Image stabilization for scanning laser ophthalmoscopy

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Abstract: A scanning laser ophthalmoscope with an integrated retinal tracker (TSLO) was designed, constructed, and tested in human subjects without mydriasis. The TSLO collected infrared images at a wavelength of 780 nm while compensating for all transverse eye movements. An active, high-speed, hardware-based tracker was able to lock onto many common features in the fundus, including the optic nerve head, blood vessel junctions, hypopigmentation, and the foveal pit. The TSLO has a system bandwidth of ∼1 kHz and robustly tracked rapid and large saccades of approximately 500 deg/sec with an accuracy of 0.05 deg. Image stabilization with retinal tracking greatly improves the clinical potential of the scanning laser ophthalmoscope for imaging where fixation is difficult or impossible and for diagnostic applications that require long duration exposures to collect meaningful information.

OCIS codes: (170.3880) Medical and biological imaging, (170.4470) Ophthalmology.

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1. Introduction

Eye motion has long been recognized as a problem for both therapeutic and diagnostic applications.[1–3] Laser photocoagulation for the treatment of age-related macular degeneration or retinal detachments requires precise targeting to treat the retinal periphery while preventing damage to the fovea.[4] Lasik and other corneal refractive surgeries also require precise alignment of the instrument with the eye, especially for those systems that "write" complex maps to the corneal stroma in order to correct ocular optical aberrations. Diagnostic applications also require correction of eye movements, especially emerging technologies such as scanning laser ophthalmoscopy (SLO), fluorescence imaging, and optical coherence tomography (OCT).[5,6] These techniques require long exposure durations either to collect data from a large spatial extent (e.g., three-dimensional OCT) or to collect enough photons to achieve reasonable signal-to-noise levels within the confines of laser safety requirements.

Eye motion stabilization can generally be accomplished invasively with suction cups, inaccurately with fixation, more precisely at slower speeds with a passive image processing approach, or at high speeds with active tracking. Fixation requires patient cooperation and is difficult in patients with poor vision due to the very diseases that need to be imaged. Passive eye tracking approaches use image information to correct eye movements from one frame to the next and thus are limited to correction of motion that does not exceed the detector frame rate.[7] Other active tracking techniques, such as Purkinje reflectors, are specifically designed for the anterior segment and are not easily modified to get an accurate measurement of retinal position.[2,8] Thus there is currently no straightforward method for image stabilization for diagnostic ophthalmic applications.

The TSLO was designed to provide retinal tracking in a scanning laser ophthalmoscope with a novel optical arrangement. Imaging is accomplished by scanning an illumination line with a single galvanometer, which results in a simple, more compact SLO.[9] The retinal tracking is accomplished with an active, hardware-based system that uses a confocal reflectometer for low-power tracking beam detection and phase-sensitive electronics to achieve high overall system bandwidth.[10]

2. Materials and Methods

The general optical layout of the new instrument to stabilize a raster on the retina during imaging is shown in Fig. 1. The instrument consists of a custom-designed confocal scanning laser ophthalmoscope (SLO) and a retinal tracker. The SLO portion of the system consists of an illumination source (LD) and detector, imaging galvanometer (IG), and imaging lenses (scan lens, SL and ophthalmoscopic lens, OL). The retinal tracking portion consists of a confocal tracking reflectometer (TR), dither scanners (DS), and tracking galvanometers (TG). The illumination source is a single-mode fiber-coupled, 2.5-mW, 780-nm, laser diode (Thorlabs Inc.). The imaging galvanometer (Cambridge

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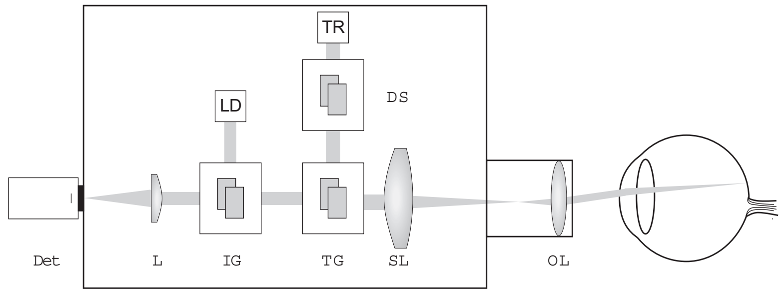

Fig. 1. General optical layout of TSLO. OL: ophthalmic lens, SL: f/2 scan lens, TG: tracking galvanometers, DS: dither scanners, TR: tracking reflectometer, IG: imaging galvanometer, LD: laser diode imaging source, L: f/2 lens, det: detector.

