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Investigation of the Effects of Myocardial Anisotropy for Shear Wave Elastography using Impulsive Force and Harmonic Vibration

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Abstract

The myocardium is known to be an anisotropic medium where the muscle fiber orientation changes through the thickness of the wall. Shear wave elastography methods use propagating waves which are measured by ultrasound or magnetic resonance imaging (MRI) techniques to characterize the mechanical properties of various tissues. Ultrasound- or MR-based methods have been used and the excitation frequency ranges for these various methods cover a large range from 24-500 Hz. Some of the ultrasound-based methods have been shown to be able to estimate the fiber direction. We constructed a model with layers of elastic, transversely isotropic materials that were oriented at different angles to simulate the heart wall in systole and diastole. We investigated the effect of frequency on the wave propagation and the estimation of fiber direction and wave speeds in the different layers of the assembled models. We found that waves propagating at low frequencies such as 30 or 50 Hz showed low sensitivity to the fiber direction but also had substantial bias in estimating the wave speeds in the layers. Using waves with higher frequency content (> 200 Hz) allowed for more accurate fiber direction and wave speed estimation. These results have particular relevance for MR- and ultrasound-based elastography applications in the heart.

Keywords

anisotropy; transverse isotropy; shear wave; frequency; myocardium

Introduction

The term “shear wave elastography” (SWE) covers a wide set of methods that have been developed to measure the mechanical properties of soft tissues (Sarvazyan *et al.*, 2011). Many different diseases cause alterations in mechanical properties of tissues, so SWE offers a way to quantitatively measure these alterations that could assist in diagnosis or evaluation of therapeutic interventions. These methods use either external vibration or acoustic radiation force to generate shear waves in the tissue and commonly ultrasound or magnetic resonance imaging techniques are used to measure the resulting motion. Many organs have been the subject of investigation including the liver, breast, and thyroid (Sandrin *et al.*, 2003;

Muller *et al.*, 2009; Tanter *et al.*, 2008; Sebag *et al.*, 2010). Over the past several years, significant effort has been put forth by numerous groups towards quantifying mechanical properties of the cardiovascular tissues (Couade *et al.*, 2010; Couade *et al.*, 2011; Bernal *et al.*, 2011; Nenadic *et al.*, 2011c; Kolipaka *et al.*, 2010).

Magnetic resonance elastography (MRE) has been used to measure shear waves from an external vibration source to evaluate wave propagation in the myocardial wall (Kolipaka *et al.*, 2010; Kolipaka *et al.*, 2011; Sack *et al.*, 2009; Elgeti *et al.*, 2009; Tzschätzsch *et al.*, 2012; Elgeti *et al.*, 2012; Elgeti *et al.*, 2014). The frequencies used for generating the shear waves for these cardiac applications using MRE typically range from 24-200 Hz. MRE has the advantage of measuring three-dimensional motion, but requires several heart cycles to conduct the experimental measurements.

Multiple ultrasound-based methods that use propagating waves have been applied for measuring the mechanical properties of the heart. Kanai recognized that the aortic valve closure induced a propagating wave in the ventricular septum (Kanai, 2005). He used a Lamb wave model to estimate the viscoelastic properties of the myocardial tissue. The interactions of shear waves propagating in a bounded medium cause mode conversion at the boundaries giving rise to shear and longitudinal waves, which influences the observed wave velocities. Nenadic, *et al.*, built upon that model to develop the Lamb wave dispersion ultrasound vibrometry (LDUV) method that used external vibration to create harmonic anti-symmetric Lamb waves (Nenadic *et al.*, 2011c; Nenadic *et al.*, 2011a; Nenadic *et al.*, 2011b). The phase velocities at several frequencies ranging from 50-500 Hz were fit to the same antisymmetric Lamb wave model originally proposed by Kanai (Kanai, 2005). The LDUV method was used for studies in normal pigs and pigs with myocardial infarctions to quantify the viscoelastic properties of the left ventricular heart wall in an open heart preparation (Urban *et al.*, 2013; Pislaru *et al.*, 2014).

