Angle-resolved Optical Coherence Tomography with sequential angular selectivity for speckle reduction

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Abstract

We present a novel method for rapidly acquiring optical coherence tomography (OCT) images at multiple backscattering angles. By angularly compounding these images, high levels of speckle reduction were achieved. Signal-to-noise ratio (SNR) improvements of 3.4 dB were obtained from a homogeneous tissue phantom, which was in good agreement with the predictions of a statistical model of speckle that incorporated the optical parameters of the imaging system. In addition, the fast acquisition rate of the system (10 kHz A-line repetition rate) allowed angular compounding to be performed in vivo without significant motion artifacts. Speckle-reduced OCT images of human dermis show greatly improved delineation of tissue microstructure.

1. Introduction

Optical coherence tomography (OCT) has emerged as a powerful tool for non-invasively probing the microstructure of biological tissue at high-speed. As with other imaging modalities that employ coherent detection, such as synthetic aperture radar and B-mode ultrasound, OCT images are confounded by speckle noise. Speckle imposes a grainy texture on images that reduces the signal-to-noise level to near unity values [1]. In medical imaging contexts, speckle can reduce the effective spatial resolution, concealing subtle differences in scattering properties known to be crucial for differentiating normal from diseased tissue states.

Speckle results from the coherent addition of waves with a random phase distribution. Photons propagating along different paths within biological tissue acquire random phase shifts as a result of scattering by a wide variety of structures including cell membranes, mitochondria, and nuclei. One fundamental difference between speckle and other noise sources of OCT systems such as detector noise and shot noise is that, in the absence of sample and probe motion, the speckle noise pattern is static in time.

The crux of the speckle reduction problem in OCT consists of obtaining uncorrelated measurements of speckle without significantly affecting spatial resolution. Whilst digital image processing methods [2-3] have met with some success, they are fundamentally limited by their reliance on statistical relationships between neighboring pixels. As such, they typically involve a compromise between the extent of speckle reduction and the loss in spatial resolution. Spatial averaging in the transverse dimension could be performed in conjunction with a corresponding decrease in the focused spot size, but with this method speckle reduction would be achieved at the expense of a decreased confocal parameter. The angular compounding method of speckle

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reduction exploits the decorrelation of speckle with respect to the angle at which light is backscattered, thereby presenting the potential for skirting spatial resolution compromises [4-7]. This method involves incoherently averaging images that are acquired from different backscattering angles. It was recently demonstrated that angular compounding allows for speckle reduction levels of 8 dB, as implemented with a massively-parallel detection system [7]. This system had one main drawback, however: the slow (25 Hz) A-line rate of the system rendered it unsuitable for imaging in vivo.

We present a novel method for acquiring OCT images that allows for both high levels of speckle reduction by means of angular compounding and high A-line rates. The number of angular images acquired by the system could be tuned in order to achieve a compromise between imaging speed and the level of speckle reduction. As a system suitable for imaging in vivo, it represents a significant development from the parallel-detection approach to angle-resolved OCT.

2. Imaging System

2.1 Interferometer

The angle-resolved OCT system consisted of a fiber-optic interferometer, a light source, and detection electronics [Fig. 1(a)]. The interferometer directed light simultaneously to a transmissive sample arm, which delivered and received light from the sample, and to a transmissive reference arm. The light source was a wavelength-swept laser. Interference between light from the reference and sample arms was detected as a function of source wavelength with a balanced, polarization-diverse detection circuit [8].

Optics at the head of the sample arm allowed for light from different backscattering angles to be detected selectively (Fig. 1b). The light beam incident on the sample was focused by an achromatic doublet lens with a diameter of 25 mm and a focal length of 35 mm. The spatial intensity profile of the collimated beam originating from collimator C1 in the sample arm was measured directly using a CCD. From this profile, the transverse spot size ($1/e^2$) and confocal parameter in air were calculated to be 14.3 μm and 0.984 mm, respectively. Light reflected from the sample was diverted by a beam splitter; subsequently, it was deflected by the galvanometer mirror M1 (Cambridge Technology) that was placed at the foci of two achromatic doublet lenses. A collimator placed after the second lens served to couple light back into the interferometer. As such, there was a direct correspondence between the angle of M1 and the backscattering angle from which the received light originated. The angular deflection of M1 was chosen so that light backscattered in the range of 172 to 188 degrees was collected. Deflection of the sample arm light transversely across the sample was performed by a second galvanometer mirror M2 (Cambridge Technology) that was scanned with a saw-tooth waveform. The deflections of M1 and M2 were synchronized with data acquisition so that individual cycles of M1 and M2 corresponded to integral numbers of A-lines.

