FEA Simulation of the Heart

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Abstract: At the heart of every simulation is a question about the natural world. The ultimate success of the simulation is then measured by how satisfactory the answer is afforded by the simulation. In particular, many numerical simulations of engineering systems seeks to provide a good approximation to an engineering quantity that is frequently difficult and costly to ascertain in the physical world, such as core temperature of a nuclear reactor when a certain design element is changed, or the maximum blood pressure attainable if the ascending aorta of a human heart is occluded. The answers to these questions provide insight to the operating condition of the engineering design or the natural system, and therefore are essential to the successful execution of an engineering design endeavor.

The objective of this paper is to present a case study of the simulation of a heart to illustrate the core issues to obtaining useful simulation results using finite element based numerical modeling and simulation. This paper will discuss the question on the interaction forces between the human heart and the valvular prosthesis to illustrate how the core issues are dealt with in the biomedical device numerical simulation practice.

Keywords: Heart Valve Modeling, Ventricular Mechanics, Heart Simulation, Verification and Validation, FEA, CFD, Device-Anatomy Interaction, Design Optimization, Hyperelasticity, Implantable Medical Device, Tissue, Holzapfel Model, Tissue Activation Model.

1. Introduction

In cardiovascular medicine, the field of heart valve replacement has evolved significantly over the past half century to a level where prosthetic valves are highly reliable and effective in restoring patients' health and saving lives. However, the development of mechanical and biological tissue valves still falls short of the ultimate goal of mimicking human native valve performance throughout a patient's entire life span. Mechanical prostheses impart outstanding durability but require life-long anticoagulation therapy. Problems associated with thromboembolism and hemorrhage in mechanical valve recipients also affect long-term morbidity and mortality, and significantly impact the quality of life for patients. Biological tissue heart valves made of porcine xenograft and bovine pericardial tissue have emerged to improve the patient's overall outcome. In particular, bovine pericardial valves have become the most frequently used heart valve prosthesis in the clinical practice due to their superior hemodynamics and device hemocompatability. The long term durability for tissue valves have also been improved significantly in recent years due to advances in tissue treatment related to calcification mitigation. Most recently, the development of a less-invasive transcatheter heart valve replacement system revolutionized the valvular disease management by providing greater accessibility to patients that would otherwise not be candidates for traditional open-heart surgery. In this new treatment, the heart valve implantation is performed

by threading the collapsed heart valve prosthesis through a small incision in the groan via the patient's artery, thus avoiding the mortality and morbidity associated with more invasive open heart procedure.

Throughout the history of the heart valve development, computational mechanics has played an important role in the mechanical reliability assessment, hemodynamic optimization, and understanding of the device-blood interaction in preventing blood damage and tissue degradation. While significant progress has been made in the modeling of the implantable device, interaction between the device and the anatomic structure and blood-device interaction continue to present new challenges in the device simulation. Challenges in the development of practical accurate tissue material models and the need for validation of the computational models have also received increased attention over the recent years.

The objective of this paper is to present a case study of the simulation of heart to illustrate the core issues to obtaining useful simulation results using finite element based numerical modeling and simulation. The critical elements in simulation of heart include geometrical representation and approximation, material constitutive representation and approximation, numerical simulation strategy, and relationship between simulation results and the fundamental question of the simulation task. This paper will discuss the question on the interaction forces between the human heart and the valvular prosthesis to illustrate how the core issues are dealt with in the biomedical device numerical simulation practice.

2. Method

2.1 Description of General Modeling Procedure

The first step in planning a complex FEA simulation project is to identify the potential challenges in both the physical problem formulation and the numerical implementation of the simulation. Human heart anatomy is very complex. The geometric representation must retain sufficient detail to keep the structural response intact, but also need to be general such that the conclusion can have more general applicability, rather than to the specific heart modeled. In our study, the initial geometry was obtained based on actual CT scans and reconstruction of the heart geometry. Even though the interest was initially limited to the left ventricular mechanics, it was quickly realized that the proximity of the right ventricle, the atria and the major cardiac vessels emanating from the heart would play an essential role in the device-anatomy interaction. The model, therefore, needs to include the whole cardiac structure.

The constitutive behavior of the heart tissue is also complex. The myocardium not only exhibits anisotropic nonlinear stress-strain response, but also behaves as an actuator when electrically activated. A user-defined anisotriopic hyperelastic material model has been developed to model skeletal muscles [1], but application of the material model to cardiac muscle requires careful material calibrations; a method was developed in this investigation to simulate the tissue contraction with thermal strain activation. For this purpose, a versatile tissue constitutive model, the Holzepfel anisotropic hyperelastic model[2], was combined with an orthotropic thermal expansion model to simulate the anisotropic nonlinear tissue mechanics with anisotropic active tissue contraction.

