Study of Mechanical Properties of Left Ventricle using Finite Element

A thesis submitted in partial fulfillment of the requirements
For the degree of Master of Science in Mechanical Engineering

By
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Abstract

Study of Mechanical Properties of Left Ventricle using Finite Element

By
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Master of Science in Mechanical Engineering

According to the CDC, 735,000 people in the United States experience a heart attack (myocardial infarction) every year. 30% of these have previously suffered a myocardial infarction (MI). With each successive event, mortality rates drastically increase. These episodes generate damaged tissue in the heart which adversely affect heart function and make diagnosis and treatment options difficult [6]. This study presents a finite element model of left ventricular function to gain insight into the mechanical changes that result from an infarct that may lead to increased risk of subsequent heart failure. This information would be a useful adjunct for physicians treating patients suffering from cardiovascular disease with a prior episode of infarct. Magnetic resonance imaging (MRI) was used to create a symmetric and geometrically realistic (natural) three-dimensional left ventricle (LV) computer model. Simulations were conducted using the finite element method to predict the mechanical behavior of the LV. Regions of simulated infarct damage were incorporated into the analyses by altering material properties for a specific region. Both symmetric and natural models were considered in the analysis.
Infarct cases 1-6 corresponding to a infarct percent by volume of 8, 9, 4, 16, 25 and 51% had a reduction in end-diastolic volume of 5, 5, 3, 8, 11, 17 mL respectively. The pressure increase required to restore EDV for Infarct 1-6 were 3, 3, 1, 5, 9 and 12 mmHg. Three natural infarct cases denoted case A, B and C consisting of a 7%, 15% and 50% infarct region by volume respectively were evaluated. The decrease in EDV for cases A, B and C were 6, 9 and 17 mL requiring an increase in LV pressure (hypertension) of 2.5, 5 and 12 mm Hg to restore normal end-diastolic volume (EDV) [32]. The simulated decrease in EDV for the infarct cases was consistent with patient’s experiencing decreased tissue compliance [21]. The higher LV pressure resulted in an increase in wall stress opposite the infarct for the symmetric and natural infarct cases. The stress distribution in the natural model had large spatial variation compared to the symmetric model, which has a smooth stress distribution from base to apex. The natural model was considered the most useful as LV geometry was found to greatly influence stress distribution and magnitude. The relative magnitude of the stress in the fiber and longitudinal direction was consistent with tensile testing of myocardial tissue [43]. The results are potentially useful in determining a relative severity in MI patients and identifying high stress locations in the LV.
1. Background

The purpose of this chapter is to provide an overview of research in the area of cardiovascular mechanics. Most of the studies discussed were conducted during the last two decades. The topics are heart geometry, experimental testing, material models, infarcts and finite element studies of the left ventricle (LV).

1.1 Geometry

The left ventricle geometry was generated in different ways among studies. This includes symmetric models created from general dimensions of the LV and detailed models produced from medical imaging such as magnetic resonance imaging (MRI) [24, 33]. Numerous studies have been conducted to capture the general shape of the heart [3]. This information is useful in developing a 3 dimensional model, however a number of studies did not create a computer aided design model from the collected data. One study used MR images to develop a statistical atlas of the heart from twenty five healthy patients. A software tool was used to manually trace the heart at different points in the cardiac cycle. Long-axis and short-axis views were used to develop the 3 dimensional representation. The resolution of the data sets were limited by the space between slice thicknesses. Another study used turbo gradient echo (TBE) and steady-state free precession (SSFP) pulse sequences and of sixty patients to establish normal ranges of end diastolic volume (EDV). This study was gender specific and statistically significant differences in size were observed from the results. Again, the contours from the medical imaging were manually traced [17, 24]. A similar study used MRI to generate dimensions for the left and right ventricle from one hundred eighty healthy patients [1]. The EDV was estimated to be lower using TGE. This suggests there are variations between
obtained values between medical imaging techniques. The wall thickness and mass was measured in another study using echocardiography. This information was relevant for comparison with both symmetric and natural CAD models. The published values were compared to the wall thickness of the 3 dimensional model developed from the MR images [24]. Another study was conducted to compare between MRI and echo in myocardial infarction (MI) patients. Echo was found to underestimate several metrics including LV volumes. The MRI provided improved detection and quantification of segmental function after MI compared to echo. This study supports the use of MRI for studies related to the mechanical behavior of the LV after MI [11]. The majority of finite element studies used MR images to develop the heart geometry. This involved the manual selection of points defining the inner and outer layer of the heart [40]. Figure 1.1 displays an example of the data points produced from MR images.

Figure 1.1 Data points for LV
The myocardium is composed of muscle fibers that are known to vary transmurally. The orientation of the fiber at any point through the wall of the LV is known as fiber direction. A constant fiber direction was used in the research study. This was a simplification and there is significant variation in fiber direction in the LV. Several studies have been conducted to capture the transmural variation in fiber direction for the LV. One study found the circumferential fiber direction to vary from -41 to 66 degrees from epicardium to endocardium in humans. The radial fiber was found to be around 0 with a slight rise moving from epicardium to endocardium. DT-MRI was used ex-vivo to obtain the cardiac fiber architecture [23]. Another study used DT-MRI on anesthetized rabbits and obtained a helical fiber variation of -60 to 40 degrees from epicardium to endocardium. These values show a similar pattern with the previous study, but demonstrate there is variation in fiber direction between species [35]. Due to limited data and variation between samples, accurately representing the fiber direction in the LV is a challenge. One study demonstrates the variation in fiber direction at different locations in the LV for the same transmural point. For example, a point on the inner surface near the base versus a point on the inner surface near the apex. This is in addition to transmural variations in fiber direction and demonstrates the convoluted nature of the LV fibers [30].

1.2 Experimental Testing

In order to analyze the mechanical properties of the passive myocardium, physical testing must be conducted on myocardial tissue. The testing protocol determined the type of material law it can be implemented into. The material models will be discussed in further detail later in this chapter. Examples of tissue testing are uniaxial, biaxial and shear. A study used bovine heart tissue for uniaxial tensile testing. The stress-strain
relationship generated was potentially useful for an isotropic model of the myocardium. However, it is known there is a significant difference in mechanical properties between the fiber and cross-fiber direction that is not considered in this study. It is not clear from the study how the tensile tests were performed in relation to the fiber direction [43]. To capture the effects of fiber direction, a biaxial tensile test must be conducted. Another study used a biaxial testing protocol to capture the effect of fiber direction. The study used canine LV tissue and significantly different behavior was found between the fiber and cross-fiber direction.

Another study conducted shear testing on cubes of porcine myocardial tissue. The results suggested anisotropic behavior of the LV. The cubes used in the study were 3x3x3 mm thick. A mean fiber direction was used to identify sample orientation for placement on the testing apparatus. This could be problematic as the fiber direction is known to vary significantly over a 3 mm transmural distance, altering the shear data. All the studies mentioned show the highly nonlinear behavior of the myocardium. The studies also yielded an exponential stress-strain curve. This suggests there was consensus on the myocardium behaving nonlinearly and having an exponential stress-strain relationship [43]. The studies were performed under quasi-static conditions

1.3 Material Models

The material model used to represent the passive myocardium is an important part of analyzing its mechanical behavior. A brief description of mechanical properties of the passive myocardium are given to provide insight into the benefits and limitations of each material model. The myocardium is composed of muscle fibers which are known to vary transmurally. As with other soft biological tissues, the myocardium is considered an

4
incompressible material [25]. Experimental testing suggest highly nonlinear behavior with an exponential stress-strain relationship. There is residual stress in the passive myocardium which has an effect on stress distribution. However, this value is difficult to quantify and consequently must be incorporated with caution. Although the myocardium appears to have viscoelasticity, the cardiac cycle has a short time scale compared to the viscoelastic response. Experimental data is also limited relating to viscoelasticity. It is reasonable to model the myocardium as elastic, meaning stress is only dependent on the strain [16, 28, 34]. The discussed models used strain energy density functions to represent the myocardium as hyper elastic. Several of the material models were discussed in further detail in the next chapter. The isotropic model was the simplest representation of the myocardium. One study analyzed several isotropic models and chose the Yeoh model as the best representation of the experimental data. This model used a polynomial strain energy density function [19]. Another study used an exponential isotropic model. A Mooney-Rivlin model has also been used to model soft tissue. This model can be used to model the nonlinear behavior of the myocardial tissue. Mooney-Rivlin is an extension of the neo-Hookean model. A more sophisticated model is transversely isotropic, which accounts for the fiber structure of the myocardium. The transversely isotropic models discussed used an exponential strain energy density function [40]. One study suggested the use of a linear isotropic term in the strain energy density function to simplify the model. The orthotropic model attempted to model the myocardium as having different properties in three orthogonal planes. This analysis was based on the shear testing data previously mentioned [9].
1.4 Infarct

Several studies have been conducted to determine the variation in size of infarct in MI patients. In one study, one fourth of the patients had an infarction size ≤ 10%. 33% of patients had an infarct region > 20% and 13% of the patients had an infarct size of >40% [11]. In another study the median size of the infarct regions was 89g with a range of (54-160 g) [26]. Figure 1.2 shows the distribution of infarct sizes between patients for the first study mentioned. The infarct regions were measured in percent infarct by volume.