Acoustic radiation force (ARF) has also been used to generate shear waves in the myocardium. Couade, *et al.*, used the supersonic shear imaging (SSI) technique to generate and measure propagating waves in the left ventricle of *in vivo* sheep in open chest preparation (Couade *et al.*, 2011). Measurements of dynamic changes through the cardiac cycle were made in different scanning orientations revealing anisotropic wave propagation properties. In addition, variation of the wave speed through the thickness of the myocardium at different angles in an explanted heart sample was also reported. A study in open chest dogs using waves generated using ARF was reported by Bouchard, *et al* (Bouchard *et al.*, 2009). Variation of shear modulus in Langendorff perfused isolated rat and rabbit hearts was also explored using ARF methods (Pernot *et al.*, 2011; Vejdani-Jahromi *et al.*, 2015). Intracardiac transducers have also been used to produce shear waves in normal pigs and pigs with infarcted regions (Hollender *et al.*, 2012; Hollender *et al.*, 2013).

Most of the previously described work with ultrasound-based SWE has been done with invasive preparations or procedures. Transthoracic measurements in humans have recently been reported and are an initial step towards translating SWE methods for clinical applications in humans (Song *et al.*, 2013).

Several of the previously described studies have noted spatial variation of the wave speed through the thickness of the myocardium. These variations can be explained by the intrinsic anisotropic architecture of the myocardial tissue and the rotation of the muscle fiber directions through the thickness of the ventricular wall (Streeter *et al.*, 1969). Two distinct studies have investigated the anisotropy of the heart wall using waves generated using ARF in *ex vivo* and *in vivo* hearts (Lee *et al.*, 2012b; Lee *et al.*, 2012a). The elastic tensor imaging (ETI) method that was developed for analysis of the anisotropy of the heart took the approach of modeling the heart as a series of transversely isotropic layers (Lee *et al.*, 2012a). In a transverse isotropic medium like skeletal muscle which has muscle fibers running in a certain direction, the wave will travel fastest when propagating along the fibers and slower when propagating perpendicular to the fiber direction (Gennisson *et al.*, 2003; Aristizabal *et al.*, 2014; Wang *et al.*, 2013). The fiber direction was estimated using ETI at different depths in the myocardium by rotating the transducer and obtaining data at multiple angles and identifying the angle which provides the fastest wave velocity. This was performed in numerous heart samples and the results were validated against histology of the samples or measurements made using magnetic resonance diffusion tensor imaging (Lee *et al.*, 2012b; Lee *et al.*, 2012a).

The effects of myocardial anisotropy on the measurements of wave velocity can be quite significant and need to be taken into account for clinical investigation with SWE techniques. However, the various techniques that have been developed cover a wide range of frequencies, 24-500 Hz, particularly when comparing results from MRE studies and studies using ARF excitations. This study will examine the effects of frequency of the propagating wave and myocardial anisotropy in a numerical model of the heart. To our knowledge, this is the first systematic study that examines the role of the excitation frequency on the waves generated in an anisotropic medium such as the heart and the wave velocities.

We will describe a finite element model that follows the idea of ETI where transversely isotropic layers are oriented at different angles. These models will be excited either using harmonic excitations or impulsive excitations that mimic harmonic external vibration or ARF, respectively. We will estimate the fiber direction by finding the direction with the maximum wave velocity. The finite element model is attractive because the true values of fiber angle and wave speed are specified and can be used for comparisons to evaluate errors in estimation. These types of models also offer flexibility for evaluating the complicated wave propagation in such a layered material for which no phantoms have been reported outside of using myocardial tissue in which the true material properties are not known and the fiber direction would have to be evaluated by other methods such as histology or diffusion tensor imaging (Lee *et al.*, 2012b; Lee *et al.*, 2012a). We will compare measured and known fiber directions and the wave speeds estimated along and across the fibers of the transverse isotropic layers. We will conclude with a discussion and concluding remarks.

