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Simulation of structure - fluid interaction in left side of human heart

Masoud Hemmati [1,2], Seyyed Amir Abbas Oloomi [2,*], Siamak Khorami Mehr

[1]*Department of Mechanical Engineering, Yazd Science and Research Branch, Islamic Azad University, Yazd, Iran*
[2]*Department of Mechanical Engineering, Yazd Branch, Islamic Azad University, Yazd, Iran*

Abstract: According to reports of WHO, 17.3 million people all over the world die of heart disease annually and this rate is growing. The purpose of the present research is to simulate the left side of human heart in terms of geometric structure and also the features of its texture and also to analyze the current field in left side of human heart choosing real boarder situation and structure – fluid interaction. Using empirical results existing in articles (the results of one – sided tension tests) the coefficients of different hyper – elastic models, which are usually used for description of behavior of left side of heart, are calculated. In order to accurately simulate the left side of heart, a delicate geometric model is built from the left side of heart (including atrium and left ventricle) in MIMIX software. After producing the network, blood current field was analyzed in FLUENT software and the wall texture of heart in ANSYS software. Real boarder situation is imposed on entrances of left atrium and the exhaust of left ventricle.

Key words: Human heart; Structure – fluid interaction; Simulation

1. Introduction

Today, modeling human organs receives lots of attention in a study field named biomechanics which has daily increase in its applications. This approach can largely contribute predicting behaviors and human body organs reactions. Heart modeling has lots of importance regarding high rate of death caused by the diseases of this organ and different methods are used in order to create these models. Referring to the reports of WHOM, 17.3 million people die of heart disease annually and this number is increasing. This organization predicted that up to 2030, 23 million people will die of heart diseases. As pediatricians claim, even children are imposed to heart diseases. The beginning of this diseases can attributed to the time before birth and during growth time of fetus. These diseases become severe with unhealthy diet during childhood and inactive life style. Then they come into peak using drugs in adolescence. Most of these heart attacks are results of perfusion artery disease (CAD). Heart attacks are most common reason of death for men and women in US. But 70% of these attacks happen without warning and the symptoms do now show until the patient falls down under the pressure of pain in chest and pursiness. In Iran, this disease has become the main cause of death while the life time increases among patients happens during adolescence. Despite of complexities of heart, merely with a mechanical view, heart is just like a pump. With this view point, we are seeking an appropriate simulation of heart (Amidi, 2009) (Rostami and Nemat bakhsh, 2009).

1.1. Background of study

(Nikoo et al., 2001) used combination of MRI and CFD in order to develop blood current simulation in heart chambers. MRI photography has been used in order to create geometric models and CFD solution techniques have been used to solve current equations. CFD model analyzed the current along with vortexes in a 3D model with a deliberate view and using contraction and dilatation of walls, but in valves we can observe gross differences due to inaccurate simulation. Saber et al, using 8 phases in each heart cycle from MRI imaging and CMRTOOLS software, succeeded to create the geometrical model. In the present research, instead of modeling aorta and mitral valves, we considered both in one short cut of heart with two holes easily which is the major problem in the current model. It is claimed that although mitral valve has skews compared to aorta, this slope is not critical until this valve is just as an exhaust. In this simulation, blood is considered and a Newtonian fluid with constant viscosity and density. Mitral valve is modeled with diameter of d_m=2.61 and area of 5.36 cm^2 and also aorta valve with diameter of 2.64 cm and area of 3.27 cm^2. Instead of choosing constant pace in valve, we used a constant pressure as a boarder condition. With this selection, adjustment of current with inlet site and adjustment of a sort of pace distribution with inlet valve will become possible. Of course it is clear that this condition is faulty for times that both valves are closed, because during contraction and dilatation of a constant volume, valves bare different amount of pressures. It is evident that the features of blood current field in numerical simulation repeat after

* Corresponding Author.

four cycles, but at the end of diastole they need at least six cycles in order to achieve a fixed repetitive process due to the vertical motions. In accurate determination of internal surface position, the error possibility is 5 pixels. For each error pixel 1.56 mm or about 8 mm error is expected which is a large fraction of short axis dimension (short axis in continue of heart cut). In this simulation, the vertical motion of ventricle is not modeled. This might be ignored because of small size of this motion compared to general motions of heart. Because of limitations in simulation and inability in scanning the motions of valves, no effort has been made in order to create a model of them so far. Instead of modeling, each valve is modeled in open and close mode instantly at the beginning and the end of opening and closing.