It was also recognized that the contact is a significant challenge in this case study due to the complex geometry, large local curvature variation and device-anatomy interaction.

The physiological pressure and cardiac cycle are well documented in the physiology textbook. It is also recognized that the large variations exists among the population and affected by the pathophysiology. For illustrative purpose, we adopted a pressure waveform from standard textbook [3]. For realistic simulation these conditions must be obtained with the application in mind.

Finally, once all the challenges were identified, several sub-models were built to tackle each of the individual challenging questions before the full model was integrated.

2.2 Geometrical Consideration of the Heart

Advances in three dimensional cardiac imaging in recent years have provided the biomedical engineers with rather precise 3-D geometry of the heart. However, the raw dataset created for image rendering is often noisy and not amenable for finite element modeling. Often significant surface smoothing is required to make the model friendlier to the finite element meshing algorithm. Meshing of the heart is accomplished using a versatile quadratic tetrahedral element. Two meshes with different densities were generated. The fine mesh model did not show significant difference from the standard mesh model (Figure 2.2.1).
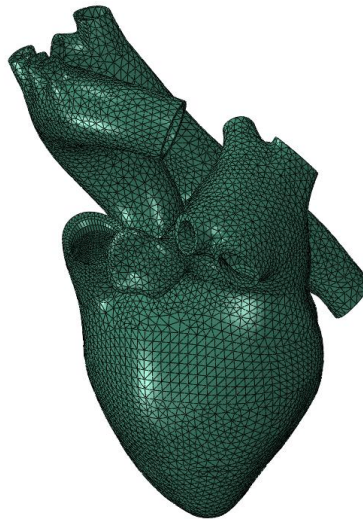

Figure 2.2.1 – Final Mesh

2.3 Tissue Constitutive Property

As discussed previously, the tissue material model was a hybrid model of the Holzapfel anisotropic hyperelastic model with anisotropic thermal expansion activation. The Holzapfel model, initially developed for artery walls, describes the stain energy potential in terms of strain invariants[2]. It is based on the tissue fibrous ultra-structure. In this case study, one family of myocardium fibers was specified following the direction of a helical spline surrounding the heart geometry (Figure 2.3.1).
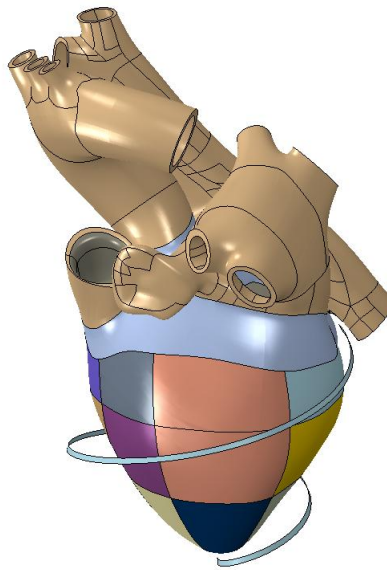
Figure 2.3.1 – Partitions Shown with Helical Spline

2.4   Ventricular Contraction Mechanics

The thermal contraction provides a simple approximation of the myocardium contraction. The specification of the thermal expansion coefficients has no physical meaning. However, the combination of the expansion coefficient and the temperature differential results in a constant tissue contraction. This contraction is calibrated to match the desired left ventricular volume contraction. The fiber orientation also controls the direction of the myocardium contraction.

2.5   Boundary Conditions and Loading Prescription

The wet surfaces of the heart model were all prescribed with pressure boundary conditions. The external surface of the heart model was supported with an elastic foundation to simulate the interaction with other organs in the thoracic cavity.

The ventricular surfaces were modeled with surface-based fluid cavities. They are connected to the distal cardiac vessels with fluid link elements that represent the function of the cardiac valves.

During systole, the activation of the ventricles is achieved by prescribing a temperature waveform to all the nodes on the ventricular walls. The near-incompressibility of the blood results in the rapid pressure rise within the ventricles during the initial iso-volumic contraction phase. Once the ventricular pressure reaches the same pressure as the distal cardiac vessels, the aortic and pulmonic valves open, resulting in cardiac ejection. The blood flow during systolic ejection is modeled by the fluid link elements with a small pressure gradient across the aortic and pulmonic valves. Diastolic relaxation of the ventricles is achieved with the temperature activation returning to the initial condition. No atria contraction was modeled in this study, since we are mostly interested in the maximum interaction force between the cardiac anatomy and the implantable device during the vigorous contraction phase of the cardiac cycle.

