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A Surface Topology and Motion Compensation System for Microsurgery Guidance and Intervention based on Common-Path Optical Coherence Tomography

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Abstract

A surface topology and motion compensation system for microsurgery guidance and intervention is developed based on common-path optical coherence tomography. One-dimensional erosion based edge-searching method and autoregressive predictor is applied to A-scan data for real-time depth-tracking. Images using the topological and motional compensation technique are obtained. In addition, the motion compensation properties are studied. The system can be easily integrated with microsurgery tools and can be used for various clinical applications.

Index Terms

Surgical guidance and intervention; Optical coherence tomography; Surface topology and motion compensation

I. INTRODUCTION

Microsurgeries involving blood vessels, nerves and tissue reparations are traditionally guided by surgical microscopes, which limits the surgeon's operation view to the enface level. As a new method of microsurgical guidance, optical coherence tomography (OCT) has been proposed, which is a noninvasive biomedical imaging modality capable of providing real-time subsurface 3D intraoperative imaging with ultra-high resolution in the order of a few microns. This can lead to less iatrogenic injury and better patient outcome [1]. Compared to other image-guiding modalities such as MRI, CT and ultrasound, OCT is inexpensive, compact, portable, and can be integrated with many kinds of surgical tools. Various OCT guided neurosurgical tools have been tried in small rodents by Jafri et. al [2]. However, most OCT systems generally suffer from limited imaging depth range in the order of only 1~3mm, which restricts its clinical applications when a tissue surface's topological variance is larger than the imaging depth range [3]. Moreover, involuntary beat and motion of subject tissues may cause OCT imaging artifacts [4] and can misguide surgical procedures. This is especially critical in surface operations such as cerebral cortex neurosurgery [5] and retina vitreous surgery [6]. In these operations, tissue's axial involuntary motion is one of the primary concerns and high dexterity from an experienced surgeon is required since beating or motion even in hundreds of microns can cause serious complications. A simple and efficient way to deal with such issues is to use an adaptive ranging technique based on depth-tracking. This is achieved by first locating the subject tissue surface using an OCT system and then using this information to adjust the coherence gate and range on the reference arm [7], [8]. The topological variance and motion is thus compensated and the image is obtained on a virtually "plain and static" surface. Motion compensation of surgical

tools has been also achieved in ultra-sound guided beating heart mitral valve surgery, which showed the benefits of requiring less dexterity and applied force compared to using solid tools [9], [10].

Common-path OCT (CPOCT) is a rational and reasonable approach for topology and motion compensation because the reference and sample signals share the same path [11], [12] thus the reference offset can be changed directly by adjusting the distance between the probe and the tissue surface. No additional synchronization control is needed in the reference arm as in prior work [7], [8]. In addition, the CPOCT approach requires no alignment and has higher imaging stability [13]; it is relatively insensitive to vibration as well as fiber induced polarization and dispersion mismatch [14]. These properties logically predict ideal features when implementing an all-fiber CPOCT probe for integration with different microsurgical tools, which then can be used to adjust the tools axial motion in real time to compensate for the topological variance and motion of the tissue.

In this work, we developed a CPOCT system capable of surface topology and motion compensation (STMC) in axial direction. In the CPOCT-STMC system, one-dimensional erosion based edge-searching algorithm and autoregressive predictive filter are applied to A-scan data for real-time depth-tracking. Images were obtained while utilizing the topological and motional compensation technique. In addition, the motion compensation properties was studied, which predicts high feasibility of better STMC performance.

II. CPOCT-STMC Methods

A. CPOCT-STMC Theoretical Overview

Fig. 1 shows the schematic of the CPOCT-STMC system and the two basic functions of the system: (a) Topological compensation, when the tissue's surface varies larger than the OCT imaging depth, and (b) Motion compensation, when the tissue has involuntary motion (here we mainly consider the axial motion). A CPOCT fiber-optic probe is integrated into a surgical tool and both are driven together by a motorized actuator. Each set of A-scan data obtained from the probe is used as an input to the CPOCT-STMC system and analyzed to locate the target surface. Subsequently the system sends out the feedback control signal to the actuator for the position correction.

