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Motion-compensated Optical Coherence Tomography based on higher-order regression for real-time volumetric imaging

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Optical Coherence Tomography (OCT) has evolved into a powerful clinical tool, with a wide range of applications in ophthalmology. However, most OCT systems for real-time volumetric (3D) and *in vivo* imaging suffer from image distortion due to motion artifacts induced by involuntary and physiological movements of the living tissue. Several methods have been proposed to obtain motion-free images, yet they are generally limited in their applicability due to long acquisition times, requiring multiple volumes [1], and/or the need for additional hardware [2]. Here we propose and analyze a motion-compensated 3D-OCT imaging system that uses a higher-order regression analysis and show that it can effectively correct the motion artifacts within 0 to 5 Hz in real time without requiring additional hardware.

Method

The scanning protocol of the proposed motion-compensated 3D-OCT system is shown in Figure 1. Three reference images in the slow-axis are utilized to obtain reference images within the C-Scan volume and compensate for the motion artifacts. Note that mainly axial motion is considered in this analysis, such as the one induced by involuntary vascular pulsation and respiration. We define d as the distance between adjacent references. L is the width of the B-scan raw image. α is defined as the ratio between d and L . The system is implemented in an in-house built 100kHz swept-source OCT centered at 1060nm with 110 nm tuning range.

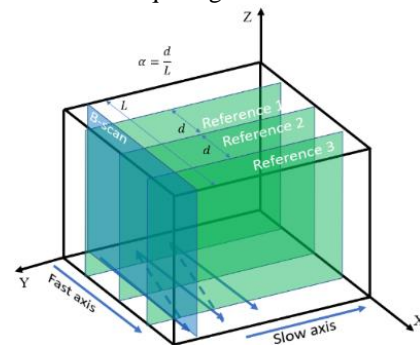


Figure 1. Schematic of the scanning protocol of the motion-compensated 3D-OCT system.

The flowchart of the system algorithm is shown in Figure 2. The algorithm consists of two main steps, namely pre-processing and motion compensation. First, we apply a gaussian filter to every B-Scan of the raw C-scan data and remove the saturation artifacts introduced by light reflected from a highly specular surface generating signals that are over the dynamic range of the data-acquisition system. We simplified the saturation removal method based on minimum variance mean-line subtraction (MVMS) [3, 4], to reduce the computation cost. Specifically, the A-line value of position (x, y) will be assigned as subtraction of the mean value of raw data $A_{\Omega}(x, y)$ from $A_{\Omega}(x, y)$. After thresholding and the connect component analysis, we remove any noisy portion, including tissue inside the cornea, then generate the initial segmentation. We revise the surface using the pixel with the largest gradient around the initial segmentation and fit it with a polynomial curve. To test this pre-processing algorithm, we applied it to clinical OCT images of corneas with normal and abnormal surface topography (e.g. from bullous keratopathy and keratoconus). As shown in Figure 3, the results indicate that the 4th order polynomial is accurate enough to represent the topological surface of both the healthy and pathological corneas.

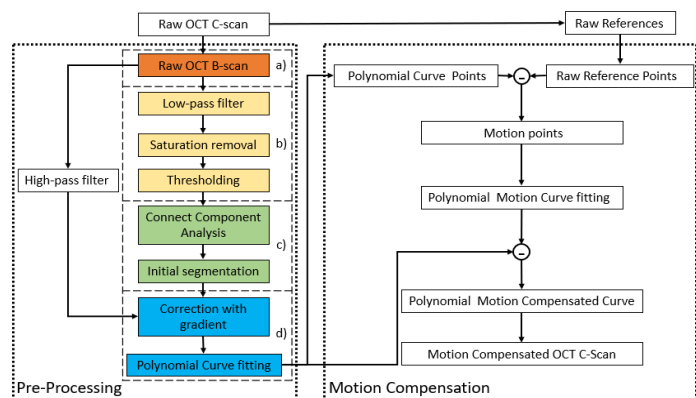


Figure 2 Flowchart of system algorithm, with (left) pre-processing (a through d) and (right) motion-compensation steps.