Technology Inc.) simultaneously scans a line across the fundus and de-scans the backscattered return onto a digital line array detector (Dalsa Inc.). Since the SLO uses an illumination line rather than a point, it is confocal in one dimension since backscattered light is rejected in only one transverse dimension rather than radially.[9] The frame acquisition rate can be adjusted in the software and is generally set to 15 or 30 frames/sec.

The retinal tracking system works by steering the entire image raster produced by the imaging galvanometer with the motion of the eye using the tracking galvanometers (Cambridge Technology Inc.). A tracking beam, locked onto a retinal feature, senses the motion of the eye. Thus the retinal tracker described in this paper is an active, hardware-based, high-speed tracker. A confocal reflectometer is used so that only reflected light from the plane of the fundus determines eye position. The error signals generated by the system are therefore not affected by reflections from the cornea and lens. The source for the tracker beam is an 880-nm light-emitting diode (PD-LD Inc.) and the tracker beam power measured at the cornea was $\sim$25 μW. An avalanche photo-diode (APD, Hamamatsu Inc.) is used to detect the extremely weak signal return from this low-power source. The tracker beam is dithered in a circle with dither scanners (Electro-Optical Products Corporation) driven at their resonant frequency of 8 kHz and with 90° phase separation between x and y scanners. When the tracking beam passes over a retinal feature with brightness different from the background, the APD signal will contain an 8-kHz signal (and harmonics), the phase of which is proportional to the distance between the tracker beam and the target. Phase-sensitive detection with a lock-in amplifier is employed to create error signals, which are then fed into a DSP feedback control loop. The control loop commands the tracking galvanometers according to the processed error signals to keep the imaging raster locked with the motion of the eye. Complete details of the TSLO can be found in a forthcoming paper.[11]

Five volunteers were evaluated in initial human subject tests. All subjects had normal healthy eyes except one who had central serous retinopathy and hypopigmentation in his left eye. The hypopigmentation was used successfully as a tracking feature. Several different experiments were performed including comparison to fixation (quantified with cross-sectional analysis), measurement of tracking velocity and accuracy during large and rapid saccades, and tracking on various natural features in the retina.

3. Results and Discussion

Most implementations of corneal tracking, for example, those that are used during laser photorefractive surgery, use algorithms that compute eye motion from successive frames. These versions use sophisticated image processing routines to identify and track any changes in an image landmark (e.g., the pupil). Hardware implementation of tracking

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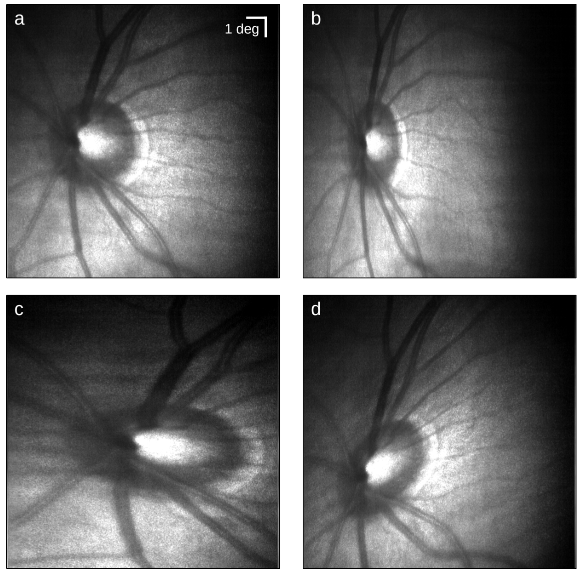

Fig. 2. Rapid eye motion can create problems for software-based retinal trackers. (a) Single frame of fundus when eye is stationary, (b) single frame when eye slews in the opposite direction as the image scanner, (c) single frame when eye slews in the same direction as the image scanner, and (d) single frame when eye slews in a direction perpendicular to the image scan. Acquisition rate was 30 frames/sec.

described herein is quite different from those tracking systems because it can achieve bandwidths far exceeding those currently achieved in software implementations of eye tracking (>1 kHz). An example of the effect rapid eye motion can have on a software retinal tracker is shown in Fig. 2. At video rates of 30 frames/sec, image compression, expansion, or distortion can occur as the eye slews with, against, or in a direction perpendicular to the image scan. Although the distortion in Fig. 2d appears to be from motion at an oblique angle, since each illumination line is detected simultaneously, the motion is actually exactly perpendicular to and faster than the line scan.

