Multiparametric optical coherence tomography imaging of the inner retinal hemodynamic response to visual stimulation

Harsha Radhakrishnan
Vivek J. Srinivasan
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University of California Davis, Department of Biomedical Engineering, Davis, California 95616

Abstract. The hemodynamic response to neuronal activation is a well-studied phenomenon in the brain, due to the prevalence of functional magnetic resonance imaging. The retina represents an optically accessible platform for studying lamina-specific neurovascular coupling in the central nervous system; however, due to methodological limitations, this has been challenging to date. We demonstrate techniques for the imaging of visual stimulus-evoked hyperemia in the rat inner retina using Doppler optical coherence tomography (OCT) and OCT angiography. Volumetric imaging with three-dimensional motion correction, en face flow calculation, and normalization of dynamic signal to static signal are techniques that reduce spurious changes caused by motion. We anticipate that OCT imaging of retinal functional hyperemia may yield viable biomarkers in diseases, such as diabetic retinopathy, where the neurovascular unit may be impaired. © The Authors. Published by SPIE under a Creative Commons Attribution 3.0 Unported License. Distribution or reproduction of this work in whole or in part requires full attribution of the original publication, including its DOI. [DOI: 10.1117/1.JBO.18.8.086010]

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1 Introduction

In the central nervous system, neurovascular coupling is initiated by neurotransmitter release at the synapse, which eventually leads to the control of contractile elements that surround vasculature, typically arterioles and capillaries.1 The resulting local increase in blood flow is called functional hyperemia. While the brain has been used extensively to study functional hyperemia, the inner retina represents an under-utilized model system for investigating this phenomenon. Laser Doppler flowmetry has been applied to study retinal blood flow2 and neurovascular coupling in the optic nerve head.3 Recently, laser speckle imaging and confocal microscopy were shown to image stimulus-induced blood flow and diameter changes in the inner retina.4 Fundus imaging studies, while lacking depth resolution, have revealed reflectance changes suggesting that blood volume increases in response to a light stimulus.5 A retinal blood flow response to a flicker stimulus was suggested using cross-sectional Doppler optical coherence tomography (OCT) in human subjects.6 Using a volumetric imaging protocol in vivo, OCT signal from the photoreceptors was shown to change in response to a light stimulus; but so far, no robust inner retinal changes have been conclusively demonstrated with OCT in vivo.

In this work, we demonstrate multiparametric imaging of visual stimulus-evoked hyperemia including blood flow, vessel diameter, and vascular red blood cell content/speed. We perform asynchronous volumetric imaging during repeated stimulus trials, and use time-series analysis techniques derived from functional magnetic resonance imaging (fMRI)7 to determine trial-locked responses.

2 Materials and Methods

Animal protocols were approved by the animal care and use committee where these experiments were performed. Long–Evans rats (250 to 320 g) were anesthetized with 40 to 50 mg/kg/h alpha-chloralose i.v., maintained at 37°C, cycled with 1% tropicamide (Mydriacyl), and paralyzed with rocuronium. Periocular irrigation with 2% lidocaine was further used to reduce motion artifacts. During imaging, a contact lens was gently placed on the cornea with goniostat. Rats were ventilated at 0.7 Hz with 85% air and 15% oxygen, and arterial blood pressure was monitored (90 to 130 mmHg). End-tidal CO₂ was monitored with a capnometer and maintained below 40 mmHg. Rats were immobilized in a custom-made stereotactic frame during imaging.

A 1310 nm spectral/Fourier domain OCT microscope was constructed for in vivo imaging of the rat’s retina9 with 7.5 μm transverse resolution, as shown in Fig. 1. A longer wavelength was used to avoid possible stimulation of the retina by the imaging system.10 The light source consisted of two superluminescent diodes combined, using a 50/50 fiber coupler to yield a bandwidth of 150 nm. The measured axial (depth) resolution in tissue was 3.8 μm. A spectrometer with a 1024-pixel InGaAs line scan camera (Goodrich Sensors Unlimited, Inc., Princeton NJ) was operated at 47,000 axial scans/s and controlled by a first computer (PC1). The power of the OCT beam on the sample was 1 mW, and the sensitivity was 96 dB. A square-wave flickering light stimulus was generated for 10 s with a fiber light source (Hg:Xe lamp) and a mechanical shutter controlled by a second computer (PC2). The maximum flicker intensity at the retina was estimated to be 2 mW/cm², while the minimum intensity between stimuli was <20 nW/cm². The stimulus spectrum is shown in Fig. 1. Comparable stimulus protocols and experimental designs are frequently used in the literature to study functional hyperemia.
in the rat retina.\textsuperscript{4,11,12} Uniform illumination of the retina was confirmed by imaging the retina onto a CCD camera connected to PC2. Repeated OCT volumes for angiography (128 $y$-locations $\times$ 256 axial scans per frame $\times$ 2 frames per $y$-location) or Doppler OCT (96 $y$-locations $\times$ 1024 axial scans per frame $\times$ 1 frame per $y$-location) were acquired during runs of 10 stimuli separated by 25 s. In one run, a stimulus frequency of 12 Hz was used, while in the other run, stimulus frequencies between 6 and 18 Hz were generated in a random order to test for frequency dependence of the response.\textsuperscript{13} All triggers were collected along with the blood pressure and capnometer readings by a third computer (PC3). Figure 2 shows the asynchronous stimulus and scanning protocol for angiography. Jittered acquisition and stimulation protocols, similar to the ones employed here, are routinely used in fMRI to achieve a higher effective temporal resolution when responses are repeatable\textsuperscript{14} across trials.

