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Speckle Size in Optical Coherence Tomography

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ABSTRACT

Speckle is inherent to any Optical Coherence Tomography (OCT) imaging of biological tissue. It is often seen as degrading the signal, but it also carries information about the tissue microstructure. One parameter of interest is the speckle size. We study the variations in speckle size on optical phantoms with different density of scatterers. Phantoms are fabricated with a new approach by introducing silica microspheres in a curing silicon matrix, providing phantoms with a controlled density of scatterers. These phantoms are also solid, deformable, and conservable. Experimental results are obtained with Time-Domain OCT (TD-OCT). Modeling is performed by simulating a phantom as a random distribution with discrete scatterers. Both experimental results and modeling show that the speckle size varies when there are few scatterers contained within the probed volume, the latter being defined by the coherence length and the spot size of the focusing optics. As a criterion to differentiate tissues, the speckle size has the same sensitivity as the contrast parameter that is studied in Ref. 1. This work also contributes to a better understanding of speckle in OCT.

Keywords: Optical coherence tomography, speckle, biomedical imaging

1. INTRODUCTION

Speckle is omnipresent in Optical Coherence Tomography (OCT) imaging. It results from the multiple contributions of the scatterers contained within the probed volume and from the multiple pathways that photons can travel before being detected.¹ Speckle is often seen as degrading the signal but it also contains information about the tissue microstructure. Recently, a few papers attempted to use speckle field parameters in order to perform tissue differentiation. One approach relies on first-order speckle statistics, using speckle contrast.² This approach is of interest mainly when there are few scatterers within the probed volume. A broader range of first order and second order statistical parameters were also studied in Ref. 3. Results are promising but the renormalization of the speckle field and the artificial intelligence analysis do not lead to a simple physical interpretation of the observed variations.

In the current paper, we pursue similar interests by studying the variation of speckle size. To a first approximation, the axial speckle size is determined by the OCT resolution length and the transverse speckle size by the focusing optics illuminating the sample. This is illustrated in Figure 1 where a coronary artery of a rabbit is imaged with the same OCT source but different spot sizes. The axial speckle size (vertical on the images) is about the same in both images, but the transverse speckle size (horizontal on the images) is much smaller with the smaller spot size. The question we want to answer is: Is the speckle also determined to some extent by the tissue microstructure?

To address the problem, we work with optical phantoms. It is quite a challenge to fabricate solid phantoms with a controlled density of scatterers. We describe how this challenge is solved in Section 2. In Section 3, we provide some details about the experimental setup used to perform OCT measurements, about the model used to perform simulations, and about the way speckle size is extracted from the OCT cross-sections. In Section 4, experimental results along with modeling results are presented. They show that both the axial and transverse speckle sizes do vary with the density of scatterers. A conclusion is provided in Section 5.

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Figure 1. Cross-section images of the coronary artery of a rabbit. Images are 0.2 mm wide in the axial direction (vertical) and 0.3 mm wide in the transverse direction (horizontal). The same Time-Domain-OCT system was used for both images. The left image was obtained with a spot size of 21 microns while the right image was obtained with a spot size of 5 microns.

2. OPTICAL PHANTOMS

It is quite a challenge to fabricate a solid and deformable optical phantom with a controlled density of scatterers that are homogeneously distributed. The use of microspheres in a phantom gives the best control over the shape, size, and density of scatterers. Figure 2 presents an electron micrograph of silica microspheres. However, microspheres are usually used in liquid phantoms, which are not suited for the study of the speckle size due to the brownian motion of the scatterers during the transverse scan. To produce a solid phantom, we mix dry silica microspheres in a room temperature vulcanizing (RTV) silicone. To insure good uniformity and dispersion of the microspheres in the silicone matrix, the mixture is dissolved and treated by ultrasonic bathing before polymerisation of the silicone. As shown in Figure 3, scanning electronic microscope (SEM) characterisation is performed to assess the success of dispersion. Three series of phantoms were fabricated using two different microsphere diameters: 1.86 microns and 2.34 microns. Each series contains 7 or 8 phantoms of various densities of scatterers, ranging from 2.1×10^8 to 1.1×10^{10} part/ml. A picture of a phantom is presented in Figure 5. Our innovative combination of silica microspheres in a silicone matrix produces phantoms with a great control over the identity, shape and density of scatterers that are also solid, deformable, and conservable over a very long period of time.

Figure 2. Electron micrograph of the silica microspheres that are introduced in a silicon matrix to fabricate solid phantoms.

Figure 3. Electron micrograph of the silica microspheres in the silicon matrix.

Figure 4. Solid, deformable, conservable phantom.

3. EXPERIMENTAL SETUP, MODELING, AND SPECKLE SIZE DETERMINATION

In order to determine if the speckle size is affected by the tissue microstructure, we perform measurements of speckle size on phantoms with different densities of scatterers. The measurements are performed with a Time-Domain OCT system that provides an axial resolution of about 14 microns. The focusing optics used in the sample arm provides a spot size of 7.4 microns, the focal point being located about 200 microns below the front face of the optical phantom. The corresponding depth of field is about 90 microns in the phantom. The simulations are performed by modeling the phantom as an ensemble of discrete scatterers. The focusing and detection optics are modeled with the ABCD matrix formalism.⁶ To a first approximation, the scattered waves are modeled as a spherical waves.

The important parameter for all the measurements performed is the Effective Number of Scatterers (ENS), which is the average number of scatterers contained within the probed volume defined by the coherence length and the spot size. This is the same parameter as the one used in Ref. 2, and we use the same definition of the coherence length as in Ref. 2.

