Coupled Fluid-Structural Analysis of Heart Mitral Valve

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Abstract: The mitral valve apparatus is a complex and refined mechanism located between the left atrium and the left ventricle of the heart which can manifest various kinds pathologies.
In order to support identification of potentially critical conditions resulting from some typical cardiosurgery operations it is important to develop models of the mitral valve that enable prediction of both stress in the leaflets and blood flow pressure gradient and velocity across the valve during the cardiac cycle.
At present most of the studies focused on mitral valve have considered structural and fluid-dynamic aspects separately.
In the present paper a modeling approach considering fluid-structure interaction using Comsol Multyphysics is described.
In particular the use of Structural Mechanics module to define complex mechanical behavior of leaflets tissue is discussed. Then by means of predefined Fluid-Structure Interaction Module load transfer from fluid to structural domain has been implemented on a simplified valve model. In this model modification of fluid domain boundary conditions as a result of structural deformation is also considered using Moving Mesh application mode, which allows accounting for large mesh deformation in the fluid domain.

Keywords: Mitral valve, biological tissue, fluid-structure interaction.

1. Introduction

The mitral valve apparatus is a complex mechanism located between the left atrium and the left ventricle of the heart. It mainly consists of two valve leaflets, anterior and posterior, connected along valve perimeter to a fibrous ring, called annulus. On the surface facing the ventricle, leaflets are also attached to marginal and strut chordae tendinae that are further attached to papillary muscles, inserted in left ventricle wall.

In a normal heart during the diastolic phase the valve opens to allow blood flow from atrium to ventricle with minimal difference of pressure. During systole the valve closes, thus occluding the mitral orifice and avoiding backflow, and leaflets coaptation prevents local blood regurgitations. At the beginning of diastole valve is closed. Then its progressive opening leads to considerably changing blood flow conditions across the valve orifice during diastolic phase. Blood flow across valve orifice causes valve leaflet deformation as a consequence of pressure applied by the fluid, which in turn can modify orifice shape and dimensions, thus influencing velocity field and pressure gradient between atrium and ventricle.

Most of the model developed up to now to study mitral valve behavior are concerned with structural aspects only or with fluid-dynamic considerations on healthy or repaired valves [1,2,3]. However correct prediction of stresses in the leaflets and blood flow parameters ideally requires a coupled fluid-structural analysis, where the interactions between blood flow and mitral apparatus deformation are considered simultaneously. Recently some works have been published considering fluid-structure interaction for mitral apparatus [4,5].

This kind of analysis can also be carried out in Comsol Multiphysics by means of predefined fluid structure-interaction which allows combining different application mode to create a model in which solid parts can interact with fluid domain. In particular, Application modes involved are Structural Mechanics module, Incompressible Navier-Stokes module and Moving mesh.

In this paper the application of this approach to mitral valve is discussed with reference to the definition of complex soft biological tissue behavior and to the implementation of Fluid-structure interaction on a simplified mitral valve model.

2. Governing Equations and Application Modes

The model consists of a fluid part, solved with the Navier-Stokes equations, and a structural mechanics part, which solves for valve leaflet deformation. A Moving Mesh application mode ensures that fluid domain is deformed along with leaflets. The corresponding governing equations and the FSI (Fluid Structure Interactions) settings on subdomains are available directly when using the predefined multiphysics coupling for fluid structure interaction.

In the present model the diastolic part of the cardiac cycle is considered, when under the action of fluid flow valve leaflets open starting from a closed position. The valve leaflets are assumed to be immerged in a fluid channel. Although valve opening is a typical example of transient phenomenon, the analysis is static and parametric, so that stationary conditions at increasing level of inlet fluid velocity are considered by means of a parametric study.

The fluid flow in the channel is then described by the Navier-Stokes equations, solving for the velocity field **u** = (u,v) and the pressure, p:

$$\rho \frac{\partial u}{\partial t} - \nabla \cdot \left[- pI + \eta \left(\nabla u + (\nabla u)^T \right) \right] + \rho (u \cdot \nabla) u = F$$
$$- \nabla u = 0$$

In this equation I is the unit diagonal matrix and F is the volume force affecting the fluid, which assuming no gravitation or other volume forces affecting the fluid, is equal to zero.