In all ocular imaging techniques that do not employ some type of motion tracking for image stabilization, eye motion is minimized via fixation. Fixation works reasonably well for young, healthy subjects with eyes absent of retinal disease. However, as one ages, there is a decrease in ability to rapidly accomplish eye movement tasks.[12] Fixation is also inaccurate and not centrally located in patients with eye disease that affects central vision. Fixation is therefore not an option in these cases. Even in healthy subjects, fixation can seldom achieve the degree of stability necessary for applications that require collection at low light levels via frame co-addition. In Fig. 3, image co-addition is compared for cases with and without fixation and with and without tracking for a healthy 46-year old subject. Comparison of Figs. 3a and b illustrate marked improvement in motion stabilization with fixation. However, the improvement in stabilization with fixation cannot compare to the degree of improvement with tracking seen in Figs. 3c and d. There is little difference between Figs. 3c and d, apart from the region of maximum brightness since the tracking system bandwidth is large enough to correct for any and all transverse eye motion.

To quantify the image smear in Fig. 3, a cross-sectional analysis was performed to measure the width of a large retinal blood vessel. The region in which the cross-section was taken is denoted with a line in Fig. 3d. Table 1 lists the width for the four images of 90 co-added frames of Fig. 3, as well as for a single frame of one of the videos. While the difference between the single frame and Figs. 3c and d was less than 1 pixel, for

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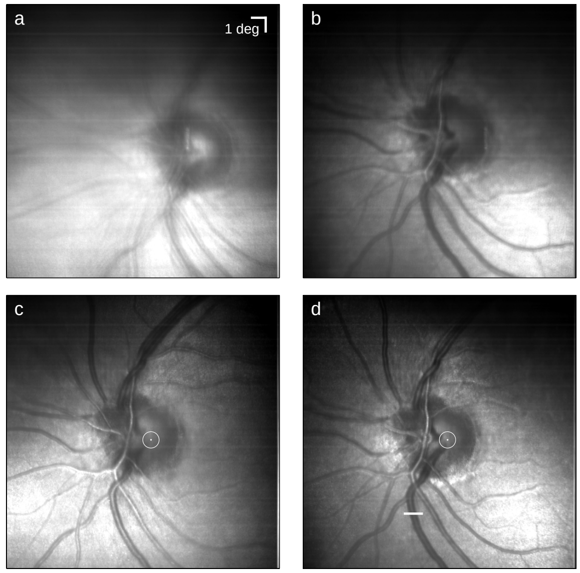

Fig. 3. Comparison of 90 frames (6 sec) co-added for a single subject (a) without tracking and fixation (moderate saccades), (b) without tracking but with fixation, (c) with tracking and fixation, and (d) with tracking but without fixation. Line indicates cross-sections used for measurement of motion blur.

Fig. 3b it was greater than 1 pixel and for Fig. 3a it was nearly 5 pixels.

In order to test the limits of the tracking system, experiments were performed to quantify the tracking velocity and accuracy. In these experiments, tracking was initiated in one eye while the subject viewed a target ~30–50 inches away with the contralateral eye. The tracking target was the bright lamina cribrosa in the optic disc. Subjects were instructed to shift their gaze rapidly between a target of four points separated by ~12 inches. In several trials, the tracked eye motion was nearly 500 deg/sec (e.g., 21.3 deg in 46 ms, 0.291 μm/deg on the retina). The measured RMS tracking accuracy was ~0.05 deg or <1 pixel for the 28.6 deg field (66-diopter Volk lens). Rapid eye motion stabilization is illustrated dynamically in Fig. 4. Since the imaging aperture is viewed through the tracking mirrors in this system, the video image will display very little motion despite the large and rapid saccades seen in the position signals collected from the tracking mirrors. The amount of eye motion in Fig. 4 can be understood qualitatively by observation of the subject's iris, which vignettes the image in the corners of the video frame when the eye has moved a significant amount.

In both Figs. 3 and 4, the utility of retinal tracking is apparent for applications in

Table 1. Cross-sectional analysis of vessel width.

image	vessel width [pixels, FWHM]
Fig. 3a	19.20
Fig. 3b	16.28
Fig. 3c	15.28
Fig. 3d	14.83
single frame	14.66