All measurements of speckle size are performed over a rectangular area as illustrated in Figure 5 to insure good statistics. The width of the window in the axial direction (depth direction) is chosen rather short to track the variation of speckle size along the axial optical axis. This impacts on the precision of the axial speckle size values. In the transverse direction, the window width is much larger thus providing more precise values of speckle size.

The speckle size can be evaluated from the Fourier transform (Figure 6) or the normalized autocorrelation function (Figure 7). The Fourier Transform has a gaussian shape to which is superimposed short scale oscillations. The speckle size is obtained by extracting the width of the gaussian shape and converting it to a speckle size. For the normalized autocorrelation, the speckle size is also obtained by fitting the gaussian shape and converting the width to a speckle size. Both methods are equivalent. For the axial speckle size, due the shortness of the window used in the evaluation, the autocorrelation provides the best results. For the transverse speckle size, better results are obtained with the Fourier transform.

4. SPECKLE SIZE VARIATION

Figure 8 presents the axial speckle size measured in the focal region for the various phantoms as a function of the ENS value. In the low density limit (low ENS value), the axial speckle size is essentially determined by the OCT resolution length. As the number of scatterers increase, the interference between their contributions to the OCT signal reduces the speckle size. The high density limit can be estimated following a treatment similar to the development performed by Wagner et al. for the speckle in ultrasound.⁵ The simulated values decrease smoothly from the low ENS region to the high ENS region. The experimental results also show a decrease in speckle size for ENS values larger than unity. As noted in the previous section, there is a large uncertainty in the speckle size determination due to short window used in the treatment. An error of 0.8 micron is estimated from the standard deviation of the values obtained from a large number of simulated speckle measurements. For the very low values of ENS, the measured speckle sizes are not significant. In addition to the uncertainty inherent to the axial speckle size determination, the signal coming from the few scatterers is so weak that the measurements become sensitive to the small signal coming from the silicon matrix. In fact, for these very low values of ENS, one can hardly speak of a speckle field since the individual scatterers are well separated and easily identified.

Figure 9 shows the variation of the axial speckle size versus the position relative to the focal point (0 mm). Only two simulated and two experimental results are provided. For the ENS value of 1, there is some discrepancy between the experimental and simulated values. This is attributed again to the error introduced by both the shortness of the window used in the analysis and to the small signal coming from the silicon matrix. Nevertheless, both show that there is a decrease in speckle size as we move away from the focal point. This is explained by the

Figure 5. Speckle field with two examples of successive rectangular windows (one with full lines, one with dotted lines) used to evaluate the speckles sizes at various optical depths.

Figure 6. Fourier transform of the speckle field in one direction.

Figure 7. Normalized autocorrelation of the speckle field in one direction.

fact that the probed volume increases due to an increase in spot size away from the focal point. Consequently, the ENS value increases which reduces the speckle size, as shown by the results of Figure 8.

Figure 10 presents the transverse speckle size measured in the focal region for the various phantoms as a function of the ENS value. In the low density limit (low ENS value), the transverse speckle size is essentially determined by the point-spread-function of the focusing optics. As for the axial case, the transverse speckle size decreases as the number of scatterers increase. In the high density limit an asymptotic value can again be estimated following Wagner et al.⁵ The window used for the evaluation of the transverse speckle size is much larger than the one used for the axial speckle size, which results in a reduced uncertainty of 0.1 micron. Both the experimental and simulated results follow a similar trend, evolving between the calculated lower and upper limits.

Figure 8. Variation of the axial speckle size in the focal region as a function of the Effective Number of Scatterers (ENS) measured on various phantoms (S1,S2, and S3) along with simulation results.

Figure 9. Variation of the axial speckle size as a function of the position relative to the focal point (0 mm) measured on various phantoms along with simulation results.

Figure 11 shows the variation of the transverse speckle size versus the position relative to the focal point (0 mm). For the ENS value of 1, unlike for the axial case, the transverse speckle size increases away from the focal point. This is mainly due to an increase in spot size, since the latter directly impacts the transverse speckle size. But as the spot size increases, the probed volume also increases, so does the ENS value. This leads to a decrease in speckle size as seen on the simulation 100 microns away from the spot size. This was also observed

Figure 10. Variation of the transverse speckle size in the focal region as a function of the Effective Number of Scatterers (ENS) measured on various phantoms (S1,S2, and S3) along with simulation results.

Figure 11. Variation of the transverse speckle size as a function of the position relative to the focal point (0 mm) measured on various phantoms along with simulation results.

experimentally on measurements performed with Swept-Source OCT but not presented here. Away from the focal point, there are thus two competing effects, the increase in spot size which tends to increase the speckle size and the increase in ENS value which tends to decrease the speckle size. Away from the focal point, there is also a transverse phase to take into account for the focusing optics and for the the detection optics. This will also affect the observed transverse speckle size and could explain the flatness of the value observed for the high ENS case as we move away from the focal point in Figure 11.

5. CONCLUSION

By performing measurements on optical phantoms, we have shown that the speckle size is not only determined by the focusing optics and the OCT resolution, but also by the tissue microstructure. Like the contrast measurements reported in Ref. 2, the speckle size measurement is sensitive to the low densities of scatterers and could be used for tissue differentiation in specific cases. We have also developed a new approach to fabricate solid, deformable, and conservable optical phantoms with a controlled density of scatterers.

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