The structural deformations are solved for using an elastic formulation and a non linear geometry formulation to allow large deformations. For boundary conditions the outer edge and surface of the leaflets is fixed to channel walls and cannot move in any direction. All other boundaries experience a load from the fluid given by

$$F_T = -n \cdot \left(- pI + \eta \left(\nabla u + (\nabla u)^T \right) \right)$$

where **n** is the normal vector to the boundary. This force represents the load exerted by the fluid as a sum of pressure and viscous components.

Finally it is convenient to use a deformed mesh when the boundaries of computational domain are moving in time or as a function of a parameter, as in the present case. This technique is implemented in Comsol Multiphysics by perturbing the mesh nodes so they conform to the moved boundaries.

The technique for mesh movement is called an arbitrary Lagrangian-Eulerian (ALE) method. The ALE method is an intermediate between the Lagrangian and the Eulerian method, and it combines the best features of both, allowing moving boundaries without the need for the mesh movement to follow the material.

In the present model the motion of the deformed mesh is implemented by using Winslow smoothing and the boundary conditions control the displacement of the moving mesh with respect to the initial geometry. At the boundaries of the leaflets this displacement is set to be the same as the structural deformation, whereas at the exterior boundaries it is set to zero in all directions.

3. Mitral Valve Model

As already observed the mitral valve apparatus is a very complex mechanism. Modeling valve behavior involves various different aspects but basically most critical issues include definition of valve geometry, constitutive modeling of leaflet tissue mechanical behavior and definition of fluid flow conditions.

Valve geometry is highly irregular and subject to significant statistical variations, leaflets are made of soft biological tissue whose properties are not easily modeled and finally valve hemodynamic is not yet considered as fully understood. As a result different models of mitral valve have been proposed, often based on specific assumption in order to tackle complications deriving from geometrical or material non linearities or from the need to improve computational efficiency by introducing some simplification.

In the following paragraphs a summary of basic assumptions, schemes and idealizations adopted to implement the model in Comsol Multiphysics is presented.

All of these assumptions of course have a strong influence on results.

The present study is part of an ongoing research on mitral valve and mitral valve repair techniques and further research steps will be focused on progressively removing such simplifying assumptions.

3.1 Valve geometry

In order to correctly reproduce main features of complex mitral apparatus and, at the same time, to guarantee computational efficiency and accuracy, a three-dimension model of the valve has been developed under some simplifying assumptions.

The first assumption concerns initial leaflet position. In the present model it was chosen as initial position the transition between systolic and diastolic phase, when the valve is still closed but pressure gradient between left atrium and ventricle is very low. In this position therefore stresses in the leaflets due to pressure may be considered negligible. On the other hand this also represents a good starting point to analyze both diastolic and systolic phase. This position is represented in Fig. 1, where different parts of leaflets and annulus are also depicted.

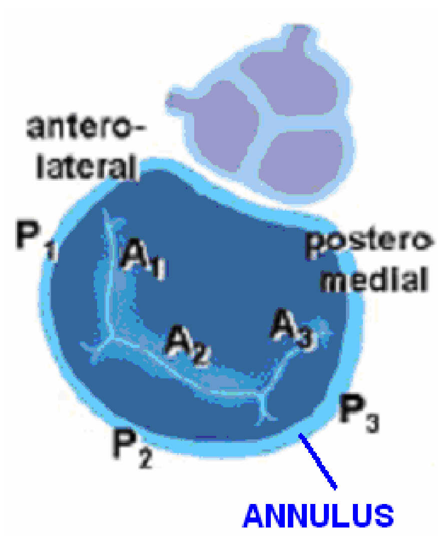

Figure 1. Mitral valve annulus and leaflets

In order to reproduce the valve appearance reported in literature, the geometry of valve annulus and leaflets shape and dimensions were derived based literature data concerning leaflet profile as a function of position along annulus perimeter, as qualitatively show in Fig. 2. However, to simplify the model geometry, leaflets were considered as initially lying flat on annulus plane.