All cardiac valves are modeled with shell elements for simplicity. The mitral valve and tricuspid valve are also modeled with chordae and papillary muscle for the proper systolic functioning of these valves (Figure 2.5.1). The chordae were modeled with link connector elements [4].
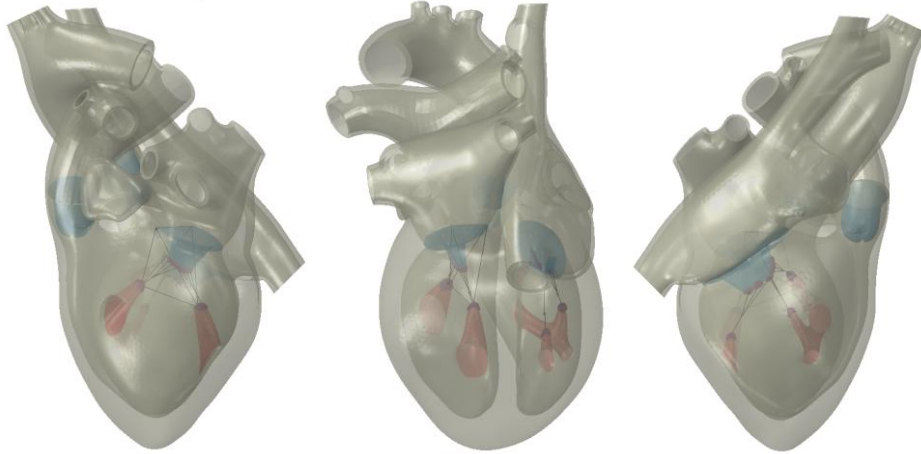

Figure 2.5.1 – Heart Model Showing Papillary, Valves and Chordae (connector elements)

2.6   Solver Selection

Abaqus/Explicit was used for robustness, scalability and the dynamic nature of the problem. In order to remove low-frequency vibrations, moderate amount of mass-proportional damping was specified. The analysis takes about 4 hours to complete using a computer of 12 core cpus.

2.7   Model Calibration and Verification

As we focus on the main objective of the simulation to investigate the anatomy-device interaction, proper calibration is essential. As a starting point, we compared the model simulation output with the measured dimensional quantities of a porcine study. As shown in Figures 3.3.1 and 3.3.2 below, the Abaqus results align well with documented porcine study data. Measurement data for implanted markers was obtained during the systolic compression phase. These marker locations were incorporated in to the Abaqus model via reference points and the distances between them were monitored with connectors and their associated output requests [5, 6].   The results are shown in the following table.

Systole Dist. (mm)	Abaqus run	Documented data	%diff
APM_2_PPM	24.1	22.6	6.5%
PPM_2_ATRIG	41.5	35.8	14.7%
PPM_2_PTRIG	38.6	33	15.7%
APM_2_ATRIG	26.5	23.3	12.7%
APM_2_PTRIG	35.7	33.5	6.5%
ATRIG_2_PTRIG	21.6	20.1	7.0%