Fig. 2(a) shows A-scan images obtained by scanning a live fish eye submerged in the water. Note that there is a backscattering peak caused by suspending particles between the probe and tissue surface, which occurs occasionally but may lead to misjudgment of the edge location. To get rid of such "ghost edge", 1-D erosion algorithm is applied to the raw data first, as shown in Fig. 2(b), which is defined as [15]:

$$(f \ominus b)(s) = \min \{f(s+x) - b(x) | (s+x) \in D_f, x \in D_b\}. \quad (1)$$

Where f is the raw A-scan data, b is the local-minimum operator. D_f and D_b are the domains of f and b . In the simplest case, b is flat structuring element which equals on D_b , so (1) can be simplified as:

$$(f \ominus b)(s) = \min \{f(s+x) | (s+x) \in D_f, x \in D_b\}. \quad (2)$$

Here we set D_b between the widths of “ghost edge” (typically $\sim 15\mu\text{m}$) and the real edge. Thus the “ghost edge” can be erased and real edge remains after erosion operation. Then the new A-scan data is processed by using a certain threshold level to avoid the noise effect, as in Fig. 2 (c). Finally the edge position is given by the first zero-crossing of the differential of the post-thresholding data, as shown in Fig. 2(d). The surface location algorithm in [7], [8] is based on the first and second moment calculation of the A-scan data, which depends largely on the gain factor distribution inside the sample and thus, cannot provide the accurate surface position. Compared to moment calculation, the edge-searching method provides highly accurate surface location.

The CPOCT-STMC system depth-tracking flowchart is shown in Fig. 3. In the experiment, the probe was required to maintain a fixed distance D from the sample surface. This was set $D=200\mu\text{m}$. Each A-scan is processed by the depth-tracking algorithm to find the actual probe-sample distance, noted as $d[n]$. The new probe position is thus given by

$$x[n+1]=d[n]+x[n]-D \quad (3)$$

and the control system subsequently sends the corresponding controlling command to the actuator to adjust the probe axial position before the next A-scan. To enhance the robust of the depth-tracking and further eliminate rogue jumps caused by the misjudgment of $d[n]$, a temporal comparison critic $\|x[n+1]-x[n]\|<S$ is applied before starting the actuator, where S is a safety distance pre-estimated from the properties of topology and motion as well as the temporal response speed of the STMC system. This was set at $S=100\mu\text{m}$ during our experiment. The probe position for each A-scan is saved and used to reconstruct the correct image from the raw data after a complete B-scan.

To improve the response of the actuator to the rhythmic motion, we applied the k-order autoregressive (AR) predictor to actively track the motion of the tissue target [10]:

$$\begin{aligned} x[n+1] &= \sum_{i=0}^{k-1} \alpha_i (y[n-i]) \\ &\quad - D \\ &= \sum_{i=0}^{k-1} \alpha_i (d[n-i] + x[n-i]) - D \end{aligned} \quad (4)$$

where $y[n-i]=d[n-i]+x[n-i]$ are the tissue target position, α_i are the model coefficients which can be estimated by the least square method from the previous k values of $y[n]$. A buffer sequence is used to store $x[n]+d[n]$, and a certain step number N ($N>k$) is set to be the starting point of the AR tracking. In our experiment, we set $N=300$, $k=250$.

B. CPOCT-STMC Experimental Setup

The schematic of the CPOCT-STMC experimental setup is shown in Fig. 4, which uses a Fourier domain CPOCT system. An SLED (EXS8410-2413) with 840nm central wavelength and $\sim 40\text{nm}$ spectral FWHM is used as the light source, which gives a theoretical in-air resolution of $\sim 8\mu\text{m}$. A 50/50 coupler works as the beam splitter and only one branch on the right side is used as the common path for signal and reference. A right-angle cleaved fiber probe P is fixed on a controllable 3-D moving stage M, with A-scan (axial) in X direction and

B-scan (lateral) in Y direction. All three axis are driven by high precision motorized actuators (Newport 850G). The reference signal comes from the Fresnel reflection at the fiber probe end, and the sample signal and the reference are received by H, a high speed spectrometer (Ocean Optics HR-4000) with a CCD detector array with 3648 pixels covering 699nm to 891nm range. The A-scan signals are processed by a laptop computer which then sends the control signal to M through LABVIEW GPIB interface. The Fast Fourier Transform (FFT) is used to convert A-scan signal from λ -space to k-space.

