Published in final edited form as:

Opt Lett. 2014 August 1; 39(15): 4392–4395.

Generic pixel-wise speckle detection in Fourier-domain optical coherence tomography images

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Abstract

We present a generic phase-domain processing method for detecting speckles in Fourier-domain optical coherence tomography (OCT) images. The physics behind the interferometry is revisited and analytically along with simulation results it indicates that the speckle formation comes with phase distortion to the complex OCT signal. The first and the second derivatives of phase along the imaging depth are then calculated for speckle identification. The phase-domain processing method was applied to images acquired by both spectral-domain OCT and swept-source OCT systems, and the experimental results show that this method enables pixel-wise speckle identification.

Optical coherence tomography (OCT) is a powerful imaging modality that utilizes temporal coherence gating to achieve high axial resolution. Due to the nature of temporal coherence interferometry, speckle is inevitable in OCT images. On one hand, speckle overlays a grainy feature on the original structural image which degrades image quality; on the other hand, speckle pattern changes with respect to various samples, carries information of the sample, and is thus favored in some applications, such as flow imaging. It has been recognized that OCT speckles originate from the mutual interference among all the OCT interferometric signals resulting from the multiple backscattered fields within the sample [1]. Since these interferometric signals also carry the structural information and constitute OCT imaging contrast, it is difficult to delineate individual speckle on OCT images. Methods have been sought to study the statistical properties of speckles [2,3]. However as indicated in [3], the statistical property of speckles studied locally can be substantially different from that studied globally over a large number of samples, and it remains an open question whether each individual pixel on an OCT image is speckle noise or bears structural information of the sample.

Yung *et al.* discussed a Z-transform adjustment procedure in time-domain OCT [4], which was previously applied to medical ultrasound [5], to enable speckle identification. In [4], speckle detection only targeted for where maximum destructive interference occurs, i.e.,

when two backscattered events were apart by half of the central wavelength of the OCT source. Besides, complicated system modification was required to construct complex demodulated OCT signals for Z-transform. In Fourier-domain OCT (FD-OCT), however, such a method cannot be directly applied for direct identification of speckle.

In FD-OCT, the phase information extracted from OCT signals has led to several applications, such as phase-resolved Doppler imaging, which relies on the relative phase change between temporal scans [6], and spectral-domain optical coherence phase microscopy, which relies on the absolute phase corresponding to the probed location [7]. In this Letter, we propose a new phase-domain processing method that uses the phase information of the complex OCT signals for speckle detection. This method enables speckles to be identified pixel-wise on OCT images, hence addressing the question whether a given pixel on the image is a speckle or not. As detailed later, this method identifies speckle by excluding those “nonspeckle” pixels and is not limited to the case of maximum destructive interference. Furthermore, it requires no additional system modification, is easy to implement with little computational cost, and is generic to any FD-OCT system.

In FD-OCT, the signal at the detector for a particular wavenumber k can be expressed as

$$I(k) = \langle |\mathbf{E}_R(k) + \mathbf{E}_S(k)|^2 \rangle \\ = \langle |\mathbf{E}_R(k)|^2 \rangle + \langle |\mathbf{E}_S(k)|^2 \rangle + \langle \mathbf{E}_R^*(k) \mathbf{E}_S(k) \rangle + \langle \mathbf{E}_R(k) \mathbf{E}_S^*(k) \rangle, \quad (1)$$

where $E_S(k)$ and $E_R(k)$ represent the electrical field reflected from the sample and reference arms, respectively, and the symbol $\langle \cdot \rangle$ denotes a time average of the electrical field over the detector integration time. For a backscattered event in the sample arm with an optical path length difference $2z_1$ with respect to the reference arm, the cross-interference term denoted as $I_{\text{OCT}}(k)$ can be written as from Eq. (1):

$$I_{\text{OCT}}(k) = 2r_R r_S(z_1) S(k) \text{Re}[\exp(-i2kz_1)], \quad (2)$$

where r_R and $r_S(z_1)$ denote the reference and sample arm electrical field reflectance, respectively, and $S(k)$ the source power spectral density. In writing Eq. (2) it has been assumed that only the temporal coherence needs to be considered. This assumption is valid for most OCT systems implemented with a single-mode fiber-based interferometer, where the spatial coherence is high [8]. For OCT systems with extended source dimension at the entrance pupil (e.g., full-field OCT) [9], the spatial coherence also needs to be considered for Eq. (2). Taking inverse Fourier transform with respect to wavenumber k and ignoring the $\text{Re}[\cdot]$ operator, Eq. (2) leads to