In other words we can say that the effective factors on disorders (irregularities) in simulation are as followings:

1. No absolute planning of exhausts
2. Motions caused by respiration
3. Inability to follow 3D surface

(Lucy et al., 2005) tried to present a comprehensive data base of left and right ventricle and left atrium so that they could use this database both in clinical and research works. These researches were carried out on 108 healthy volunteers (63 male and 45 female). Studying on six persons and using MRI imaging and dissolution CFD techniques analyzed current patterns in left ventricle. In this study, they have tried to show that in diastole time, inlet current of mitral valve moves directly toward the apex and after passing 2 third of the way, the current moves toward the apex with a vertical motion.

2. The results

1. The left and right ventricle among men is larger in dimensions compared to women.
2. With increasing age in men, the volume and mass of ventricles decreases significantly. While among women, increase in age does not change the volume or dimensions of ventricles.
3. The fraction of discharge from left and right ventricles increases by the increase of age.
4. There is no difference in fraction of discharge among men and women in left atrium.
5. The volume of the end of diastole in the left ventricle is 160±29 mm among men and 135±26 among women
6. The volume of the end of systole in the left ventricle is 50±16 mm among men and 42±12 among women
7. The volume of the end diastole in the right ventricle is 190±33 mm among men and 148±35 among women
8. The volume of the end systole in the right ventricle is 78±20 mm among men and 58±18 among women
9. Especial density of internal heart mass is 1.05 g/cm^3.
10. Discharge fraction of left ventricle among men and women is similar. But discharge fraction in right ventricle among women is 7% larger.
11. The volume of end of systole and pounding in left atrium is so much more among men compared to women.
12. The left ventricle mass 22% and the right ventricle mass 15% among men are larger than those among women respectively.
13. About 15% of end volume of diastole and systole for left ventricle and about 25 % in right ventricle is less in women compared to men.

(Humphrey et al., 1987) used functions of strained energy such as Mooney - Rivlin and Ogden in order to model the hyper – elastic texture characteristics of heart. In order to determine the coefficients of each hyper – elastic model, we should extract the mechanical behavior of hart texture through empirical tests on samples of heart muscle.

(Ghaemi et al., 2009) carried out their studies on cow's heart and suggested a hyper – elastic model considering heart muscle as an isotropic crisscross. One of useful tests of obtaining features related to soft tissue times is tension release test.

(Demer and Yin, 1983) used dog's heart to carry out his examinations. His suggested function of strained energy density is a figurative function of main strain. The tissue of heart is hyper – elastic, non- isotropic and almost incompressible and homogeny. Modeling highly dilatation is so much harder because measuring tension in an active form is not possible. Contraction and dilatation of heart can be a combination of inactive elastic of contraction and dilatation resulted from blood pressure and active release – contraction resulted from fiber release – contraction (neurotic tissues). Structure – fluid interaction has an important role in this process. During charging phase, right ventricle neurotic tissues rest and increase in diastolic blood pressure causes dilatation of ventricle. During discharging phase, the ventricle tissue is hardened and this hardening causes increase in ventricle pressure more than lung pressure and opening the lung valve. In clinical applications, separating active / inactive and tensions / strains are difficult. Nevertheless, we considered variable hardness parameters with material time in order to model ventricle hardening and active contraction. In Fig. below, strain and tension changes are illustrated for active non – isotropic model.

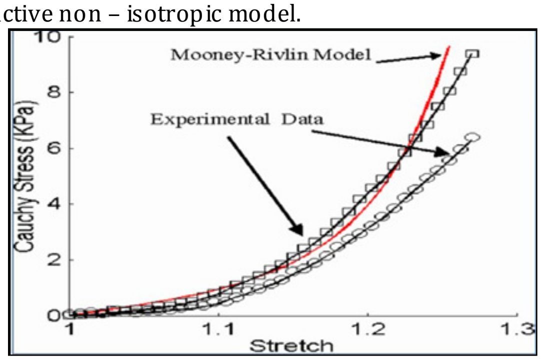

Fig. 1: Above- the results of empirical test and consistency of Mooney - Rivlin model with it

Different studies have been done on heart tissue in order to determine mechanical features and identification of type of this material. Generally heart tissue in inactive mode, is crisscross isotropic, and heterogeneous. In past times, discussions have been done under this topic that which Viscoelastic, hyper – elastic or hypo – elastic model better illustrates features of heart tissue and which model should be used in simulation. In hyper elastic model, strained energy function is used which usually is in form of a figurative function of multi sentential based on main strains. Hypo elastic model is a non – linear elastic model.