Further assumption involves the presence of clefts in the posterior leaflet, which has been considered by introducing a physical separation between the three posterior clefts. Also in the commissural region, that is the lateral area at the transition between anterior and posterior leaflet, a physical separation in included, where in reality a redundancy area is present. The shape of the leaflet was also simplified as shown in Fig. 3.

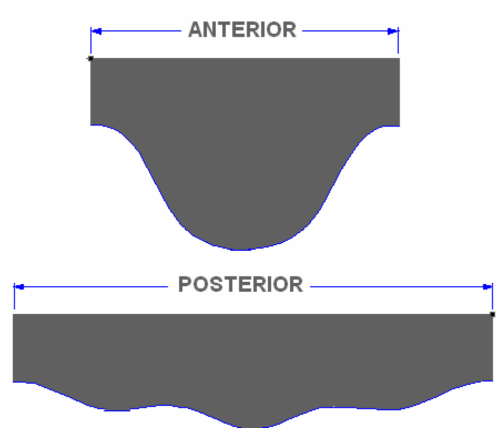

Figure 2. Leaflets profile

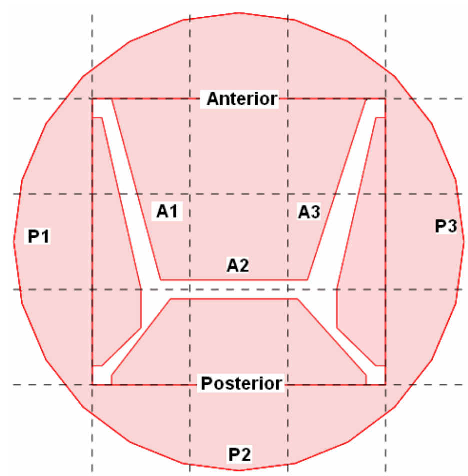

Figure 3. Scheme assumed for mitral valve annulus and leaflets

Finally since only the diastolic phase is considered, the chordae tendinae have not been modeled. While this is justified by the evaluation purpose of the present model, it is clear that when considering prediction of leaflet stresses caused by pressure exerted by the fluid during systolic phase, this hypothesis should necessarily be removed.

3.2 Leaflets tissue

Mitral valve leaflets consist of a soft biological tissue, which exhibits a complex mechanical behavior, resulting from the coupling between different constituents. Heart valve tissues are essentially made of a fibrous tissue network, mainly collagen and elastin, saturated with a fluid that is mostly water.
The fibrous network of leaflet tissue is characterized by waviness, which in turn affects stress-strain response. At low strains wavy tissues can be extended by relatively low stresses but as strains increase, fibers get progressively straightened and the overall response of the tissue becomes much stiffer.
From the point of view of biomechanics, the properties of a tissue are known if its constitutive equation is known. Many constitutive models for biological tissues are derived by extending theories originally developed for rubber. One of the best-known approaches to the elasticity of bodies capable of finite deformation is provided by the hyperelasticity models in which the form of an elastic potential, or strain energy function, W is postulated. Once the strain-energy function is known the stress state can be determined by taking the derivative of W with respect to a strain measure:

$$\sigma = \frac{\partial W}{\partial \varepsilon}$$

where σ and ε are any work conjugate stress and strain measures.
For many soft biological tissues however spatial distribution of collagen fibers may results in a preferred fiber alignment orientation, at least on a statistical base. This is also the case of mitral valve leaflets in which collagen fiber distribution, which has been studied in [6], results in anisotropic stress-strain relation.
There exist different approaches to extend the use of traditional hyperelastic models to anisotropic soft tissues. In particular extensive reviews of these kinds of approach can be found in [7] and more specifically for cardiovascular tissues and heart valves in [8].
Implementation of a constitutive equation, for example into a finite element code, necessarily also requires determination of constitutive parameters, peculiar for each model, which can only be found by experiments.
For mitral valve experimental data obtained from biaxial test on porcine mitral valves are reported in [9]. These data were also used to identify constitutive law in [10].
In this constitutive model, based on introducing the concept of pseudo-invariants I_4, the form of the strain energy is:

$$W = c_0\left(\exp^{Q} - 1\right)$$

where Q is a quadratic function of two strain invariants:

$$Q = c_1\left(I_1 - 3\right)^2 + c_2\left(\alpha - 1\right)^4$$

The term α is related to pseudo-strain invariant I_4 by the relationship:

$$I_4 = N \cdot C \cdot N = \alpha^2$$

in which **C** is the Right Cauchy-Green deformation tensor and **N** is a vector denoting the preferred fiber direction of the material.
Material parameters c_0, c_1 and c_2 based experimental data reported in [9] are also available in [10] for both posterior and anterior leaflet.
This constitutive model can be implemented in Comsol Multiphysics by modifying Strain energy function Ws_smsld via the menu Physics / Equation System / Subdomain Settings.
Instead of the more traditional approach in which the conjugate pairs second Piola-Kirchhoff stress and Green-Lagrange strain are used, Comsol uses the first Piola-Kirchhoff stress P and its conjugate strain, the displacement gradient F.
It is then possible to utilize the capability of Comsol Multiphysics to automatically differentiate an expression, thus it is sufficient to modify only the strain energy function.
In the model implemented in Comsol strain energy form is also modified to take into account nearly incompressible behavior.

The capability of the software to reproduce this complex material behavior were then verified by means of a simple model in which experimental biaxial tests used to determine material properties were reproduced. These tests consist in stretching a strip of tissue in two perpendicular directions, either imposing the same deformation (equi-biaxial test) or different deformation ratio (off-biaxial test). Both equi-biaxial and off biaxial tests were compared, with good agreement between experimental and predicted values.

Model used for this purpose is reproduced in Fig. 4. in the case of off-biaxial test.

As an example results referring to anterior leaflet off-biaxial test are reported in Fig. 5 for both direction parallel and perpendicular to fibers.

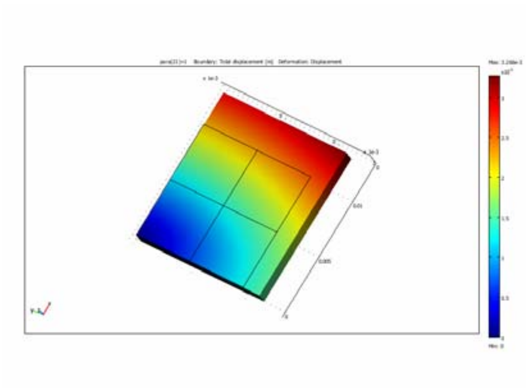

Figure 4. Model reproducing biaxial test on leaflet tissue

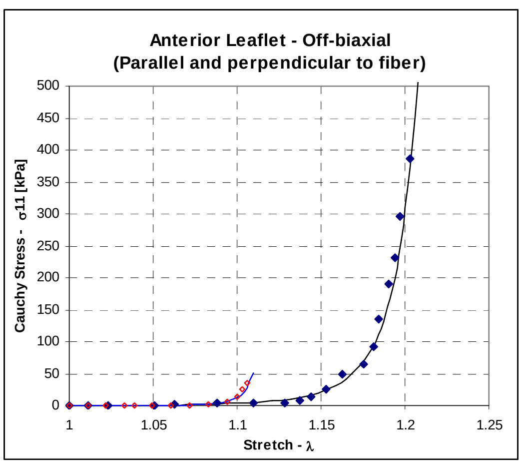

Figure 5. Tissue mechanical behavior – Predicted vs. experimental data

These results demonstrate the feasibility of implementation in Comsol of constitutive models in which anisotropic hyperelasticity of mitral valve tissue is considered.

It should be noticed however that when modeling real component, as valve leaflets, knowledge of tissue properties in directions parallel and perpendicular to fiber has to be combined with some information about preferred collagen fiber orientation within the leaflet. This orientation has been demonstrated to be rather variable depending on leaflet region considered, so in the present valve model, which is mainly aimed to evaluate applicability of multiphysics analysis to this specific problem, the tissue has been simply assumed has elastic linear, with an extremely low Young Modulus (2 MPa) for both leaflets. However, basing on good result obtained, implementation of hyperelastic anisotropic model represents the next step for future model development.