The laser source employed a semiconductor optical amplifier and an intra-cavity filter [9]. It generated polarized light with an output power of 25 mW averaged across the tuning range of 1212 to 1347 nm. The intra-cavity filter of the laser light source contained a 48 facet polygon mirror with a rotation rate of at 12500 rpm (Lincoln Laser), resulting in an A-line rate of 10 kHz. The filter employed a double-pass configuration in which light was deflected from the polygon mirror onto a stationary mirror prior to retracing its path [10]. The average reference arm power incident on individual detectors within the balanced receivers (Thorlabs PDB-110C) was 20 μW. The system sensitivity was 101 dB, as measured from an axial reflectivity profile of an attenuated mirror corresponding to a single backscattering angle. The penetration depth corresponding to this image is expected to be lower than that expected from a conventional OFDI system with a sensitivity of 110 dB [11], due in part to the loss of 6 dB in sensitivity.
from the beam splitter within the sample arm. The Nyquist-limited ranging depth was 2.85 mm in air.

Analog input was digitized at 12 bits by a digital acquisition card operating at 10 MS/s per channel (National Instruments PCI-6115). Triggers for A-line acquisition were obtained optically: a fraction of the light from the reference arm was diverted to a fiber Bragg grating, and the reflected light was detected by an InGaAs photoreceiver (New Focus 1811). For each A-line, 950 wavelength samples were acquired. The signals from the photoreceiver were converted to trigger pulses using a custom circuit that allowed for variation of the pulse delay. Imaging was performed with 40 angular samples per spatial volume element. With a total of 86,400 A-lines acquired for the image, each angle-resolved image comprised 2,160 A-lines.

2.2 Image Reconstruction

For each backscattering angle, axial reflectivity profiles were obtained by Fourier transform of the interference signals. The data processing steps that were performed prior to Fourier transform were background subtraction, mapping from wavelength- to $k$-space, digital dispersion correction [12], and multiplication with a Hamming window [11,12]. These steps were performed using compiled code written in C; subsequent processing was performed with Matlab (Mathworks).

Images corresponding to different backscattering angles were offset from one another in the axial dimension, due to slight misalignments in the sample arm. As a calibration step, this offset was measured for each angular image. Prior to angular compounding, the inverses of the measured offsets were digitally applied to the angular images so that they were aligned in the axial dimension with the reference image. The collection efficiency of the sample arm corresponding to each backscattering angle was measured, and was employed as a normalizing factor before angular compounding was performed.

Angular compounding consisted of averaging the squared magnitudes of the reconstructed complex reflectance signals acquired at different backscattering angles. Reflectivity magnitudes were mapped to a logarithmic scale to generate two-dimensional images. For all angular compounding operations, angles that were evenly spaced across the full angular range of the system were chosen. For instance, for compounding across 3 angular samples, angles of -172 degrees, 180 degrees, and 188 degrees were chosen. Images derived without angular compounding corresponded to a backscattering angle of 180 degrees.

3. Signal-to-Noise Ratio and Angular Compounding

The signal-to-noise ratio (SNR) relates the mean power reflectance, $\langle I \rangle$, to the standard deviation of the power reflectance, $\sigma_I$. SNR = $\langle I \rangle / \sigma_I$, where the average is performed over pixels from an optically homogeneous region [7]. We note that the SNR differs from the system sensitivity, which is the minimum detectable reflectance in the absence of speckle. The dominant contribution to signal variation within scattering regions is typically speckle. When angular compounding is performed, the SNR is increased, and the magnitude of the increase is determined by the extent to which angular samples are correlated. We emphasize that the SNR differs from the system sensitivity, as the latter is a measure of the minimum detectable reflectance in the absence of speckle.