![Figure 1.2 Infarct size by % of LV volume [24]](image)

1.5 Finite Element Analysis of Left Ventricle

Finite element studies have been conducted on the LV of the heart with varying geometries and material representations. The stress distribution appeared to be consistent
with the highest stress on the inner wall and decreasing transmurally in the fiber direction [14]. Large spatial variation in stress on the inner wall was found for the various studies. Figure 1.3 displays the stress distribution in the fiber direction for two different exponential, transversely isotropic models. The defined local coordinate system was chosen to represent the fiber and cross fiber structure of the myocardium. The left image used the Costa constitutive law. The right image used the Guccione constitutive law. The parameters for the two models were obtained from different experimental data sets [39].

![Stress distribution in fiber direction using two different constitutive models](image)

**Figure 1.3 Stress distribution in fiber direction using two different constitutive models [39]**

Peak stress values were found at various locations spatially on the LV wall. There were regions of high and low stress within close proximity along the inner wall. The study suggests variation in geometry has a significant effect on the prediction of stress values. Another study use a similar material model with different LV geometry predicted higher stress values. Although the magnitude of stress changed between studies, there was generally agreement in the distribution of transmural and inner surface stress in the fiber direction [19, 36, 39]
2. Introduction

The human heart is a muscular organ which acts similar to a mechanical pump and is responsible for circulating blood throughout the body. The heart is composed of four chambers: left ventricle, left atrium, right ventricle and right atrium. The left ventricle (LV) is of special interest because it experiences the highest pressure and stress of the four chambers of the heart. The heart contains four valves, all of which allow blood flow in one direction. The flow of blood in and out of the LV is controlled by the mitral and aortic valve. The mitral valve allows oxygenated blood to enter LV from the left atrium. The aortic valve allows oxygenated blood from the LV to exit into the ascending aorta. The left side of the heart receives oxygenated blood from the lungs and pumps it systemically. The flow of blood in and out of the right ventricle is controlled by the pulmonary and tricuspid valve. The tricuspid valve allows deoxygenated blood to flow from the right atrium into the right ventricle. The pulmonary valve allows blood to flow from the right ventricle to the pulmonary trunk. The right side of the heart receives deoxygenated blood from the body and pumps it to the lungs where the blood is oxygenated. Strong fibers called the chordae tendineae connect the cusps of the mitral and tricuspid valve, or atrioventricular valves, to the papillary muscles. The papillary muscles are located on the inner surface of the ventricle and prevent inversion or prolapse of the valves [4]. Muscular tissue, or the myocardium, makes up the walls of the heart. The myocardium is the thickest layer of the heart and is composed of fibers. The fiber direction changes through the wall and from base to apex [23, 30, 35]. Figure 2.1 displays the general anatomy of the human heart.
The cardiac cycle is the complete process by which blood is circulated throughout the body by the heart. It has two main phases, systole and diastole. Systole is the contraction phase in which blood is forced out of the heart. Diastole is the relaxation phase in which the heart passively fills with blood. Diastole has four sub-phases: isovolumetric relaxation, rapid filling, reduced filling, and atrial contraction. During isovolumetric relaxation, the four valves are closed and pressure within the atria begins to rise. When the atrial pressure rises above the ventricles pressure, the atroventricular valves open and blood rushes into the ventricles. This is the rapid filling phase. The ventricles continue to relax with an initial decrease in ventricle pressure. The ventricle pressure eventually increases as blood continues to flow from the atria into the passive ventricles through the open atroventricular valves. 80-90% of the ventricle filling occurs during this phase. The
last 10-20% of blood is forced into the ventricles during atrial contraction [30]. The state following atrial contraction is considered end-diastole. The ventricles have reached their maximum volume and end-diastolic volume (EDV) is measured. At this point, the ventricles are experiencing the highest diastolic internal pressure and stress. End-diastole is a critical point in the cardiac cycle because ventricle filling directly affects the LV ability to pump blood systemically [2].

According to the CDC, 735,000 people in the United States experience a heart attack (myocardial infarction) every year. Of these, 210,000 have already experienced a first myocardial infarction (MI). With each successive event, mortality rate drastically increases [6]. These episodes generate tissue damage resulting in decreased compliance in the heart. Diastolic dysfunction can be caused by decreased compliance in the myocardium, leading to reduced EDV and compromised muscular contraction efficiency in effected tissue. A common cause of MI is a blockage in the coronary arteries. Prolonged oxygen deprivation can leave regions of the myocardium damaged and stiffened. The research study focused on understanding the LV prior to and following a MI. The LV experiences the highest pressure and stress and is therefore susceptible to failure. A better understanding of the mechanical behavior of the LV may provide insight into potential treatment options and failure mechanisms associated with heart failure. Such knowledge could help reduce the mortality rate of people who have suffered a MI [21].

The research study evaluated the mechanical behavior of the LV with and without the presence of an infarct. Cardiac tissue geometry was obtained from magnetic resonance images (MRI) of an explanted heart. The use of the MRIs for the research study was
approved by the institutional review board at California State University, Northridge and provided by The Visible Heart Laboratory. A geometrically realistic (natural) and symmetric LV geometric model was created from the MRIs in SOLIDWORKS [17, 24]. Past studies indicate MRI to be most accurate way of representing the complex LV geometry [33]. Statistical maps have been created which give average values at different locations in the LV [24]. This data can be used to develop general LV geometry, but does not capture the unique variation in wall thickness and shape found amongst specimens. A systematic method was developed for generating a computer model of the LV, and implementing mechanical property variations to represent the presence of an infarct. Infarct regions were defined by shape, extent, and location in the ventricular wall. The infarct regions were modeled as transmural, meaning they extended through the complete thickness of the myocardium. The size of the infarcts in this study were selected to cover a range of sizes based on the distribution observed in clinical studies of MI patients [11, 26]. Material properties were assigned to the myocardium based on experimental tensile testing on canine ventricular tissue [43]. This testing method provided a straight forward testing protocol with axial tensile tests in both the fiber and two orthogonal directions. The stress-strain curves produced were easily implemented into the material model. The experimental data allowed for the muscle fibers to be represented. One study performed axial tensile tests without considering the orientation of the fibers. This provided a stress-strain curve that could be implemented into an isotropic model. However, the data was not useful when attempting to represent the muscle fibers with different properties from the orthogonal directions, which has been demonstrated to be the case in experimental testing [19]. Shear testing on myocardial tissue cubes has been performed, providing
additional data on the myocardial tissue properties [9]. The physical meaning of these
data sets is less intuitive and the data presents challenges when attempting to incorporate
them into a material model. The finite element software, COMSOL Multiphysics, was
used to model the myocardium as an incompressible, exponential, transversely isotropic
material with a constant fiber direction through the myocardial wall for the normal and
infarct cases. This material model was selected because it represented the fibrous
structure of the myocardium and was implementable into COMSOL Multiphysics [38]. A
constant fiber direction was chosen due to the complex nature of the fibers in the LV and
lack of data relating to the fiber structure of the specimen analyzed in the MRI study.
Other studies have approximated the fiber structure using a rule based method [10, 39,
41]. Although this was expected to have an impact on the results, this study analyzed the
effects of varying infarct sizes rather than changes in fiber architecture. The material
models initially considered included: Mooney-Rivlin, Yeoh, and two transversely
isotropic. The material models were initially considered in the symmetric LV only. The
Mooney-Rivlin and Yeoh models were both isotropic. The two transversely isotropic
models had different first terms with the first model having a linear term and the second
an exponential. The transversely isotropic model with the exponential term was chosen
for the normal and infarct cases because it most closely represented the experimental data
and known properties of the LV. The transversely isotropic model with the linear first
term did not represent the experimental data as closely as the other transversely isotropic
model. The isotropic models were not capable of representing the stiffer behavior of the
myocardium in the fiber direction. An orthotropic model was used in a previous study to
describe the passive myocardium. This material representation was not considered in this
study because the variation in properties between the orthogonal directions. The infarct regions used the same material model but the parameters were chosen to be ten times stiffer. This was consistent with previous studies and produced strain values in the simulations consistent with medical imaging of infarct tissue [11]. EDV and wall stress were considered for different pressures on the inner wall of the symmetric and natural model. Similar parameters were considered in other studies [16, 38]. This study analyzed the stress in the longitudinal and fiber direction. The longitudinal stress was not presented in previous finite element studies. A constant pressure was applied to the inner wall of the model to simulate blood pressure. This condition assumed blood to be incompressible. Although blood is not completely incompressible, assuming incompressible was considered sufficient for this application [5]. The pressure load was applied in the same manner in previous finite element studies [16, 38]. Normal internal LV pressure and EDV were obtained from published literature [22]. The boundary conditions used a spring damper system on the base of the LV. This boundary condition allowed for some translation and dilation of the base, providing a more realistic representation of the restriction on the LV than a fixed base, which has been used in other finite element studies [10]. The results gave insight into the effects of infarct regions and hypertension on EDV. The effects of varying size of infarct on diastolic function were analyzed. Previous finite element studies have analyzed the LV with infarct, but did not consider varying location and size [36,41]. Other studies have attempted to analyze the stress and EDV in a healthy LV [16, 38]. This study also provided stress distribution and possible locations of stress concentration within the LV wall.
3. Modeling

The research study analyzed the mechanical behavior of the left ventricle (LV) of the human heart. This chapter describes the modeling process in detail. The MR images provided three dimensional coordinates of locations on the inner and outer surface of the LV. The coordinates were then used to develop a computer-aided design (CAD) model of the LV. A natural and symmetric LV model were created. The CAD models of the LV were imported into COMSOL Multiphysics where material properties were implemented into the LV geometry and finite element analysis was performed.

