Published in final edited form as:

Proc SPIE Int Soc Opt Eng. 2019 February ; 10880: . doi:10.1117/12.2503809.

Adaptive Doppler analysis for robust handheld optical coherence elastography

Farzana Zaki, Yahui Wang, Chizhong Wang, and Xuan Liu

Dept. of Electrical and Computer Engineering, New Jersey Institute of Technology, University Heights, Newark, New Jersey, USA, 07102

Abstract

Optical coherence tomography (OCT) allows structural and functional imaging of biological tissue with high resolution and high speed. Optical coherence elastography (OCE), a functional extension of OCT, has been used to perform mechanical characterization. A handheld fiber-optic OCE instrument allows high sensitivity virtual palpation of tissue with great convenience and flexibility and can be used in a wide range of clinical settings. Moreover, fiber-optic OCE instruments can be integrated into a needle device to access deep tissue. However, the major challenge in the development of handheld OCE instrument is non-constant motion within the tissue. In this study, a simple and effective method for temporally and spatially adaptive Doppler analysis is investigated. The adaptive Doppler analysis method strategically chooses the time interval (δt) between signals involved in Doppler analysis, to track the motion speed $v(z, t)$ that varies as time (t) and depth (z) in a deformed sample volume under manual compression. The aim is to use an optimal time interval to achieve a large yet artifact free Doppler phase shift for motion tracking.

Keywords

Optical coherence tomography; Optical sensing and sensors; Tissue characterization; Doppler; Optical coherence elastography

1. INTRODUCTION

Biological tissue has altered mechanical properties under different physiological and pathological status. For example, cancerous tissue has different stiffness compared to surrounding normal tissue. Conventionally, diagnosis of cancerous tissues through manual palpitation provides a qualitative assessment of tissue mechanics. In recent decades, several elastography methods such as ultrasound elastography and magnetic resonance imaging (MRI) elastography [1–4] have been used to quantify the tissue properties in macroscopic scale. However, deformation of the mechanical properties of the cancerous tissue occurs in the microscopic scales. Therefore, early detection of micro-scale change in tissue mechanical properties will enable early detection and treatment of cancer.

Optical coherence elastography (OCE), based on optical coherence tomography (OCT), is a non-invasive medical imaging platform to assess mechanical characterization of biological tissues in microscale level through the measurement of depth-resolved displacement under

external or internal mechanical excitation. OCE offers high spatial resolution and high speed elastography measurement and is a promising technology for the clinical diagnosis and surgical guidance, due to its sensitivity to detect the sample displacements as small as tens of picometres [5–6].

In compression OCE, the displacement of the sample at different depth is measured to quantify the deformation and to derive sample mechanical properties. For the sample under compression, the displacement at different depth (z) was tracked by performing Doppler analysis on the complex valued Ascan OCT signal ($I(z, t)$) (Eq. (1)). In Eq. (1), z represents the axial coordinate, t denotes a time interval between Ascan signals involved in Doppler analysis, $\delta\phi(z, t)$ is the phase shift between two Ascans respectively acquired at time t and $t + \Delta t$, and Im, Re, and $*$ indicates to take the imaginary part, real part and complex conjugate of a complex signal.

$$\delta\phi(z, t) = \tan^{-1} \{ \text{Im} [I^*(z, t)I(z, t + \Delta t)] / \text{Re} [I^*(z, t)I(z, t + \Delta t)] \} \quad (1)$$

An estimation of displacement ($\delta L(z)$) within time interval Δt can be calculated by using Eq. (2) where λ_0 is the central wavelength of the light source used in OCT imaging. With the displacement, the deformation of the sample can be quantified using the strain (Eq. (3)).

$$\delta L(z, t) = \lambda_0 \delta\phi(z, t) / (4\pi) \quad (2)$$

$$\epsilon(z, t) = d\delta L(z, t) / dz \quad (3)$$

On the other hand, analysis by Yazdanfar *et al* suggests that the variance of Doppler phase ($\text{Var}(\delta\phi)$) is inversely proportional to the signal to noise ratio (SNR) of the OCT signal (magnitude) as shown in Eq. (4) [7].

$$\text{Var}(\delta\phi) \propto (\text{SNR})^{-1/2} \quad (4)$$

Within a very small time interval (Δt), the incremental displacement is small. Hence the Doppler signal is overwhelmed by noise. When the time interval Δt is too large, the phase shift between two Ascans can be larger than 2π , causing phase wrapping artifact and false estimation of the displacement [9]. In summary, displacement tracking based on Doppler analysis often suffers from either low SNR value or phase wrapping artifact, limiting its capability in accurately characterizing the mechanical properties of biological tissue.

