Tracking Optical Coherence Tomography

R. Daniel Ferguson  
Daniel X. Hammer  
Lelia Adelina Paunescu  
Siobahn Beaton  
Joel S. Schuman


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Abstract

An experimental tracking optical coherence tomography (TOCT) system has been clinically tested. The prototype instrument uses a secondary sensing beam and steering mirrors to compensate eye motion with a closed loop bandwidth of 1 kHz. The retinal tracker improved image registration accuracy to <1 pixel yielding co-added OCT images averaged over multiple visits with sharp fine structure. As the resolution of clinical OCT systems improves, tracking accuracy will be reduced to its theoretical limit, less than the OCT beam diameter. The capability to reproducibly map complex structures in the living eye at high resolution will lead to improved understanding of disease processes and improved sensitivity and specificity of diagnostic procedures.

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Optical coherence tomography (OCT) has emerged over the past decade as an outstanding tool for imaging biological structures. Ultra-high resolution, Doppler, spectroscopic, polarization-sensitive, and three-dimensional OCT imaging modes, among others, have been demonstrated by numerous researchers\textsuperscript{1-5}. These tools have found important applications in quantitative retinal imaging, and detection and characterization of retinal pathology. Clinical OCT instruments have been shown to produce highly accurate maps of the retinal nerve fiber layer (RNFL)\textsuperscript{6}. We have modified a commercial clinical OCT system by integrating a prototype of an active, hardware-based retinal tracker for OCT image stabilization. The new experimental system, called tracking optical coherence tomography (TaCT), uses a secondary sensing beam and steering mirrors to compensate eye motion\textsuperscript{7} with a closed loop bandwidth of up to 1 kHz. Reproducible mapping and visualization of retinal structure and function with increasing resolution will lead to improved understanding of disease processes, and to improved sensitivity and specificity of diagnostic procedures. Retinal tracking-stabilized OCT will greatly facilitate progress toward these goals by enabling precise and ever more complex scan sequences.

At present, the speed of OCT imaging techniques is insufficient to realize their full potential in the clinic without image stabilization, due to both accessible scan rates and the ANSI eye safety standards for maximum permissible light exposure. There are fundamental tradeoffs between acquisition time, spatial resolution, incident power and noise; the OCT beam must dwell on each voxel long enough, or repetitively scan each voxel enough times, to allow the detector to collect sufficient energy to overcome the system noise, including shot and speckle noise. When imaging large tissue volumes, acceptable signal-to-noise ratios and image quality can be obtained only at the expense of long scan durations that are frequently longer than a patient’s ability to fixate. The third generation clinical OCT system (Stratus\_OCT, Carl Zeiss Meditec Inc.,
CZMI) can capture a 1024 (depth) × 500 (transverse) voxel image with a scan duration of 1.25 sec, for a total voxel scan rate of over 400K voxels per second. The incident SLD beam power is 750 µW, with ~8 µm depth resolution and ~20 µm transverse resolution. Recently, 3-D transversal OCT imaging has been demonstrated with 1.2 sec scan times at 2 million voxels per second scan rates, though with less depth resolution and several times the incident power. These examples are very near the fundamental limits, and scan dimensions and durations have been designed, in effect, to avoid the microsaccadic eye motion “barrier” near one second.

Involuntary eye movements during image acquisition, and particularly between scans, currently limit the multi-image acquisition potential of OCT, and thus its viability for several advanced clinical applications. To correctly build an image voxel-by-voxel or co-add multiple images, the coordinates of each voxel relative to a fixed reference point must be known. Even in patients that are stable fixators, eye motion amplitude may be several hundred microns, much larger than the resolution of the second and third generation CZMI OCT instruments. Because there is currently no straightforward means of re-synchronizing the scanning data acquisition system coordinates (θ, φ, and z) with the physical coordinates of the retina, eye motion during extended OCT image acquisition may create errors and artifacts that cannot be compensated by post-processing. As OCT image volume or resolution increases (resulting in longer scan durations), a means of retinal tracking is increasingly important. Retinal position tracking enables OCT to repetitively scan locations accurately during an extended scan period and permits averaging of multiple scans. Frame averaging techniques reduce OCT speckle noise with minimal impact upon transverse or longitudinal resolution. Detailed 2-D scans can be concatenated into error free 3-D image cubes. Such capabilities may be expected to provide greater diagnostic sensitivity and specificity for glaucomatous changes in the eye, for example.
Because image registration is based on retinal landmarks, precisely aligned scans can be compared weeks, months, or years apart. Retinal morphology can thus be accurately recorded and monitored over time to detect subtle changes.