3.1 MRI Study

In order to analyze the mechanical behavior of the LV, an accurate geometric representation was created. To acquire the three dimensional coordinates of the inner and outer surfaces of the LV, an MRI study was performed. The MR images were provided by The Visible Heart Laboratory. The use of the MR images were approved by the IRB at California State University, Northridge. The MR images were acquired from a deceased patient heart of which the patient was not known to have any medical issues related to the heart.

The imaging technicians for The Visible Heart Laboratory had a protocol which was used to prepare and image the explanted human heart. Prior to imaging, the great vessels of the specimen were cannulated and the heart was perfusion fixed in 10% buffered formalin for a period of 24-48 hours under 40-50 mmHg such that the heart would retain a diastolic shape. After perfusion fixation, the heart was placed in a polymer
container containing 10% formalin and all air was removed in order to reduce imaging artifacts.

A Siemens 3 Tesla MRI was used to capture the heart images. The MR images provided short-axis and long-axis views of the deceased patient heart. Two different imaging settings were used to capture both the long-axis and short-axis images. A T1 weighted imaging sequence was used with a repetition time and echo time of 300 ms and 2.5 ms respectively. The imaging frequency was 123.26 Hz. The acquisition type was two dimensional. These scanning parameters were useful for demonstrating anatomy. White matter such as high fat tissue appear bright and compartments filled with water appear dark. The slice count for the T1 images were 35 with a slice thickness of 4mm. The second imaging protocol was a magnetization-prepared rapid gradient-echo (MP-RAGE) sequency. This sequence captures high tissue contrast and provides high spatial resolution of the whole heart. The repetition time and echo time for this sequence was 1810 ms and 4.3 ms respectively. The imaging frequency was 123.26 Hz. The acquisition type was three dimensional. The slice count was 120 and the slice thickness was 1.1 mm, providing greater image resolution. The images provided using the MP-RAGE protocol were used for the modeling process.

The MR images were viewed using RiadiAnt Digital Imaging and Communications in Medicine (DICOM) Viewer software. This tool allowed for the measurement of heart anatomy within the MR images by manual selection. The contrast of the images could also be adjusted. Angles and elliptical shapes could be measured within the DICOM viewer. This tool allowed for the images to be scaled based on length and to define a two dimensional point in spacing which could be expanded into three
dimensional space as the same point was marked in multiple layers. The accuracy of marked points was limited by the user, who manually selected the points. This error was reduced using the zoom feature to enhance the ability of the user to select the correct location.

3.2 CAD Model

Two models were created for the study; a symmetric model and a natural model using the MR images. The symmetric model was generated from average values of the LV. The natural model used data points acquired from the explanted heart. The myocardium wall thickness and curvature of the LV varies spatially, with the thickest region at the base and thinnest at the apex. There is also temporal variation in wall thickness through diastole. The LV has three distinct layers: endocardium, myocardium and epicardium. The endocardium lines the inner surface of the LV and provides a smooth, non-adherent surface for interaction with blood. The myocardium is the thick, muscular layer of the LV and plays an important role in diastolic function. The epicardium is the outermost layer and serves to protect the heart. This research study focused on the myocardium. The other two layers were expected to have less of an effect on the mechanical behavior of the LV during diastole. The wall of the symmetric model were given a smooth transition between wall thickness, the greatest wall thickness near the base and the smallest wall thickness near the apex. Two models were created to allow for comparison of results.

3.3 Symmetric Model

The symmetric model was created in SOLIDWORKS using average dimensions of the LV [17]. A cross section of the LV in two dimensions was traced and revolved
around a center axis to create a symmetric three dimensional LV model. Figure 3.1 showed the symmetric LV model.

Figure 3.1 Symmetric LV Model created in SOLIDWORKS
3.4 Natural Model

The natural CAD model was developed using MR images from an explanted heart. Images from the short-axis view of the MP-RAGE imaging sequence were imported into Solidworks from RadiAnt DICOM Viewer. Each image was marked at a predefined origin in the 2D frame of the image. This point was identified using pixel coordinates within the Radiant DICOM Viewer. The monitor used had a resolution of 1440x900 on a 15 inch screen. A measuring tool within Radiant DICOM was used to provide a scale for the image. After each image was imported into Solidworks, the origin was set for the predefined point on the image. The scale on the image was used to define the appropriate measurement of the image in SOLIDWORKS. This process was completed for each image. Each image corresponded to a distance below the base of the heart. When the process was completed it provided a series of stacked images in 3D space. Twelve short-axis images were used spanning the distance from base to apex of the LV model. A more accurate representation of the MR images in the CAD model could have been achieved by increasing the number of short-axis images imported from the MRI. This would decrease the spacing between the images and allow for more three dimensional data points. Long-axis views were added to reduce the error induced by the spacing between the MR images imported into SOLIDWORKS.

The inner and outer edge of the myocardium were traced using a style spline tool in SOLIDWORKS. The style spline allowed for manual manipulation of a curved line in a two dimensional plane. The accuracy of this process was limited by the resolution of the MR images and the ability of the user to trace the myocardium. This procedure was performed for each image. The completed splines provided data points representing the
geometry of the LV. A lofting tool was used to generate a volume enclosed by the inner and outer traced lines. This volume represented the LV myocardium. Anatomical features that were excluded from the model include: mitral valve, aortic valve, papillary muscles and chordae tendineae. The study did not consider the mechanical behavior of the valves. The boundary conditions placed on the LV model were chosen to simulate the behavior of the mitral valve during diastole. A spring damper system was applied to the base of the LV to restrict free body motion, while still allowing some translation and dilation. The papillary muscles and chordae tendineae are responsible for controlling inversion and prolapse of the mitral valve during ventricular contraction. These components are in a relaxed state and expected to have little effect on the mechanical behavior of the myocardium during diastole. Figure 3.2 and 3.3 show the completed natural model.

Figure 3.2 Anterior view of natural heart model
3.5 Infarct Regions

During a myocardial infarction (MI), regions of tissue in the LV can be deprived of oxygen, leading to infarct. This damage leads to decreased tissue compliance in the infarct. Infarct regions were added to the natural and symmetric CAD model within SOLIDWORKS. The infarcts were transmural, meaning they extended the full thickness of the myocardium. Transmural infarcts were chosen as they are common pathological features in MI patients. A cut tool was used to define an infarct domain within the CAD model. The infarct size was chosen to cover the spectrum of sizes observed in clinical studies of MI patients \([11, 13, 26]\). The shape of infarct can vary significantly between patients. This study used general shapes chosen to be similar to
those observed from medical imaging and biopsy. Six infarct regions were modeled for the symmetric model. Table 3.1 showed the percent infarct by volume for the symmetric model.

<table>
<thead>
<tr>
<th>Symmetric</th>
<th>Percent Infarct (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normal</td>
<td>0</td>
</tr>
<tr>
<td>Infarct 1</td>
<td>8</td>
</tr>
<tr>
<td>Infarct 2</td>
<td>9</td>
</tr>
<tr>
<td>Infarct 3</td>
<td>4</td>
</tr>
<tr>
<td>Infarct 4</td>
<td>16</td>
</tr>
<tr>
<td>Infarct 5</td>
<td>25</td>
</tr>
<tr>
<td>Infarct 6</td>
<td>51</td>
</tr>
</tbody>
</table>

Table 3.1 Infarct size for six cases in symmetric model

The location, size and shape of the six infarct cases for the symmetric model were shown in figure 3.4.

![Infarct cases](image)
Figure 3.4 Six infarct cases for symmetric model

Infarct 1-3 had similar size to eachother, but location was varied from the base to the apex. Three infarct regions were added to the natural model. Table 3.2 showed the percent infarct by volume for the natural model.