3.3 Subdomain settings and boundary conditions

In order to carry out an analysis with fluid structure-interaction both solid and fluid domains need to be modeled. As for tissue properties and valve geometry, also the definition of proper blood flow conditions represents a challenge to the analyst. Blood flow through mitral orifice during cardiac cycle is very complex due to varying hemodinamic conditions resulting from movement and deformation of left atrium and ventricle walls as well as of leaflets.

In the present model it was chosen, again as a starting point in view of future developments, to consider the valve as immerged in a channel with a regular geometric shape. At the inlet of this channel a fully developed laminar flow with a maximum fluid velocity is imposed, whereas at the outlet a neutral condition is assumed. No slip boundary conditions are used for all other boundaries. Again, while these conditions are not completely representative of blood flow through mitral orifice, they are more easily adaptable to in-vitro experimental conditions and could be more useful to validate the model before attempting further steps.

A scheme of the model and the subdivision in solid and fluid domain adopted is given in Fig. 5. The analysis consists of a parametric study in which maximum inlet velocity is progressively increased up to 1 m/s.

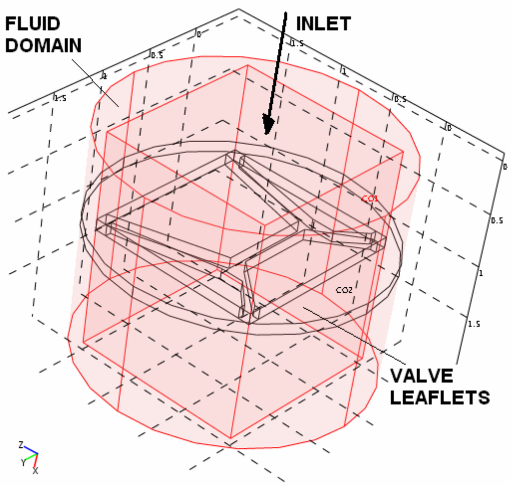

Figure 5. Scheme of subdomains for fluid-structure interaction

4. Results and Discussion

Compared to a more traditional structural analysis the main results that can be obtained by means of a fluid structure interaction analysis are pressure distribution and velocity field in the fluid subdomain together with stresses in the solid subdomain originating from pressure and viscous forces exerted by the fluid.

An example of results for the present model is given in Fig. 6 where deformed valve is represented. Using streamline, slice and boundary plots simultaneously both structural and fluid flow variables can be represented.

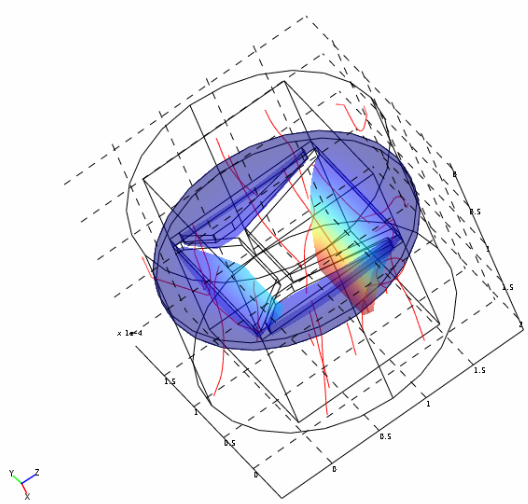

Figure 6. Valve deformed by fluid flow

Overall results in terms of transvalvular pressure gradient, maximum fluid velocity and stresses in the leaflets were found to be comparable with studies reported in literature on the same subject, although the model need to be refined to allow a more precise evaluation upon a quantitative point of view.

5. Conclusions

A finite element analysis of heart mitral valve considering fluid-structure interaction has been carried out. In particular Multiphysics approach has been found to be useful in order to evaluate with a single analysis the most important surgical parameters: pressure gradient between left atrium and ventricle during diastolic phase, maximum fluid velocity and stresses in the valve leaflet.

While the model need some refinement, it appears as a good starting point to develop model able to describe valve behavior and then to simulate non-invasively surgical repair of pathological valves.

Also the possibility of implementing anisotropic hyperelastic laws in Comsol has been evaluated positively.

Next steps will include extension of this constitutive model to the valve, more detailed modeling of chordae tendinae apparatus and modeling of systolic phase, when valve is closed and pressure is applied on the ventricular face of the leaflet.

8. References

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