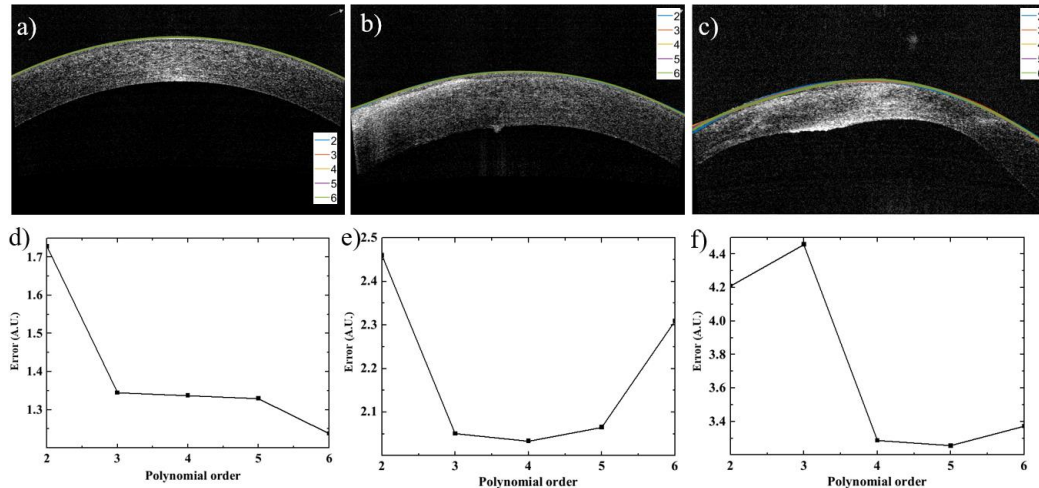


Figure 3. Polynomial fitting results of corneal boundary. a), b), and c) shows the different polynomial order fitting results and corresponding fitting error d), e) and f) of the normal cornea, cornea with bullous keratopathy, and cornea with keratoconus, respectively.

We compare the B-scan surface detection result with the three reference images and obtain the motion during the three intersecting planes. Then, the movement of every A-scan within the B-scan is predicted and compensated by the polynomial fitting.

Results

Because most involuntary motion frequency of the human body ranges from 0 to 5Hz [5], we explored the movement over this frequency range in this study. Figure 4. a)-c) shows an example of the motion-compensation result during corneal C-scan volume imaging with axial movement around 1Hz. As another example, Figure 4. d)-f) shows volumetric imaging of a phantom with motion around 3.6Hz. The algorithm successfully corrects the motion artifacts in these two frequency settings.