<table>
<thead>
<tr>
<th>Natural</th>
<th>Percent Infarct (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normal</td>
<td>0</td>
</tr>
<tr>
<td>Infarct A</td>
<td>7</td>
</tr>
<tr>
<td>Infarct B</td>
<td>15</td>
</tr>
<tr>
<td>Infarct C</td>
<td>50</td>
</tr>
</tbody>
</table>

Table 3.2 Infarct size for infarct cases in natural model
Figure 3.5 showed the location, size and shape of the three infarct cases for the natural model.

![Infarct A, Infarct B, Infarct C]

**Figure 3.5 Infarct size and location for natural model**

The locations of infarct A-C were chosen on the left side of the natural model for the three cases.

### 3.6 Material Properties

The LV mechanical properties observed in vivo were used to determine the material model that should be used to represent the myocardium. This research study focused on the mechanical behavior of the myocardium. The LV myocardium is known to have muscle fibers. The fibrous tissue displayed a different stress-strain relationship in different directions, indicating it behaves in a more complex fashion than a simple isotropic material. The myocardium is a soft biological tissue and it has been established experimentally the myocardium can be regarded as an incompressible material. This was determined by applying various amounts of hydrostatic stress to tissue specimens [37]. The myocardium has been known to be residually stressed following unloading after systole. Although this
was expected to have an effect on stress, quantifying residual stress is difficult in three
dimensions [27, 42]. For this study, residual stresses were not considered as a stress free
reference configuration must be established before residual stresses can be incorporated
into the model. The myocardium also appears to have viscoelastic behavior [12, 18]. Here,
viscoelastic effects were neglected as the viscoelastic response is long compared to the
short time scale of diastole. Little attention has been given to viscoelasticity in literature
and few data sets are available on the viscoelastic behavior of the LV myocardium.

Four models were initially considered for the analysis of the mechanical behavior
of the heart. The models included: Mooney-Rivlin, Yeoh, and two different exponential
models. The Mooney-Rivlin and Yeoh models were isotropic. The isotropic models were
incapable of capturing the fibrous nature of the LV wall. The exponential models allowed
for the fiber structure to be represented in the model. The two exponential models were the
same, aside from the first term. A linear and exponential first term were analyzed.

The material properties were represented using a strain energy density function.
The deformation gradient of the material was related to strain energy given by the function.
The myocardium undergoes large deformations and therefor hyperelastic material models
were considered. The derivative of the function with respect to strain gave the stress. The
Mooney-Rivlin and Yeoh model were material models commonly used for hyperelastic
materials with isotropic properties. The strain energy density functions for the Mooney
Rivlin and Yeoh model were visible in equations 1 and 2 [19].

\[ W = C_{10}(I_1 - 3) + C_{01}(I_2 - 3) \]  \hspace{1cm} \text{Eq. 1}

\[ W = \sum_{i=1}^{3} C_{i0}(I_1 - 3)^i + \sum_{i=1}^{3} \frac{1}{D_i}(J - 1)^{2i} \]  \hspace{1cm} \text{Eq. 2}
The transversely isotropic models were given by the strain energy density functions given in equation 3 and 4.

\[
W = \mu (I_1 - 3) + \frac{a}{2b} \left[ e^{b(I_4 - 1)^2} - 1 \right] + \frac{c}{2} (J - 1)^2
\]  

Eq. 3

\[
W = \frac{a}{2b} e^{b(I_4 - 1)^2} + \frac{af}{2bf} \left[ e^{bf(I_4 - 1)^2} - 1 \right] + \frac{c}{2} (J - 1)^2
\]  

Eq. 4

The first term represented the non-collagenous and non-muscular matrix within the tissue, including fluid. The second term represents the properties of the muscle fiber structure. The third term represented the compressibility factor of the tissue. The invariants found in the strain energy density functions were a function of the Cauchy-Green deformation tensor [14]. The right Cauchy-Green deformation tensor was found in equation 5.

\[
C = F^T F
\]  

Eq. 5

\[
J = \det(F)
\]  

Eq. 6

Where F was the deformation gradient tensor. The three invariants were found in Equation 7-9.

\[
I_1 = tr(C)
\]  

Eq. 7

\[
I_2 = \frac{1}{2} [I_1^2 - tr(C^2)]
\]  

Eq. 8

\[
I_4 = f_0 \cdot (Cf_0)
\]  

Eq. 9

Where \( f_0 \) represents the fiber direction in the local coordinate system. The stress was found from equation 10.

\[
\sigma = F \sum \frac{\partial W}{\partial I_i} \frac{\partial I_i}{\partial F} - pI
\]  

Eq. 10

Where p is a Lagrange multiplier related to the compressibility of the material [16].
The material constants were obtained from curve fitting to experimental results for axial stress-strain testing conducted in a previous study [43]. The study performed tensile testing on fresh canine tissue. In the previous study, fiber direction was determined by eye and 4x4 cm slices were excised from the canine myocardium. Thread was woven through the four sides to apply the necessary loading. The tissue was submerged in a potassium concentration to minimize unwanted contraction during testing. Two independent linear motors were used apply the loading on the tissue in two orthogonal directions. The applied force was measured by two transducers (Konigsberg F5-A). The data was collected from the behavior of the center one-third of the slice using video dimension analyzers which tracked four targets glued on the surface of the specimen. The outer edge displacements were measured using linear displacement transducers (Transtek). The tissue was preconditioned by subjecting it to several slow stretchings and unstretchings. This was done until the results were reproducible, usually requiring 5-7 cycles of a maximum load of 200 g. The initial position was unstretched and began when the tissue was relaxed to the point of slackening. The $E_{\text{ff}}/E_{\text{ss}}$ ratio of 1.0, meaning the tissue was equally strained in both orthogonal directions seems to be the most realistic to the loading protocol and also allowed for the largest range of data to be collected. A curve fitting was done for the fiber direction with the $E_{\text{ff}}/E_{\text{ss}}$ ratio of 1.0 to determine the parameters similarity to published values [43]. The constants for the transversely isotropic models were generated by curve fitting to the biaxial data for the $E_{\text{ff}}/E_{\text{ss}}$ ratio of 1.0. The experimental data was limited for lower strain values. Figures 3.7 and 3.8 display the stress-strain relationship for the experimental results. The curve fit for the material model and infarct stress-strain relationship was also visible in figures 3.6 and 3.7. The infarct regions in the model were modeled with the same
material properties, but as ten times stiffer. This was accomplished by multiplying the coefficient in front of the first two terms in the strain energy density function by ten. This stiffness was chosen to produce simulated strain values consistent with strain values determined from medical imaging of infarct regions [11,26].

Figure 3.6 Stress-strain relationship in the fiber direction (triangle: stress-strain curve obtained from experimental results of canine heart [43]; Square: Stress-strain relationship for the exponential, transversely isotropic model which was curve fit to the experimental data; Circle: Stress-strain relationship for the exponential, transversely isotropic model with increased stiffness to simulate infarct)
Figure 3.7 Stress-strain relationship in the longitudinal and radial direction
(triangle: stress-strain curve obtained from experimental results of canine heart [43];
Square: Stress-strain relationship for the exponential, transversely isotropic model which
was curve fit to the experimental data; Circle: Stress-strain relationship for the
exponential, transversely isotropic model with increased stiffness to simulate infarct)

The material parameters used for each model were found in Table 3.3.

<table>
<thead>
<tr>
<th>Mooney-Rivlin</th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>C10</td>
<td>12.71 kPa</td>
<td></td>
</tr>
<tr>
<td>C01</td>
<td>4.36 kPa</td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Yeoh</th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>C10</td>
<td>1.77 kPa</td>
<td></td>
</tr>
<tr>
<td>C20</td>
<td>26.32 kPa</td>
<td></td>
</tr>
<tr>
<td>C30</td>
<td>92.51 kPa</td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Linear Transversely Isotropic</th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>μ</td>
<td>3.5</td>
<td></td>
</tr>
<tr>
<td>a</td>
<td>2.28 kPa</td>
<td></td>
</tr>
<tr>
<td>b</td>
<td>9.726</td>
<td></td>
</tr>
<tr>
<td>c</td>
<td>28000 kPa</td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Exponential Transversely Isotropic</th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>a</td>
<td>2.28 kPa</td>
<td></td>
</tr>
<tr>
<td>b</td>
<td>9.726</td>
<td></td>
</tr>
<tr>
<td>a_f</td>
<td>1.685 kPa</td>
<td></td>
</tr>
<tr>
<td>b_f</td>
<td>15.779</td>
<td></td>
</tr>
<tr>
<td>c</td>
<td>28000 kPa</td>
<td></td>
</tr>
</tbody>
</table>

Table 3.3 Material parameters for strain energy density functions [16, 19]
The exponential, transversely isotropic function was the material model used to represent the infarct tissue. The material constants used in the infarct cases are shown in Table 3.4.

