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Complex decorrelation averaging in optical coherence tomography: A way to reduce the effect of multiple scattering and improve image contrast in a dynamic scattering medium

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Abstract

We demonstrate that complex decorrelation averaging can reduce the effect of multiple scattering and improve OCT imaging contrast. Complex decorrelation averaging calculates the product of an A-scan and the complex conjugate of a subsequent A-scan. The resultant signal is the product of the amplitudes and the phase difference. All these resulting complex signals at a particular location are then averaged. We take advantage of the fact that complex averaging, in contrast to conventional magnitude averaging, is sensitive to phase decorrelation. Sample motion that increases signal phase variance results in lower signal magnitude after complex averaging. Such motion preferentially results in a faster decorrelation of the multiple scattering signal when compared to the single scattering signal with each scattering event spreading the phase. This indicates that we may reduce multiple scattering by implementing complex decorrelation averaging to preferentially reduce the magnitude of the multiply scattered light signal in OCT images. By adjusting the time between phase-differenced A-scans, one can regulate the amount of measured decorrelation. We have performed experiments on liquid phantoms that give experimental evidence for this hypothesis. A substantial improvement in OCT image contrast using complex decorrelation averaging is demonstrated.

The extensive development of frequency-domain optical coherence tomography (OCT) for more than a decade has enabled A-scan rates in the MHz range. At the same time, frequency-domain OCT gives easy access to the amplitude and phase of the OCT signal. These characteristics together have opened the possibilities of doing different kinds of averaging in order to improve OCT imaging.

The conventional OCT averaging method ('magnitude averaging') calculates the magnitude-squared Fourier-transformed spectral fringes before averaging thus is insensitive to changes

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in phase. Recently, groups have averaged the complex-valued, Fourier-transformed, spectral-fringe signals before calculating the magnitude in a technique called complex averaging (also called coherent averaging or spectral averaging) [1,2].

In general, complex averaging increases the dynamic range by reducing the noise floor while maintaining similar signal values when compared to magnitude averaging [1,2]. At the same time, complex averaging results in increased speckle contrast [2]. Superior white Gaussian noise suppression in OCT images has been demonstrated previously with complex averaging compared to magnitude averaging [1,2]. Complex averaging has improved Doppler OCT imaging [3,4] and computational downscaling of A-scan rates in MHz-OCT demonstrating improved sensitivity [5].

It is well known that multiple scattering in OCT reduces image contrast and resolution especially at greater depths within the tissue. The existing OCT scattering model developed by Thrane et al. [6] describes the magnitude-averaged OCT signal depth profile and lateral resolution when taking multiple scattering into account. The development of methods to reduce the influence of multiple scattering in OCT images would improve image quality and enhance OCT's utility.

Here, we introduce a novel concept based on averaging for reducing the deleterious effects of multiple scattering when imaging through a dynamic scattering medium. When later used in OCT tissue imaging, one possible approach may be to introduce displacements in the tissue, which can be done in different ways as already demonstrated in the field of OCT elastography [7]. We demonstrate that complex decorrelation averaging (cda) can decrease the relative influence of multiple scattering in comparison to single scattering and improve OCT imaging contrast in addition to increasing the dynamic range. Complex decorrelation averaging calculates the product of an A-scan and the complex conjugate of a subsequent A-scan. The resultant signal is the product of the amplitudes and the phase difference. This calculation is followed by the complex averaging of all resultant signals at a particular location. This averaging method is described by Eq. (1)

$$cda = \left| \frac{1}{N} \sum_{n=1}^N \tilde{S}_n \tilde{S}_{n+dl}^* \right| \quad (1)$$

where $\tilde{S}$ and $\tilde{S}^*$ are a complex-valued A-scan and the conjugate of a complex-valued A-scan, respectively. N is the number of averages, and dl is the number of A-scans skipped ahead with $dl=1$ meaning consecutive A-scans. By adjusting the time interval between phase-differenced A-scans, one can regulate the amount of measured decorrelation. Phase variance and decorrelation time are coupled with increases in phase variance leading to shorter decorrelation times. Assuming that the particles in a highly scattering medium move independently and are uniformly distributed, random walk theory predicts that phase variance is proportional to the mean square displacement of the scatterers and the number of scattering events [8].