Methods

Finite Element Model

We developed different multilayered models to explore the effects of frequency and myocardial anisotropy using finite element modeling (FEM). Models were constructed using

Abaqus (ver. 6.12-1, Dassault Systemes, Waltham, MA) to model the heart in both systole and diastole. Each layer was disk shaped with a diameter of 100 mm and thickness of 1 mm. For the systolic model we used 20 layers and in the diastolic model we used 10 layers to mimic the myocardial thickening and thinning during systole and diastole, respectively. The different layers are combined by a “TIE” condition available in Abaqus. This constraint requires the nodal displacements to be continuous in the boundary surface between two adjacent layers. More specifically, the nodes in one surface are constrained to have the same displacements as their corresponding nodes in the coupled surface. For this model, the distribution of the mesh in each layer is the same for all the layers, so there is no mismatch between the nodes in the tied surfaces (Dassault and Systems, 2011). We rotated the fiber direction, or the principal axis of the transverse isotropic layers from -50° to $+80^\circ$ from the epicardial surface to the endocardial surface similar to previous reports (Lee *et al.*, 2012b; Lee *et al.*, 2012a). The increment of the angles for each layer for the systolic heart model was 6.8421° and for the diastolic heart model was 13.6842° . The bottom and top surfaces of the models were surrounded by water to simulate the loading effects from the blood and the outer boundaries were bordered by infinite elements available in Abaqus to minimize the wave reflections. We defined the elastic shear modulus along the fibers, $\mu_{\parallel}$, and across the fibers, $\mu_{\perp}$, to define the transverse isotropic layers. These values were taken after considering past studies from *ex vivo* and *in vivo* experiments (Lee *et al.*, 2012b; Lee *et al.*, 2012a; Nenadic *et al.*, 2011c; Urban *et al.*, 2013; Song *et al.*, 2013; Hollender *et al.*, 2012). Additionally, we defined the stiffness matrix components in Table I. Each layer had the same material property definition, but the angle of the fibers is changed with each layer and the angle was varied linearly through the thickness of the model. As a point of comparison we also constructed a transversely isotropic model with 20 layers where all the layers were oriented at 15° and each layer had the material definition of the systolic model. The Poisson's ratios (ν_{tp}) of the layers were set to 0.4996 in the systolic model and 0.4999 in the diastolic model to mimic the incompressibility of biological tissues. Other past studies have used values of Poisson's ratio close to 0.49 (Janz and Grimm, 1972; Janz and Grimm, 1973; Gotteiner *et al.*, 1995). Elements for the layers were linear hexahedron elements with reduced integration and enhanced hourglass control (C3D8R). Elements for the water media are linear hexahedron acoustic elements (AC3D8R). Table I summarizes the material properties and number of layers for each model.

The models were excited using a line source in the middle of the model that extended through all of the layers and the force was applied in the direction of the thickness, which we will define as the z-axis. The harmonic excitations were performed at 30, 50, 100, and 200 Hz. The impulse was modeled with a half cycle of a raised sine wave with frequency of 1 kHz with duration of 500 μ s. The dynamic responses of the tissue model to the excitations were solved by the Abaqus explicit solver and the particle velocities were extracted for further analysis. The data was sampled at a rate of 10 kHz. A schematic of the systolic model is shown in Fig. 1. The diastolic model is slightly modified to only have 10 layers.

Analysis

The motion in the z -direction from each layer was extracted and the wave velocity was estimated using a Radon transform-based method (Song *et al.*, 2013). The Radon transform of a function $f(x,y)$ is defined as (Bracewell, 2000)

$$g_{\theta}(R) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} f(x,y) \delta(R - x \cos \theta - y \sin \theta) dx dy \quad (1)$$

where the angle θ is direction of the projection in the image space (x, y) and R is the abscissa in that direction.

In the MATLAB implementation of the Radon transform with the function 'radon.m' (The MathWorks, Natick, MA), the transform space is given as a function of the angle, θ , and the radial coordinates, n , corresponding to each row of $g_{\theta}(R)$, which can alternatively be written as $g_{\theta}(n)$. For a 50 Hz harmonic wave traveling along the x -direction (Fig. 2(a)), outwards from a source similar to those generated in this study, through time (related to variable y in Eq. (1)) and the angle of the wavefronts in this spatiotemporal domain is related to the speed of the wave. The Radon transform of this wave image is shown in Fig. 2(b). The spatial and temporal scaling is ignored when taking the Radon transform, but the pixel resolution can then be used to determine the relationship between the angles and the speed of the wave. In other Radon-transform based methods for shear wave speed estimation the peak in the Radon space is identified and the value of θ is then related to the speed of the wave by