In this section, we present a computational framework that relates the SNR improvement resulting from angular compounding to the system optical parameters. It is constructed from more detailed analyses of OCT speckle statistics that do not incorporate angular measurements [13]. We employ a linear systems framework, because the measured backscattered field $S$ and...
the actual backscattered field $G$ corresponding to a particular spatial volume element are related by the convolution operator $\otimes$:

$$S(x) = K(x) \otimes G(x).$$  \hfill (1)

The parameter $x$ is related to the angular backscattering angle $\theta$ and the focal length of the lens $f$ as $x = f \tan(\theta)$. Simply put, it is the radial displacement of a beam that has been backscattered at angle $\theta$, as measured on the side of the imaging lens L1 opposite to the sample (Fig. 2). The angular response kernel $K$ is the field magnitude of the collimation beam, and it is responsible for the correlations between angular samples. $K$ operates separately on the real and imaginary parts of $G$, so that $S$ is complex-valued. It was measured by directing light from the collection collimator at an area camera, and calculating the square root of the measured radial intensity distribution.

Using the linear systems framework above, SNR improvements can be determined numerically using a numerical simulation consisting of the following steps:

a. Simulating the speckle field $G$ by drawing $n$ independent and identically-distributed (I.I.D.) random values from a standard normal distribution, independently for real and imaginary parts of $G$, where $n$ is the number of measured angular data points.

b. Calculating the effect of the angular response by convolving $G$ with $K$;

c. Calculating the power reflectance for each angular data point by taking the magnitude-squared of the $n$ values obtained from step b), so that the resulting probability distribution for a given backscattering angle is exponential;

d. Calculating the SNR improvements for averages over different numbers of angular data points, ranging from 1 to $n$. Averages are performed with the results of part c);

e. Repeating steps a) to d) and averaging the results. The number of repetitions was chosen to be 10,000.

The SNR is known to increase in proportion to the square root of the number of uncorrelated angular samples [1]. For the case where angular correlations are present, we define the effective number of uncorrelated angles as the square root of the SNR improvement obtained when all angular data points are included in angular compounding. In this simulation, the number of effective angles is determined by two factors: the diameter of the lens, which determined the minimum and maximum values of $x$, and the width of the collection beam, as determined by $K$. We note that there were no arbitrary parameters in the simulation.

4. Imaging

4.1 Phantom

A solid optical phantom was constructed by mixing TiO$_2$ powder (Sigma-Aldrich, 634662) with two-part silicone (GE, RTV 615). Centrifuging was performed prior to curing in order to remove large TiO$_2$ clusters. This phantom was well suited to measurements of SNR and angular correlation measurements due to its homogeneity.

A representative angular distribution for a resolution element located 20 $\mu$m beneath the surface is shown in Fig. 3(a), from which correlations between adjacent angular samples are readily observed. The corresponding cross-correlation function is shown in Fig. 3(b). The SNR obtained from reflectivities corresponding to a single polarization channel was found to increase from 1.2 with a single angular sample, to 2.6 with compounding across 40 angular
It follows that the effective number of uncorrelated angles for this system is 6.8. The relationship between the SNR and the number of compounded samples was in good agreement with the numerical simulations described in Section 3 (Fig. 4). The measured increase in SNR was slightly lower than the simulated values for small numbers of compounded angles, however, which suggests that correlations between reflectivities widely separated in angular space were higher than those predicted by the model. That the measured number of effective angles was in agreement with simulation indicates that the SNR increase was limited by the lens aperture and the collection beam width, rather than by intrinsic angular correlations of the backscattered light. We can expect, therefore, that a greater increase in SNR could be obtained either by increasing the lens aperture or by decreasing the collection beam width.

4.2 Human Skin

For imaging of human skin (finger tip) in vivo, a glass window angled at 40 degrees with respect to the incident beam was placed beneath the focusing lens to provide mechanical stabilization and to reduce specular reflections. Angular compounding resulted in a dramatic increase in image quality (Fig. 5). In terms of spatial resolution, the image acquired from a single angular sample [Fig. 5(a)] is equivalent to the one that would be obtained from a conventional OFDI system. As the number of compounded angular samples was increased [Figs. 5(b)-5(f)], the boundary between the stratum corneum and the papillary dermis became more pronounced and the sweat ducts within the stratum corneum became more clearly delineated (Fig. 5 insets). Qualitatively, there was no appreciable increase in image quality for angular compounding performed across more than 7 angles. This result was expected, given that the effective number of uncorrelated angular samples was found to be 6.7 from the measurements of the optical phantom.

