complementary metal–oxide–semiconductor (CMOS) camera (Basler acA2000). The achromatic doublet lens in front of the SLM had a focal length of 250 mm, and the achromatic doublet lens in front of the CMOS camera had a focal length of 100 mm. Identical objectives (20X Olympus, dry) were used in the reference and sample arms of the interferometer. The lateral field of view was approximately 0.5 mm by 0.5 mm, limited by the active area of the camera sensor used for imaging and the magnification from the sample to the camera. The maximum axial imaging range was estimated to be 1.2 mm, limited by the digital frequency of spectral modulation imposed by the SLM. To demonstrate 3D OC-OCT imaging within a large depth range, we used achromatic doublet as imaging objectives to obtain results from the 3D phantom made by depositing photoresist layer on silicon substrate. Notably, a polarized beam splitter (PBS) was inserted between the Michelson interferometer and the optical computation module. The PBS functions as a polarizer and an analyzer, and ensures a one-to-one mapping between the pixel value of the SLM and light reflectivity.

Prior to imaging experiments, we calibrated $K(k)$, the mapping between the pixel index (k) in a row of SLM and the corresponding wavenumber K , because the pixels in a row of the SLM generally do not sample wavenumber domain spectral data uniformly. The calibration was achieved by measuring the interferometric spectrum obtained from a specular sample and enforcing linear phase [12]. Axial resolution degradation at different lateral coordinates due to the nonlinearity of $K(k)$ is moderate because of the small FOV. We also calibrated $R(v)$, the mapping between the value v projected to SLM pixels and the actual light

reflectivity (R) of the SLM, because $R(\nu)$ depends on the wavelength and polarization of the incident light, and is generally nonlinear. When ν takes value of Fourier basis function [F_{nk} in Eq. (1)] and is directly projected to the k th pixel in a row of SLM pixels, the spectral modulation is nonsinusoidal, leading to diminished signal amplitude and ghost high harmonic peaks after optical Fourier transformation. To ensure that precise sinusoidal modulation was imposed to the interferometric spectrum, we projected the value of $R^{-1}(F_{nK(k)})$ to the k th pixel in a row of SLM pixels. Moreover, the SLM cannot directly generate complex exponential function needed in Fourier transform [Eqs. (1) and (2)]. Therefore, we projected cosine and sine patterns ($F_{\cos} = (\cos(2\pi nK(k)/N) + 1)/2$ and $F_{\sin} = (\sin(2\pi nK(k)/N) + 1)/2$) to the SLM. We temporally interlace $f_{\cos}$ ($F_{\cos}$ with $k = 1, 2, 3, \dots$) and $f_{\sin}$ ($F_{\sin}$ with $k = 1, 2, 3, \dots$) for spectral modulation, synchronized the data acquisition with the alternation of cos and sin patterns, acquired signals from cosine and sine channels ($s_{\cos} = f_{\cos}^T \mathbf{M} - s_{DC}$ and $s_{\sin} = f_{\sin}^T \mathbf{M} - s_{DC}$), and extracted the magnitude of the OC-OCT signal, $I = (s_{\cos}^2 + s_{\sin}^2)^{1/2}$. With the reference light much stronger than sample light, s_{DC} could be estimated by m_k obtained with the sample arm blocked. To simplify subsequent description, we refer the function projected to the SLM as f_n that was generated after wavenumber calibration, reflectivity calibration, and temporal interlacing.

We first experimentally validated the z sectioning capability of OC-OCT. We assessed the axial point spread function (PSF) of the OC-OCT imaging system, using A-scans obtained from a mirror with an impulse reflectivity profile. We projected a series of complex exponential functions (f_n , $n = 1, 2, 3, \dots$) to different rows (different y coordinate) of SLM pixels. As a result, different rows of the detector received signals modulated by different complex exponential functions and came from different depths of the sample. The axial PSF (a 1D vector) was then obtained by averaging the image directly obtained from the camera along the x direction. We varied the axial position of the mirror using a translation stage, and obtained axial PSFs as shown in Fig. 3(a). The horizontal axis at the bottom of Fig. 3 is the row index (n_R) of the sensor array and is linearly related to the depth z : $z = an_R + b$. We correlated the peak pixel index with the actual axial position of the sample [Fig. 3(b)] and extracted the values for a and b through linear fitting ($a \approx 0.17 \mu\text{m}$ and $b \approx 36.91 \mu\text{m}$) that allowed us to convert n_R to actual depth shown as the horizontal axis at the top of Fig. 3(a). To evaluate the axial resolution, we used a Gaussian envelope to fit the PSFs obtained at different depths, and the axial resolution of our OC-OCT system was estimated to be $5 \mu\text{m}$ according to the full width at half-maximum (FWHM) of the Gaussian function. The experimental axial resolution was slightly inferior to the theoretical axial resolution ($\delta z = 0.44\lambda_0^2 / \Delta\lambda = 3.9 \mu\text{m}$ given $\lambda_0 = 470 \text{ nm}$ and $\Delta\lambda = 25 \text{ nm}$), probably because various optical components resulted in a smaller effective spectral bandwidth.