The prototype retinal tracking instrument employed in the clinical tests described below is a high-speed, non-imaging system that uses retinal landmarks to achieve precise transverse image registration without image processing. The patented system, illustrated in Figure 1, projects a low power tracking beam onto small retinal features of modest light or dark contrast relative to neighboring areas. Such features may include blood vessel junctions, lamina cribrosa (a bright layer within the optic nerve disc), or any retinal lesions, drusen, scars or irregular pigmentation. The tracking beam is dithered on the retina with resonant scanners (DS), detected with a confocal reflectometer, and processed with phase-sensitive electronics. The tracking beam is directed onto the retina using a galvanometer-driven “tracking” mirror pair (TG). Real-time digital signal processing extracts position and orientation information from optical tracking servo signals to make precise closed-loop beam steering corrections much faster than the highest possible image-framing rate. Because the OCT beam is also reflected from the tracking mirrors, this beam automatically follows any movement of the retina with the same speed and precision as the tracking beam. RMS tracking errors are typically smaller than the OCT beam diameter. After integrating this tracking technology into a clinical OCT II (CZMI) system, precise eye-motion correction was demonstrated within and between 2-D OCT scans in both normal subjects and glaucoma patients. Tracking beam power never exceeded 25 \( \mu \text{W} \) in the clinical tests.

The clinical tests consisted of a series of OCT scans performed on 3 separate visits of more than twenty subjects with normal and glaucomatous eyes collected both with and without retinal tracking. The tracking feature selected for all scans was the lamina cribrosa, which is
generally a stable, robust tracking feature. The sequence of OCT scans used in this protocol for each visit includes circle scans around the disc (angular diameter of ~12 deg and a resolution of ~110 μm), and sequences of radial scan through the disc and the fovea (6 scans successively rotated by 60 deg with a scan length of ~20 deg and a resolution of ~60 μm). The circle scan is generally used to measure retinal nerve fiber layer (RNFL) thickness around the optic disc to screen or monitor subjects with glaucoma. The radial scan is used to generate a map of a large region of the retina for measurement of RNFL or retinal thickness. OCT scans were processed and analyzed with custom software designed to register and co-add multiple images within and between visits and to extract quantitative information on the location of retinal features with sharp edges (e.g., optic disc cup). Because image noise (dominated by speckle noise leading to significant image granularity) varies approximately inversely with the square root of the number of measurements, image co-addition is used to increase signal-to-noise ratio. In the co-added OCT scans, qualitative criteria for improvement due to eye motion stabilization were the number and width of blood vessel shadow edges, and the clarity of clearly identifiable features such as distinct retinal and sub-retinal layers. Quantification of tracking performance in comparison to fixation by measurement of transverse image registration is limited by the OCT transverse pixel resolution. The total peak-to-peak amplitude of eye motion for a good fixator, including involuntary microsaccades and drift, is ~0.5 deg (~150 μm), or approximately 1.5 and 3 pixels for circle and radial OCT II scans respectively. The accuracy of a similar tracking system in other higher resolution imaging implementations was found to be <0.05 deg (~15 μm). In the clinical trials, a qualitative improvement in OCT circle disc scans was found in 97% of all subject visits. Analysis of the mean variation in disc edge position in OCT radial disc scans found an improvement in every subject, with a reduction from greater than 2 to approximately 1
Figures 2 and 3 illustrate the tracking performance for 2 subjects. A detailed analysis of the results for all subjects will be published elsewhere.