$$c = \frac{\Delta x}{\Delta t} \tan(\theta), \quad (2)$$

where x and t are the dimensions of the pixels in the spatial and temporal directions in the spatiotemporal image (Urban and Greenleaf, 2012; Song *et al.*, 2013). One application of Eq. (2) is that we can limit the angles which we examine depending on anticipated speeds that will be measured and not use data in the Radon space related to speeds or values of θ outside of these speeds. Additionally, for a harmonic wave, there are multiple peaks at a given angle. One approach to take this characteristic into account is to take the square of the amplitude of the data in the Radon space and perform a summation over the vertical dimension, n , such that

$$R_{SSQ}(\theta) = \sum_{n=1}^N g_{\theta}^2(n), \quad (3)$$

and N is the number of rows generated by the MATLAB function radon.m.

The peak of the R_{SSQ} function can be found and the speed of the wave can be calculated. The R_{SSQ} function for Fig. 2(b) is shown in Fig. 2(c). The profile in Fig. 2(c) has also been computed using a speed limit set of $c_L = [0.5, 8]$ in m/s, so any angles corresponding to speeds outside of this range had the data in the Radon space set to zero. The sum-squared Radon transform method (R_{SSQ}) was used for all wave speed estimations. The wave

velocities for the harmonic excitations are to be regarded as phase velocities and the wave velocities from the impulsive excitation are regarded as group velocities.

For each layer we obtained a distribution of wave velocities with respect to angle. We fit this distribution to an elliptical model, which is consistent with a transverse isotropic material (Qiang *et al.*, 2015; Wang *et al.*, 2013), to extract the angular orientation of the ellipse, and from that ellipse fit we estimate the wave speed along and across the fibers.

Experiment

In order to compare our simulations with a real myocardial sample, an experiment was performed in an *ex vivo* porcine heart with a transesophageal (TEE) probe that has a rotating linear array head that could perform ARF-based SWE measurements. The heart that was used had been frozen so the mechanical properties could be different from a perfused *in vivo* heart, but the fiber orientation is still consistent. The TEE probe can focus close to the transducer surface and the ability to rotate the transducer makes it advantageous for exploring the anisotropy of tissues. The TEE probe was coupled to the heart with ultrasound gel in air. The TEE probe (MPT7-4, Philips Healthcare, Andover, MA) was driven with a Verasonics system (V1, Verasonics, Inc., Kirkland, WA) to perform the SWE measurements. Acquisitions were made at 9° increments over the range of 0-170°. The ARF push duration was 600 μ s. The push beam was focused in the middle of the myocardium and extended through nearly the whole wall. A plane wave compounding detection method with a frame rate of 3.33 kHz was used to detect the wave motion (Montaldo *et al.*, 2009). Three plane waves were used at -2°, 0°, +2°. The motion was calculated from the in-phase/quadrature (IQ) data using autocorrelation (Kasai *et al.*, 1985). The wave motion data was averaged in axial windows of length 1.23 mm and 0.62 mm overlap. The impulsive FEM and experimental data was then analyzed using a two-dimensional Fourier transform (2D-FT) method to extract the wave velocity at the specific frequencies of 50, 100, and 200 Hz (Bernal *et al.*, 2011). The data at 30 Hz was not extracted because artifacts can occur in this type of analysis at low frequencies.

Results

We analyzed the 20 layer transversely isotropic model and found that the angle that was estimated for each layer was within 0.5° of the specified value of 15°. The wave speeds identified along and across the fibers for each layer are shown in Figs. 3 and 4, respectively. There is a bias observed in the wave speed at the edges of the model, and there is also a noticeable overall bias at 30, 50, and 100 Hz excitations.

An example of the motion at 9 ms is shown in Fig. 5 for the 50 Hz, 100 Hz, 200 Hz, and impulse excitations in the systolic model. The 50 Hz wave shows some small variation in speed through depth. The 100 and 200 Hz waves are moving slower at the upper and lower boundaries and faster in the center. The impulse excitation creates waves with a wide variation of speeds through the thickness.