$$\mathfrak{F}^{-1}\{I_{\text{OCT}}(k)\}\{z\} = 2r_R r_S(z_1) \mathfrak{F}^{-1}\{S(k)\}(\hat{z})|_{z=z-2z_1}, \quad (3)$$

where $\mathfrak{F}^{-1}\{F(\xi)\}(x)$ denotes inverse Fourier transform of function $F(\xi)$ with variable x , and for simplicity, r_S is assumed to be wavelength independent. In Eq. (3), the modulus of $\mathfrak{F}^{-1}\{I_{\text{OCT}}(k)\}(z)$ consists of the OCT image. It should be noted that in reality the measurements

are real-valued spectral signals as indicated in Eq. (2). Taking inverse Fourier transform of the measurements leads to the complex conjugate ambiguity in FD-OCT images.

According to the Wiener–Khinchin theorem, the spectral density $S(\omega)$ and the temporal autocorrelation function $G(\tau)$ form a Fourier transform pair, i.e., $G(\tau) = \mathfrak{F}\{S(\omega)\}(\tau)$. Following the notation in [10], $G(\tau)$ is defined as

$$G(\tau) = \langle \mathbf{E}_0^*(t+\tau) \mathbf{E}_0(t) \rangle, \quad (4)$$

where τ is the time delay between two temporal measurements. Further, the general form of $G(\tau)$ can be expressed as

$$G(\tau) = I_0 \gamma(\tau) \exp\{-i[\omega_0 \tau - \alpha(\tau)]\}, \quad (5)$$

where $I_0 = G(0)$ is a constant normalization factor based on the assumption that the light source is statistically stationary; $\gamma(\tau)$ is the normalized modulus of $G(\tau)$, which denotes the coherence gating ability of the source and provides the axial resolution of OCT; $\alpha(\tau)$ depends on the nature of the light source spectrum. By introducing the Wiener–Khinchin theorem, substituting Eq. (5) into Eq. (3), and writing in term of kz instead of $\omega\tau$, we have

$$\mathfrak{F}^{-1}\{I_{\text{OCT}}(k)\}(z) = 2r_R r_S(z_1) I_0 \gamma[2(z-z_1)] \cdot \exp\{-i\{2k_0(z-z_1) - \alpha[2(z-z_1)]\}\}, \quad (6)$$

where the factor 2 in the arguments comes from the fact that the optical path length in the sample is denoted by $2z$ with z representing the depth.

Equation (6) explicitly shows the phase of the OCT signals for a single backscattered event from the sample, which is a linear term $2k_0(z-z_1)$ with respect to depth z plus a modification function $\alpha[2(z-z_1)]$.

For the case where multiple reflected fields are detected from the sample, Eq. (2) has to be modified. The electrical field $\mathbf{E}_S(k)$ in Eq. (1) is now the sum of all the reflected fields from the sample arm. Considering the inverse Fourier transform applied on Eq. (2) is a linear operation, Eq. (6) can be rewritten for n simultaneously detected backscattered events as

$$\mathfrak{F}^{-1}\{I_{\text{OCT}}(k)\}(z) = 2r_R I_0 \sum_{j=1}^n \{r_S(z_j) \gamma[2(z-z_j)] \cdot \exp\{-i[2k_0(z-z_j) - 2\alpha(z-z_j)]\}\}. \quad (7)$$

For a light source with a Gaussian shape power spectral density $S(\omega)$, $\alpha(\tau)$ in Eq. (7) becomes zero, which can be easily shown by taking the inverse Fourier transform of a Gaussian $S(\omega)$. Thus, for a Gaussian light source in an FD-OCT system, Eq. (7) becomes

$$\mathfrak{F}^{-1}\{I_{\text{OCT}}(k)\}(z) = 2r_R I_0 \sum_{j=1}^n v_j(z), \quad (8)$$

$$v_j(z) = r_s(z_j) \gamma[2(z - z_j)] \exp[-i2k_0(z - z_j)], \quad (9)$$

where $v_j(z)$ denotes the interferometric signal formed by the backscattered field at depth z_j within the sample and the backscattered field from the reference arm. The right-hand side of Eq. (8) is the complex FD-OCT signals obtained after Fourier transform of the acquired raw data. In writing Eq. (8), it is assumed that within one A-scan the phase differences of all the reflected fields from within the sample are caused only by their optical path lengths. This assumption, while generally not explicitly stated, is implied in most literature [11,12].

