Upper limit for angular compounding speckle reduction

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Yonatan Winetraub,^{1,2,3,4} Chris Wu,^{2,3} Graham P. Collins,^{2,3} Steven Chu,^{1,4,5} and Adam de la Zerda^{1,2,3,4,6,a)}

AFFILIATIONS

¹Biophysics Program at Stanford, Stanford, California 94305, USA

²Department of Structural Biology, Stanford University, Stanford, California 94305, USA

³Molecular Imaging Program at Stanford, Stanford, California 94305, USA

⁴The Bio-X Program, Stanford, California 94305, USA

⁵Departments of Physics and Molecular and Cellular Physiology, Stanford University, Stanford, California 94305, USA

⁶The Chan Zuckerberg Biohub, San Francisco, California 94158, USA

^{a)}Author to whom correspondence should be addressed: adlz@stanford.edu

ABSTRACT

Angular compounding is a technique for reducing speckle noise in optical coherence tomography that is claimed to significantly improve the signal-to-noise ratio (SNR) of images without impairing their spatial resolution. Here, we examine how focal point movements caused by optical aberrations in an angular compounding system may produce unintended spatial averaging and concomitant loss of spatial resolution. Experimentally, we accounted for such aberrations by aligning our system and measuring distortions in images and found that when the distortions were corrected, the speckle reduction by angular compounding was limited. Our theoretical analysis using Monte Carlo simulations indicates that “pure” angular compounding (i.e., with no spatial averaging) over our full numerical aperture (13° in air) can improve the SNR by not more than a factor of 1.3. Illuminating only a partial aperture cannot improve this factor compared to a spatial averaging system with equivalent loss of resolution. We conclude that speckle reduction using angular compounding is equivalent to spatial averaging. Nonetheless, angular compounding may be useful for improving images in applications where the depth of field is important. The distortions tend to be the greatest off the focal plane, and so angular compounding combined with our correction technique can reduce speckle with a minimal loss of resolution across a large depth of field.

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Optical coherence tomography (OCT) is a powerful tool for noninvasive probing of the microstructure of biological tissues. Because the technique relies on coherent detection of scattered light, however, OCT images are confounded by speckle noise: a grainy texture that reduces the signal-to-noise ratio (SNR) and effective spatial resolution. A widely used method to reduce speckle noise is angular compounding,^{1,2} which averages results obtained with the imaging beam probing the sample at different angles. Both the imaging system and the sample can remain stationary during the scan, allowing high imaging throughput with high image quality.^{3,4} Angular compounding has been reported to significantly decrease speckle, with the SNR of angular compounded images as much as 6.5 times that of a corresponding single-angle image.⁴ Furthermore, several studies^{5,6} found that angular compounding increases SNR significantly more than spatial averaging. Recent work has suggested combining image

processing with angular compounding^{7,8} to further reduce speckle. A key benefit claimed for angular compounding (e.g., compared to spatial averaging) is that it may achieve speckle reduction with minimal to no degradation of spatial resolution.^{1,3–6} As angular compounding becomes more widely used, it is important to study how it differs from spatial averaging and to what extent it removes speckle without impairing the spatial resolution.

In this work, we employed a common setup [see Fig. 1(a)] for angular compounded OCT, in which a galvo-controlled scanning mirror is offset a varying distance h from the optical axis.^{3,5} In classical OCT, the galvo mirror remains centered on the optical axis (i.e., $h = 0$). To acquire a single pixel of an A-scan, a ray reflects off the galvo mirror, passes through the optical system at point (s, θ) , and scatters from the sample at point (x, y, z) , as depicted in Fig. 1(b). The full A-scan is built up from a series of such pixels due to the ray

FIG. 1. (a) Angle-resolved OCT system schematic. RM, reference mirror; BS, beam splitter; G, galvo-controlled scanning mirror; L, objective lens ($f = 18$ mm); and SX and SY, translation stages moving sample and optics together with respect to the laser source and galvo mirror. Translation by SX produces the offset h . (b) Sample arm optics showing the ray trajectory from the galvo mirror to the sample. A ray reflects off the galvo mirror at point $(x' = h, y' = 0)$, passes through the optical system's aperture at point (s, θ) , and is scattered by the sample at point (x, y, z) .

scattering at different depths z . To acquire a B-scan, the galvo mirror rotates, sweeping s from -1 to 1 while keeping θ constant. Varying the distance h provides scans of (x, y, z) using rays at different angles, which are averaged for the angular compounded image.^{3,5}