In this work, we developed an adaptive Doppler tracking strategy by adaptively selecting optimal time interval (Δt) between Ascans involved in Doppler analysis (Eq. (1)). Imaging

experiments were conducted on homogenous and heterogeneous structured PDMS phantoms.

2. EXPERIMENTAL SETUP AND ALGORITHM

A spectral-domain Doppler OCT system (Figure 1) was used for all the measurements and the detailed description of the system can be found in our previous publication [8]. To be brief, the light source of our system is a superluminescent diode (SLD1325 Thorlabs, 100 nm bandwidth, corresponding to a 7.4 μm axial resolution) and we use a CMOS InGaAs camera (SUI1024LDH2, Goodrich) to capture the interference signal. By using a fiber-optic coupler (FC), the output of the light source can illuminate the reference and sample arm of a fiber-optic Michelson interferometer. To perform compression OCE measurement, we sandwiched the sample between two rigid plates. One of the plates was a window that allowed optical access of the sample, and the other plate was mounted to a motor (Thorlabs Z825B).

We fabricated polydimethylsiloxane (PDMS) samples to evaluate the adaptive Doppler tracking method. We combined PDMS base with curing agent at different volumetric ratios (20:1 or 10:1) to achieve different stiffness. In addition, we added titanium dioxide to provide light scattering. The OCT image of a PDMS phantom is shown in Figure 2. The phantom has two layers with different stiffness (with 20:1 and 10:1 base to agent ratio).

Doppler OCT signals were obtained from sample under compression. To obtain the optimum time interval at a specific depth for Doppler analysis, we first computed the n Doppler phase shifts ($\delta\phi_1, \delta\phi_2, \delta\phi_3, \dots, \delta\phi_n$) using n pairs of Ascans signals ($I(z,t)$ and $I(z,t+\delta t)$, $I(z,t)$ and $I(z,t+2\delta t)$, $I(z,t)$ and $I(z,t+3\delta t)$, ..., $I(z,t)$ and $I(z,t+n\delta t)$). Here δt indicates the time interval between the acquisition of adjacent Ascans and is determined by the camera used in the OCT system. SNR kept increasing with increasing time interval until the phase wrapping occurred. Therefore, we determined whether there was phase wrapping artifact for $\delta\phi_1, \delta\phi_2, \delta\phi_3, \dots, \delta\phi_n$. If no phase wrapping was detected in all the phases, we used $\delta\phi_n$ to calculate the average displacement within a time interval $t_{\text{opt}}=n\delta t$. Otherwise, we identified the smallest time interval ($n_{\text{wrap}}\delta t$) that caused phase wrapping, and chose $t_{\text{opt}}=\alpha n_{\text{wrap}}\delta t$ where α is a constant (α).

3. RESULTS

We acquired a frame of complex OCT data with 1024 Ascans. For each Ascan, we calculated Doppler phase shift between this Ascan and another Ascan with a time interval of $\delta t, 7\delta t$ and $14\delta t$. The results shown in Figure 3 suggest that the Doppler signals obtained with different t have similar level of noise and the mean Doppler signal increases as t .

We further demonstrated how the choice of time interval affects the quality of Doppler signal for OCE measurement. Figure 4 shows the mean Doppler phase for different depths. For both curves, the Doppler phase increases with the time interval, and then the phase wrapping occurs and results in decreased Doppler phase, because the estimation of Doppler phase shift (Eq. (1)) uses an inverse tangent function that outputs value between $-\pi$ to $+\pi$. When the time interval t is too large and the phase shift becomes larger than π , Doppler

analysis cannot provide correct displacement estimation. We performed zero-crossing detection on phase signals shown in Figure 4 and obtained the optimal time interval for Doppler analysis. At a smaller depth, both the displacement and Doppler phase have smaller values. Therefore, it takes a longer time interval for phase wrapping to happen, and t_{opt} should be larger. This further justified the need for an adaptive Doppler tracking method for OCE.

