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FLUID-STRUCTURE INTERACTION ANALYSIS OF VENOUS VALVE HAEMODYNAMICS

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SUMMARY

Venous valves play an important role in ensuring that blood returns to the heart. Nevertheless there are few quantitative reports of venous valve function. This study reports results from a 3D fluid-structure interaction model of a venous valve, including the valve sinus, for varying Reynolds number. Analysis of valve orifice deformation, trans-valvular pressure drop and velocity increase through the valve demonstrate the influence of changes in venous flow on valve geometry and the variation in pressure drop with flow rate. These results inform understanding of the role of venous valves in disease processes and their contribution to circulatory haemodynamics.

Key words: *venous valve, fluid-structure interaction*

1 INTRODUCTION

Venous valves assist in venous return, preventing flow towards the extremity when the proximal venous pressure becomes larger than the distal pressure, ensuring that blood returns to the heart. The venous valves may open and close under a range of physiological conditions including changes in posture and activation of the skeletal muscle and respiratory pump mechanisms. Whilst Lurie et al [1] have reported the behaviour of valves in the greater saphenous and superficial femoral veins, detailed 3D studies of valve geometry and deformation are lacking. This is due to the challenge of imaging valves in the smaller peripheral vessels and the location of valves deep within the musculature, particularly in the lower limb.

In contrast to the significant literature addressing cardiac valve function, there have been few attempts to study venous valves using 3D numerical models, which include reports by Buxton and Clarke [2], Narracott et al. [3] and Tien et al. [4].

A more detailed understanding of venous valve function, and the variation in valve behaviour under different conditions, is required to allow modelling of the venous system to reach the same level of maturity as existing arterial models. Potential clinical applications of models of venous haemodynamics include; initiation of deep vein thrombosis [5] and venous return following creation of an arterio-venous fistula for haemodialysis treatment [6].

This study reports results from a 3D fluid-structure interaction model of a venous valve including the local geometry comprising the valve leaflets and sinus region. These results are discussed in the context of both understanding of local valve fluid dynamics and to provide detail of the trans-valvular pressure drop to inform 1D and 0D models of venous haemodynamics [7].

2 METHODOLOGY

A 3D fluid-structure interaction model of the local valve geometry was constructed using ANSYS APDL (ANSYS Inc.) as a pre-processor and LS-DYNA (Livermore Software Technology

Corporation) to solve the fluid-structure interaction problem. In the following description LS-DYNA input cards are referred to directly (e.g., *CARD_NAME). The valve geometry was represented in parametric form using a similar approach to previous work [3], with the extension of the model to include the sinus geometry and the simulation of the full valve geometry without symmetry assumptions. The geometry of the valve leaflets and vessel wall are shown in Figure 1 with vessel radius (r_{vein}) 2.5mm, sinus radius (r_{sinus}) 3.93mm, leaflet length (l_{leaf}) 5mm and leaflet radius (r_{leaf}) 5.89mm. A gap of 0.25 mm was defined between the valve leaflets to ensure fluid was able to flow through the valve from the start of the analysis. The solid domain extended 2.5 mm before and 8.8 mm after the sinus region which was 12.5 mm in length (l_{sinus}).