We demonstrated the capability of OC-OCT for depth resolved *en face* imaging. To achieve *en face* slicing of the sample at depth z_0 , we projected the same modulation pattern (f_{n0}) to different rows of the SLM. We brought the sample, a USAF1951 resolution target, to depth z_0 ($z_0 = 32.30 \mu\text{m}$) and obtained the image shown in Fig. 4(a). The area within the red square is enlarged in Fig. 4(b), in which the smallest discernable structure is the sixth element of the eighth group, suggesting a lateral resolution of $2.2 \mu\text{m}$, similar to theoretical estimation of $2 \mu\text{m}$ that considers the convolution of multiple factors (camera sampling, diffraction limit of

the imaging objective, spectral resolution of the grating). To validate the image in Fig. 4(a) was indeed sectioned through optical computation, we projected f_{n0} to the top rows of the SLM and projected a constant value to pixels at the bottom rows of the SLM. The sample remained at depth z_0 , and other settings remained unchanged. The resultant OC-OCT image [Fig. 4(c)] shows large brightness at the top and appears to be completely dark at the bottom. In Figs. 4(d)-4(f), we further compare *en face* images obtained from the resolution target when the SLM modulation pattern selected different imaging depths ($z_0 = 32.30 \mu\text{m}$, $z_0 + 1.25 \mu\text{m}$, $z_0 + 2.5 \mu\text{m}$). When the virtual plane determined by the SLM moved away from sample surface, the brightness of OC-OCT image decreases.

We also demonstrated OC-OCT for 3D imaging using onion skin cells. We projected the same Fourier basis function to different rows of the SLM to obtain *en face* OCT image at a specific depth. By varying the Fourier basis function, we obtained *en face* images at different depths in Figs. 5(a)-5(c), with $5 \mu\text{m}$ axial displacement in between. Figure 5(d) is the image obtained by averaging OC-OCT signals within a depth range of $15 \mu\text{m}$.

We also obtained 3D rendered volume through OC-OCT imaging. We designed lateral patterns on a laser-plotted polyester-based photomask, and fabricated a 3D phantom by depositing photoresist layer (SU-8 2035) with $37 \mu\text{m}$ elevation on silicon substrate using the photolithography facility at Brookhaven National Laboratory. We changed the modulation function projected to the SLM to acquire *en face* OC-OCT data from different depths for volumetric imaging. With 2D images obtained from different depths (29 *en face* images obtained with a $1.25 \mu\text{m}$ axial interval), a 3D rendered volume is obtained [Fig. 6(a)]. Figures 6(b) and 6(c) show images corresponding to the surface of the silicon substrate and the top of the deposited pattern. Figure 6(d) is the image generated by averaging OC-OCT signal at different depths. Along the red lines in Fig. 6(d), we generated cross-sectional images [Figs. 6(e)-6(g)] using the volumetric data. The rectangles in Figs. 6(e)-6(g) correspond to depth profiles for the areas within rectangles of Fig. 6(d), from which the top of the letters “I” and “T,” middle of the letter “N,” and bottom of the letter “J” are discernable.

The OC-OCT system described in this manuscript enabled optically computed 3D OCT imaging for the first time to the best of our knowledge. OC-OCT is fundamentally different from existing technologies that take the transverse plane as the preferential scanning dimension. For optical coherence microscopy (OCM) and full-field OCT, mechanical scanning cannot be eliminated. One significant advantage of OC-OCT is its flexibility in data acquisition. In this study, we performed 3D imaging by projecting the same Fourier basis function to different rows of the SLM and sequentially acquiring *en face* images at different depths. If fast imaging is needed in an oblique plane, the OC-OCT system can project a different Fourier basis to different rows of the SLM and make the oblique plane the dimension for preferential data acquisition. For structural OCT imaging, we measured the real and imaginary parts of the complex OCT signal with the SLM generating cosine and sine modulations and calculated the amplitude of the OCT signal. The real and imaginary parts of the complex OCT signal can also be used for phase resolved imaging that is sensitive to nanometer scale displacement, in applications such as optical coherence elastography and imaging cell dynamics. The current OC-OCT system generated temporally

interlace cosine and sine patterns for spectral modulation. Hence, its imaging speed was limited by the speed of the SLM (60 Hz refreshing rate). Dispersion mismatch can also be compensated through optical computation by introducing a nonlinear phase term to the Fourier basis projected by the SLM. To fully utilize the speed of the camera, complex modulation of the interferometric spectrum can also be achieved by projecting spatially interlaced cosine and sine patterns to the SLM. We estimated the sensitivity of our OC-OCT system to be 87 dB, limited by the small dynamic range of the camera. The imaging quality is also affected by lateral cross talk, which can be effectively eliminated by structural illumination/detection using the SLM.

Acknowledgment.

We thank the Brookhaven National Laboratory for the use of their equipment.

Funding. National Cancer Institute (1R15CA213092-01A1).

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Fig. 1. Data flow in (a) conventional FD-OCT and (b) in OC-OCT; (c) Fourier transform of spectral interferogram.

Fig. 2. Configuration of the OC-OCT system for 3D imaging. BS, beam splitter; PBS, polarization beam splitter; OBJ, objective.

Fig. 3.

(a) A-scan obtained from a mirror at different depths; (b) linear relationship between the n_R and the actual depths.

Fig. 4.

(a) *En face* image of USAF 1951 resolution target; (b) the sixth element of the eighth group in the resolution target can be resolved; (c) top part of SLM was programmed to obtain OCT signal from the resolution target; (d)–(f) *en face* images of the resolution target when the plane chosen by the SLM moved away from the sample surface.

Fig. 5.

(a)–(c) OC-OCT images of onion skin cells at different depths; (d) image generated by averaging signal at different depths. Scale bars represent 50 μm .

Fig. 6.

(a) 3D rendering image of 3D phantom; (b) *en face* image from the substrate and (c) the top of photoresist layer; (d) *en face* image generated by averaging OC-OCT signal within a depth range of 36 μm ; (e)–(g) cross-sectional images of three positions indicated by red lines in (d). Scale bars represent 300 μm .