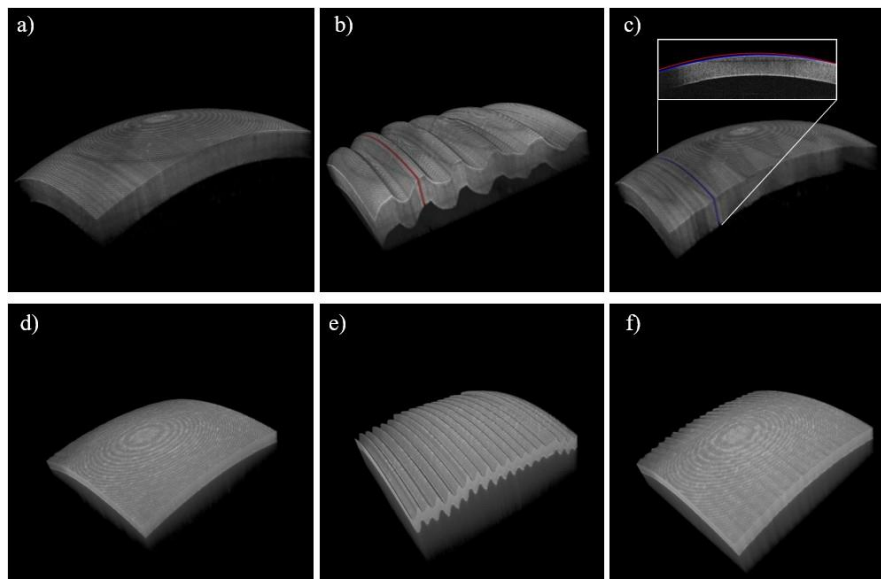


Figure 4. Motion-compensation result of (top) cornea and (bottom) phantom C-scan volume. a) is the static reference of the cornea, b) is the cornea with motion around 1Hz. c) is the reconstructed corneal image after motion compensation. The red line in b) and blue line in c) represents the B-scan location, representing the corneal surface before and after compensation, respectively. Both are also shown in the inset of c), which provides a direct comparison between pre- and post-compensation. d), e) and f) are the static reference of the phantom, phantom with motion around 3.6Hz, and reconstruction of phantom after motion compensation.

The Root Mean Square Error (RMSE) is used as our quantitative metric of the result, to measure the difference between the reference and compensated sample in micrometers. To test the performance of the system and the algorithm, we compensated for the motion around 1 Hz using different order polynomial fittings with four cornea samples as shown in Figure 5. The result shows that 4th order fitting achieves the best fitting accuracy, with minimal RMSE.

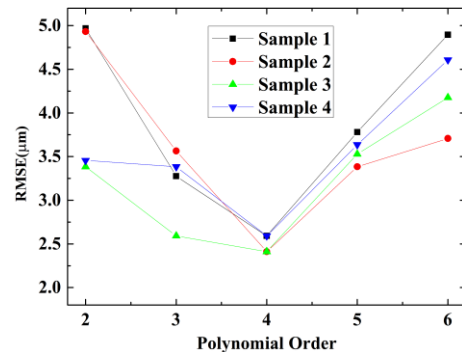


Figure 5. RMSE between four compensated cornea samples and the static reference with different polynomial orders.

The compensation result could be affected by the distance between the adjacent reference planes, which is defined as α . Since the middle reference is fixed in the center of the x-direction, α can range from 0 to 0.5. Figure 6 shows that α around 0.37 achieves the best result, that is, a minimal RMSE, below around $2.6 \mu\text{m}$.

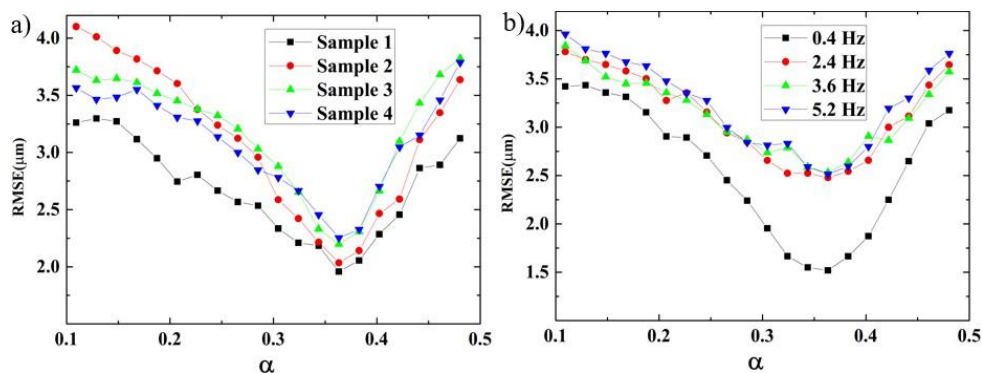


Figure 6. Error between motion compensation and static reference as a function of α for 4 different a) corneas with 1 Hz axial motion and b) phantoms with diverse motion frequencies.

In conclusion, we propose and analyze a motion-compensated 3D-OCT system and show that it can compensate motion over the frequency range from 0 to 5 Hz effectively with less than $2.6 \mu\text{m}$ error. Implementing such motion compensation into clinical OCT systems has the potential to improve the reliability of 3D imaging and reduce associated computational cost.

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