III. RESULTS AND DISCUSSION

A. Topological Compensation

The topological compensation is tested in a static condition using a phantom sample with 8-layers of highly curved surfaces. At first we obtained a B-scan 2-D image by conventional fixed-reference method, shown as Fig. 5(a). The red arrows indicate the motion of the probe as well as its position. As one can see from the left part of Fig. 5(a), the 840nm Fourier domain CPOCT has an effective working depth of ~ 1 mm and the layer structure on the “hill top” is very clear within this range. However, due to the limited depth scanning range as expected, the OCT image fades away as the probe is moved away from the top. Fig. 5(b) shows an improved image using depth-tracking topology compensation. As shown by the red arrows, the probe follows the falling of the surface as it obtains A-scans. The moving trace of probe is recorded and overlapped on Fig. 5(b) in red line, and the trace is consistent with the surface profile. By using the topology compensation, the probe was able to track the sample surface variance and the effective imaging depth was largely extended to the probe’s free-moving range.

B. Motion Compensation

Then we tested the motion compensation properties of the CPOCT-SMTC system. First, we tested the SMTC temporal response by tracking the surface of a galvanometer-driven phantom sample, which moves back and forth in axial direction periodically at the amplitude of $110\mu\text{m}$. Fig. 6(a) shows the real-time response of CPOCT-SMTC to the sample motion directly using (3) without predictor. The period of target motion is 5 second, and as one can see, due to limited actuator speed ($500\mu\text{m/s}$), there is phase delay of approximately 20 degrees between the response and the motion. In contrast, Fig. 6(b) shows the CPOCT-SMTC response with the AR predictor using (4). The phase delay is significantly decreased to about 5 degrees. Fig. 6(c) shows that the increasing phase difference between the response and the motion with increased frequency and that the AR predictor can largely decrease the phase delay. The phase difference tends to go to zero as the motion frequency is decreased, therefore the motion compensation can be further improved by using a faster motorized actuator to effectively compensate for the faster and more complex real tissue motion as in [7]–[10]. Fig. 6(d) shows the response-to-motion amplitude ratio as a function of frequency. The response amplitude was almost the same for the both cases.

The phase difference effect on the image quality is shown as Fig. 7. A 3-layer flat phantom sample is first B-scanned by a traditional OCT system when the phantom is either static or moving with 10 second period, respectively as Fig. 7(b) and (c). As we can see, there is a large fluctuation artifact on the image even when the sample is flat. We then applied the STMC technique to re-image the moving phantom as Fig. 7(d) and (e), and the artifact can be significantly decreased. By comparing Fig. 7(d) and (e), we can see that using the AR predictor can bring a huge improvement in motional depth-tracking and this agrees well with the data shown in Fig. 6(c).

IV. Conclusion and future work

In this work, we developed a CPOCT-STMC system for applications in microsurgical guidance and intervention. A 1-D erosion based edge-searching algorithm and autoregressive predictor were applied to A-scan data for real-time accurate depth-tracking. OCT images were obtained while utilizing the topological and motion compensation technique which shows a significant improvement in the imaging capability. The all-fiber CPOCT-STMC system was shown to be stable, compact, and can be integrated into microsurgery tools with relative technical ease. Although limited by the speed of the motorized actuator used in the experiment, the CPOCT-STMC method demonstrates high potential and feasibility for the further development. In our future work, we plan to use faster actuators and a higher speed OCT system to achieve better STMC performance.

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Fig. 1. Schematic of CPOCT-STMC system. (a) Topological compensation. (b) Motion compensation.

Fig. 2. Edge-searching method. (a) Raw A-scan data, "Ghost edge" exists. (b) A-scan data after 1-D erosion algorithm, "Ghost Edge" is erased and real edge remains. (c) Thresholding of A-scan. (d) First zero-crossing for edge location (depth scale changed).

Fig. 3. CPOCT-STMC system flowchart. ASA: A-scan acquisition; SR: surface recognition; AR: autoregressive predictive filter; PC: probe control.

Fig. 4.
CPOCT-STMC experimental setup.

Fig. 5. Images of a phantom sample by CPOCT-STMC system. (a) Traditional OCT with limited imaging depth. (b) Topological compensation with extended imaging depth.

Fig. 6. Motion compensation properties of the CPOCT-SMTC system. (a) Response of CPOCT-STMC to a target motion without predictor. (b) Response of CPOCT-STMC to a target motion with AR predictor. (c) Phase difference between response and motion increases with frequency. (d) Response-to-motion amplitude ratio with frequency changing.

Fig. 7. Motion compensation of a 3-layer flat phantom sample by CPOCT-STMC system. (a) Experiment details. M: moving stage with motorized actuator; P: fiber CPOCT probe; (b) Traditional B-scan image of static phantom. (c) Traditional B-scan image of periodically moving phantom without depth tracking. (d) STMC B-scan image of periodically moving phantom without predictor. (e) STMC B-scan image of periodically moving phantom with AR predictor.