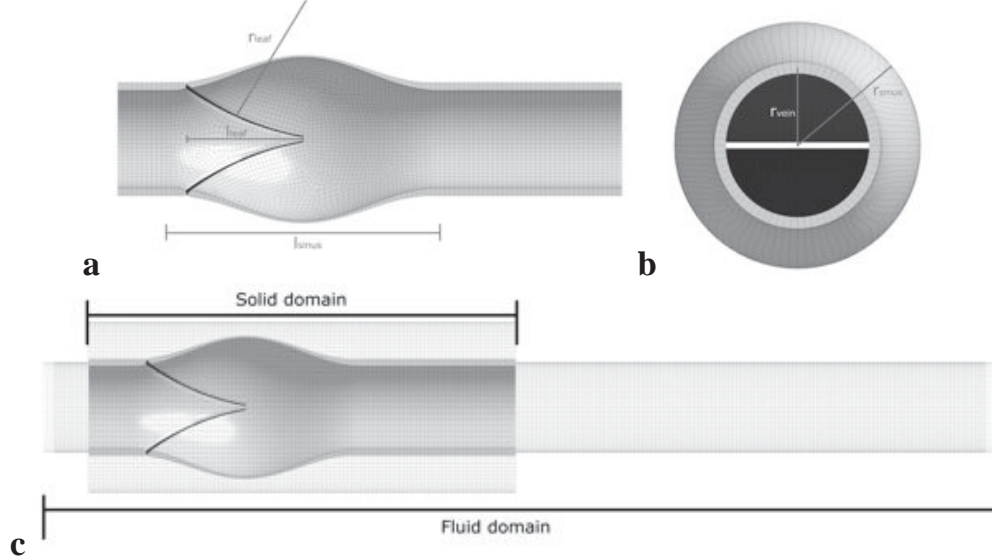


Figure 1: (a) YZ view of solid model mesh (vein wall, valve leaflets and sinus region) (b) XY view of solid model showing leaflet gap (c) Location of solid model within fluid mesh

A Boolean operation was used in ANSYS APDL to define the intersection between the valve leaflets and the valve sinus. The solid components of the model were represented using shell elements with a varying thickness defined for the sinus wall to represent greater distensibility of the sinus region [1] (as shown in Figure 1). The thickness of the vein wall (h_{vein}) and valve leaflets (h_{leaf}) were defined as 0.375 mm and 0.050 mm respectively and the form of thickness variation within the sinus region was defined by Equation 1.

$$h(z) = h_{vein} * (1 - 0.75 * \sin(\pi z / l_{sinus})) \quad \text{EQ. 1}$$

where z is the axial distance from the start of the sinus region. In this study the valve was assumed to be stress free in the closed position and the vessel was assumed to be stress free under zero pressure. The valve leaflets and vessel wall were simulated using a linear elastic material model (*MAT_ELASTIC) with Youngs modulus 50.7 MPa and 0.507 MPa respectively, Poissons ratio 0.499 and density $1e^5 \text{ kg.m}^{-3}$ using type 16 fully integrated shell elements. The density of the solid elements was higher than physiological values to reduce the timestep of the solution with mass scaling defined using *CONTROL_TIMESTEP to ensure a minimum timestep of 5×10^{-6} seconds. The fluid domain was defined to extend beyond the region of the solid geometry, as shown in Figure 1, the radius of the fluid domain was defined as 2.5 mm in the inlet and outlet regions and 4.75 mm in the region of the valve. Solution in LS-DYNA results in an Eulerian representation of the fluid domain with fluid-solid interaction included using a penalty method to impose the velocity of the solid onto the fluid domain (CTYPE = 4). Fluid elements were defined using a null material (*MAT_NULL) with density 1050 kg.m^{-3} and viscosity 0.0035 Pa.s and a Gruneisen equation of state (*EOS_GRUNEISEN) with speed of sound, $c = 10 \text{ ms}^{-1}$.

Ambient inlet and outlet regions were defined at the extremes of the fluid domain with zero pressure applied at the outlet and a parabolic velocity profile defined at the inlet of the domain. The time variation of the inlet velocity profile was defined by Equation 2.

$$V(t) = V_{max} (1 + e^{-(t-t_0)/T})^{-1} \quad \text{EQ. 2}$$

where V_{max} is the desired parabolic velocity profile, $t_0 = 0.1695$ and $T = 0.017$. Ten analyses were undertaken to compare the steady-state equilibrium condition of the valve with V_{peak} from 0.091

ms^{-1} to 0.823 ms^{-1} corresponding to Reynolds number from 68 to 617, in increments of 68. The analyses were run for a time of 0.5 seconds to allow equilibrium to develop, with solution timestep of 3.5×10^{-6} seconds determined by solid mass scaling as described above. Typical element size for both solid and fluid domains was specified to be $\sim 0.13 \text{ mm}$, resulting in 36 elements across the vein diameter and a mesh of 10,240 shell elements and 461,340 solid elements. The simulation was run on a single core of an Intel X5690 3.47GHz processor, with typical run times of the order 65 hours per analysis.

Results were saved in increments of 5ms and the deformed geometry of the valve orifice was compared at 0.45 seconds for the lowest and highest Re value. Velocity augmentation and pressure changes through the valve region were assessed by exporting nodal velocity and pressure results on the centreline and plotting as a function of nodal z coordinate.