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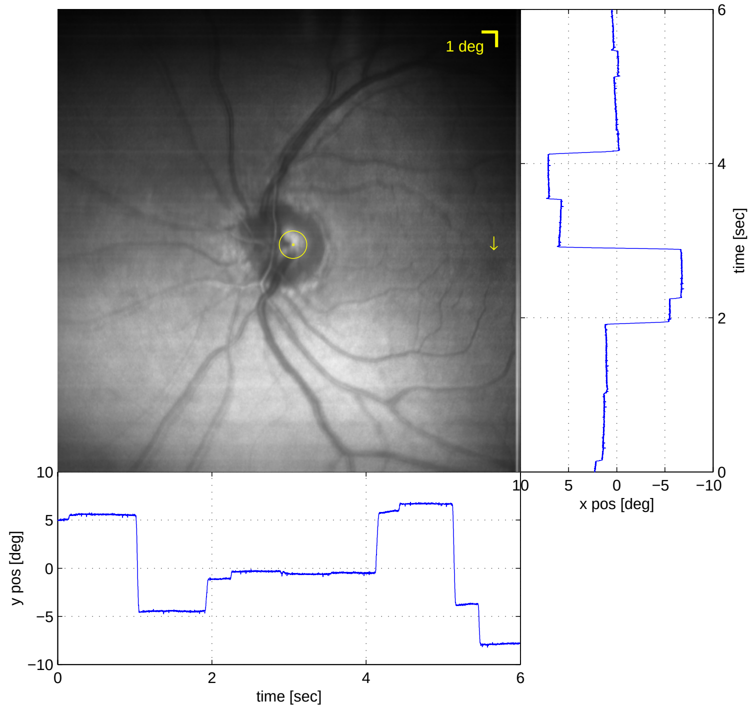

Fig. 4. (2.3 Mb) Tracking during large, fast saccades (∼300 deg/sec). Eye x- and y-positions acquired from galvanometers are shown in graphs beside video. Tracking point (circles indicate dither beam radius and amplitude) and fovea (arrow) are denoted. Still image is co-added frames over the duration of the video. Acquisition rate was 15 frames/sec.

which long exposure times or frame co-addition must be employed in order to collect low light levels back-scattered from the eye. In these images, higher order motion artifact can also be seen. For example, as one moves away from the tracking point (optic disc), torsional motion is more pronounced. This can be observed as an increase in the blur of a vessel as the distance from the optic disc increases. However, involuntary torsional eye motion and its effects on position accuracy is much smaller than that for voluntary transverse saccades. A potential clinical limitation for the TSLO is retinal tracking in the presence of opacities in the anterior segment. However, since both tracking and imaging is accomplished in a confocal optical arrangement, low to moderate homogeneous opacities may reduce the light collection efficiency but will not seriously degrade tracking fidelity. Dense, inhomogeneous lens opacities will cause problems for any system that collects information from signals backscattered from the retina.

In initial human subject trials, retinal tracking using the optic nerve disc (lamina cribrosa) as the target could be achieved in all subjects. In subjects where optic disc tracking cannot be accomplished, for example when spontaneous venous pulsation in vessels passing through the optic disc cause dynamically-varying contrast, tracking can be performed using other natural features in the eye. Figs. 5, 6, and 7 illustrate tracking on a blood vessel junction, hypopigmentation, and the fovea.

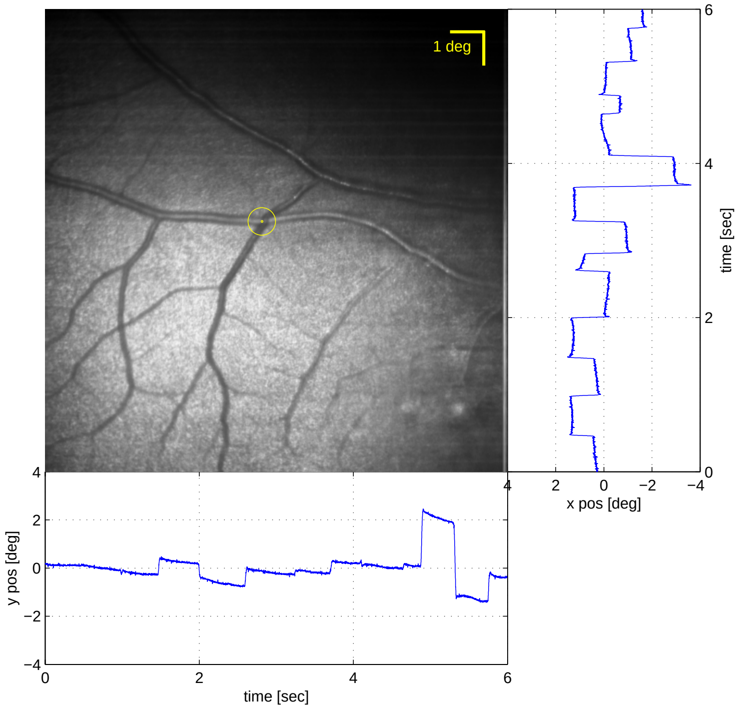

Fig. 5. (2.7 Mb) Tracking on retinal blood vessel junctions. Annotated as Fig. 4.