<table>
<thead>
<tr>
<th>Infarct</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>a</td>
<td>22.8 kPa</td>
</tr>
<tr>
<td>b</td>
<td>9.726</td>
</tr>
<tr>
<td>a_f</td>
<td>16.85 kPa</td>
</tr>
<tr>
<td>b_f</td>
<td>15.779</td>
</tr>
<tr>
<td>c</td>
<td>28000 kPa</td>
</tr>
</tbody>
</table>

**Table 3.4 Material parameters for infarct tissue**

### 3.7 Finite Element Implementation

The finite element (FE) software COMSOL Multiphysics (CM) was used to analyze the mechanical behavior of the LV with and without the presence of an infarct. The simulations were constructed to represent similar conditions experienced by the LV in vivo. It was expected the results would be influenced by parameters which have a direct physiological representation. For example, the general behavior of the mitral valve at the base of the LV was considered to define the boundary conditions imposed on the LV base in the simulations. These parameters were implemented into CM. Other parameters in the simulations were related to the FE analysis. Parameters such as the mesh and solver properties affected results without having a direct physiological relationship to the LV. This chapter details the FE analysis of the previously mentioned geometric models. Implementation of the natural model into the FE software is discussed in detail. The symmetric model was analyzed using the same process.
3.8 Coordinate System

The material model used for analysis of the LV was transversely isotropic. This made it necessary to define a local coordinate system in the model. The myocardium is known to have transmural variation in fiber orientation. Due to the convoluted nature of the fiber direction variation, a constant fiber direction was chosen for the FE analysis. This study analyzed the effects of infarct size rather than variations in fiber orientation. This allowed for the fiber structure to be represented in the simulations without overcomplicating the defined coordinate system. CM allowed for user defined coordinate systems based on the geometry of the model. The diffusion method was used to define the fiber direction of the LV. A scalar potential was automatically created by CM which forms the first base vector following the curvature of the geometry in a defined direction. This method was chosen for its computational efficiency and ease of implementation. Two orthogonal directions were manually selected within CM defining the first two coordinate system vectors. The first vector moved from the inner surface to the outer surface of the LV. This coordinate direction was defined as the radial direction. The second vector moved from base of the LV to the apex. This coordinate direction was defined as the longitudinal direction. The final coordinate direction was chosen as orthogonal to the first two defined directions. This vector was labeled as the fiber direction and moved helically around the LV. The three orthogonal vectors represented a local coordinate system which varied at every point in the myocardium of the LV. Figure 3.8 and 3.9 gave a visual representation of the defined local coordinate system. The three directions were denoted with a number. The fiber direction was number one. The longitudinal direction was number two. The radial direction was number three.
Figure 3.8 Defined local coordinate system in CM

Figure 3.9 Local coordinate system for point within the LV model
(Corresponding direction: Red with radial, green with longitudinal and blue with fiber)
The variables assigned by CM to the three orthogonal directions were changed in the local variables menu to f1, f2 and f3 corresponding to the radial, longitudinal and fiber direction respectively. The variable names were shortened to allow for a more compact representation of the parameters. Other CM assigned variables were changed in the local variables menu.

3.9 Mesh Generation

Mesh generation was an important part of the FE analysis. Poor mesh properties can make solutions error-prone. A combination of user-defined and automated tools were used to create the LV mesh. Important considerations were element size, distribution, quality and convergence of the solution. Refinement of the mesh was an iterative process.

CM provided a user-controlled mesh interface which allowed for maximum element size, minimum element size, maximum element growth rate, curvature factor and resolution of narrow regions to be manually defined. The software also offered predefined settings for various mesh densities. The selected element type was tetrahedral. Tetrahedral elements were chosen to capture the complex LV geometry while maintaining adequate element quality.

Prior to mesh refinement of the LV models, convergence was tested. The symmetric LV model was used to determine convergence. A rough mesh density was assigned to the symmetric LV model and the simulation was completed. The fiber stress value at several points on the inner wall of the LV model were recorded. The mesh density was increased and the same process was performed. This was done to ensure the values were converging and the program was working properly.
Further refinement of the mesh was performed on the natural LV model. To produce a mesh with similar sized elements throughout the geometry, the minimum and maximum element sizes were chosen to have a difference of 0.5 mm. The element growth rate limited the size change between adjacent elements. The effect of this parameter was limited by the minimal difference between the minimum and maximum element size. The curvature factor limited the size of elements along curved surfaces. The curvature factor and resolution of narrowing regions were found to again have limited effect on the mesh due to the predefined minimum and maximum element size. The mesh was generated automatically using the user-defined parameters listed in table 3.5. Figure 3.10 showed the mesh for the natural LV model.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum element size</td>
<td>.0045 m</td>
</tr>
<tr>
<td>Minimum element size</td>
<td>.0040 m</td>
</tr>
<tr>
<td>maximum element growth rate</td>
<td>1.6</td>
</tr>
<tr>
<td>curvature factor</td>
<td>0.7</td>
</tr>
<tr>
<td>resolution of narrowing regions</td>
<td>0.4</td>
</tr>
</tbody>
</table>

*Table 3.5 User-defined mesh parameters*

![Figure 3.10 Front view of natural LV model with mesh](image)
The mesh near the sharp edges of the LV geometry along the top of the model in figure 3.10 were not refined. The edges were in contact with the free body constraint and unrealistic artifacts were expected in the region. CM provided a statistical analysis of the generated mesh, giving values corresponding to mesh quality and size. The element quality parameter for the mesh was related to aspect ratio of the elements. The general distribution of the element quality was considered when refining the mesh. The distribution of element quality found in the mesh was weighted to the side of the ideal quality. Table 3.6 provided the element number and mesh statistics for the natural model.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of Elements</td>
<td>18844</td>
</tr>
<tr>
<td>Minimum element quality</td>
<td>0.1473</td>
</tr>
<tr>
<td>Average element quality</td>
<td>0.759</td>
</tr>
<tr>
<td>Element volume ratio</td>
<td>0.08595</td>
</tr>
<tr>
<td>Mesh volume</td>
<td>$1.188 \times 10^4 \text{ m}^3$</td>
</tr>
<tr>
<td>Maximum growth rate</td>
<td>3.189</td>
</tr>
<tr>
<td>Average growth rate</td>
<td>1.682</td>
</tr>
</tbody>
</table>

**Table 3.6 Mesh statistics for natural model**

### 3.10 Boundary Conditions

The boundary conditions were chosen to approximate the free body constraints and loading conditions the LV experiences in vivo. A common translational constraint used in heart modeling is a fixed base [10]. The surface where the translational constraint was applied is shown in figure 3.11. The blue slashes indicate the surface where the translational constraint was applied.
Medical imaging shows there is both translation and radial expansion at the base of the LV. This behavior was taken under consideration when establishing the boundary conditions. This research study constrained free body motion using a spring damper system for the base. This was considered a better representation of the constraints on LV motion and reduced artifacts induced by the fixed constraint. The spring damper constraints were applied to the whole surface of the LV base because this region is known to be constrained to some degree in vivo. The walls of the LV are free to expand during diastole with minimal constraint applied by the pericardial sac surrounding the heart. The initial spring damper parameters were selected to create a critically damped system, producing displacements which were physiologically reasonable. The critically damped parameters were chosen based on the simulated displacement and reduction of...
artifacts around the base, rather than physiological behavior. Figure 3.12 shows the displacement of the base of the natural model under normal conditions. The black line in figure 3.12 represents the undeformed base of the LV. The base translated 1mm vertically for 8 mm Hg internal pressure loading. Although not shown, the radial displacement of the inner edge of the base of the LV was 0.25 mm. The spring damper constants were visible in table 3.7. The stiffness was higher in the radial direction as the LV base was expected to expand radially less than it translated vertically.

![Figure 3.12 Translation of LV base under 8 mm Hg loading](image)

<table>
<thead>
<tr>
<th>Direction</th>
<th>Spring ((\text{N/(m}^2\text{m}^2)))</th>
<th>Damper ((\text{N}^\ast\text{s/(m}^2\text{m}^2)))</th>
</tr>
</thead>
<tbody>
<tr>
<td>radial</td>
<td>10,000,000</td>
<td>2366</td>
</tr>
<tr>
<td>vertical</td>
<td>1,000,000</td>
<td>748</td>
</tr>
</tbody>
</table>

Table 3.7 Spring Damper constant for constraint on LV base

The LV experiences an internal pressure from blood as it fills during end-diastole. LV pressure can fluctuate depending on the activity level of the individual or condition of the heart. The research study analyzed various LV pressures ranging from 8 mmHg to 20 mmHg to determine the effects of hypertension. Hypertension is a condition where blood pressure is consistently above normal levels. The normal hemodynamic parameters of 8-12 mmHg for the LV were obtained from literature [22]. The LV filling was simulated by
applying a uniform pressure over the entire inner surface of the LV. To represent the LV internal pressure as uniform it was assumed blood was incompressible. However, blood does have a degree of compressibility as it contains plasma, blood cells and platelets. It was considered reasonable to assume that in large vessels such as the LV, blood incompressibility was a good assumption [5, 8, 15].