Again, Eqs. (8) and (9) indicate that for a single backscattered event within the sample, the phase of the OCT signal is linear with respect to the imaging depth. While for multiple backscattered events, the phase at a certain depth z is determined by the mutual interference of all the interferometric signal $v_j(z)$ for $j = 1, 2, \dots, n$, representing the contributions from all the backscattered events from z_1 to z_n . In the case of speckle, the linearity of phase along the depth is disturbed. Hence by detecting the pixels that have phase distortion, the speckle areas can be identified. Figures 1(a)–1(c) illustrate the simulation of speckle formation and the corresponding phase distortion in one A-scan. In the simulation, the coherence length was set at 3 μm . The OCT signal was assumed to result from two backscattered events at depth 9 and 10.9 μm in the sample, denoted as $v_1(z)$ and $v_2(z)$, respectively. Figure 1(a) illustrates the modulus of $v_1(z)$ and $v_2(z)$, and the modulus of the complex OCT signal resulting from the interference between $v_1(z)$ and $v_2(z)$, which were calculated using Eqs. (8) and (9). It is seen that a speckle was formed at depth around 10 μm . Figures 1(b) and 1(c) plot the first and second derivatives of the phase of the complex OCT signal along the depth, respectively. It is clear that at the depth where the speckle was formed, the phase distortion occurred.

In this work, we use both the first and second derivatives of the phase along the depth to determine where phase distortion occurs. By examining the second derivative in addition to the first derivative, the sensitivity of this phase-domain processing method for speckle detection is increased. As illustrated in Figs. 1(b) and 1(c), at depths where the second derivative of the phase along the depth shows maximum deviations from that of the nonspeckle area, the deviation of the first derivative of the phase along the depth has decreased from its maximum (peak) value and might be hardly differentiated from that of the nonspeckle area in the presence of system phase noise.

In practice, the phase is first unwrapped over the depth profile after Fourier transform is performed to the raw data of each A-scan. The first derivative of the phase along the depth is then calculated. The pixels whose phase first derivatives are outside the range (denoted as $\Phi / z_m \pm \Delta$) are considered to have phase distortion, where Φ / z_m is the median of the phase first derivative for the entire A-line. In order to determine the range parameter Δ , a relatively “homogeneous” region on the image is selected. The intensity variance of the pixels within this region after excluding the as-identified speckles is calculated with respect to the range parameter Δ . The intensity variance is ideally zero for a homogeneous region; however, it is influenced by tissue heterogeneity and speckles. By excluding speckles, the variance will decrease. Hence the range parameter Δ is chosen when the intensity variance

of the selected region after excluding the as-identified speckles reaches the minimum. To increase the sensitivity of speckle detection, the second derivative of the phase along the depth is also calculated and the same procedure is applied, as a supplement to the results obtained from the first derivative of phase along the depth. All the pixels whose either the first or the second derivative is outside the predefined range will be marked as speckles.

It is noticed that this approach is purely processed in the phase domain and thus is termed as the phase-domain processing method. Figure 2 illustrates part of an A-scan signal measured from a resin phantom embedded with 25 mg/mL silica nanospheres by an 800 nm spectral-domain OCT (SD-OCT) system. The measured axial resolution of the system was 3.36 μm in air and the lateral resolution was 7 μm . In Fig. 2(a), the detected speckle pixels using the above method are plotted with red circles together with the A-scan data. Figures 2(b) and 2(c) illustrate the first and second derivative of phase along the depth corresponding to the same pixels as those in Fig. 2(a), respectively. It is noticed that the baseline median Φ/z_m in the first derivative of phase along the depth displayed in Fig. 2(b) was removed. The ranges denoted by the blue dashed boxes in Figs. 2(b) and 2(c), respectively, were used to indicate the thresholds for the phase derivatives, outside which the significant phase change was regarded as distortion that comes with speckles at the corresponding depth (or pixel). This phase-domain processing technique can also boast the subnanometer sensitivity to depth that is similar to the spectral-domain optical coherence phase microscopy [7] and enables speckles to be delineated pixel by pixel. It should be mentioned that pixels whose intensities are within the noise level tend to have random phases. Thus, this phase-domain processing method tends to lead to “false” speckle at areas where signal is low (e.g., deeper depths where the backscattered fields are severely attenuated). One solution to this problem is to set a threshold for the intensity. The speckle pixels with intensities below the threshold are regarded as “false” speckles. Furthermore multiple A-lines can be acquired and averaged for each location to reduce the random noise, which in turn will reduce the chance of getting false speckles.