Using a lens model with axially symmetrical aberrations, we can calculate (x, y) as a function of h, s , and θ ,⁹

$$\begin{aligned} x &= A_1 s \cos \theta + A_2 h + B_1 s^3 \cos \theta + B_2 s^2 h (2 + \cos 2\theta) + \dots, \\ y &= A_1 s \sin \theta + B_1 s^3 \sin \theta + B_2 s^2 h \sin 2\theta + \dots, \end{aligned} \quad (1)$$

where constants A_1 and A_2 describe the first-order imagery and B_1 and B_2 describe primary aberrations. Both x and y depend on h . Thus, changes in h not only result in reflection of light from the sample at different angles (angular compounding) but also move the beam with respect to the sample, which introduces spatial averaging. A pure angular compounding setup should correct for these displacements.

Recent angular compounding work has suggested the use of image registration by global translation estimation prior to averaging, to reduce spatial averaging.^{7,10,11} These corrections do remove some of the spatial averaging, but are insufficient since x and y are nonlinear functions of h and s . Furthermore, h introduces distortions out of the B-scan plane, so a full 3D volume scan is required to perform image registration.

To account for aberrations in our model, we first align the galvo B-scan direction with the x' -axis (i.e., set $\theta = 0$). This confines distortions to the B-scan plane (now the x - z plane), simplifying Eq. (1),

$$\begin{aligned} x &= A_1 s + A_2 h + B_1 s^3 + 3B_2 s^2 h + \dots, \\ y &= 0. \end{aligned} \quad (2)$$

We then introduce nonaxially symmetrical aberrations and neglect higher order terms, writing the result in terms of the system's "distortion field," U, W , and V ,

$$\begin{aligned} U(x, y, z; \phi) &= x(\phi) - x = \phi \times (u_0 + u_1 x + u_2 z), \\ W(x, y, z; \phi) &= y(\phi) - y = 0, \\ V(x, y, z; \phi) &= z(\phi) - z = \phi \times (v_0 + v_1 x + v_2 z), \end{aligned} \quad (3)$$

where ϕ is the angle of the beam from the z -axis in the sample. For small h , $\phi \approx h/nf$, where f is the objective focal length and n is the sample's refractive index. u_i and v_i are aberration parameters summarizing all optical distortions, which are unique to each optical setup. We also use the notation $x = x(\phi = 0)$, $y = y(\phi = 0)$, $z = z(\phi = 0)$. To understand the true extent of speckle reduction due to pure angular compounding, we must account for this distortion field.

Note that it suffices to measure the distortion field in the B-scan plane provided that the galvo B-scan direction is carefully aligned with the x' -axis along which the angular compounding offsets are performed. As noted above, in the absence of this alignment, a full 3D scan would be required to measure the distortion field, which would then include aberration parameters w_i associated with distortions out of the B-scan plane.

To test our model experimentally, we proceeded in three stages: first, we used a V target to align the B-scan direction to $\theta = 0$. Next, we imaged a phantom sample with standard angular compounding, and measured the distortion field. Finally, we used the distortion field to obtain a corrected angular compounded image of the phantom. In all experiments, we used a commercial Spectral Domain OCT (SD-OCT) system (Ganymede HR SD-OCT using LSM02-BB lens, Thorlabs, Newton, NJ) with a FWHM optical pixel size of $4 \mu\text{m}$ (the beam diameter at the focus). The SD-OCT light source was a superluminescent diode (SLD) with a center wavelength of $\lambda = 900$ nm and a spectral bandwidth of 200 nm. The A-scan rate was 30 kHz. B-scans consisting of 1000 A lines were acquired at about 30 Hz. The A-scan transverse sampling was $1 \mu\text{m}$, for an oversample ratio of 4 . Our standard preprocessing for each B-scan consisted of dispersion compensation and then Fourier transformation of the interferogram to produce a complex B-scan. For each B-scan image, we retained the absolute value and averaged across 20 B-scans to remove photon shot noise from the data.