2.1. Commanding equations

Commanding equations on blood current are Navier–Stokes and Attachment which change into equations for non – contractible and Newtonian fluids. The current is considered slow. Also the equality condition of wall displacement and fluid grains attached to it should be observed to simulate wall – blood interaction.

2.1.1. Mooney - Rivlin model

Strained energy function of Mooney - Rivlin is defined as below:

$$W = C_{10}(\bar{I}_1 - 3) + C_{01}(\bar{I}_2 - 3) + \frac{1}{D_1}(J_{el} - 1)^2 \quad (1)$$

In which C_{10}, C_{01} and D_1 are time dependent parameters. In this situation, Initial shear Modulus and Initial Bulk Modulus are defined as below:

$$K_0 = \frac{2}{D_1}, \ \mu_0 = 2(C_{10} + C_{01}) \quad (2)$$

3. Research method

Reconstruction of real geometric model has three main elements:

1- Data record
2- Cutting
3- Reconstruction of model according to the general sample

In order to reconstruct a real geometry of human heart we should first provide medical images of this organ. We use x ray photography by C.T. Scan tool in order to provide the required images for production of 3D shape. MIMIX software is an image analyzer software which has the ability of reading 2D images of C.T. Scan, MRI and C.T. angiography tool which have dicom format. Heart related images are in 21 phases that each one illustrates a special time in electrocardiogram diagram. Each phase includes 397 slices and three views in different dimensions.

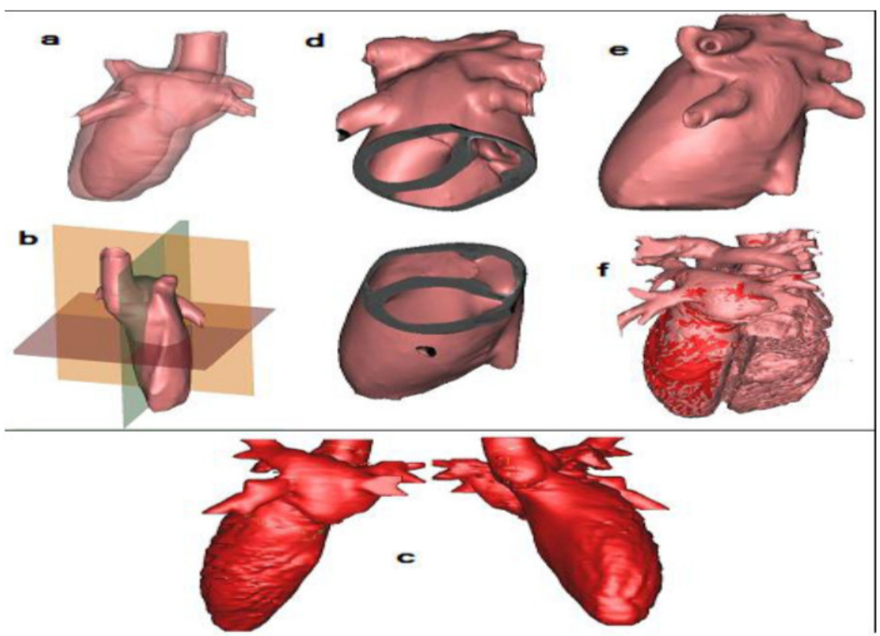

Fig. 2: A. Left wall of human heart, B. Special position of left side of human heart in MIMIX software, C. Blood volume in left side of human heart, D. A slice of heart produced in MIMIX software, E. Produced heart in MIMIX software, F. Illustration of blood volume in human heart

4. Results

The boarder conditions used for blood current field and boarder conditions used to solid field:

In the present research, the left side of human heart is modeled during ventricle diastole. During diastole, the mitral valve was open and aorta valve was completely closed. For four inlet vessels to atrium we used boarder condition of pace variable with time and aorta valve is modeled as a wall. We used boarder condition of fixed base for four inlet vessels to atrium and aorta. For solid and liquid surfaces which have energy transmission, we used boarder condition of shared solid – fluid surface. Describing hyper elastic material using different models, we need empirical data of mechanical behavior of soft tissue, and the diagram of tension and strain changes in order to obtain coefficients of hyper elastic models. In this thesis we used the results of study carried out by Chung Yang et al. these results are related to one – axis heart muscle.

In the Fig. below, we illustrate the results of this test and its consistency with Mooney - Rivlin model.