3.11 Geometric Labeling System

Points were defined along the inner surface of the LV models. The defined points for the symmetric model were visible in figure 3.13. Points labeled with a number corresponded to a stress on the symmetric model. The blue region in figure 3.14 represented an infarct. Points 1-3 were on the side of the infarct for all symmetric infarct cases.

![Figure 3.13 Defined points for symmetric model](image)

The points for the natural model were visible in figure 4.1. Points labeled with a letter corresponded to a point on the natural model. The blue region in figure 3.14 represented an infarct. Points A-C were on the side of the infarct in the labeled
locations for all natural infarct cases. Stress values given at Points A-F were an average of ten values falling in a one centimeter radius around the defined point. This was chosen due to high spatial variation of stress in the natural model.

Figure 3.14 Defined points for natural model
The regions and cross-sections used for reference later in the chapter were defined in figure 3.15 and 3.16.

**Figure 3.15 Labeling system by region**

**Figure 3.16 Labeling system for natural model**
The defined planes allowed for a better understanding of the cross-sectional views of stress distribution. The stress in the fiber, longitudinal and radial direction was identified interchangeably with the variables shown in table 3.8.

<table>
<thead>
<tr>
<th>Variable</th>
<th>Direction</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\sigma_1$</td>
<td>Fiber</td>
</tr>
<tr>
<td>$\sigma_2$</td>
<td>Longitudinal</td>
</tr>
<tr>
<td>$\sigma_3$</td>
<td>Radial</td>
</tr>
</tbody>
</table>

Table 3.8 Defined variables for stress in the fiber, longitudinal and radial direction
4. Analysis

The results for this study are presented in this chapter. The first section described the mesh convergence process. A study was conducted to determine the spring damper coefficients for later implementation in the second section. The third section covered the material model comparison. The fourth chapter gave the results for the infarct simulations on the symmetric model. The fifth section provided the results for the natural model.

4.1 Mesh Convergence

A convergence test was performed on the symmetric model to determine if COMSOL Multiphysics was running properly and producing reasonable results. This involved a series of mesh refinements. The stress at three points in the fiber and longitudinal direction were recorded for each simulation. The stress values at a single point were expected to be closer as the mesh was refined. Five trials were conducted. The characteristics of the mesh for the five trials were visible in Table 4.1.

<table>
<thead>
<tr>
<th>Trial</th>
<th>Element #</th>
<th>Min Quality</th>
<th>Average Quality</th>
<th>Min Size (mm)</th>
<th>Max Size (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>1074</td>
<td>0.248</td>
<td>0.7061</td>
<td>2.33</td>
<td>12.5</td>
</tr>
<tr>
<td>2</td>
<td>3139</td>
<td>0.1493</td>
<td>0.7798</td>
<td>1.5</td>
<td>8.3</td>
</tr>
<tr>
<td>3</td>
<td>7173</td>
<td>0.2352</td>
<td>0.7451</td>
<td>0.831</td>
<td>6.65</td>
</tr>
<tr>
<td>4</td>
<td>22920</td>
<td>0.2166</td>
<td>0.7711</td>
<td>0.333</td>
<td>4.57</td>
</tr>
<tr>
<td>5</td>
<td>17273</td>
<td>0.2635</td>
<td>0.765</td>
<td>4.5</td>
<td>5</td>
</tr>
</tbody>
</table>

Table 4.1 Characteristics of the five meshes used to determine convergence for symmetric model
The stress values in the fiber and longitudinal directions for the five meshes were visible in Table 4.2.

<table>
<thead>
<tr>
<th>Trial</th>
<th>σ₁ (Pa)</th>
<th>Point 1</th>
<th>Point 2</th>
<th>Point 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>2423</td>
<td>4753</td>
<td>4379</td>
<td>973</td>
</tr>
<tr>
<td>2</td>
<td>2350</td>
<td>5736</td>
<td>4928</td>
<td>939</td>
</tr>
<tr>
<td>3</td>
<td>2373</td>
<td>5741</td>
<td>5662</td>
<td>814</td>
</tr>
<tr>
<td>4</td>
<td>2356</td>
<td>6151</td>
<td>5447</td>
<td>861</td>
</tr>
<tr>
<td>5</td>
<td>2354</td>
<td>6079</td>
<td>5430</td>
<td>894</td>
</tr>
</tbody>
</table>

Table 4.2 Fiber and Longitudinal stress values for the convergence test for symmetric model

4.2 Boundary Condition Considerations

A common boundary condition to restrict free body motion is to fix the base of the ventricle. For this study, a different approach was considered which used a spring damper system applied to the base of the LV. The symmetric model was used analyze the effects of varying spring damper parameters on the displacement and reduction in artifacts near the base of the LV. A pressure of 8 mmHg was applied to the inner wall of the symmetric model for the three spring damper trials and the fixed trial. The fixed trial completely restricted the motion of the surface of the base shown in figure T. The initial parameters were chosen to produce a critically damped system. These parameters were then altered to make the spring damper system more and less stiff. Table 4.3 showed the considered parameters and the corresponding displacement from the undeformed state of the base of the symmetric LV. The base of the LV for the three tests and the fixed boundary condition were visible in figure 4.1. The parameters listed for test 2 were used for the analysis of the symmetric and natural models in the simulations that follow.
<table>
<thead>
<tr>
<th>Trial</th>
<th>k radial</th>
<th>k vertical</th>
<th>b radial</th>
<th>b vertical</th>
<th>Vertical Displacement (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Test 1</td>
<td>1,000,000</td>
<td>100,000</td>
<td>748</td>
<td>236</td>
<td>7.5</td>
</tr>
<tr>
<td>Test 2</td>
<td>10,000,000</td>
<td>1,000,000</td>
<td>2366</td>
<td>748</td>
<td>1</td>
</tr>
<tr>
<td>Test 3</td>
<td>100,000,000</td>
<td>10,000,000</td>
<td>7483</td>
<td>2366</td>
<td>0.3</td>
</tr>
</tbody>
</table>

Table 4.3 Parameters for spring damper system

Figure 4.1 Displacement and stress for boundary condition parameters

4.3 Material Model Comparison

Four material models were implemented into the symmetric geometric model in COMSOL Multiphysics (CM). The models included: Mooney-Rivlin, Yeoh, linear transversely isotropic and exponential transversely isotropic. The stress values in the fiber, longitudinal and radial direction were displayed for each model. The spring damper constraint with the critically parameters were applied to the base of the LV. A pressure of 8 mmHg was applied to the inner surface of the symmetric model. The fiber stress values for the four models were found in table 4.4, the longitudinal stress values were found in table 4.5.
<table>
<thead>
<tr>
<th>Material Model</th>
<th>Point 1</th>
<th>Point 2</th>
<th>Point 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mooney-Rivlin</td>
<td>1590</td>
<td>2481</td>
<td>2195</td>
</tr>
<tr>
<td>Yeoh</td>
<td>1881</td>
<td>3847</td>
<td>3571</td>
</tr>
<tr>
<td>Linear Transversely Isotropic</td>
<td>1628</td>
<td>3016</td>
<td>2905</td>
</tr>
<tr>
<td>Exponential Transversely Isotropic</td>
<td>2353</td>
<td>6062</td>
<td>5428</td>
</tr>
</tbody>
</table>

**Table 4.4 Stress in fiber direction for four material models**

<table>
<thead>
<tr>
<th>Material Model</th>
<th>Point 1</th>
<th>Point 2</th>
<th>Point 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mooney-Rivlin</td>
<td>225</td>
<td>622</td>
<td>418</td>
</tr>
<tr>
<td>Yeoh</td>
<td>704</td>
<td>1215</td>
<td>1739</td>
</tr>
<tr>
<td>Linear, Transversely Isotropic</td>
<td>375</td>
<td>933</td>
<td>1419</td>
</tr>
<tr>
<td>Exponential, Transversely Isotropic</td>
<td>894</td>
<td>2498</td>
<td>3110</td>
</tr>
</tbody>
</table>

**Table 4.5 Stress in longitudinal direction for four material models**

The highest stress magnitude was in the fiber direction at the middle region for the four models. There was greater difference between the stress in the fiber and longitudinal direction for the transversely isotropic models compared to the isotropic models. The largest stress in the longitudinal direction was near the apex. The exponential, transversely isotropic model predicted higher stress values than the other models. The stress distribution images were shown for the four material models in figures 4.2-4.5.
Figure 4.2 $\sigma_1$ (left), $\sigma_2$ (middle) and $\sigma_3$ (right) for Mooney-Rivlin model

Figure 4.3 $\sigma_1$ (left), $\sigma_2$ (middle) and $\sigma_3$ (right) for Yeoh model
Figure 4.4 $\sigma_1$ (left), $\sigma_2$ (middle) and $\sigma_3$ (right) for linear, transversely isotropic model

Figure 4.5 $\sigma_1$ (left), $\sigma_2$ (middle) and $\sigma_3$ (right) for exponential, transversely isotropic model
The stress distribution in the fiber and longitudinal direction was highest on the inner surface of the LV for the four models. The stress artifacts near the base were effectively reduced in the fiber direction for the four models.

