Supplemental Document

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High performance OCTA enabled by combining features of shape, intensity, and complex decorrelation: supplement

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Animal preparation

Five 8~10 week-old adult male C57BL/6J mice (~20±2 g weight) were used in this study. The mouse was anesthetized with pentobarbital (1%, 0.01 ml/g) and fixed in a custom-designed animal holder with integrated bite bar and ear bar system to mitigate the eye movements caused by breath and heartbeats. Before OCT imaging, the pupil was dilated with eye-drops (10% phenylephrine hydrochloride ophthalmic solution, USP, AK-Dilate) to allow OCT beam access to the posterior segment.

System Setup and Implementation

The imaging system used in this study was based on a typical configuration of spectral domain optical coherence tomography (SD-OCT). The system used a superluminescent diode (SLD, Superlum, Ireland) with a central wavelength of 840 nm and a full width at half maximum bandwidth of 100 nm, which provides a theoretical axial resolution of ~3.1 μm. The light from the SLD was split into two paths with a 70:30 fiber based beam splitter (Thorlabs, Inc. USA) where the 70% power path goes to the sample arm while the 30% power path goes into the reference mirror. The sample arm beam was scanned by a galvo-mirror (Thorlabs, Inc. USA) in both x and y directions. After the scanner, an objective lens with a 50 mm focal length and an ocular lens (Volk, USA, 90D) were employed to adjust the beam size and enable large field of view. The theoretical lateral resolution was estimated to be ~10 μm using existing mouse eye models. The OCT detection unit was a high-speed spectrometer equipped with a fast line-scan camera, providing a 100kHz line-scan rate with 2048 active pixels.

In this study, a stepwise raster scanning protocol was used to acquire volumetric dataset (z-x-y). The slow-scanner (y direction) was driven by a step waveform with a total of 400 steps. At each step, three repeated B-frames were successively acquired for analyzing dynamic flow signals, with each B-frame composed of 400 A-lines in the fast-scan (x) direction. Each 3D data set need a total acquisition time of ~6.0 s and covers a 2×2 mm spatial region in x-y plane. Finally, a raw data cube of spectral interferogram $S(k,x,y,t)$ was generated for each 3D volume scan, where k is the wavenumber, and t is the number of the repeated B-frame at each scanning step y.

Detailed derivation of Eq. (2)

Local complex decorrelation, as an estimator of the motion magnitude, was calculated between successive B-scans obtained at the same location as well as surrounding voxels with a 4D spatio-temporal average kernel, i.e.:

$$D = 1 - \frac{1}{T-1}\sum_{t=1}^{T-1}\frac{C(t)}{I(t)} = 1 - \frac{1}{T-1}\sum_{t=1}^{T-1}\frac{\sum_{s=1}^{S}A(t,s)A^*(t+1,s)}{\sum_{s=1}^{S}\frac{[A(t,s)A^*(t,s)+A(t+1,s)A^*(t+1,s)]}{2}}, \quad \text{(S1)}$$

where $C(t)$ is the local first-order auto-covariance and $I(t)$ is the local zeroth-order auto-covariance or generally called intensity. $A(t,s)$ denotes the resultant complex OCT signal, t and s is the index in temporal and 3D spatial dimensions with a total kernel size of T and S, respectively, and the kernel was fixed to $5 \times 1 \times 1 \times 3$ ($z \times x \times y \times t$) in this study.

Based on the multi-variate time series (MVTS) model mentioned in Ref. [1], the mathematical expectation of decorrelation can be derived as following:

$$1 - D \rightarrow \frac{E[A(t)A^*(t+1)]}{E[A(t)A^*(t)]} = \frac{(1-\alpha)^2 E[A_s(t)A_s^*(t)]}{\alpha^2 E[A_d(t)A_d^*(t)] + (1-\alpha)^2 E[A_s(t)A_s^*(t)] + E[n(t)n^*(t)]}, a.s., \quad \text{(S2)}$$

where $a.s.$ means convergence with probability one. To simplify the formula, the concept of iSNR was involved,

$$iSNR = \frac{E[n(t)n^*(t)]}{E[A(t)A^*(t)]} = \frac{E[n(t)n^*(t)]}{\alpha^2 E[A_d(t)A_d^*(t)] + (1-\alpha)^2 E[A_s(t)A_s^*(t)] + E[n(t)n^*(t)]}, \quad \text{(S3)}$$

$$D \rightarrow 1 - \frac{(1-\alpha)^2}{\alpha^2 + (1-\alpha)^2}(1 - iSNR), a.s.. \quad \text{(S4)}$$

where the noise level $E[n(t)n^*(t)]$ was determined by averaging the air region and the bottom noise region in tomograms and the local signal intensity $E[A(t)A^*(t)]$ was calculated based on the OCT data.

References

1. L. Huang, Y. Fu, R. Chen, S. Yang, H. Qiu, X. Wu, S. Zhao, Y. Gu, and P. Li, "SNR-Adaptive OCT Angiography Enabled by Statistical Characterization of Intensity and Decorrelation With Multi-Variate Time Series Model," IEEE Trans Med Imaging **38**, 2695-2704 (2019).