Noncontact quantitative biomechanical characterization of cardiac muscle using shear wave imaging optical coherence tomography

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Abstract: We report on a quantitative optical elastographic method based on shear wave imaging optical coherence tomography (SWI-OCT) for biomechanical characterization of cardiac muscle through noncontact elasticity measurement. The SWI-OCT system employs a focused air-puff device for localized loading of the cardiac muscle and utilizes phase-sensitive OCT to monitor the induced tissue deformation. Phase information from the optical interferometry is used to reconstruct 2-D depth-resolved shear wave propagation inside the muscle tissue. Cross-correlation of the displacement profiles at various spatial locations in the propagation direction is applied to measure the group velocity of the shear waves, based on which the Young’s modulus of tissue is quantified. The quantitative feature and measurement accuracy of this method is demonstrated from the experiments on tissue-mimicking phantoms with the verification using uniaxial compression test. The experiments are performed on ex vivo cardiac muscle tissue from mice with normal and genetically altered myocardium. Our results indicate this optical elastographic technique is useful as a noncontact tool to assist the cardiac muscle studies.

OCIS codes: (170.6935) Tissue characterization; (170.4500) Optical coherence tomography.

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1. Introduction

Heart failure after ischemic heart disease is a leading cause of death in the United States [1]. The major limitation for the recovery from the ischemic heart disease is the poor regeneration of cardiomyocytes [2]. A number of research groups have been working on the understanding of cardiomyosite renewal mechanisms and developing of cardiac regeneration strategies in transgenic mouse models [3, 4]. Since the heart is a dynamic organ with highly regulated function, noninvasive assessment of the localized biomechanical properties of mouse cardiac muscle is required to provide useful information regarding the potential mechanical function of the heart [5, 6] and thus can be an essential tool for such studies.

Atomic force microscopy (AFM) has been widely used to probe the elasticity of single cells at nano-scale spatial resolution [7], and the quantitative elastic moduli of isolated cardiac cells are available through force spectroscopic analysis [8]. However, the AFM imaging protocol lacks the capability to conduct measurements in situ or in vivo. Elastographic techniques based on ultrasonic imaging and magnetic resonance imaging are also well established for in vivo detection of tissue mechanical properties [9, 10]. Based on the monitoring of shear wave propagation inside tissue, the quantification of both elasticity and
viscosity can be performed with in-depth resolution. Though, the limited spatial resolution of these two imaging modalities restricts the elastographic measurement to the organ level. Thus, methods that enable biomechanical measurement at the micro scale (single or group of a few cells) with potential to perform in vivo measurements are of great importance and could lead to more thorough and direct analysis of normal, diseased, and regenerated cardiac muscle tissues and help in guiding therapies targeted into restoring cardiomyocyte activity after, e.g., myocardium infarction.

Optical coherence elastography (OCE) is an emerging technique that provides qualitative and quantitative noninvasive imaging of tissue elasticity [11–16], which can potentially meet this demand. Reinforced from the high spatial resolution of optical coherence tomography (OCT), OCE is able to provide micro-scale depth-resolved mapping of tissue elasticity [17–20]. Generally, OCE relies on dynamic or static/quasi-static loading methods to induce tissue deformation and OCT-based displacement-sensing techniques to detect the tissue response [21–23]. The tissue mechanical contrast can be obtained from the strain-related parameters [24–26], the information of resonant frequency [13, 27], and also the propagation of the generated tissue deformation [28, 29]. Quantitative assessment of the elastic modulus of tissue has been demonstrated based on the use of OCE to monitor the propagation of the induced elastic waves [30–32] in cornea [33], skin [34, 35] and soft-tissue tumor [36].

Recently, Song et al. [31] and our group [37] have developed OCE imaging method that enables depth-resolved visualization of the elastic wave at an ultra-fast frame rate. Here, we present the further development of the shear wave imaging OCT (SWI-OCT) [37] for noncontact quantitative biomechanical characterization of mouse cardiac tissues. The group velocity of shear wave is calculated based on the cross-correlation of the tissue temporal displacement profiles at various locations in the wave propagation direction. The quantitative elasticity of sample is achieved based on the shear wave velocity through Kelvin-Voigt model. Results (Young’s modulus) from the experiments on tissue-mimicking phantoms are verified with uniaxial compression tests, which indicate the measurement accuracy of our quantitative SWI-OCT method. The feasibility of using this elastographic technique as a tool in cardiac research is demonstrated by analyzing control and mutant mouse hearts (CKO Lats1/2) with genetically altered cardiomyocytes that are expected to have lower stiffness of myocardium [3]. The results suggest that this quantitative SWI-OCT method holds great potential for in situ noncontact biomechanical assessment of live heart muscle tissue during, e.g., open-chest surgery, and can greatly assist in cardiac tissue research, specifically in cardiac regeneration studies.