In the alignment step, our goal was to align $\theta = 0$ to an accuracy of one optical pixel over the course of a $500 \mu\text{m}$ scan (i.e., $\Delta\theta = 0.46^\circ$). We fabricated a V target on a silicon wafer using standard

lithographic and dry etching processes. The target consisted of two perpendicular trenches, each $50\ \mu\text{m}$ wide $\times$ $28\ \mu\text{m}$ deep [see Fig. 2(a)]. We mechanically fixed the V target to a translation stage along the y -axis. We applied a few microliters of gold nanorod solution (OD 50)¹² to increase contrast.

We imaged the V target such that both trenches were visible in the B-scan [see Fig. 2(b)]. We estimated the center position of each trench and observed how these changed when the V target was moved along the y -axis, and adjusted our system accordingly. When the centers moved by equal and opposite amounts during y -translation, the system was aligned to $\theta = 0$. We computed our system's alignment to be $|\theta| < 0.43^\circ$.

After alignment, we imaged a 2% w/v Intralipid phantom (solid gel made from 20% w/v stock solution in agarose) at 17 angles at equal intervals ranging from -6.3° to $+6.3^\circ$ in air, which corresponds to our system's numerical aperture $\text{NA} = 0.11$. The angles were set by adjusting the position of the sample stage along the x -axis. In the phantom, which had $n = 1.33$, the range was $|\phi| \leq 4.7^\circ$. Figure 3(a) shows an OCT B-scan of the phantom acquired from a single angle. The speckle contrast is defined as $\sigma_I/\langle I \rangle$, the standard deviation of the intensity over the mean linear intensity, in an area of uniform scattering such as the imaged phantom in Fig. 3(a). The SNR is the inverse of this quantity, $\langle I \rangle/\sigma_I$. For the B-scan in Fig. 3(a), $\text{SNR} = 2.76$. When we performed standard angular compounding using all 17 angles [Fig. 3(b)], the SNR increased to 7.45, that is, by a factor of 2.70 compared to that of a single-angle image. (We call this a relative SNR, or an RSNR, of 2.7.) This result is comparable to values reported elsewhere.

Next, we assessed how much of this improvement may be due to unintended spatial averaging caused by optical distortions. We

FIG. 2. (a) Bright field microscopy image of the calibration target showing two trenches forming a V. (b) OCT B-scan taken along the blue line in (a). Alignment of the B-scan with the x -axis is measured by observing the movement of the trench centers when the target is moved along the y -axis. Scale bars: $50\ \mu\text{m}$.

FIG. 3. Angular compounding of Intralipid phantom. (a) B-scan image (x - z plane) acquired from a single angle. $\text{SNR} = 2.76$. Relative SNR (RSNR) = 1, by definition. (b) Image obtained by standard angular compounding of scans at 17 angles ϕ , with $|\phi| \leq 4.7^\circ$. $\text{SNR} = 7.45$ and $\text{RSNR} = 2.7$. (c) Estimated distortion field at $\phi = 4.7^\circ$ shows significant movement in both x and z directions. Inset: 2D color scale for the displacements. (d) Image using angular compounding corrected for the distortion field. $\text{SNR} = 4.57$ and RSNR is significantly reduced to 1.65. Scale bars: $30\ \mu\text{m}$.

estimated our system's distortion field by measuring the translational displacement of 100 random patches (each $30\ \mu\text{m} \times 30\ \mu\text{m}$) in our images using a subpixel registration algorithm,¹³ then fitting the patch displacements to Eq. (3) using least squares to estimate $u_0, u_1, u_2, v_0, v_1, v_2$. As can be seen in Fig. 3(c), the distortion field was significant (as large as $12\ \mu\text{m}$) and very different from a translation-only registration error. After this distortion was accounted for [as shown in Fig. 3(d)], the RSNR was reduced to 1.65. The residual error of the patch motion fit was ~ 1 – $2\ \mu\text{m}$, suggesting that small uncorrected displacements remain in the data.

We noticed that in the dark area in Fig. 3(c), near the lens's optical axis, U and $V < 500\ \text{nm}$. To try to obtain a corrected image with even less residual distortion than Fig. 3(d), we used this region to acquire 17 B-scans in the y - z plane ($\theta = 90^\circ$). (For these scans, we translated the sample along the y -axis.) A single-angle image [as shown in Fig. 4(a)] had an SNR of 2.75, essentially identical to the

FIG. 4. Angular compounding of Intralipid phantom at a position where the distortion field is $< 500\ \text{nm}$. (a) B-scan image (y - z plane) acquired from a single angle. $\text{SNR} = 2.75$. (b) Image obtained by angular compounding of 17 scans, corrected for the small distortion field. $\text{SNR} = 4.10$ and $\text{RSNR} = 1.49$, very close to our theoretical limit. (c) Area bounded by red box in (a), enlarged for angles $\phi = -4.7^\circ$ to -0.6° (top row) and 0° – 4.2° (bottom row). The similarity of the pattern at many angles explains why compounding achieves only a modest speckle reduction. Scale bars: $30\ \mu\text{m}$.