Figure 3 shows the effects of motion correction on the angiograms. Motion correction was achieved by volumetric
For absolute blood flow changes\textsuperscript{15}, here, we argue that filtered by a 2-s sliding window, and then averaged over all trials a comparison of Fig. 4(c) and 4(d) suggests that the measuring change in the blood pressure during the stimulus. Furthermore, measurements of hemodynamics. The first advance is that transverse pulsatile flow variation associated with the heartbeat. Hence, for relative blood flow calculated in the B-mode plane

\[
\frac{df_{rel,z}}{d\theta} = \frac{d}{d\theta} \cot(\theta) = -\csc^2(\theta). \tag{1}
\]

Thus, as shown in Fig. 5(b) and 5(c), cross-sectional area and mean velocity axial projection change in the same direction when \(\theta\) changes, thus exacerbating relative flow noise. The effect of angular changes on flow calculated in the B-mode plane is most pronounced near \(\theta = 0^\circ\). By comparison, for relative blood flow calculated in the \textit{en face} (xy) plane, the vessel cross-sectional area \(A_{xy}\) varies as \(1/\sin(\theta)\), and the mean velocity axial projection \(v_z\) varies as \(\cos(\theta)\), causing relative flow to have an angular dependence of \(\cot(\theta)\). Hence, for relative blood flow calculated in the \textit{en face} plane

\[
\frac{df_{rel,xy}}{d\theta} = \frac{d}{d\theta} [1] = 0. \tag{2}
\]

For these reasons, we found that relative \textit{en face} flow had a standard deviation of 10\%, whereas relative B-mode flow had a standard deviation of 28\% for a transverse shift of 100 \(\mu\)m near the optic nerve head (\(N = five vessels\)).

The second advance is that OCT signal can be separated into static and dynamic components\textsuperscript{16} via a complex high-pass filter along the slow axis. More specifically, we estimate the dynamic signal \(I_d\) by taking a point-wise complex difference between consecutive frames acquired at the same location:

\[
I_d = I - I_{0}\text{.}
\]
Fig. 5 (a) Geometric drawing showing how velocity axial projection and cross-sectional area depend on vessel angle. (b, c) For relative flow determined in the B-mode plane, the velocity axial projection and vessel area change in the same direction when the vessel angle changes due to a 100-μm transverse shift. However, for relative flow determined in the en face plane (d, e), the velocity axial projection and vessel area change in opposite directions. Due to this counterbalancing effect, relative en face flow had a standard deviation of 10%, whereas relative B-mode flow had a standard deviation of 28% for a transverse shift of 100 μm near the optic nerve head (N = five vessels). Color-coded boxes are used to designate the same vessels in different planes.

Fig. 6 OCT angiography of functional hyperemia caused by a 10-s, global 12-Hz light stimulus. (a) The dynamic signal (red-dashed line) is corrupted by motion artifacts. A vascular mask is created to distinguish static and dynamic signals, and the dynamic signal is normalized to the static signal (blue-dotted line) to remove motion artifacts revealing intrinsic dynamic scattering changes (black-solid line). Comparison of averaged angiograms at baseline (b) and during activation (c) shows evidence of arterial dilation (d). The baseline and activation time windows are marked in (a). (e) Diameter changes were manually quantified along the arterial tree.
\( I_d(x, y, z) = \frac{1}{2} |A_{2n}(x, y, z) - A_{2n-1}(x, y, z)|^2. \)  

(3)

\( A_{2n} \) and \( A_{2n-1} \) represent the axial phase-corrected complex OCT signal amplitudes at the \( n \)th \( y \)-position. We estimate the static signal \( I_s \) by excluding pixels where the dynamic signal exceeds a threshold \( I_t \).