2. Materials and methods

2.1 Tissue-mimicking phantoms

Tissue-mimicking phantoms with known stiffness are used in this study to demonstrate the feasibility of using SWI-OCT to quantitatively measure the elasticity of soft samples. Phantoms with three different concentrations (0.50%, 0.75%, and 1.00%, weight/weight) of agar powder are made with the thickness of ~10 mm. For each concentration, three phantoms are prepared and utilized in the experiments for statistical purpose.

2.2 Ex vivo mouse cardiac muscle

To study the feasibility of using SWI-OCT for noncontact quantitative characterization of the mechanical properties of mouse cardiac muscle tissues, we performed measurements on mutant mouse hearts (CKO Lats1/2) with genetically altered cardiomyocytes, which are expected to have lower stiffness. Specifically, eight weeks to 4 months old animals of the conditional knockout (CKO) line of Lats1/2 (Lats1/2F/F; Myh6-cre/Esr1*) [3] are used. On days 4 and 5 before the experiments, 100 µl of 10 mg/ml tamoxifen (Sigma) in peanut oil (Sigma) is injected in each animal to activate Cre activity. Peanut oil injected animals with the
same genotype are used as control. To eliminate the effect of the tissue contraction on
elastographic characterization, right before the measurements, all the animals are perfused
with 30 mM potassium chloride to ensure cardiac tissue relaxation. Freshly dissected pieces
of the myocardium from the left ventricular wall are used for the SWI-OCT imaging. The
samples were dissected as cuboids with the length of ~10 mm, the width of ~6 mm and the
thickness of ~2.5 mm. The number of mice used for the analysis is three and four for the
control group and the CKO Lats1/2 group, respectively. All animal manipulation procedures
have been approved by the Institutional Animal Care and Use Committee of the Baylor
College of Medicine.

2.3 SWI-OCT system

The SWI-OCT system combines a focused air-puff device and Fourier domain OCT [37]. The
schematic of the system is shown in Fig. 1(a). The home-built air-puff device provides a
short-duration (Gaussian shape, 0.8 ms FWHM), low-pressure air stream on the sample
surface with an excitation pressure that can be controlled and estimated based on the loading
parameters [38]. The phase-sensitive OCT system employs a Titanium:Sapphire laser source
(Micra-5, Coherent, Inc.) with the central wavelength of ~808 nm and the bandwidth of ~110
nm. The collimated laser beam is coupled into a single-mode fiber with the setup shown in
Fig. 1(a), which enables adjustable power delivered to the sample arm of the OCT system.
The interference of light from the reference arm and the sample arm through a 50/50 fiber
optic coupler is spectrally resolved through a high-resolution spectrometer with a CMOS
camera (Basler, Inc.) running at 62.5 kHz line rate. This provides the OCT system with the
temporal resolution of 16 μs for M-mode imaging. The axial resolution of the system is
measured to be ~5 μm in tissue, and the FWHM of the OCT beam at focal plane is ~4 μm.
The sensitivity of the system to sample displacement reaches ~11 nm (measured based on the
standard deviation of the phase values over time on ex vivo cardiac muscle tissue) with the
phase-resolved detection that provides sufficient SNR for the monitoring of the induced
sample deformation. Also, this high sensitivity of detection makes it possible to reduce the
required deformation amplitude in the sample, which helps to maintain the structural and
functional properties of the sample during measurement.

2.4 Data acquisition

The SWI-OCT method enables the visualization of the elastic wave propagation at the
equivalent frame rate of the A-line acquisition speed through the synchronization between the
air-puff excitation and the OCT M-mode imaging [37]. Briefly, the OCT beam controlled by a
galvanometer-mirror forms a 1-D scanning line on the tissue surface covering a particular
distance to monitor the elastic wave propagation. At each position of this scanning line, M-
mode imaging is performed to record localized displacement profiles of the sample at all
available imaging depths (up to ~0.7 mm in mouse cardiac muscle tissue). The air-puff
excitation is maintained at a constant position within the scanning line and the opening of the
air gate is synchronized with the start of each M-mode imaging through sharing the same
triggering signal from the computer, as shown in Fig. 1(a). Thus, with respect to the air-puff
loading, the detection of the sample deformation at all scanning positions is performed
simultaneously, which allows depth-resolved 2-D imaging of the wave propagation at the
frame rate of 62.5 kHz. Figure 1(b) shows the typical loading and imaging setup with an OCT
en-face structural image of the mouse cardiac muscle from the left ventricle wall.