SNR of the x - z plane scans. The angular compounded image using all 17 scans and correcting for the small distortion field [Fig. 4(b)] had an SNR of 4.10, for an RSNR of 1.49—even smaller than that of Fig. 3(d). On examining small regions across a series of single-angle images [e.g., as in Fig. 4(c)], we saw that the speckle pattern did not change significantly when ϕ changed by relatively large angles. These results suggest that even slight unwanted translation can reduce speckle, and studies of pure angular compounding ought to account for movements of $\sim 1/10$ pixel (much lower than the limit suggested by Ref. 14). Conversely, by using small subpixel shifts, spatial compounding alone might significantly reduce speckle with only slight loss of resolution.

We now turn to a theoretical analysis of the ideal effect of angular compounding (see the [supplementary material](#) for additional details). We model the speckle contrast that would occur with our system if the distortion field was fully accounted for, to derive a theoretical upper limit for speckle reduction by pure angular compounding for our system. We have previously shown¹⁵ that a simple model assuming N identical isotropic scatterers randomly distributed in an imaging voxel can describe speckle behavior in OCT images. For the present analysis, we begin by considering a single point scatterer at an arbitrary location in the voxel. This scatterer generates a complex electric field $g + ih$, which changes if the imaging beam rotates by an angle ϕ around the voxel center.

Our Monte Carlo simulation randomizes the particle position in the voxel and computes $g(\phi)$ and $h(\phi)$. By randomizing over many particles at different positions, we estimate the covariance $\text{Cov}[g(\phi), g]$ and compute the correlation R between $g(\phi)$ and $g = g(\phi = 0)$, and similarly for h . Surprisingly, this numerical result for $R(\phi)$ fits very well with a Gaussian

$$R(\phi) = \frac{\text{Cov}[g(\phi), g]}{\sqrt{\text{Var}[g(\phi)] \text{Var}[g]}} \approx \exp\left(-\frac{\phi^2}{4b\sigma_\phi^2}\right), \quad (4)$$

where $\sigma_\phi = 0.5 \text{ NA}/n$ and $b = 1.00$ from the fit (see the [supplementary material](#)). Furthermore, $R(\phi)$ drops to 0 only at high angles (i.e., $|\phi| > 2\sigma_\phi$).

For the case of N independent scatterers in the voxel producing a total electric field $G + iH$, it follows from the linearity of the covariance function that the correlation between $G(\phi)$ and $G = G(0)$ is the same as that of individual particles: $R(\phi)$. We can show by numerical integration that the correlation $\rho(\phi)$ of the speckle intensity $I = \sqrt{G^2 + H^2}$ is very close to the square of G and H 's individual correlations, $R^2(\phi)$. Substituting our Monte Carlo result, Eq. (4), this yields

$$\rho(\phi) \approx \exp\left(-\frac{\phi^2}{2b\sigma_\phi^2}\right). \quad (5)$$

Next, we estimate the angular compounding signal B . In an ideal case, we could average the signal acquired from all possible angles, attenuated by the optical support or NA of the system

$$B = \frac{1}{\sqrt{2\pi\sigma^2}} \int_{-\infty}^{\infty} \exp\left(-\frac{\phi^2}{2\sigma^2}\right) I(\phi) d\phi, \quad (6)$$

where $\sigma = 0.5 \text{ NA}/n$, because NA is the $1/e^2$ radius of the lens. (Clearly $\sigma = \sigma_\phi$, but it is convenient to keep track of the two places where it enters the computation.) Finally, we compute the relative SNR for this ideal case, which we call $\tilde{S}$

$$\tilde{S} = \left(\frac{\text{Var}(I)}{\text{Var}(B)}\right)^{1/2} = \left(1 + 2\frac{\sigma^2}{b\sigma_\phi^2}\right)^{1/4} = 1.31. \quad (7)$$

We conclude that 1.31 is the upper limit for the RSNR achievable by pure angular compounding. Any additional speckle reduction observed should be attributed to spatial averaging. Note that 1.31 is lower than $\sqrt{2}$, the RSNR to be expected by averaging just two completely uncorrelated speckle patterns. This is not surprising as the lens aperture is a circ-function, and the employed beams are approximately Gaussian, moving the beam in the back aperture will tilt the beam somewhat before being clipped.