Table 2.7.1 – Abaqus Results vs. Documented Marker Distances

3.   Results and Discussion

3.1   Global deformation

Before making detailed quantitative comparison between FEA simulation and measurements, the general motion of the ventricles were examined using deformation map (Figure 3.1.1). While the general motion is consistent with typical observation from clinical imaging of patient data, some of the local deformation pattern is not physiological. This can be partly attributed to the over-simplified fiber orientation map, as well as the single fiber family prescribed in the tissue constitutive model. Since the interest of this phase of the project is not concerned with the accurate physiological function of the heart, the global mechanical equilibrium can still allow us to study the overall deformation pattern, ejection fraction and mechanical interaction with implant devices.
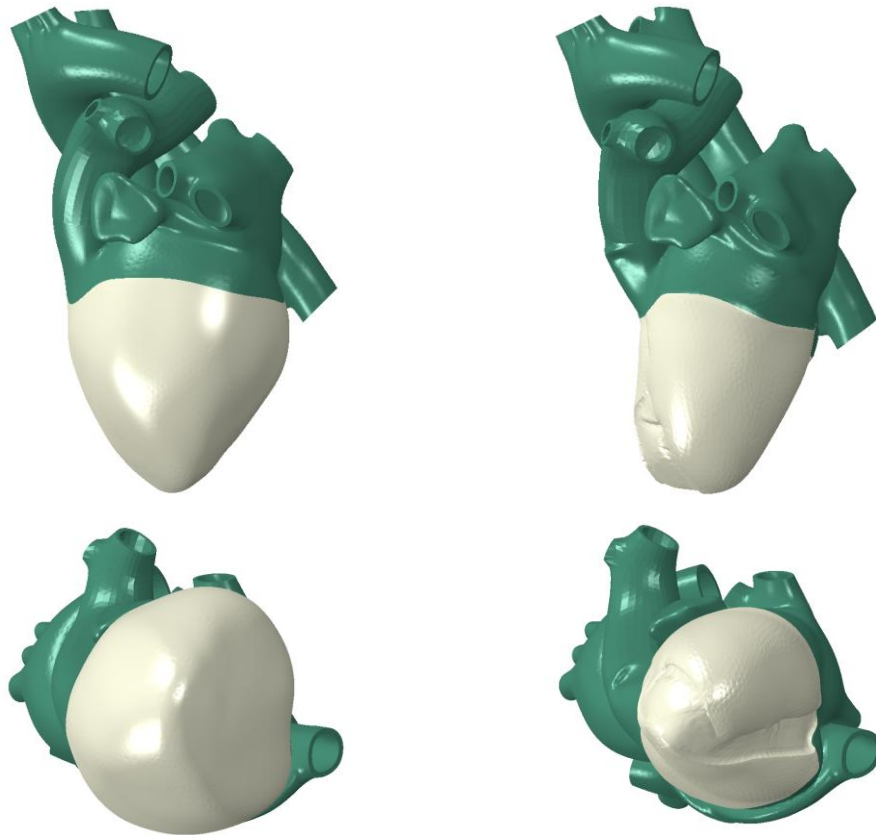
Figure 3.1.1 – Undeformed and Fully Deformed Heart Model

3.2 Left Ventricle Ejection Fraction and Pressure Response

The cardiac ejection fraction can be directly deduced from the ventricular volume output from the simulation. Figure 3.2.1 shows the time trace of the ventricular volume as overlaid on the prescribed waveform. Figure 3.2.2 shows the left ventricular pressure during the systolic contraction phase of the cardiac cycle, again compared against the prescribed waveform.
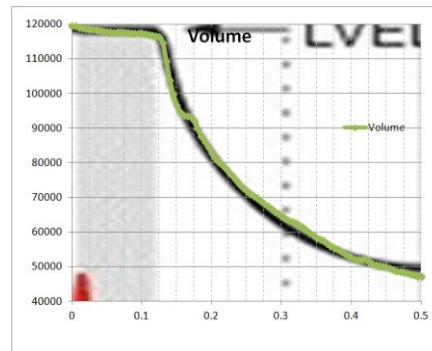

Figure 3.2.1 – Abaqus Ejection Fraction Results Superimposed on Test Data
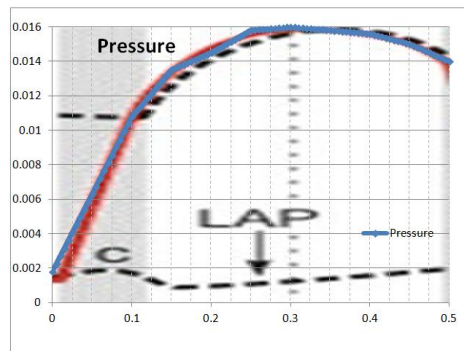

Figure 3.2.2 – Abaqus Pressure Response Results Superimposed on Test Data

4. Concluding Remarks

Considerable knowledge was acquired during the process of developing this heart simulation model which may be applied to future biomedical device numerical simulation practices. The final deformed shape is not as physiologically accurate as one would like, however it was not expected to be obtained with the initial modeling approach and the overall results are satisfactory. While the ejection fraction, pressure response and marker displacement results obtained align well with documented test data, an ongoing effort is being conducted to include a multi-step approach with slightly different boundary conditions along with a new partitioning and element orientation strategy. This new approach along with other "lessons learned" from the initial model will

provide for an even more realistic heart simulation. Once completed, the final step will be to reproduce this effort with an ischemic heart and introduce a mechanical device to gain insight.

There is limitation for this model if it is to be used to study the detailed cardiac mechanics. To complete the development of this model, we also need to perform additional model validation using animal study and clinical imaging. Once completed, model validation can be performed using implantable transducers to measure the desired interaction forces and motions. These models will allow the device engineers to improve the current design and may also help to ensure the safety of the device through computer simulation.

5. References

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6. Acknowledgment