During experiments, all the samples are oriented similarly to have the muscle fibers
aligned with the OCT beam scanning line, as shown in Fig. 1(b). The orientation of the
muscle fibers is determined by visual observation under a microscope (as well as clearly seen
from OCT en face images). This procedure minimizes the influence of muscle anisotropy on
calculated values. The direction of the air-puff port is set perpendicular to the direction of the
imaging line and the air-puff delivery angle is kept as ~20° relative to the sample surface.
normal for all measurements. To avoid dehydration of the tissues, at least half of the sample is submerged inside the 30 mM potassium chloride during measurements. The data acquisition with SWI-OCT for each sample takes 30 seconds. The spatial interval between adjacent scanning points is set to be ~12.5 μm.

2.5 Quantification of shear wave velocity and sample elasticity

The reconstruction method for the elastic wave propagation is detailed in our previous work [37], including the unwrapping and alignment of the phase signal, the correction of the phase error caused by the deformation at the sample surface [39], and the masking of the phase image with binarized OCT structural image. For the quantification of the shear wave velocity inside the sample, a 2-D transverse-depthwise region is selected within the sample with ~1 mm away from the air-puff excitation position to avoid the phase artifacts [37] and the possible longitudinal mechanical waves [32] close to the source of deformation. At each depth inside the selected region, temporal displacement profiles at all the transverse spatial locations are cross-correlated with the displacement profile at the position that is nearest to the air-puff excitation point. Through the indexing of the maximum value from the correlation function, the time delay formed at each transverse measurement location during the propagation of the shear wave is obtained. With a linear fit of the data points in the domain of the time delay versus the propagation distance, the group velocity of the shear wave can thus be calculated. The typical temporal displacement profiles from one particular depth in a tissue-mimicking phantom with 0.75% agar concentration is shown in Fig. 2(a), where the time delay can be clearly observed along the shear wave propagation. As an example to show the quantification of velocity, Fig. 2(b) indicates the plot of the data from cross-correlation of the signals in Fig. 2(a). The linear fit results in the group velocity of ~1.9 m/s. Figure 2(c) shows a typical

![Diagram](image-url)
example of cross-correlation of the wave propagation signals from cardiac muscle tissue. In the data analysis for cardiac muscle tissues, we select the 2-D region with the depth range of ~0.3 mm and the transverse range of ~0.9-1.3 mm based on the availability of sufficient SNR. Here, the sufficient SNR is defined when the amplitude of the deformation signal (obtained from the phase-resolved detection) is higher than the phase noise level of the system.

Fig. 2. Quantification of shear wave velocity. (a) The typical temporal displacement profiles from one particular depth in a tissue-mimicking phantom with 0.75% agar concentration. The magnitude of deformation at each spatial location is normalized to the absolute value of its minimum over time. (b) The plot of the data from cross-correlation of the displacement profiles in (a) in the domain of the time delay versus the shear wave propagation distance. Linear fit is applied to the data for the quantification of wave group velocity. (c) A typical example of the cross-correlation of the wave propagation signals from the mouse cardiac muscle tissue.

With the assumption of isotropic homogeneous medium of the sample, and taking use of a Voigt model to consider the shear elasticity, the shear modulus \( \mu \) can be related to the shear wave group velocity \( C \) through the equation [40],

\[
\mu = \rho C^2,
\]

where \( \rho \) is the mass density of the sample. Also, under the same assumption, the Young’s modulus \( E \) and the shear modulus \( \mu \) have the relationship of

\[
E = 2(1 + \nu)\mu,
\]
where \( \nu \) represents the Poisson’s ratio of the sample. Assuming the incompressibility of the soft tissue samples [41], the Poisson’s ratio \( \nu \) is approximately 0.5, which relates the Young’s modulus and the shear wave group velocity through

\[
E = 3\rho C^2.
\]  
(3)

In our quantification, we use 1000 kg/m\(^3\) and 1060 kg/m\(^3\) for the density of the tissue-mimicking phantoms and the mouse cardiac muscle tissues, respectively. Our calculation for the Young’s modulus of the sample is based on the averaged shear wave group velocity from all the involved depths. The statistics shown in the results indicate the mean value and the standard variation among the samples of the same category (agar concentration for phantom and control or mutant group for cardiac muscle).