Fig. 1 shows two simulated complex-number datasets with different phase variance. The red stars mark the result of complex averaging and indicate that higher phase variance leads to a smaller magnitude. This is further illustrated at the bottom of Fig. 1 where higher phase variance leads to lower magnitudes for complex averaging but not magnitude averaging. For a given number of scattering events, any motion of the scatterers will lead to a higher phase variance and consequently a shorter decorrelation time resulting in a lower magnitude when performing complex averaging. In addition, motion of the scatterers will result in a larger reduction of the multiple scattering signal when compared to the single scattering signal because the phase variance is directly proportional to the number of scattering events. Our hypothesis is that by proper selection of the A-scan interval we are able to obtain a substantially greater reduction in the multiply scattered versus singly scattered components of the signal. We have performed experiments, described below, on liquid phantoms (22°C) that give experimental evidence for our hypothesis. We use a dynamic scattering medium consisting of freely moving particles in a liquid. First, A-scans are recorded and the complex OCT signal is determined for each A-scan. Next, the product of an A-scan and the complex conjugate of a subsequent A-scan is calculated. Finally, complex averaging is performed in order to obtain the averaged OCT depth profile as described by Eq. (1).

The results presented in Fig. 2 and 3 below were recorded using a 47-kHz spectral-domain 1310nm OCT system with 10 μm axial resolution (in air) and either 24 μm (Fig. 2) or 17 μm (Fig. 3) lateral resolutions [9]. Magnitude- and complex-decorrelated-averaged OCT depth profiles of 5% Intralipid are shown in Fig. 2. An M-mode scan was recorded with a total of 350,000 A-scans. Complex decorrelation averaging was performed on the dataset. The time interval between the phase-differenced A-scans sets the amount of decorrelation with longer time intervals leading to a larger amount of decorrelation and a greater reduction in the magnitude. The overall form of the magnitude-averaged OCT depth profile presented in Fig. 2 is in agreement with the OCT scattering model developed by Thrane et al. [6] that predicts a reduced slope at deeper depths due to multiple scattering. With this as a reference, the experimental results presented in Fig. 2 provide experimental evidence to our hypothesis that complex decorrelation averaging can reduce multiple scattering. The complex-averaged curves fall below the magnitude-averaged curve, and the difference between these curves increases with depth as the phase variance increases due to multiple scattering. As a result, complex decorrelation averaging lengthens the linear portion of the curve, representing the regime dominated by single scattering. When we increase the time interval between two A-scans for our complex decorrelation averaging calculation, we increase the phase variance due to greater particle movement and further reduce the effect of multiple scattering. A reduction of multiply scattered light is demonstrated at the deep side of the reflection from the bottom of the plastic Petri dish, i.e., inside the plastic, where there is no backscattering and the only signal is multiply scattered light. As the time interval for the phase difference is increased, the multiply scattered light signal is further reduced. Furthermore, the strong peak signal at the static bottom of the Petri dish is expected to be dominated by single scattering, and as expected, the peak value is relatively unchanged by the complex decorrelation averaging.

To demonstrate improved OCT image contrast with complex decorrelation averaging, we performed a volume scan of a 1.27 mm thick layer of 10%-Intralipid in water on top of a

resolution test target. The volume scan consisted of 50 cross-sectional planes 2 mm wide. At each plane, we recorded 1000 frames (170 A-scans/frame) at a frame rate of 188 frames/s (dead-time between frames 1.7 ms). The entire dataset consisted of 16.2 GB.

The averaged OCT images are presented in Fig. 3. Magnitude averaging is performed together with intra-frame complex decorrelation averaging using the phase difference between adjacent A-scans in a B-scan (time interval between A-scans $\Delta t = 21.28 \mu\text{s}$), and inter-frame complex decorrelation averaging using the phase difference between corresponding A-scans in adjacent B-scans skipping over three frames ($\Delta t = 21.28 \text{ ms}$). The complex decorrelation averaging results have been normalized to the maximum signal of the magnitude averaging result. Multiply scattered photons induce a background haze and loss of contrast, especially when the scattering medium is composed of particles with diameters smaller compared to the wavelength resulting in somewhat isotropic scattering, which is the case with the Intralipid used here [10]. A substantial improvement in image contrast is observed going from magnitude averaging to intra-frame complex decorrelation averaging to inter-frame complex decorrelation averaging. Clearly, as the time interval for the phase difference is increased by a factor of 1000 for the inter-frame complex decorrelation averaging, multiple scattering is further suppressed due to increased particle motion.