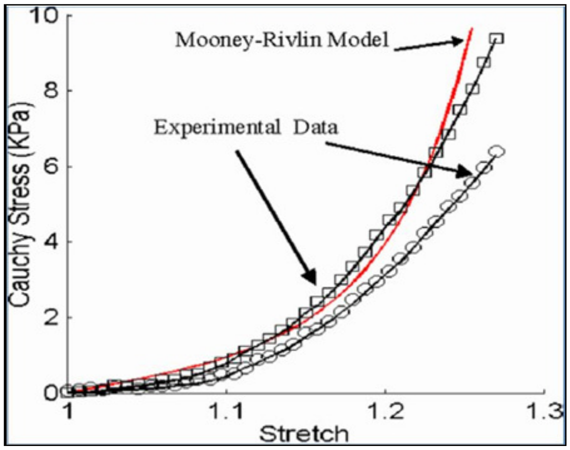

Fig. 3: Above- the results of empirical experiment and their consistency with Mooney - Rivlin model

Having the results of empirical experiment, we analyze different hyper elastic material in order to select the best model to estimate the best heart tissue. The amounts of strain and tension are inserted in form of two data columns. The models which have the most consistency with empirical results curve are appropriate for describing the behavior of hyper elastic material.

Myocardium density equals 1.05 g/ml. in this project we used third – phase Ogden model in order to describe the behavior of hyper elastic soft tissue.

4.1. Investigation of structure – fluid interaction in left ventricle

Structure – fluid interaction is analyzed according to separate solution algorithm (dividing into smaller or separate parts) or simultaneous solution algorithm. Separate solution method is such that each structure and fluid equation is solved separately. If solvents are called just one time per time span, explicit solution method will be used. Explicit solution method, is so useful in cases in which coupling is weak (such as small replacements of fluid). There are other methods for solving structure – fluid interaction rings in which the solution of structure and fluid is repeated up to the point that convergence of interaction forces and replacement happens.

Table 1: Obtained coefficients for third – phase Ogden in ANSYS software

Ansys Con.	α_3	μ_3	α_2	μ_2	α_1	μ_1	α_3
Myocardium	19.49	43.456	19.49	43.456	19.49	43.456	19.49

This solution algorithm is called implicit. For mega replacements of structures, the non – contractible fluids which are surrounded by deformable structures or in cases which high volumes of fluid moves along with structure, we should at least use separate implicit solution methods with repetition cycles of structure – fluid interaction. In ANSYS software two types of structure – fluid interaction are supported which are titled as below:

- One – way solution
- Two – way solution

In the first method which is the same separate explicit solution method, the results obtained from CFD analysis (including forces, temperature and burdens caused by displacement thermal transition) are practiced in structure – fluid interaction level as an external burden in structure analysis. In this method, displacements obtained from structure analysis will not transmit to CFD analysis. In the second method which is the same separate implicit solution method, the results obtained from structure analysis (ANSYS Mechanical), such as displacement, pace and velocity are transmitted to fluid analysis software (ANSYS Fluent) as an external burden through shared boarder. So the results of fluid analysis, like fluid pressure or temperature, are returned as a burden through shared boarder to structure analysis software. This process will continue up to the point the convergence is achieved. We first validate some of results obtained in different modes and then the results go into process of one – way and two – way analysis.

4.2. Investigation of structure – fluid interaction in left part of heart in one – way analysis

First, using explicit solution, we will analyze the method of structure – fluid interaction. In this mode of fluid current field, with the assumption of fixed ventricle wall and solved atrium and the pressure caused by fluid on fluid wall, we exert the boarder condition to the internal heart tissue and solve the solid question. As we mentioned before, in one – way analysis, the data is transmitted just one time at the beginning of solid solution from fluid analysis to solid analysis. Here the imposed pressure is exerted as a ramp function slowly. This method enables us to decrease the high pressure in each time span by truncation of loading spans while loading in order to achieve to a more constant solution.

Table 2: Boarder condition in one – way solution

		Inlet	Outlet
One – way	Fluent Unsteady-dp	Inlet Pressure	Pressure Outlet(0 pa)
	Solid Static Structural	Platform and four rings including inlet vessels to atrium	Interface Wall
		Fix Support	Load Pressure