To demonstrate the performance of this pixel-wise phase-domain processing method, we applied the algorithms to a fingertip OCT image acquired by the SD-OCT system mentioned above. Figure 3(a) illustrates the OCT image. Two small regions were randomly selected on the OCT image as indicated by the two yellow boxes in Fig. 3(a), and their 7 $\times$ magnified versions are shown in Figs 3(b) and 3(c), respectively, where the detected speckle areas are plotted in red and overlaid onto the gray-scale structural images for direct comparisons. In order to test the above phase-domain processing method on an SD-OCT system with different coherent lengths (and different speckle patterns), a Gaussian window was applied to the raw data to bandpass the source spectrum and reduced the axial resolution to 8 μm in air. The results of the same areas as marked in Fig. 3(a) are illustrated in Figs. 3(d) and 3(e). Comparing the speckle areas in Figs. 3(b) and 3(c) with those in Figs. 3(d) and 3(e), we observe denser speckles with finer structures along the axial direction due to the shorter coherence length. As the lateral resolution keeps the same, the speckle patterns in the lateral direction are similar in Figs. 3(b)–3(e).

In addition to SD-OCT images, a fingernail image acquired by a 1300 nm swept-source OCT (SS-OCT) system with a 9 μm axial resolution in air and 16 μm lateral resolution was

also processed with the same algorithm to demonstrate that the phase-domain processing method is generic for Fourier-domain OCT images. The results are illustrated in Figs. 4(a) and 4(b), where Fig. 4(b) corresponds to the area marked by the yellow box in Fig. 4(a), which is magnified by seven times.

In summary, this Letter reports a phase-domain processing method to pixel-wise identify speckles on FD-OCT images. The pixels whose first or second derivatives of the phase along the depth are outside a preset range are regarded as speckles. This method does not rely on the absolute phase value, and hence it is immune to the bulk phase change among individual A-scans. The impact of this work is manifold. One direct application is speckle reduction for FD-OCT images. By correcting only the speckle areas while leaving the rest unaltered, the speckle noise can be greatly reduced without compromising the image quality. Another application is to differentiate various tissue types based on the speckle patterns, which is similar to the one reported in [13]. As this work proposes a method to directly identify speckles, further quantitative metrics such as speckle density can be obtained. The detected pixel-wise speckle is also potentially beneficial to speckle variance OCT [14] for flow imaging by only evaluating the speckle areas.

Acknowledgments

This research was supported in part by the National Institutes of Health under grants R01EB007636, R01CA153023.

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Fig. 1.

Simulation of speckle formation. (a) The intensity of the OCT signal in one A-scan where speckle is formed by the interference of two backscattered events $v_1(z)$ and $v_2(z)$. (b) The first derivative of the phase along depth for the OCT signal in (a). (c) The second derivative of the phase along depth for the OCT signal in (a).

Fig. 2.

A-scan signal along depth measured from a silica-nanosphere embedded resin phantom by an SD-OCT system. (a) A-scan data and the detected speckle pixels indicated with red circles. (b) The first derivative of phase along depth. (c) The second derivative of phase along depth. The blue boxes in (b) and (c) denote the phase derivative thresholds, and values outside the boxes are regarded as distortion that comes with speckles.

Fig. 3.

Detected speckles from the image acquired by an SD-OCT system with different coherence lengths. (a) A representative fingertip OCT image. (b), (c) The detected speckles (in red) corresponding to the areas marked in (a) with a $3.36\ \mu\text{m}$ coherence length. (d), (e) The detected speckles corresponding to the areas marked in (a) with an $8\ \mu\text{m}$ coherence length. The speckles were plotted in red and overlaid onto the gray-scale OCT images.

Fig. 4. Detected speckles from the image acquired by an SS-OCT system. (a) A representative fingernail image. (b) The detected speckles corresponding to the area marked in (a). The speckles were plotted in red and overlaid onto the gray-scale OCT image.