4.4 Symmetric Model

Simulations were performed on the symmetric model to analyze the behavior of the infarct and normal cases for the normal pressure loading of 8 mm Hg. The spring damper system was applied to the base of the LV using the critically damped parameters. The EDV was recorded for each case at a pressure of 8 mmHg. The results showed a decrease in EDV for the infarct cases. An additional simulation was performed for each infarct case to determine the pressure required to return the EDV in the infarct cases to the EDV in the normal case. The new EDV for the increased pressure was denoted the restored EDV. The largest infarct case resulted in the greatest decrease in EDV. Table 4.6 gave the EDV, restored EDV and pressure results for the simulations.

<table>
<thead>
<tr>
<th>Symmetric</th>
<th>EDV at 8 mm Hg (mL)</th>
<th>Restored EDV (mL)</th>
<th>Pressure (mmHg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normal</td>
<td>128</td>
<td>128</td>
<td>8</td>
</tr>
<tr>
<td>Infarct 1</td>
<td>123</td>
<td>128</td>
<td>11</td>
</tr>
<tr>
<td>Infarct 2</td>
<td>123</td>
<td>128</td>
<td>11</td>
</tr>
<tr>
<td>Infarct 3</td>
<td>125</td>
<td>128</td>
<td>9</td>
</tr>
<tr>
<td>Infarct 4</td>
<td>120</td>
<td>128</td>
<td>13</td>
</tr>
<tr>
<td>Infarct 5</td>
<td>117</td>
<td>128</td>
<td>17</td>
</tr>
<tr>
<td>Infarct 6</td>
<td>111</td>
<td>123</td>
<td>20</td>
</tr>
</tbody>
</table>

Table 4.6 EDV at 8 mmHg, restored EDV and pressure for symmetric model

The stress in the LV wall was analyzed for the infarct cases at the pressure in table 4.7. The stress values were determined for three different regions on the inner wall of the LV. The three points on the infarct side and three points opposite the infarct
previously defined. The stress in the fiber direction for the normal and infarct cases at the selected points were shown in table 4.7.

<table>
<thead>
<tr>
<th>Model</th>
<th>Point 1</th>
<th>Point 2</th>
<th>Point 3</th>
<th>Point 4</th>
<th>Point 5</th>
<th>Point 6</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normal</td>
<td>2773</td>
<td>5976</td>
<td>5448</td>
<td>2753</td>
<td>5995</td>
<td>5492</td>
</tr>
<tr>
<td>Infarct 1</td>
<td>1705</td>
<td>9080</td>
<td>5320</td>
<td>3420</td>
<td>8329</td>
<td>7890</td>
</tr>
<tr>
<td>Infarct 2</td>
<td>1840</td>
<td>8704</td>
<td>6984</td>
<td>3108</td>
<td>8432</td>
<td>7459</td>
</tr>
<tr>
<td>Infarct 3</td>
<td>2873</td>
<td>6334</td>
<td>5594</td>
<td>4039</td>
<td>7234</td>
<td>6085</td>
</tr>
<tr>
<td>Infarct 4</td>
<td>6750</td>
<td>9645</td>
<td>7333</td>
<td>5475</td>
<td>10565</td>
<td>9175</td>
</tr>
<tr>
<td>Infarct 5</td>
<td>7778</td>
<td>11929</td>
<td>12098</td>
<td>6800</td>
<td>15461</td>
<td>11323</td>
</tr>
<tr>
<td>Infarct 6</td>
<td>4428</td>
<td>10280</td>
<td>9839</td>
<td>7440</td>
<td>20516</td>
<td>18037</td>
</tr>
</tbody>
</table>

Table 4.7 Stress in the fiber direction for symmetric model with infarct cases

The stress distribution in the fiber direction for the normal and infarct cases were shown in figure 4.6.
The highest stress in the fiber direction was found in the middle region of the ventricle on the inner surface. The stress was increased on the inner surface of the wall away from the infract region compared to the normal model. The artifacts near the base of the LV were effectively reduced in the fiber direction, even for high pressures. Artifacts were present near the border where the material properties change between normal and infarct. The stress in the longitudinal direction for the normal and infarct cases at the selected points were shown in table 4.8.

Figure 4.6 $\sigma_1$ for symmetric model (a. normal; b-h corresponded to infarct cases 1-6 respectively)
The stress distribution in the fiber direction for the normal and infarct cases was shown in figure 4.7.
Figure 4.7 $\sigma_2$ for symmetric model (i. normal; j-o corresponded to infarct cases 1-6 respectively)

The greatest stress in the longitudinal direction was found near the apex on the inner surface. This was true for the normal and infarct cases 1-6. Artifacts were present near the base of the LV in the longitudinal direction.

4.5 Natural Model

Simulations were performed on the natural model to analyze the behavior for infarcts A-C and the normal case for the pressure loading of 8 mmHg. The spring damper system was applied to the base of the LV using the critically damped parameters. The EDV was
recorded for each case at a pressure of 8 mmHg. The results showed a decrease in EDV for the infarct cases. An additional simulation was performed for each infarct case to determine the pressure required to return the EDV in infarct cases A-C to the EDV in the normal case. The new EDV for the increased pressure was denoted the restored EDV. The reduction in EDV increased with increased infarct size for the simulations. Table 4.9 gave the EDV and pressure results for the simulations.

<table>
<thead>
<tr>
<th>Natural</th>
<th>EDV (mL)</th>
<th>Restored EDV (mL)</th>
<th>Pressure (mm Hg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normal</td>
<td>143</td>
<td>143</td>
<td>8</td>
</tr>
<tr>
<td>Infarct A</td>
<td>137</td>
<td>143</td>
<td>10.5</td>
</tr>
<tr>
<td>Infarct B</td>
<td>134</td>
<td>143</td>
<td>12.75</td>
</tr>
<tr>
<td>Infarct C</td>
<td>126</td>
<td>140</td>
<td>20</td>
</tr>
</tbody>
</table>

Table 4.9 EDV, restored EDV and pressure for the natural model with infarcts

Increased size of infarct resulted in a decrease in EDV. The EDV for infarct C was not restored even with the high internal pressure of 20 mmHg. The stress in the LV wall was analyzed for the infarct cases at the pressures shown in table 4.11. The stress values were acquired from three different regions on the inner wall of the LV. Three points on the infarct side and three points opposite the infarct defined previously. Due to large variation in stress spatially for the natural model, ten stress values from a 10 mm radius around the defined points were used and the averages were presented in table 4.10.
The stress in the fiber direction were shown in figure 4.8. The cross sections were along the 0*-180* plane.

<table>
<thead>
<tr>
<th></th>
<th>Point A</th>
<th>Point B</th>
<th>Point C</th>
<th>Point D</th>
<th>Point E</th>
<th>Point F</th>
</tr>
</thead>
<tbody>
<tr>
<td>Natural</td>
<td>3500</td>
<td>7100</td>
<td>8500</td>
<td>2700</td>
<td>6500</td>
<td>8100</td>
</tr>
<tr>
<td>Infarct 1</td>
<td>2000</td>
<td>9100*</td>
<td>11200</td>
<td>2900</td>
<td>8500</td>
<td>10500</td>
</tr>
<tr>
<td>Infarct 2</td>
<td>3000</td>
<td>9400*</td>
<td>4400*</td>
<td>6000</td>
<td>9000</td>
<td>10000</td>
</tr>
<tr>
<td>Infarct 3</td>
<td>3900*</td>
<td>13000*</td>
<td>12100</td>
<td>6500</td>
<td>18000</td>
<td>22500</td>
</tr>
</tbody>
</table>

Table 4.10 Stress in the fiber direction for the infarct cases in the natural model

(* represented stress values found inside the infarct region)
The peak stress values were found on the inner surface for both the normal and infarct cases A-C in the middle region. The stress values had large spatial variation moving from the base to the apex. Minimal artifacts were seen near the base. Artifacts were present near the border where the material properties changed between normal and infarct. The stress was highest in the direction of the fibers. The longitudinal stress values for the normal and infarct cases at the selected points were presented in table 4.11.

<table>
<thead>
<tr>
<th>Natural</th>
<th>Point A</th>
<th>Point B</th>
<th>Point C</th>
<th>Point D</th>
<th>Point E</th>
<th>Point F</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normal</td>
<td>800</td>
<td>3300</td>
<td>4100</td>
<td>800</td>
<td>3100</td>
<td>3200</td>
</tr>
<tr>
<td>Infarct A</td>
<td>600</td>
<td>5100*</td>
<td>4800</td>
<td>1000</td>
<td>3800</td>
<td>5100</td>
</tr>
<tr>
<td>Infarct B</td>
<td>3300</td>
<td>8000*</td>
<td>3000*</td>
<td>1400</td>
<td>5500</td>
<td>6800</td>
</tr>
<tr>
<td>Infarct C</td>
<td>1500</td>
<td>6500*</td>
<td>8500</td>
<td>4500</td>
<td>12000</td>
<td>13200</td>
</tr>
</tbody>
</table>

Table 4.11 Stress in longitudinal direction for the infarct cases in the natural model

(* represented stress values found inside the infarct region)
The longitudinal stress images were presented in figure 4.9. The cross sections were along the $0^\circ$-$180^\circ$ plane.