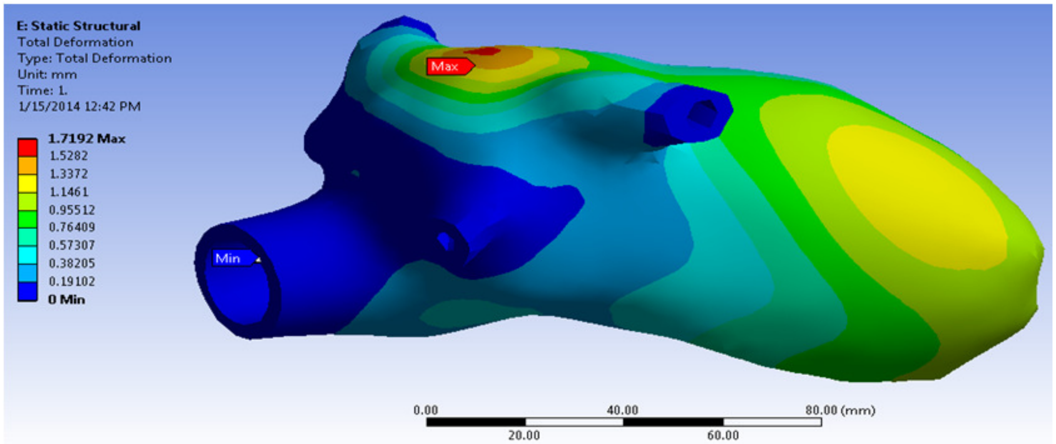

Fig. 3: Deformation degree (highest degree of deformation) in one – way solution

4.3. Investigating structure – fluid interaction in left ventricle with two – way analysis

In this case, both solid and fluid analyses are coupled and in each coupled repetition from each coupled time span, first the solid analysis (or fluid) and then the fluid analysis (or solid) is carried out. In order to analyze the solid in each repetition, pressure and cutting tension data on fluid analysis is conveyed to solid analysis and in each fluid analysis solution; displacement information is conveyed from solid to fluid analysis. It is worth mentioning that for more constancy of question, the imposed force from fluid and displacement resulted from solid analysis is divided linearly during several coupled spans. If, for example, let the least amount of coupling repetition equal to 5, in the first repetition, one fifth of force and displacement are imposed to each analyzers and also exerting the lower discount coefficient to force, more constancy is achieved. We carried out this simulation at the time of heart's diastole. This means that in time of diastole, atrium – ventricle valves are open. So at the time of diastole, the mitral valve is open and blood current enters to left ventricle from the side of atrium. Time taken by diastole is 0.5 seconds. The blood enters to atrium through four vessels and enters to the left ventricle through mitral valve. When mitral valve is open, the blood current enters to the left ventricle with high speed due to the pressure of atrium (the first peak in the Fig.). Then, with stimulation of atrium, the remained blood in the atrium is pumped to the left ventricle (the second peak in the Fig. 4).

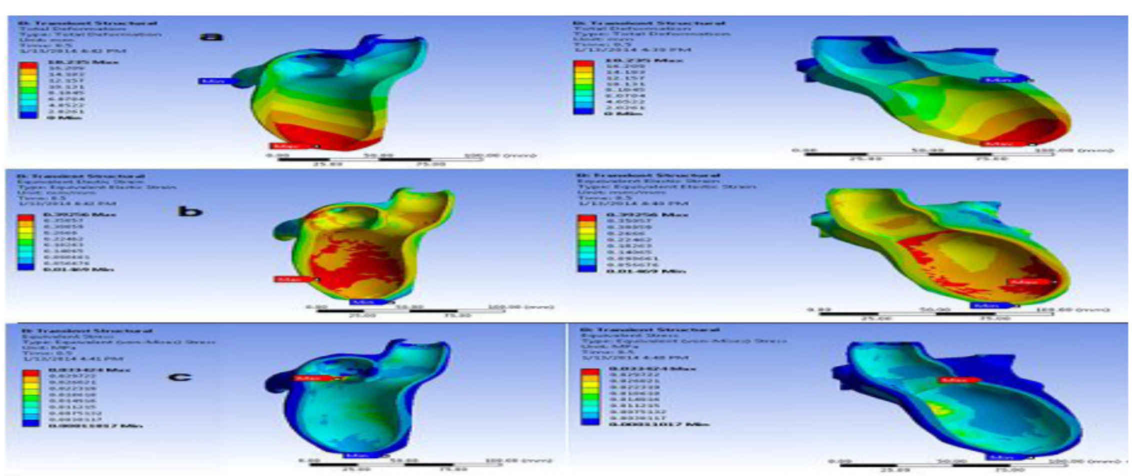

Fig. 4: A. Deformation of left wall in heart with two – way analysis, b. Changes in strain degree of left wall in heart with two – way analysis, c. Changes in tensions of left wall in heart with two – way analysis

5. Conclusion

In this research, the geometry of human heart, the left part of human heart, was simulated deliberately in MIMIX simulator software. Wall – blood interaction in one – way analysis was investigated in left part of human heart. Wall – blood interaction in two – way analysis was investigated in left part of human heart and the results were compared with one – way analysis results. As we expected, the best degrees of deformation happened around the apex.

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