Polar plots of the estimated wave speeds are shown in Fig. 6 for layers 3, 8, 13, and 18 for the different excitations in the systolic heart model. The impulse results show clear rotation

of the wave speed distribution which is expected due to the specification of the model. The results using 100 and 200 Hz also show some rotation but not as much as the impulse results. The 30 and 50 Hz results show very little rotation. We analyzed the data in the systolic heart model and estimated the angle of the fibers for each layer. The results are shown in Fig. 7 with a reference line for comparison related to the specified values in the FEM simulation. The results for 30 and 50 Hz are directly on top of each other and show very little variation through the different layers and provide an average of the fiber angle. The 100 and 200 Hz results show better agreement but still have a biased result. The impulse excitation provides good agreement for layers 4-17 but has more bias at the edges of the model, which is probably due to the fluid boundary condition. The impulse excitation contains information at multiple frequencies and the center frequency is around 400 Hz. This center frequency was found by performing Fourier transforms on the wave motion signals (data not shown) and finding the frequency when the spectra had their peaks.

From the elliptical fitting of the estimated wave speed distributions, we found the wave speeds along and across the fibers in the transverse isotropic material definition. The wave speeds found along and across the fibers are shown in Figs. 8-9, respectively. The reference values in the systolic model for the layers were 5 m/s and 2.5 m/s along and across the fibers, respectively. These were calculated using $c = \sqrt{\mu/\rho}$ where μ is the shear modulus and ρ is the mass density. For the case along the fibers, the 30 and 50 Hz results show a very large bias ($> -60\%$). (Bias was calculated by finding the difference between the true and estimated values and dividing by the true value and then multiplying by 100 to express as a percentage.) The bias at 100 and 200 Hz is reduced but is still large (-35 to -20%). The impulse excitation has very small bias except at the top and bottom edges of the model. In the case measurements of the speeds across the fibers, the 100 and 200 Hz show an overestimation in the middle of the model, and the impulse shows low bias. The 30 and 50 Hz still have large bias across the fibers.

Polar plot results from the diastolic heart model are shown in Fig. 10. The 30 and 50 Hz results do not show much change through the thickness. The amount of rotation of the wave speed plots for the 100, 200 Hz, and impulsive excitations were similar. Figure 11 shows the results of the fiber direction estimation. The 30 and 50 Hz excitations do not estimate the fiber angle accurately, and the excitations with higher frequencies have better agreement with the reference curve. The estimated speeds along and across the fibers for the diastolic model are shown in Figs. 12-13, respectively. The level of bias in the estimated speeds along and across the fibers is less than in the systolic heart although for 30 and 50 Hz the bias is about -30% . There is a small level of overestimation, up to $+20\%$ for the 100 Hz and impulse cases, for across the fibers.

Lastly, we conducted an experiment on an excised porcine heart in the left ventricular wall. We rotated the head of a transesophageal probe and performed acoustic radiation force experiments at each angle. We analyzed the data to estimate speeds at different angles through the thickness of the myocardium. Figure 14 shows results of the wave propagation speeds estimated at 50, 100, and 200 Hz extracted from the impulsive systolic heart model (top row) and from the experimental data (bottom row). The 50 Hz results show very little

variation in the angular location of the peaks of the shear wave speed. The 100 and 200 Hz results show that the angles of the peaks in the wave speed distribution shift through depth.

Discussion

We have created finite element models that were composed of multiple transversely isotropic layers that were each oriented at different angles. We explored the effects of the frequency of the excitation on the wave propagation in this complicated medium. One interesting finding was that lower frequencies like 30 and 50 Hz which are used in MRE of the heart had very little sensitivity to the fiber angles. The waves had their highest wave speed at an angle related to the middle layers or the average value of the angular range. However, it should be noted that a relatively strong bias in the measured wave speed was observed. Therefore, using this type of excitation should not be expected to obtain an accurate value of the underlying wave speed or shear modulus of the material or tissue. This substantial bias may partially be related to Lamb wave behavior of an antisymmetric mode that we have studied for the characterization of the myocardium in previous work. At these low frequencies, the wave speeds are markedly lower than the wave speed at higher frequencies for a uniform plate of a given stiffness (Nenadic *et al.*, 2011c; Nenadic *et al.*, 2011b). However, using lower frequencies typically provides larger wave amplitudes, and if a measurement is desired which is not affected by the fibers, a lower frequency should be used.