Figure 2 shows average OCT disc circle images co-added from three successive visits to the clinic (~60 total scans) with and without tracking. Note the absence of speckle noise in the averaged OCT images. Even the image co-added without tracking (Fig. 2c) shows substantial signal-to-noise improvement over a single scan, though with significant motional blurring of fine detail. Horizontal retinal layers are generally less affected because the shallow gradients of these layers, even in subjects with glaucoma, renders the sharpness of such boundaries (once vertically aligned) less sensitive to in-plane motion. Nevertheless, the sharpness of the retinal layers is clearly improved, as presumably is the precision of the delineation of these boundaries with automated analysis. It is in the appearance of non-layered fine structures that the advantage of tracking is immediately apparent in the co-added cross-sectional images. Blood vessel shadows (caused by the relatively high light extinction by blood at these wavelengths) show good contrast and are precisely aligned. Details of the vessel cross-section, including lumen, emerge with tracking. Fine structure within the choroid is evident in the tracking average. The limitation in image detail is determined by pixellation rather than motional broadening.

Figure 3 shows a set of three single OCT radial disc scans for one orientation (of six) acquired on three successive visits with (left column) and without tracking (right column). The average OCT image is shown in the final frame of each column (Fig. 3j and t). With tracking, the morphology of the disc and cup appear to be quite stable. Without tracking, not only is the in-plane transverse alignment and distortion problematic, but out-of-plane motion increases the probability that a different section of the disc is scanned. These motions are uncorrectable by post-processing, and further distort the image when executing the vertical alignment software by
coupling transverse alignment error into vertical distortion. The average radial scans with tracking do not merely preserve the disc shape of the individual scans, they actually refine it. With the custom vertical alignment software, small slopes or vertical variations in the nerve head contour due to depth motion or vibration uncorrected by the transverse retinal tracker tend to average away. The resulting average is asymptotically a truer representation of disc and cup geometry. Without tracking, the average, while smooth, is without diagnostic significance. The mean variance in disc cup edge position for the individual visits for this subject, a good fixator, was 0.35 pixels with tracking and 1.22 pixels without tracking. One of the significant improvements expected from TOCT is the improved ability to measure the same retinal location every time a patient is examined to monitor longitudinal progression of a disease or defect. To test the hypothesis that tracking improved the ability to return to a fixed retinal position, the variance in disc cup edge position for multiple visits was examined and found to increase to 0.45 pixels with tracking and 3.76 pixels without tracking.

In the Stratus_OCT system, there is significant improvement in both transverse and longitudinal OCT image resolution. This will inevitably yield higher precision of thickness maps, and generally superior image quality. With the additional capabilities provided by TOCT, retinal boundaries will become even better defined by averaging because high image spatial frequencies are preserved. Direct, high resolution 2-D and 3-D imaging of the disc and other retinal structures and pathologies with steeper vertical features will benefit even more dramatically from retinal tracking.

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Full References


Short References

Figure Captions

Figure 1: Optical schematic of TOCT. Three-dimensional layout flattened to two dimensions for clarity. D: dichroic beamsplitter, L1-3: fiber collimating lenses, TG: tracking galvanometer-driven mirrors, OCT G: OCT beam galvanometer-driven mirrors, DS: dither resonant scanners, S1: split aperture.

Figure 2: Composite circle disc OCT images co-added from three visits of 20 scans each. a) Fundus image (single frame). b) With tracking. c) Without tracking. N: nasal, T: temporal, S: superior, I: inferior. Scale bar is 1.0 mm.

Figure 3: Radial disc OCT scans from one subject on three successive visits. a-i) Scans taken with tracking. j) Composite image created from scans a-i. k-s) Scans taken without tracking. t) Composite image created from scans k-s. V: visit #, R: run #. Scale bar is 0.5 mm.
Figure 1: Ferguson et al.
Figure 2: Ferguson et al.
Figure 3: Ferguson et al.