To visualize the quality of OCE image processed by different methods, we compared the Bscan Doppler images of the two-layer phantom (Figure 2). The same raw spectral data were used to obtain Figure 5 (a), (b) and (c). We used an interval of 1 ms to obtain Figure 5 (a), and 30 ms obtain Figure 5 (b). The adaptive Doppler tracking was implemented to obtain Figure 5 (c). In Figure 5, motion along +z direction is coded as blue and along -z direction is coded as red. As shown in Figure 5 (a), with small time interval, the Doppler image is overwhelmed by random noise. With large time interval, phase wrapping occurs and leads to incorrect estimation of the motion along -z direction (Figure 5 (b)). In comparison, adaptive Doppler tracking significantly improved the image quality. In Figure 5 (c), two distinctive layers with different stiffness are clearly discernable. The adaptive algorithm chooses the optimal time intervals for each sample depth to ensure the highest SNR for the corresponding sample depth.

4. CONCLUSION

In this paper, we describe an adaptive Doppler tracking method for OCE. We adaptively chose an optimum time interval t for every depth in Doppler analysis to achieve high SNR signal without phase wrapping artifact. The method was validated using experimental data.

ACKNOWLEDGEMENT

This work is partially supported by fund from National Institutes of Health (NIH) (1R15CA213092-01A1).

REFERENCES

- [1]. Ophir J, Cespedes I, Garra B, Ponnekanti H, Huang Y, and Maklad N, "Elastography: ultrasonic imaging of tissue strain and elastic modulus in vivo," *Eur. J. Ultrasound* 3, 49–70 (1996).
- [2]. Baghani A, Eskandari H, Salcudean S, and Rohling R, "Measurement of viscoelastic properties of tissue-mimicking material using longitudinal wave excitation," *IEEE Trans. Ultrason. Ferroelect. Frequency Control* 56(7), 1405–1418 (2009).
- [3]. Sinkus R, Tanter M, Catheline S, Lorenzen J, Kuhl C, and Sondermann E, "Imaging anisotropic and viscous properties of breast tissue by magnetic resonance-elastography," *Magn. Reson. Med* 53, 372–387 (2005). [PubMed: 15678538]
- [4]. Ieva AD, Grizzi F, Rognone E, Tse ZTH, Parittotokkaporn T, Baena F. R. y., Tschabitscher M, Matula C, Trattig S, and Baena R. R. y., "Magnetic resonance elastography: a general overview of its current and future applications in brain imaging," *Neurosurg. Rev* 33, 137–145 (2010). [PubMed: 20195674]
- [5]. Wang RK, Ma Z & Kirkpatrick SJ Tissue Doppler optical coherence elastography for real time strain rate and strain mapping of soft tissue. *Appl. Phys. Lett* 89, 144103 (2006).
- [6]. Kennedy KM, Chin L, McLaughlin RA, Latham B, Saunders CM, Sampson DD, and Kennedy BF, "Quantitative micro-elastography: imaging of tissue elasticity using compression optical coherence elastography," *Sci. Rep* 5, 15538 (2015). [PubMed: 26503225]

- [7]. Yazdanfar S, Yang C, Sarunic M, and Izatt J, "Frequency estimation precision in Doppler optical coherence tomography using the Cramer-Rao lower bound," *Opt. Express* 13(2), 410–416 (2005). [PubMed: 19488367]
- [8]. Qiu Y, Wang Y, Xu Y, Chandra N, Haorah J, Hubbi B, Pfister BJ, and Liu X, "Quantitative optical coherence elastography based on fiber-optic probe for in situ measurement of tissue mechanical properties," *Biomed. Opt. Express* 7(2), 688–700 (2016). [PubMed: 26977372]
- [9]. Liu Xuan, Zaki Farzana R., Wu Haokun, Wang Chizhong, and Wang Yahui, "Temporally and spatially adaptive Doppler analysis for robust handheld optical coherence elastography," *Biomed. Opt. Express* 9, 3335–3353 (2018). [PubMed: 29984101]

Figure 1.
OCT system used for adaptive Doppler algorithm scheme

Figure 2. OCT image (magnitude) obtained from a two-layer scattering phantom. Scale bars indicate 250 μm .

Figure 3.
Comparison of Doppler phases at different time intervals

Figure 4.

Doppler phase at different depths using different time intervals for Doppler analysis.

Figure 5:

Comparison of Bscan Doppler images of two-layer scattering phantom at different time intervals: (a) a small time interval ($t = 1\text{ms}$), (b) a large time interval ($t = 30\text{ms}$) and (c) an adaptive optimal time intervals. Scale bars represent $250\mu\text{m}$.