3 RESULTS

The variation in valve orifice geometry with Reynolds number is shown in Figure 2. Orifice deformation is quantified in Figure 2a where the displacement of the centre of the leaflet is plotted against analysis time, demonstrating the development of a steady state condition. The orifice geometry is shown in Figure 2b and 2c at Re values of 68 and 617 respectively.

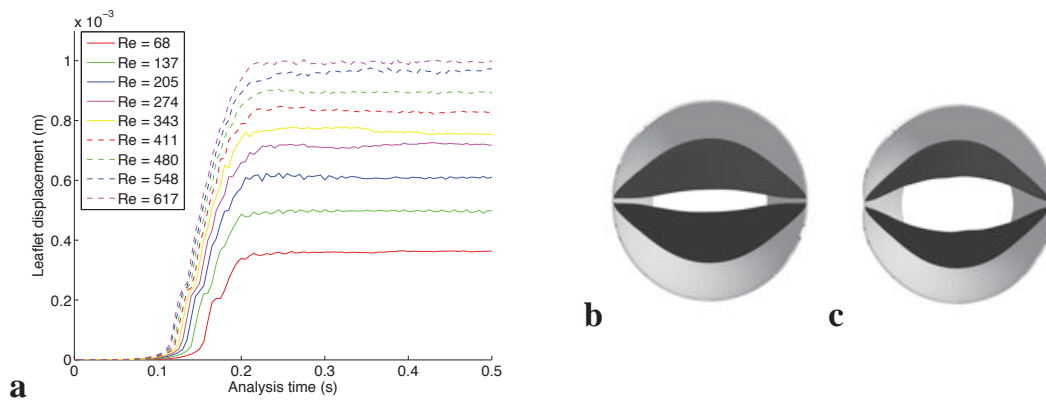


Figure 2: (a) Variation in leaflet centre displacement with simulation time for all Reynolds numbers. Valve orifice deformation at 0.45 seconds for (b) Re = 68 and (c) Re = 617.

The variation in fluid velocity with Reynolds number is shown in Figure 3. The centreline z velocity is shown in Figure 3a, normalised relative to the inlet velocity. The velocity distribution in the YZ plane at the centre of the vein is shown in Figure 3b and 3c for Re values of 68 and 617 (with colour scale from 0 to 0.457 ms^{-1} and from 0 to 1.234 ms^{-1}), respectively.

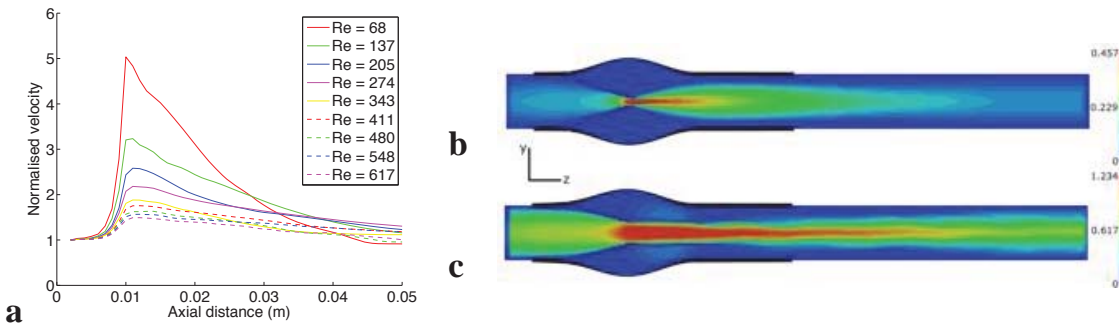


Figure 3: (a) Variation in centreline z velocity, normalised relative to inlet velocity. Velocity distribution in the YZ plane at 0.45 seconds for (b) Re = 68 and (c) Re = 617.

The variation in fluid pressure along the centreline of the vein with Reynolds number is shown in Figure 4a. Figure 4b plots the pressure drop over the fluid domain against the Reynolds number.