Some similar trends were present in the transverse isotropic and systolic models that were bounded by fluid, particularly the observation of speed bias at the fluid-loaded boundaries (Figs. 3 and 4). We also observed some bias associated with the different frequencies in Figs. 3 and 4 as shown in Figs. 8 and 9. However, in the across the fibers case, we did not observe the overestimation in the transverse isotropic case as in the systolic model.

We quantified the amount of rotation for each of the excitations. In the systolic model the amount of rotation was -10.5° , -10.5° , 55.9° , 75.2° , and 104.0° for the 30 Hz, 50 Hz, 100 Hz, 200 Hz, and impulse excitations, respectively. For the diastolic model the amount of rotation was -8.7° , 31.2° , 83.1° , 69.5° , and 88.0° for the 30 Hz, 50 Hz, 100 Hz, 200 Hz, and impulse excitations, respectively. The target for both cases was 130° , and in general the excitations with higher frequencies yielded more accurate levels of rotation.

When higher frequencies were used, they showed higher sensitivity to the fiber direction. Previous studies with ARF excitations have shown that accurate fiber directions could be estimated (Lee *et al.*, 2012b; Lee *et al.*, 2012a). Typically, ARF excitations yield waves that have bandwidths ranging from 50-500 Hz (Urban *et al.*, 2012). This frequency content can be modulated by the beam shape and excitation length (Palmeri *et al.*, 2014). In particular, ARF-induced waves may have reduced frequency bandwidth when a transthoracic approach is used. Using these higher frequencies also yielded smaller levels of bias with respect to estimating the true wave speed in the layers. We did observe some underestimation at the borders of the model where there was a fluid boundary condition, which could be related to Scholte wave propagation (Langdon *et al.*, 2015).

In both the systolic and diastolic models there was a unique point in the estimated fiber directions at which the results for all the frequencies converged. In the systolic model this was near layer 11 and in the diastolic model it was between layers 5 and 6. These points represented the middle of the models and demonstrate that regardless of excitation frequency or boundary conditions that accurate fiber direction could be estimated. This may hold promise for some simplified *in vivo* characterizations.

The overestimation of the speeds across the fibers for 100 and 200 Hz excitations may be due to the ellipse fitting process. The data in Figs. 6 and 10 are not perfect ellipses and so the fitting or the Radon transform speed estimation may have caused some overestimation.

Unfortunately, there is no closed form solution of this model to validate our results. However, the general agreement between the model and the excised porcine heart serves to validate the model. We did not quantify the fiber direction in this tissue sample, but the qualitative agreement is good.

Another aspect of the myocardium is that the tissue can be characterized as a viscoelastic medium. We have previously assumed that the wall was a homogeneous plate and characterized the viscoelastic material properties using Lamb wave model fits (Urban *et al.*, 2013; Pislaru *et al.*, 2014). We can extend this present elastic model to incorporate viscoelastic, transversely isotropic layers based on the methods described by Qiang, *et al* (Qiang *et al.*, 2015).

In addition to the viscoelastic nature of the heart, the heart is a dynamic organ that is constantly moving. These models were meant to capture the anatomy (wall thickness and material properties) at only two distinct time points in the cardiac cycle. The general model could be extended to examine other states within in the cardiac cycle. Filtering techniques may also have to be utilized to remove the underlying myocardial motion in order to examine the wave propagation motion for estimation of the wave velocities and fiber directions. For *in vivo* application the ARF beams would have to induce motion with adequate amplitude through the thickness of the wall and with sufficient frequency bandwidth for estimation of the wave velocities and fiber directions. These types of technical improvements are the topics of ongoing study within the field.