3. Results

The shear wave velocity values are measured to be 1.01 ± 0.04 m/s, 1.89 ± 0.04 m/s, and 3.13 ± 0.09 m/s for the tissue-mimicking phantoms with the agar concentrations of 0.5%, 0.75%, and 1%, respectively. The quantification of the Young’s moduli of these samples is based on Eq. (3), and the results are presented in Fig. 3. Quantitatively, for the concentrations of 0.5%, 0.75%, and 1%, the Young’s moduli measured from SWI-OCT are 3.1 ± 0.2 kPa 10.7 ± 0.5 kPa, and 29.5 ± 1.6 kPa, respectively, which indicates the increase of the sample stiffness with the increase of the agar concentration. For the verification of these data, we also performed elasticity measurements with uniaxial compression test (using Instron Model 5943) on the tissue-mimicking phantoms with the same concentrations (number of samples \( N = 3 \)). Results show the Young’s moduli of 3.8 ± 0.3 kPa, 11.0 ± 0.3 kPa, and 29.1 ± 0.4 kPa for the 0.5%, 0.75%, and 1% agar phantoms, respectively, as shown in Fig. 3. The good agreement between the SWI-OCT measurements and the uniaxial compression tests demonstrates the feasibility of using SWI-OCT for quantitative assessment of the elasticity of soft samples.

![Graph showing Young's moduli quantified from SWI-OCT for the tissue-mimicking phantoms with 1%, 0.75% and 0.5% agar concentrations, compared with the results from uniaxial compression tests. Number of samples \( N = 3 \) for all measurements.](image)
Fig. 4. The 2-D depth-resolved reconstruction of shear wave propagation in the cardiac muscle tissues from (a) control and (b) CKO Lats1/2 mice (Media 1). The magnitude of deformation at each spatial location is normalized to the absolute value of its minimum over time. Scale bars correspond to 0.5 mm. The red dots represent the air-puff excitation positions on both tissue samples.

The reconstructed shear wave propagation in the typical cardiac muscle tissues from the control and the CKO Lats1/2 mice are shown in Fig. 4(a) and 4(b), respectively, at three representative temporal points. In both samples, the shear waves propagate from the left side to the right side, with the red dots indicating the air-puff excitation positions. The deformation magnitude at each spatial location is normalized to the absolute value of its minimum over time to provide better visualization of the shear wave propagation in the tissues. The 2-D depth-resolved images of tissue displacements are superimposed on the OCT structural images. It can be seen that over the same period of time, the shear wave propagates over a longer distance in the control sample than in the sample with genetically altered cardiomyocytes, indicating a lower shear wave velocity and thus the lower stiffness of the CKO Lats1/2 mutant cardiac tissue.

The plots of the shear wave propagation distance versus the time delay are shown in Fig. 5(a) and 5(b) for the typical cardiac muscle tissues from the control and CKO Lats1/2 mice, respectively. The dashed arrows indicate the directions of the maximum cross-correlation coefficients, showing slower shear wave propagation in the mutant muscle tissue with altered cardiomyocytes. The quantitative results are shown in Fig. 6(a) and 6(b) with the group velocities of the shear waves and the Young’s moduli of the tissues, respectively. The group velocities of the shear waves are measured to be 7.0 ± 2.1 m/s and 2.9 ± 1.3 m/s, and the Young’s moduli are quantified as 165.3 ± 89 kPa and 30.4 ± 22.6 kPa for the left ventricle cardiac muscle tissues from the control and CKO Lats1/2 mice, respectively. The unpaired two-sample student’s t-test applied on the results of Young’s moduli shows statistical significance of the difference in the elasticity of muscle tissues with normal and genetically altered cardiomyocytes. It is well known that the stiffness of the cardiac muscle tissues increases with age [42]. The Lats1/2 deficient cardiomyocytes are thought to have more fetal-like characteristics [3, 4], suggesting lower stiffness of the hearts as compared to the control ones, which is consistent with our experimental results characterized by SWI-OCT. This suggests that the quantitative SWI-OCT method can be used to assist cardiac muscle studies with respect to the assessment of the sample mechanical properties.
Fig. 5. Plots of the shear wave propagation distance versus the time delay at typical depths for the myocardium from (a) control and (b) CKO Lats1/2 mice. The magnitude of deformation at each spatial location is normalized to the absolute value of its minimum over time. The time axes indicate the shear wave propagation with the time delay forming over spatial locations.