To quantify this improvement in image contrast, the contrast-to-noise ratio (CNR) [11] has been calculated for the region of interest (ROI) indicated by the red box in Fig. 3. The CNR for a ROI is defined as $\text{CNR} = 10\log_{10}\left[\frac{(\mu - \mu_b)^2}{\sigma^2 + \sigma_b^2}\right]$, where μ and σ are the mean and standard deviation of the pixel values in the ROI, respectively, and μ_b and σ_b are the mean and standard deviation of the pixel values in a background region of the image (yellow box in Fig. 3), respectively [11]. A higher value of CNR means better image contrast. All CNR calculations are based on linear scaled data. The CNR values obtained for the averaged *en face* images in the plane of the test target lines as shown in Fig. 3 are 4.287 dB for magnitude averaging, 4.401 dB for intra-frame complex decorrelation averaging, and 5.118 dB for inter-frame complex decorrelation averaging. For comparison, the complex averaging was performed without computing the phase difference beforehand using all 1000 frames as shown in Fig. 3. The CNR value for the resulting averaged *en face* image in the plane of the test target lines is only 0.877 dB. The complex averaging has low contrast because both the multiple and single scattering events have time to decorrelate so both the signal and background are greatly reduced. It is important to choose a phase-difference time delay where the phase variance of the single scattering signal only increases modestly compared to significant increases for the multiple scattering signal (see bottom panel of Fig. 1). Finally, the results for the case of no averaging are shown in Fig. 3 for the sake of completeness. The test target lines are barely visible in this case, and the CNR value is as low as 0.210 dB.

In Fig. 4, the CNR is shown as a function of the number of frames averaged for inter-frame complex decorrelation averaging ($\Delta l = 4$), with a higher number of frames averaged meaning a higher CNR as expected. A power law equation has been fitted to the experimental data, and a very high quality of the fit is demonstrated.

Investigating the use of this novel image enhancement method in OCT tissue imaging will be the subject of future studies. These studies will address issues such as the introduction of

displacements in tissue, and the amount of averaging needed to see a benefit in real tissue. The relative large amount of averaging needed for complex decorrelation averaging, and the resulting relative long data acquisition time, may pose some limitations on the in-vivo application of the method, limiting it to tissues with low bulk motion. An important issue will be to determine the optimum time interval between A-scans used in the complex decorrelation averaging. It will depend on the amount of displacements introduced in the tissue, and at the same time, there may be large variations of multiple scattering in real in vivo tissue. One way of handle this could be to use an iterative method in order to optimize the image contrast by adjusting the time interval between A-scans used in the complex decorrelation averaging. This could be automated based on image contrast metrics. A potential important application of complex decorrelation averaging may be within OCT angiography [12,13]. It is a well-known problem in OCT angiography that multiple-scattering tails of a few hundred microns are detected beneath the blood vessels, thereby limiting the visualization of underlying vessels [14]. The scattering off the moving red blood cells in the blood vessels may facilitate a reduction of the multiple-scattering tails in OCT angiograms by using complex decorrelation averaging.

In conclusion, we have demonstrated experimental evidence that complex decorrelation averaging is a way to reduce the effect of multiple scattering and substantially improve OCT imaging contrast in a dynamic scattering medium in addition to the increase in the dynamic range. This represents a novel approach of reducing the deleterious effects of multiple scattering in OCT imaging.