Lastly, the model only simulates the variation of the myofiber orientation change in a discrete sense, whereas the true variation operates within a continuum. Additionally, the layers were all of uniform thickness and the model had no curvature or changes in thickness. Another simulation platform that could be used is the recently introduced Living Heart Project which is a multiphysics-based model that incorporates the electromechanical properties within the heart (Baillargeon *et al.*, 2014). This model has anisotropic material properties dictated by varying fiber directions through the myocardial thickness. We will explore superimposing wave propagation experiments upon the normal motion of the heart to evaluate the wave propagation in a more realistic geometry.

Conclusion

We have developed a model to simulate the anisotropic nature of the heart wall by using layers of transversely isotropic materials oriented at different angles. We used different excitations to explore the effects of the frequency related to accurate speed and fiber direction estimation. It was found that low frequencies yield low sensitivity to the fibers and underestimated the true wave speeds in the layers. Using excitations at higher frequencies allowed for better estimation of the fiber direction and more accurate wave speeds. These results have particular relevance for elastography techniques applied to the myocardium that use excitations at different frequencies such MR- and ultrasound-based techniques.

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Figure 1. Schematic of systolic model. The three-dimensional model consists of multiple layers of 1 mm thickness. The diameter of each layer is a disk with a diameter of 100 mm. The model is bounded on the top and bottom by water. The line of excitation is in the middle of the model.

Figure 2.

Analysis of harmonic wave motion with R_{SSQ} algorithm. (a) Spatiotemporal wave motion data for a 50 Hz wave. The solid line is the line related to the wave speed estimated. Note that for proper angle analysis the image must be specified and plotted in units of pixels. (b) Radon transform space for wave motion, where a large amplitude is observed near $\theta = 52^\circ$. (c) The RSSQ function with speed limits applied with $c_L = [0.5, 8.0]$ m/s.

Figure 3.
Estimated wave speeds along the fibers using different excitations in the full thickness transversely isotropic model compared to the reference value of $c = 5$ m/s.

Figure 4.
Estimated wave speeds across the fibers using different excitations in the full thickness transversely isotropic model compared to the reference value of $c = 2.5$ m/s.

Figure 5. Wave propagation in systolic model for different excitations. (a) 50 Hz, (b) 100 Hz, (c) 200 Hz, (d) impulse excitation.

Figure 6.

Polar plots of estimated wave velocities for different layers (3, 8, 13, and 18) in the systolic heart model for the different harmonic excitations at 30, 50, 100, and 200 Hz and the impulsive excitation.

Figure 7.

Fiber angle estimation for systolic heart model for different excitations. The reference line is the angle specified for each layer in model.

Figure 8.
Estimated wave speeds along the fibers using different excitations in the systolic model compared to the reference value of $c = 5$ m/s.

Figure 9.

Estimated wave speeds across the fibers using different excitations in the systolic model compared to the reference value of $c = 2.5$ m/s.

Figure 10.

Polar plots of estimated wave velocities for different layers (2, 4, 6, and 8) in the diastolic heart model for the different harmonic excitations at 30, 50, 100, and 200 Hz and the impulsive excitation.

Figure 11.

Fiber angle estimation for diastolic heart model for different excitations. The reference line is the angle specified for each layer in model.

Figure 12.

Estimated wave speeds along the fibers using different excitations in the diastolic model compared to the reference value of $c = 2$ m/s.

Figure 13.

Estimated wave speeds across the fibers using different excitations in the diastolic compared to the reference value of $c = 1$ m/s.

Figure 14.

Comparison between systolic heart model and experimental data from excised porcine heart.

(a) FEM, 50 Hz, (b) FEM, 100 Hz, (c) FEM, 200 Hz, (d) Experimental, 50 Hz, (e)

Experimental, 100 Hz, (f) Experimental, 200 Hz.

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Table 1

Summary of Model Parameters

Model	Number of Layers	μ_I , kPa	μ_L , kPa	C_{11} , MPa	C_{13} , MPa	C_{33} , MPa	ν_p
Systolic	20	25.00	6.25	10.006	9.9946	10.015	0.4996
Diastolic	10	4.00	1.00	10.001	9.9986	10.004	0.4999