Figure 4.9 $\sigma_2$ for natural model with infarct cases A-C
The stress in the longitudinal direction was highest near the apex for the natural model. There were large variation in stress spatially. Artifacts were observed near the base of the LV and the border of the normal and stiffened material representations.
5. Discussion

The mechanical behavior of the LV for the symmetric and natural geometric model was analyzed with and without the presence of an infarct. The results of the two geometric models were considered separately and compared. A correlation between the results and the physiological relevance was also discussed. This was to provide insight into possible failure mechanisms in Myocardial Infarction (MI) patients associated with mechanical properties. Additional studies were performed to determine the effects of the boundary conditions, solution convergence and different material properties on results. The magnitude of stress values and end-diastolic volumes calculated in COMSOL Multiphysics (CM) were not considered to be an accurate representation of what would be seen in vivo. Rather, the results for the infarct cases in the symmetric and natural model were compared to the normal case for the symmetric and natural model.

As the mesh was refined in the convergence test for the symmetric model, the stress in both the fiber and longitudinal direction for points 1-3 appeared to converge to similar values. This was an indication the software was running properly and producing consistent results. The areas of higher stress, points 2 and 3, were more sensitive to mesh changes suggesting a more refined mesh was needed in these areas to capture stress variation. Mesh five had a reduced element count and a smoother mesh than mesh four, while still producing similar stress values in both directions. This indicated mesh refinement could be performed in CM to reduce computational demand of the simulations without compromising accuracy of results.
Unrealistic stress artifacts near the base in the fiber direction were effectively reduced for the spring damper constraint compared to the fixed constraint. This suggested the spring damper constraint may be a better free body constraint for the simulations than the fixed base constraint. The spring damper constraint had minimal effect on the stress values away from the base, further supporting this. The displacement of the base for the parameters in test 2 was physiologically reasonable, indicating it was adequate for implementation in the study. The parameters for test 1 produced a base displacement larger than was justifiable for further use. The spring damper constraint reduced stress artifacts near the base for the infarct cases suggesting it was useful in simulations where infarcts were considered.

The four material models considered to represent the myocardium in the symmetric model included: Mooney-Rivlin, Yeoh and the two transversely isotropic. Similar stress distribution was observed between models while magnitudes varied significantly. This suggested the different material models and parameters had a greater effect on stress magnitude than distribution. A greater difference in stress magnitude between the fiber and longitudinal direction in the transversely isotropic models compared to the isotropic models was observed. This suggested the isotropic models failed to capture stress magnitude variations between the fiber and longitudinal direction which is clearly displayed in experimental testing [43]. The linear, transversely isotropic model predicted lower stress magnitudes compared to the exponential transversely isotropic model. The linear isotropic term approximated the exponential stress-strain behavior less closely than the exponential transversely isotropic model suggesting the latter model was the better representation of the mechanical properties of the
myocardium. For this reason the exponential transversely isotropic model was implemented in the infarct cases for the symmetric and natural model.

For the symmetric infarct cases 1-6, table 4.6 showed the EDV was found to decrease for all infarct cases compared with the normal symmetric model. The largest infarct resulted in the greatest decrease in EDV. This indicated the presence of a larger infarct size more greatly reduce EDV. Infarct 1-3 were similar in size, but varied in location from the base to apex. The reduction in EDV was similar for the three cases. This suggested location of the infarct had less effect on EDV reduction than size of infarct for the same pressure load. The increased pressure to restore normal EDV resulted in an increase in stress in the fiber and longitudinal direction away from the infarct.

For the natural infarct cases A-C, table 4.9 showed the EDV decreased for all infarct cases compared with the normal natural model. The EDV decreased for increasing infarct size for the internal loading of 8 mmHg. This suggested larger infarct size may have a larger effect on EDV reduction. The increased pressure required to restore the simulated EDV resulted in increased stress away from the infarct. The highest stress in the fiber direction was in the middle region away from the infarct suggesting this may be an area of interest. The size of infarct in cases A-C had little effect on stress distribution away from the infarct. This indicated variation in infarct size had a greater effect on EDV and stress magnitude than stress distribution. The unrealistic stress artifacts near the material change border for the infarct cases indicated stress values near these regions were not useful for consideration in the results. The large changes in stress spatially suggested geometric features directly affected stress results.
The stress distribution for the symmetric model was smooth from the base to apex. The stress distribution for the natural model had large spatial variation, with areas of higher and lower stress in relatively close proximity. This suggested geometric features influenced both stress distribution and magnitude in the simulations, emphasizing the importance of accurately modeling the geometry from the MR images. The infarct size had a greater effect on reduction of EDV in the natural model than the symmetric model. Both the natural and infarct model had similar stress distribution, highest on the inner wall and decreasing moving to the outer wall. This suggested variations in geometric features had a greater effect on stress magnitude than transmural variation.

The results of the symmetric and natural model were compared. Several differences were found between the two geometric models. The natural model had large spatial variation, with areas of higher and lower stress in relatively close proximity. The symmetric model had a smooth transition between stress values moving from the base to apex of the LV model. This was expected to be a result of the geometric differences between the two models. Geometrically, the symmetric model was thickest at the base and decreased in thickness moving towards the apex. The natural model did not have the same smooth variation in wall thickness. Depending on the area of interest, the wall thickness varied. As a general trend, the wall thickness was greatest at the base and lowest near the apex for the natural model. The wall thickness of the base of the symmetric model was also 1.5 mm thicker than the natural model. This was expected to relate to the lower stress values predicted by the symmetric model compared to the natural. Geometry has been shown to have a significant effects on stress results [39]. This
would suggest the results from the natural model for analysis of the mechanical behavior of the LV were more useful than the symmetric model when correlating them to LV physiology.

It has been observed in clinical studies decreased compliance in LV tissue leads to a decrease in EDV [2]. The results of the study showed a decrease in EDV for the infarct cases. This suggests agreement with clinical studies of MI patients and the observed results in the study. As EDV is decreased, the stroke volume can also be decreased. The stroke volume is the amount of blood ejected from the ventricle during systole. The simulations showed a decrease in EDV for the three cases, suggesting decreased compliance in the tissue will result in a decrease in EDV. The larger the infarct region, the greater the decrease in EDV for a constant pressure of 8 mm Hg. Less ventricular filling consequently leads to a greater end-diastolic pressure. This is an attempt by the body to restore normal cardiac output [21]. The restorative pressures for infarct A-C were 10.5, 12.75 and 20 mm Hg. Infarct 2 and 3 fell outside normal LV pressure. It has been observed clinically increased diastolic pressure can result in increased left atrial and venous pressures. This can lead to additional side effects such as pulmonary congestion and edema.

In this study, a pressure was applied to restore the simulated EDV for the infarct cases. The results showed an increase in stress on the inner wall away from the infarct. Hypertension is observed clinically in MI patients with present infarcts. The increased LV pressure can restore systolic function, but it does not normalize diastolic wall stress. The stressed tissue can result in cardiomyopathy. This is a condition where over stressed tissue becomes enlarged. The LV receives most of its blood supply through the coronary
arteries during diastole. Over stressed and enlarged tissue has decreased pumping
efficiency and requires a larger oxygen supply. The stress results in this study may
provide insight into the stress distribution and concentrations associated with MI patients.

The research study had limitations that could be improved upon in future studies.
The local coordinate system represented the orientation of fibers seen in vivo as constant.
The local coordinate system could be improved by accounting for the known transmural
variation in fiber orientation. The material model considered the myocardium to be
transversely isotropic and consequently the longitudinal and radial direction were
assigned the same material properties. Experimental testing data of myocardial tissue is
limited. Additional experimentation on the stress-strain data for myocardial tissue would
allow for a better representation of the LV myocardium. The MR imaging resolution and
the ability of the user to identify cardiac geometry limited the accuracy of the geometric
model. Further attention could be placed on improving the geometric representation of
the LV from the MR images. LV motion is restricted in vivo by multiple components
including the pericardial sac, veins and arteries. In the study a spring damper system was
implemented in the model to restrict motion. This method was considered a better option
than the commonly used fixed base, but could still use improvement to more accurately
represent the free body restrictions on the LV. Fluid-structure interactions were not
considered in the study. It was expected these interactions would have some effect on the
behavior of the finite element model, specifically on the inner wall.

The concluding thoughts were infarct regions led to a decrease in EDV in the
simulations. The larger the infarct, the greater reduction in EDV. The increase in pressure
required to restore EDV significantly increased wall stress opposite the infarct. The stress
distribution varied spatially for the natural model suggesting the geometry of the model
directly influenced the behavior of the simulations. This should be taken into
consideration in future studies.
References


