DEVELOPING COMPUTATIONAL FLUID-STRUCTURE INTERACTION MODELS OF THE LYMPHATIC VALVE

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Abstract

The lymphatic system is an essential vascular network vital to proper physiologic function and is responsible for the transport of lymphatic fluid from most tissues to the venous return. At the mesenteric level, bi-leaflet valves encapsulated by a bulbous sinus serve to prevent retrograde flow between tubular vessel segments. Despite their importance, little detailed knowledge is available to truly understand how the leaflet deflections influence the surrounding fluid dynamics. To understand this, a computational study was performed using an idealised geometry of a rat mesenteric lymphatic valve. A fluid-structure interaction approach was used to determine the opening behaviour and resulting fluid dynamics of the lymphatic valve leaflets. Resulting velocity contours during opening revealed a peak velocity of 38 mm/s at the orifice of the valve with an increase in curvature of the leaflet.

Keywords: valve, lymphatic, fluid-structure interaction

1 Introduction

The lymphatic system is a vascular network that is vital to proper physiologic function and is responsible for the transport of lymph (largely aqueous with relatively low concentrations of solutes, cells and particulate matter) from tissues to the venous return [1]. In addition to maintaining tissue homeostasis, the lymphatic system also plays a vital role in immune cell trafficking, cerebro-spinal fluid/nasal drainage, and lipid transport [2].

At the collecting level, bi-leaflet valves serve to prevent retrograde flow [3], which are essential in maintaining proper un-directional flow of lymph. Further, valve defects have been shown to underlie the pathogenesis of lymphatic distichiasis, a dominantly inherited form of primary lymphedema [4,5]. Physical injury to valves occurs in lymphatic filariasis [6], which is the most common cause of lymphoedema in the world affecting over a hundred million people worldwide [7,8]. Once the healthy valve behaviour is understood, the disease states can begin to be investigated.

Experiments involving lymphatic valves are extremely difficult due to low flow rates (on the order of µL/hr) and small vessel size (100 µm characteristic diameter). Thus, computational simulations are vital to understanding the correct functionality of the vessels. In this study, we have developed a fluid-structure interaction (FSI) model of a rat mesenteric lymphatic valve using a two-way iteratively implicit approach by coupling ANSYS Mechanical APDL and ANSYS Fluent within the ANSYS Workbench suite. These results will add to the breadth of knowledge used to develop therapeutics to treat disorders of the lymphatic system.

2. Methods

2.1 Idealised Geometry and FSI Modelling Approach

An idealised geometry of a rat mesenteric lymphatic valve described previously (Wilson JT et al 2015) was used as the computational domain for simulations (Fig 1).
Due to the expected high degree of leaflet deflection over short time scales (e.g. complete opening in less than 1 second), a strong coupling between the fluid and structural equations is necessary. A two-way iteratively implicit method was employed using ANSYS Fluent as the fluid solver and ANSYS Mechanical APDL as the structural solver. The equations of the solid and fluid are solved separately but data is exchanged within a time step. Using terminology derived by ANSYS, each iteration within a time step is known as a coupling iteration. The ANSYS platform was chosen because of its mesh proficiencies and ability to remesh the fluid region at every time step.

2.2 Structural Boundary Conditions

The ¼-leaflet was meshed using the SOLID187 element, which is a 3D, 10-node element which is well-suited to modelling irregular meshes and allows for quadratic behaviour. The valve leaflets were fixed at the annulus (edge where the leaflets attach to the wall of the vessel) and large deflections were taken into account. Rather, the solver takes into account stiffness changes resulting from changes in element shape and orientations during deflection. A Hilber-Hughes-Taylor (HHT) time integration method, which is an extension of the Newmark time integration method that allows for energy dissipation and second order accuracy, was used to discretize the equations of motion and the Newton Raphson Method was applied to solve the discretized equations. For the material property, a Neo-hookean model was employed (Eqn. 1):

\[
W = \frac{G}{2} (\overline{T}_1 - 3) + \frac{K}{2} (J - 1)^2
\]

where \(W\) is the strain energy potential, \(G\) is the initial shear modulus, \(\overline{T}_1\) is the volume-preserving first invariant of the Cauchy-Green deformation tensor \(K\) is the bulk modulus and \(J\) is the determinant of the deformation gradient tensor. A value of 20kPa was used for \(G\), which is slightly lower than that of arterial elastin [9, 10, 11], assuming a nearly incompressible Poisson’s ratio of 0.45. A viscous damping matrix, \([C]\), was introduced to provide numerical stability within the structural simulation:

\[
[C] = \beta [K]
\]
where $\beta$ is the stiffness matrix multiplier for damping and $[K]$ is the structural stiffness matrix. Uncoupled structural simulations were also performed where $\beta$ was varied from 0.005 to 0.05 to determine its effect on deflection at the leaflet tip. During these simulations, a time-varying average pressure load was applied to the belly of the leaflet to induce opening.