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References

1. Tomlins PH, Wang RK. Meas Sci Technol. 2007; 18:3365.
2. Szkulmowski M, Wojtkowski M. Opt Express. 2013; 21:9757. [PubMed: 23609683]
3. Peterson LM, Jenkins MW, Gu S, Barwick L, Watanabe M, Rollins AM. Biomed Opt Express. 2012; 3:3022. [PubMed: 23162737]
4. Hong YJ, Miura M, Ju MJ, Makita S, Iwasaki T, Yasuno Y. Invest Ophthalmol Vis Sci. 2014; 55:5016. [PubMed: 25052993]
5. Pfeiffer T, Wieser W, Klein T, Petermann M, Kolb JP, Eibl M, Huber R. Proc SPIE. 2016; 9697:96970S.
6. Thrane L, Yura HT, Andersen PE. J Opt Soc Am A. 2000; 17:484.
7. Kennedy BF, Wijesinghe P, Sampson DD. Nature Photonics. 2017; 11:215.
8. Weitz, DA., Pine, DJ. Diffusing-wave spectroscopy. In: Brown, W., editor. Dynamic Light Scattering: The Method and Some Applications. Oxford; 1993. p. 652-720.
9. Fleming CP, Wang H, Quan KJ, Rollins AM. JBO Lett. 2010; 15:030516.
10. Yadlowsky MJ, Schmitt JM, Bonner RF. Appl Opt. 1995; 34:5699. [PubMed: 21060400]
11. Adler DC, Ko TH, Fujimoto JG. Opt Lett. 2004; 29:2878. [PubMed: 15645810]
12. Makita S, Hong Y, Yamanari M, Yatagai T, Yasuno Y. Opt Express. 2006; 14:7821. [PubMed: 19529151]
13. Nam AS, Chico-Calero I, Vakoc BJ. Biomed Opt Express. 2014; 5:3822. [PubMed: 25426313]
14. Srinivasan VJ, Radhakrishnan H. NeuroImage. 2014; 102:393. [PubMed: 25111471]

Fig. 1.

Top: Two simulated complex-number datasets (10,000 points each) having different phase variance. The result of complex averaging is indicated by the red stars and indicate that higher phase variance leads to a smaller magnitude. Bottom: Plot of the average magnitude against the standard deviation of the phase angle (50,000 points used) for complex and magnitude averaging. For visualization purposes, the amplitude is chosen to be $1 \pm .03$ in all cases. The phase angle at all standard deviations is normally distributed.

Fig. 2.

OCT depth profiles of 5%-Intralipid as a result of magnitude averaging and complex decorrelation averaging. The numbers in parentheses (e.g., $200 \Delta t$) indicate the A-scan time interval (e.g., $200 \Delta t$ specifies that the phase difference was calculated between A-scans 200 lines apart (Δl) with a time interval between adjacent lines $\Delta t = 21.28 \mu\text{s}$) used for complex decorrelation averaging. The relatively large signal just before $2000 \mu\text{m}$ is the reflection from the bottom of the plastic Petri dish (close-up of that signal shown in inset). The smaller peak on the blue and red curves just before $1500 \mu\text{m}$ is an artifact. Approximately 350,000 A-scans were averaged.

Fig. 3.

Averaged OCT images of a 1.27 mm thick layer of 10%-Intralipid on top of a test target (Thorlabs R2L2S1N, Negative NBS 1963A Resolution Target) from a volume scan based on 50 cross-sectional planes 2 mm wide and separated by 75 μm . The height of the en-face images is 3.68 mm. Upper three panels: (A) Magnitude averaging; (B) Complex decorrelation averaging - adjacent A-scans in B-scan (Intraframe); (C) Complex decorrelation averaging - corresponding A-scans in adjacent B-scans skipping over three frames (Inter-frame). Lower two panels: (D) Complex averaging of all 1000 frames without computing the phase difference first; (E) No averaging. Upper row in panels: Averaged B-scans (cross-sectional plane number 25); Lower row in panels: Averaged *en face* images in the plane of the test target lines visible in the bottom of the B-scans. Number of frames averaged was 1000 for magnitude averaging, intra-frame complex decorrelation averaging, and complex averaging. For inter-frame complex decorrelation averaging ($dI = 4$) the number of frames averaged was 996. Increasing the A-scan time interval in the inter-frame complex decorrelation averaging clearly improves image contrast. Values for the contrast-to-noise ratio of the *en face* images are listed. The red and yellow boxes indicate the ROI and the background region, respectively, used in the calculation of the CNR values. All results were normalized to the maximum signal of the magnitude averaging result.

Fig. 4.

The contrast-to-noise ratio as a function of the number of frames averaged for inter-frame complex decorrelation averaging ($dl = 4$). The red line is a nonlinear curve fit to the experimental data. Note that CNR is not in dB in this case.