The statistical decorrelations observed between consecutive A-lines as the orientation of the angular scanning mirror was changed derive primarily from angular decorrelations, rather than from motion of the sample. Two pieces of evidence support this assertion. First, the measured angular correlations correspond well to those expected from the statistical model. Secondly, acquiring images without scanning the galvanometer mirror, but with the same number of averaged A-lines does not result in speckle reduction for in vivo imaging of human dermis (Fig. 6). This result suggests that statistical decorrelations that result from the motion of the sample between consecutive A-lines are very small relative to those resulting from observing the sample from different backscattering angles.

5. Conclusion

The speckle-reduction capabilities of a novel angle-resolved OCT system that enabled the resolution of backscattered light within an angular range of 172 to 188 degrees were demonstrated. Discrimination of different backscattering angles was achieved by means of a galvanometer mirror placed at the focus of a telescope in the sample arm, so that the backscattering angle of the collected light varied sequentially in time as the mirror was rotated. By averaging images acquired at different backscattering angles, an SNR improvement of 3.4 dB was obtained, which resulted in a substantial improvement in image quality when applied to imaging of human skin in vivo. The magnitude of the SNR improvement and the correlations between angular samples were well predicted by a statistical model of speckle that incorporated the system optical parameters.

In this system, the number of angular samples corresponding to particular spatial volume elements was a tunable parameter, which contrasts with the previously demonstrated angle-resolved OCT system that employed parallel-detection. With the parallel-detection system, A-lines from all angles were acquired simultaneously at a rate of 25 Hz; with this sequential-detection system, individual A-lines were acquired at 10 kHz, and an effective A-line rate of
250 Hz was obtained with 40 angular samples. As a result, the artifacts and signal attenuation that result from sample arm movement were significantly reduced in the sequential-detection system, enabling imaging of biological samples in vivo. By utilizing recently developed OCT systems having A-line rates in the hundreds of kilohertz [10,14], angle-resolved OCT with sequential-detection may ultimately enable real-time speckle reduction performed at video rates. As such, angular compounding could be leveraged as a means of converting advances in A-line speed into improvements in speckle reduction without sacrificing the image frame rate.

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Fig. 1.
Angle Resolved OCT System Schematic (a) and sample arm optics (b). PC: polarization controller; Circ: circulator; C: collimator; P: polarizer; PS: polarization splitter; BD: balanced receiver; SOA: semiconductor optical amplifier; DG: diffraction grating; BS: beam splitter; M: mirror; L: achromatic doublet lens. L1: \( f = 35 \) mm; L2: \( f = 50 \) mm; L3: \( f = 35 \) mm; L4: \( f = 75 \) mm. The dashed region in (b) is oriented perpendicularly with respect to the plane of the interferometer.
Fig. 2.
Scattering geometry employed in the speckle model showing the incident beam traversing the center of the lens L1 (focal length $f$) and light backscattered at angle $\theta$. After traversing L1, the backscattered light is radially displaced by a distance of $x$. 

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Fig. 3.
A representative angular backscattering distribution corresponding to a single polarization channel, obtained from one resolution element in the tissue phantom (a), and the corresponding normalized cross-correlation function.
Fig. 4.
SNR as a function of the number of compounded angles for signals acquired from a homogeneous tissue phantom.
Fig. 5.
OCT images of human finger tip acquired in vivo with no angular compounding (a); angular compounding with 2(b), 4(c), 7(d), 16(e) and 32 (f) angular samples. Speckle reduction with angular compounding enhances the contrast between neighboring structures, as highlighted by the insets corresponding to the dashed region in (a). The scale bar corresponds to 250 μm. The transverse extension of the image is 7 mm. S: sweat duct emerging from the lower papillary dermis (PD) into the upper stratum corneum (SC).
Fig. 6. 
OCT images of human finger tip acquired in-vivo with 2160 A-lines (a); and with $40 \times 2160 = 86400$ A-lines (b). In case (b), sets of 40 consecutive A-lines were averaged, yielding 2160 A-lines in the displayed image. The angular scanning galvanometer was held fixed in both cases, so that images were obtained from a single angular sample. The scale bar corresponds to 250 μm. The transverse extension of the image is 7 mm.