Fig. 6. The SWI-OCT measurements reveal lower stiffness in Lats1/2 deficient cardiac tissues. (a) Shear wave velocities and (b) Young’s moduli of cardiac muscle tissues from control and CKO Lats1/2 mice. * $p<0.05$ from unpaired two-sample student’s $t$-test. Number of samples $N = 3$ for control and $N = 4$ for CKO Lats1/2 mice.

The uniaxial tension test (using Instron Model 5943) was performed on the control left ventricle cardiac muscle tissues from mice of similar age. The Young’s modulus was found to be $\sim 190$ kPa, which is close to the values obtained using SWI-OCT method. Due to the inhomogeneity of the cardiac muscle tissue at the measurement scale, the applied assumption of isotropic homogeneous material for the quantitative model might introduce errors for the estimation of the true Young’s modulus. To further optimize the quantification, more sophisticated analytical model is needed describing the complexity of shear wave propagation in inhomogeneous multilayered tissues.
4. Discussions and conclusions

The described method employs the multiwave imaging principle [43], where the imaging contrast and the imaging resolution are separately provided by two sources. Specifically, the induced shear wave inside the sample introduce the mechanical contrast that enables the measurement of the sample elasticity, and the use of OCT to monitor the shear wave propagation results in the improvement of the imaging resolution from millimeter scale (wavelength of the mechanical wave) to micron scale (coherence length of the optical wave). The wavelength of the induced shear wave could be controlled based on the loading parameters. For example, reducing the spot size of the air-puff stimulation will introduce more high frequency components to the shear wave, which will improve its capability to sense the mechanical gradient of inhomogeneous samples.

For quantitative SWI-OCT, the temporal resolution of the OCT system is an important parameter that determines the highest shear wave velocity this method can measure and also limits the transverse mapping scale for the potential heterogeneity detection. Considering a shear wave speed of 5 m/s in the sample, with the current 0.016 ms temporal resolving ability, the smallest transverse distance that allows the SWI-OCT system to monitor the wave propagation is 80 μm, which can be treated as the transverse scale for spatially mapping the elasticity of the sample. On the other hand, in terms of the measurement range, with increased transverse distance that is used for the quantification, the detectable wave velocity is proportionally increased. The implementation of a Fourier domain OCT system with 500 kHz A-line rate [44] will improve the capability of this SWI-OCT method for the future heterogeneity studies of mouse cardiac muscle. The smallest wave velocity that the SWI-OCT method can cover is determinat on the recording time of each M-mode OCT imaging. Longer time of M-mode data acquisition results in extended measurable range, however, sacrificing the possibility of rapid detection. Specifically, having 0.016 ms temporal resolution, 32 ms duration of each M-mode acquisition and a typical 1 mm distance, the wave velocity range that theoretically can be measured is from 31.25 mm/s to 62.5 m/s. Thus, the range of the measurable Young’s modulus is from ~3.1 Pa to ~12.4 MPa.

The sensitivity of the measurement can be defined as the smallest change of the Young’s modulus that can be detected, meaning one pixel shift from the temporal scale during the velocity measurement. Based on this definition, the sensitivity of this method depends on the system temporal resolution, wave propagation distance that is used for the quantification and wave speed. For instance, with the same temporal resolution (0.016 ms) and selected distance (1 mm), the wave velocity of ~31.25 mm/s (the lowest velocity that can be measured) results in the theoretically smallest resolvable change of wave velocity as ~15.63 μm/s, and theoretically ~0.003 Pa of the change of Young’s modulus, which can be considered as the best sensitivity that can be theoretically achieved with the current system. However, practically, measurement error may occur from the estimation of the time delay due to the discrete sampling in the temporal scale, especially for the stiffer samples that have relatively higher shear wave velocity.

In SWI-OCT, the field of view of the detection is limited by the propagation distance of the shear wave. Due to the significant attenuation of the wave amplitude, in order to cover a relatively larger region, the deformation magnitude at the shear source is required to be increased, which, however, might result in unreliable measurement of the sample response using the optical phase information from low coherence interferometry [45]. Also, the associated higher loading pressure may alter the sample properties. Conducting multiple measurements at several locations [32] can be a good way to address this issue and enlarge the detection region.