Up to this point, we have assumed that the collimated laser beam in Fig. 1(a) illuminates the full width of the lens. Speckle reduction can be further increased, however, by illuminating a small part of the lens, which increases the ratio of the compounding angles to the effective NA of the illuminated area. Assuming the relative illuminated area is p^2 ; $0 < p \leq 1$, then b in Eqs. (4) and (7) changes to bp^2 , leading to an RSNR limit of

$$\tilde{S}_{AC} = \left(1 + 2\frac{\sigma^2}{bp^2\sigma_\phi^2}\right)^{1/4} = \left(\frac{2}{p^2} + 1\right)^{1/4}, \quad (8)$$

which can be higher than 1.31. However, since the partial illumination of the lens also reduces the spatial resolution by about a factor of p , we should compare the angular compounding scenario with an alternative in which we fully illuminate the lens and perform spatial averaging that reduces the spatial resolution by the same factor. When we do these calculations (see the [supplementary material](#) for more detail), we find that the RSNR for such spatial averaging is

$$\tilde{S}_{SA} = (2/p^2 + 1)^{1/4}. \quad (9)$$

Comparing Eqs. (8) and (9), we find that $\tilde{S}_{AC} = \tilde{S}_{SA}$. In other words, when the trade-off with resolution and effective NA is accounted for, speckle reduction of angular compounding is identical to spatial averaging. Note that all our analysis is for single-axis compounding.

It is worth noting that our theoretical analysis and experimental results agree beyond the simple comparison of the theoretical RSNR limit and the actual RSNRs achieved. The probability distributions of the speckle for our three types of images—single-angle, standard angular compounded, and corrected angular compounded—follow three distinct functions (see the [supplementary material](#) for more details). The speckle in our single-angle images such as Fig. 3(a) closely match the Rayleigh distribution expected for fully developed speckle. This confirms that we have not introduced any significant decorrelation of the speckle prior to applying angular compounding. Conversely, standard angular compounded images such as Fig. 3(b) have nearly Gaussian distributions of speckle, as expected for speckle that has been significantly smoothed. Finally, compounded images where the effects of optical distortions have been corrected, such as Fig. 4(b), have a distribution intermediate between Rayleigh and Gaussian. Moreover, the intermediate distribution is well-fit with only a single parameter adjusted to match the observed RSNR. These results serve as further confirmation of our theoretical model and the assessment that pure angular compounding has a limited smoothing effect.

We have also considered the case where the idealized averaging over all angles, as in Eq. (6), is replaced by averaging over a limited

range of angles, $|\phi| < Q$ for some finite Q (see the [supplementary material](#) for more detail). We then computed versions of the corrected angular compounded image of [Fig. 4\(b\)](#) using fewer than 17 angles ϕ , with $|\phi| < Q$. Again, the prediction and experimental result show reasonable agreement, with the RSNR decreasing to 1.0 as Q decreases.

In conclusion, angular compounding may not reduce speckle compared to simple postprocessing spatial averaging. Experiments that achieve greater speckle reduction seemingly by angular compounding may be benefiting from unintended spatial averaging (and concomitant resolution loss) due to lens aberrations or partial illumination of the aperture. We have described two steps to measure and account for such aberrations: scan alignment using a V target and distortion field estimation. Once distortion is accounted for, the pure angular compounding that remains (for single-axis scanning over 13° , the full NA) is expected to increase SNR by not more than a factor of about 1.3. Nonetheless, angular compounding combined with our correction technique may be especially suited for improving images in applications where the depth of field is important. Because the lens aberrations tend to cause the greatest distortions off-focus, that is where distortion-corrected angular compounding can provide the most benefit, reducing speckle with a minimal loss of resolution across a broad depth of field.

See the [supplementary material](#) for additional details regarding the numerical modeling and theoretical analysis of speckle contrast under angular compounding, including the case in which the lens is partially illuminated. Also included are a comparison of the speckle probability distributions obtained in our experiments with theoretical distributions, and the calculation of the effect of angular compounding over a limited range of angles, again with a comparison to experimental data.

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