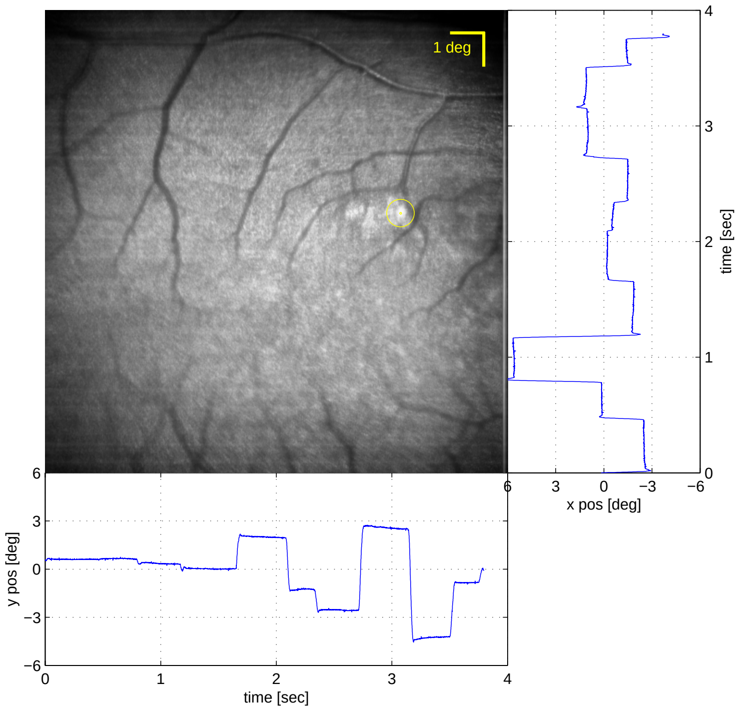

Fig. 6. (2.1 Mb) Tracking on a region of hypopigmentation. Annotated as Fig. 4.

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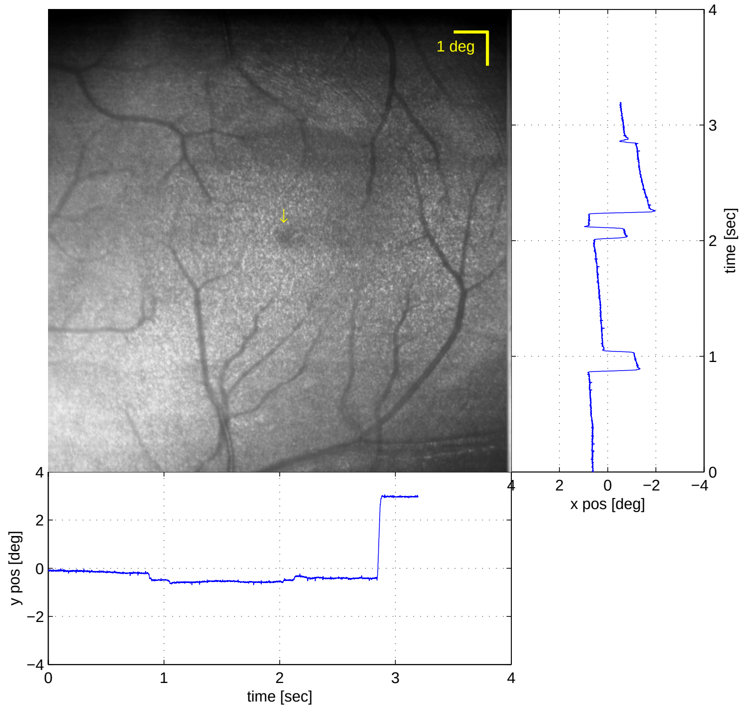

Fig. 7. (1.6 Mb) Tracking on the foveal pit. Annotated as Fig. 4.

4. Conclusion

The performance of a scanning laser ophthalmoscope with high-speed retinal tracker was tested in human subjects. The instrument performance far-exceeds passive, image processing-based eye trackers. A comparison of image stabilization in healthy subjects by fixation and retinal tracking showed the latter to significantly reduce image blur. The system was able to track large (>20 deg) and rapid (∼500 deg/sec) saccades with an accuracy of ∼0.05 deg. Although tracking was best on the optic nerve disc (lamina cribrosa), the system was able to lock onto blood vessel junctions, hypopigmentation, scleral crescent, and macular pigment in the fovea.

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