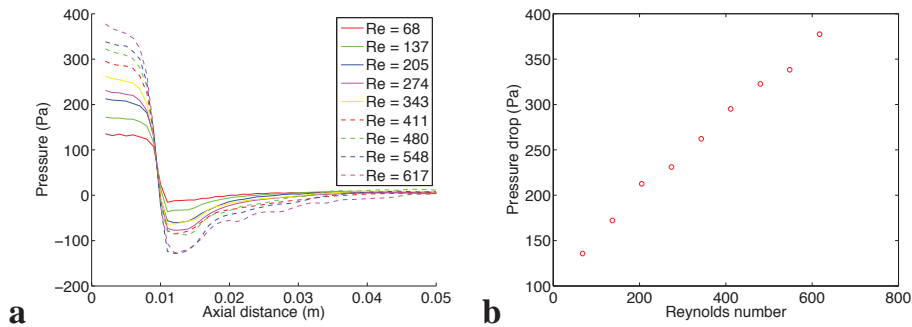


Figure 4: (a) Variation in fluid pressure along centreline for all Reynolds numbers. (b) Variation in trans-valvular pressure drop with Reynolds number.

4 DISCUSSION

These results demonstrate valve behaviour in line with *in vivo* ultrasound measurements made by Lurie et al. [1]. *In vivo* the centreline velocity between the valve leaflets is reported to be 1.9 times the value distal to the valve, with a reduction in orifice area at the leaflets to 35% that in the distal region [1], this is in good agreement with the results of the Reynolds number 272 simulation (2.2 fold increase, orifice area 46%). However, this Reynolds number represents a higher velocity magnitude, 36 cm.s^{-1} , than that observed *in vivo*, 10 cm.s^{-1} . This model allows sensitivity of valve behaviour to valve material properties and the stress free configuration of the leaflets to be explored in future work. The variation in pressure drop with flow reported in Figure 4b, and the variation of this response with valve parameters in this 3D model, can be used to inform 0D models of valve behaviour in the context of 1D modelling of circulatory haemodynamics [7] along with extension of the model to transient valve behaviour and pulsatile flows.

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COUPLING IN VIVO HUMAN MITRAL VALVE TO THE LEFT VENTRICLE

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SUMMARY

We present an integrated model of mitral valve (MV) coupled with the left ventricle (LV). The model is derived from clinical images and takes into account of the important valvular features, left ventricle contraction, nonlinear soft tissue mechanics, fluid structure interaction, and the MV-LV interaction. The integrated MV and LV model can simulate the cardiac function both in diastole and systole. Although the model is a step closer towards simulating physiological realistic situation, further work is required to ensure that the highly complex valvular-ventricular interaction, and the fluid-structure interaction, can be reliably represented.

Key words: *human mitral valve, left ventricular coupling, fluid structure interaction, immersed boundary method*

1 INTRODUCTION

Computational modelling of the MV mechanics, particularly within the context of the left ventricle (LV) environment, can enhance our understanding of the valvular-ventricular interaction, and potentially lead to more efficient MV repairs and replacement. Since the structure of the MV is closely tied to the left ventricle through the chordae connection, it is important to simulate the dynamics of MV by taking into account of LV dynamics, as well as the fluid-structure interaction (FSI) between the MV and LV. Kunzelman, Einstein and co-workers first started to simulate normal and pathological mitral function [1, 2] with FSI. Over the last few years there have been a number of FSI valvular models [3, 4], none of these included the effect of the LV motion, hence the flow field is not physiological. Yin et al. [6] modelled a chordaed MV inside a LV and identified fluid vortices associated with the LV motion. However, the LV motion was modelled as a set of prescribed moving boundary, and the MV model was simply constructed using a network of linear elastic fibres. Chandran and Kim [2] recently reported a prototype FSI MV dynamics in a simplified LV chamber model. To date, there has been no work reported that includes both the MV and LV models and the fluid-structure interaction properly.

In this study, we have developed a fully integrated MV-LV model, which is image-derived and simulated using a hybrid immersed boundary-finite element framework (IB/FE) [7], and takes into account of the important valvular features, left ventricle contraction, nonlinear soft tissue mechanics, and fluid-structure interaction.

2 METHODOLOGY

The hybrid IB/FE method employs an Eulerian description of the viscous incompressible fluid, along with a Lagrangian description of the structure that is immersed in the fluid. Interactions between the Lagrangian and Eulerian fields are achieved by integral transforms with discrete Dirac delta function kernels. Readers may refer to [7] for more details of the IB/FE method.