### 2.3 Flow Conditions

The fluid domain (Fig. 1) was imported into the commercial CFD software, ANSYS Fluent, and meshed using tetrahedral elements resulting in a computational mesh of approximately 156,500 volumetric elements. The mesh in the region below the valve leaflet closest to the region of central lymph flow was refined because these elements were expected to stretch during opening while elements between the valve leaflet and sinus were expected to compress. A fluid with dynamic viscosity, $\mu$, of 1.5 cP [12] and density, $\rho$, of 1 g/cm$^3$ was used to model lymph. An unsteady and increasing pressure boundary condition with a maximum pressure of 50 Pa was applied at the inlet of the geometry (Fig. 2) and the outlet was kept constant at 0 Pa.

Figure 2: Inlet pressure boundary conditions applied to the fluid domain. The pressure increases from 0 Pa to 50 Pa in 0.45 s.

Symmetry boundary conditions were applied at the regions where the full valve leaflet had been divided into quarters to model mirror symmetry at these locations. Diffusion-based mesh smoothing, local cell remeshing and local face remeshing were applied to rectify irregularities in the fluid mesh resulting from the deformations. The simulation was allowed to run for 0.45s using a time step of 3 ms. Three coupling iterations per time step were implemented in ANSYS Fluent with a maximum number of coupling iterations of 150.

### 3. Results

Structural simulations were performed where the stiffness matrix multiplier, $\beta$, was varied between 0.005 and 0.05 to determine its effect on the leaflet deflection results. Increases in $\beta$ resulted in decreases in the maximum tip deflection, $\delta_{\text{max}}$ (Fig. 3).
Figure 3: Effect of $\beta$ on deflection in $y$-component deflection of the leaflet, $\delta_y$. Note that the $y$-component is perpendicular to the axial direction of flow.

For example, $\beta = 0.005$ resulted in $\delta_{max} = 29.251 \mu m$ while $\beta = 0.05$ resulted in $\delta_{max} = 27.309$. To minimise the effect of $\beta$ on deflection but still preserve the numerical stability of the solution, a value of $\beta = 0.03$ was used.

For the fully-couple case, $\delta_y$ increases in a linear fashion until approximately 0.3 s (Fig. 4).

Figure 4: $\delta_y$ versus time, t, for the coupled FSI simulation.

At approximately 0.3 s $\delta_y$ begins a nonlinear monotonic increase until it reaches approximately $\delta_y = 14 \mu m$ corresponding to an inlet pressure of 50 Pa.

The peak velocity within the central jet of flow had a maximum value of 38 mm/s at 0.45 s (Fig 5).
Figure 5: Velocity contours at a plane perpendicular to the axial direction of the vessel. Velocity peaks at 38 mm/s at 0.45 s.

Additionally, the leaflet becomes substantially more curved as the inlet pressure increases. However, a numerical value to the curvature has yet to be calculated and will be performed in future studies.

4. Discussion and Conclusion

In this study, an FSI model of a rat mesenteric lymphatic valve was developed. ANSYS Fluent uses an ALE approach, which is advantageous because of the user’s ability to quantify wall-shear stress values along the belly of the leaflet. While not presented in this paper, these wall shear stress calculations will be performed in future computations. Peak velocities were found to be approximately 38 mm/s for the maximum applied inlet pressure of 50 Pa at 0.45 s.

One limitation of this study includes the simplification of the geometry to ¼ of the full valve and sinus and the implementation of symmetry boundary conditions. To the authors’ knowledge, there are no known studies that indicate the flow is symmetric at the valve. However, due to the low flow rates ($Re>1$) within the lymphatic flow regime, it is possible the flow would behave in a symmetric manner despite the swift movements of the valve leaflets. While the mesh has a relatively small number of elements (~156,500 elements) compared to some required for cardiovascular flows (>1,000,000), the valve leaflets themselves are much more flexible due to the low shear modulus. Thus, it is expected that the lymphatic valve leaflets would deflect much faster under a given load compared to other biological materials. This creates a further challenge and justification for initial simplification of this complex physiological process.

Future studies will include implementing a sinusoidal inlet pressure to open and close the valve leaflets as well as modelling the full geometry to understand whether the symmetric assumption is appropriate. Overall, this work has created a foundation for the future development of FSI models of the lymphatic valve.
References