During our quantification of the sample elasticity, by using Eq. (1) and (3), we also assume the negligible viscosity of the sample and do not take the dispersion of the shear wave into account for the group velocity calculation. The presented results demonstrate under such
assumptions (including the ones presented in Section 2.5) the estimated Young’s moduli of soft samples using SWI-OCT agree well with the uniaxial compression measurements, which indicate the quantitative capability of SWI-OCT. Future work will be focused on the spectral analysis of the detected shear waves [46] with the quantification of phase velocity to obtain both elasticity and viscosity of the mouse cardiac muscle tissues.

Compared with AFM that has been generally used in cardiac tissue research with cellular level detection [47], SWI-OCT probes the group elastic effect of the cardiomyocytes that we believe can provide new information regarding the heart biomechanics, as an important assistance for better understanding of the properties of cardiac muscle. Also, the measurement scale and the procedure of SWI-OCT enable the potential studies of myocardium elasticity under different cardiomyosite status, such as during cell contraction and with different cardiomyosite directions, which are of great importance for the mechanical tests of regenerated or engineered cardiac muscle tissues. Unlike the AFM that is difficult to be applied for in situ or in vivo detection, SWI-OCT is possible to be utilized for intraoperative measurement on live mouse, and also, the integration of this method with endoscopic techniques, such as needle OCE [15, 48], might eliminate the requirement of opening the mouse chest. Compared with the traditional ultrasound elastography or magnetic resonance elastography, SWI-OCT has superior sensitivity of detecting the tissue displacement and can be used to obtain the sample elasticity with better preservation of its structure and functions. Because OCT provides 3-D high-resolution (micro-scale) images of the tissue with its optical properties, for example, the attenuation coefficient [49] and the polarization property [50], SWI-OCT holds the potential for multi-dimensional tissue characterization with one system.

To the best of our knowledge, this study provides the first demonstration of using quantitative OCE for noncontact biomechanical characterization of cardiac muscle and opens the door for a number of studies in the area of cardiac tissue engineering and cardiac muscle regeneration. Compared with the recently reported shear-wave-based OCE techniques which utilize piezoelectric actuator [30, 31] and ultrasound waves [32, 51] for the loading of sample, SWI-OCT relies on a transient air stream to interrogate the tissue and thus requires no contact and no acoustic coupling medium for the elasticity measurement. This noncontact feature of SWI-OCT enables potentially easier application for the in situ cardiac muscle characterization with better protection of the tissue. To avoid the interference of the heartbeat motion during SWI-OCT imaging, a few potential solutions could be employed such as: (1) temporary heart stopping procedure [52], which allows the measurement performed on an arrested heart; (2) employing synchronization between the loading-measurement and the heartbeat through electrocardiogram or optical mapping; (3) using post-imaging synchronization algorithms [53, 54], or (4) using optical pacing [55]. All of these techniques can minimize the influence of the heart motion on the imaging of the shear wave propagation and improve quantification of muscle biomechanical properties.

The shear wave in cardiac muscle tissue has the frequency range of approximately 0-2.5 kHz (cut-off frequency at 20 dB drop). This results in relatively large wavelength of the shear wave (millimeter scale). Although only parts of the wavelength is presented in the field of view (shown in Fig. 4), the propagation of the wave can be clearly observed based on the time delay of the local deformation profiles and the wave group velocity can be determined through the cross-correlation approach.

Because of the air-puff stimulation, surface Rayleigh wave can also be generated. Because Rayleigh wave and shear wave have very similar wave velocities [56], in our experiments, we have not observed distinct difference between them. Faster imaging system and larger transverse scans need to be applied in order to resolve the propagation difference between the surface Rayleigh wave and the shear wave. Our ongoing study is focused on separating these two different types of elastic waves and investigating how they can be utilized together to improve the measurement accuracy of the tissue elasticity.

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In conclusion, we present a quantitative optical elastographic method based on SWI-OCT for the use of noncontact biomechanical characterization of mouse cardiac muscle tissues. The feasibility of this method in quantitatively measuring sample elasticity is verified with uniaxial compression tests on tissue-mimicking phantoms. The results from ex vivo mouse myocardium samples with normal and genetically altered cardiomyocytes demonstrate this method can be used as an essential tool in cardiac research.

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