\[ I_s(x, y, z) = \frac{1}{4} |A_{2n}(x, y, z) + A_{2n-1}(x, y, z)|^2 \times \left[ I_d(x, y, z) < I_t \right]. \]  

(4)

Here “dynamic” signal is defined as any component that varies faster than the \( \sim 7 \) ms timescale (the interframe time for angiography). For decorrelation due to transverse motion, this corresponds to a speed of \( \sim 1 \) mm/s. If changes during activation are purely hemodynamic, dynamic scattering may increase or decrease due to the increase or decrease in red blood cell content and speed, while intrinsic static scattering properties remain unchanged. Artifacts such as papillary constriction would have similar effects on detected signals from both dynamic blood cells and the adjacent static tissue. The static (nonvascular) signal thus serves as a convenient “reference” to compensate artifacts in the dynamic signal and reveals stimulus-evoked changes.

3 Results and Discussion

As residual motion is on the order of microns, masks can be defined to distinguish dynamic signal [Eq. (3), mostly due to red blood cells] from static signal [Eq. (4), due to neuronal, glial, and possibly endothelial and smooth muscle tissue]. Figure 6(a) shows that normalization of the dynamic signal to the static signal reduces artifacts and reveals intrinsic changes related to functional hyperemia. Exemplary images of the dynamic and static signals, determined from Eqs. (3) and (4), respectively, are shown. As the dynamic signal consists of scattering signal components varying faster than a 7 ms timescale, both an increase in red blood cell content and an increase in speed (through increased Doppler shift or decorrelation rate) may account for the observed changes.17 Arterial speeds are expected to be larger than \( 1 \) mm/s, meaning that after 7 ms, the complex-valued signal is already uncorrelated. Therefore, flow speed changes minimally affect the dynamic signal, as determined by Eq. (3). Averaged angiograms at baseline and during activation revealed that arterial diameter also increases, as shown in Fig. 6(b)–6(d). Diameter changes were quantified manually along the arterial tree at baseline and during activation in Fig. 6(e) (CI = 3.3% to 11.0%, \( p < 0.005 \)).

Figure 7 shows imaging of blood flow changes during activation using the method of en face integration.15 Figure 7(a) shows an angiogram with a white arrow pointing to a vessel

![Image](image-url)
whose flow was measured. Figure 7(b) and 7(c) show en face Doppler OCT images of the vessel at baseline and during peak activation. Absolute flow was calculated by integrating the velocity axial projection over the en face plane. Figure 7(d) shows the absolute flow time series during all 10 stimuli along with predictions determined by fitting a linear model to the data. A third-order polynomial was included in the linear model to account for baseline fluctuations. Figure 7(e) shows the fractional flow change averaged across all 10 trials. Relative dynamic scattering changes [Fig. 6(a)] were smaller than relative flow changes [Fig. 7(e)]. Figure 7(f) shows a preliminary investigation of the stimulus frequency dependence of the time-integrated responses between 6 and 18 Hz.

We note some limitations of our techniques here. The strategy of static signal normalization is valid only if intrinsic scattering properties of static tissue do not change appreciably due to activation. It is conceivable that if blood volume increases, retinal tissue is compressed, and therefore, scattering properties may change. However, in our previous studies of functional OCT changes in the retina, we were unable to detect any corresponding inner retinal changes. So-called “fast optical signal” changes, resulting directly from neuronal activity and membrane polarization changes, are likely too small to be significant here.

Overall, we observed repeatable blood flow, diameter, and dynamic scattering increases in the inner retina during a visual light stimulus. Our results strongly suggest that these are hemodynamic responses in the inner retina caused by the visual stimulus. Similar stimulus protocol and experimental designs have been used in the literature to evaluate visual stimulus-evoked inner retinal hemodynamic changes. Moreover, we observed no systematic physiological changes (such as blood pressure changes) synchronized with the stimulus that could account for the observed responses [Fig. 4(c) and 4(d)]. The time courses and magnitudes of the measured responses are consistent with previous studies in the same species. However, we also observed a weak flicker frequency dependence of the blood flow response time integral [Fig. 7(f)], suggesting that the evoked response may be due, at least in part, to the increase in average luminance during stimulation.

4 Conclusion

In conclusion, methodologies for imaging the inner retinal hemodynamic response to a light stimulus in the rat retina were demonstrated. Volumetric imaging and three-dimensional correlation significantly reduced motion artifacts. En face measurements of relative flow changes were found to be robust against motion. Normalization of the dynamic signal to static signal mitigated spurious changes from residual accommodation, pupil constriction, or axial motion. We anticipate that the techniques presented here can be applied to study comparable hemodynamic responses in human subjects.

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References