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Validation of a Fluid-Structure Interaction Model of an Aortic Dissection versus a Bench Top Model

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***Abstract*—**The aim of this investigation was to validate the fluid-structure interaction (FSI) model of type B aortic dissection with our experimental results from a bench-top-model. Another objective was to study the relationship between the size of a septectomy that increases the outflow of the false lumen and its effect on the values of the differential of pressure between true lumen and false lumen. FSI analysis based on Galerkin's formulation was used in this investigation to study flow pattern and hemodynamics within a flexible type B aortic dissection model using boundary conditions from our experimental data. The numerical results of our model were verified against the experimental data for various tear size and location. Thus, CFD tools have a potential role in evaluating different scenarios and aortic dissection configurations.

***Keywords*—**Aortic dissection, fluid-structure interaction, in vitro model, numerical

I. Introduction

ACUTE aortic dissection is considered to have a high death rate after the onset of the symptoms [1]. This disease starts with a tear in the inner wall of the aortic wall which allows the blood to flow within the media and eventually developing two lumina called true lumen and false lumen separated by a flap. The communication between these two lumina occurs through single and multiple tears of various sizes. Many numerical studies were conducted in the literature to study the variables that affect the development of this disease [2]-[4] and the required medical devices for its treatment [5]-[7]. Few studies have focused on modeling flow pattern and hemodynamics within the aortic dissection.

Tse et al. [8] studied numerically the hemodynamics of development of a dissecting aneurysm using a patient-specific dissected aorta. The aortic walls were assumed rigid in that study. Fan et al. [9] studied numerically the effect of various pertinent parameters such as the size and location of reentry tear and the area ratio of false lumen to true lumen on the rupture of false lumen using an idealized geometry of an aortic dissection. The authors in that study assumed rigid walls and neglected the effect of the branches. Numerical analysis of flow patterns and wall shear stress distribution in rigid walls of patient-specific aortic dissection model was investigated by Cheng et al. [10]. Their results showed that the dissected aorta was dominated by locally highly disturbed, and possibly turbulent, flow with strong recirculation. Khanafer and Berguer [11] illustrated numerically how dissections developed and advanced in an idealized simplified descending aorta. Their results showed that the media layer exhibited greater wall stress than both adventitia and intima layers which play an important role in the progression of an aortic dissection disease. Computational study of anatomical risk factors (tear size, location, and false lumen location) in idealized rigid models of chronic aortic dissection (straight and curved vessels) was studied by Ahmed et al. [12]. The results of that study showed that larger false lumen pressure occurred when distal tear was small or absent. Khanafer et al. [13] conducted a numerical study to investigate hemodynamics within a rigid wall aortic dissection model. Their results showed that the size of tear had a profound effect on the pressure difference between false lumen and true lumen.

One can note from the above cited literature that there are few studies investigating wall stresses and flow pattern in aortic dissection disease and they are all considered rigid wall models or geometries of aortic dissections without branches. Therefore, the aim of this study is to analyze correctly flow pattern and hemodynamics in an aortic dissection model of different tear sizes and locations using FSI.

II. Mathematical Formulation

The three-dimensional model of an aortic dissection used in this investigation with variable tear size and location is shown in Fig. 1. An ex-vivo dynamic flow setup was constructed to mimic the human circulatory system. The experimental setup consisted of the phantom, a compliance chamber, pulsatile pump, and a data acquisition system. True lumen wall was made of a glass, while the false lumen, flap, and branches were constructed with Polytetrafluoroethylene (PTFE) (Elastic modulus = 9 MPa). Water was used as a perfusion fluid flowing through both the numerical and in vitro models, with a density of 1000 kg/m^3 and a dynamic viscosity of 8.90×10^{-4} Pa·s, respectively [13]. Moreover, water was assumed to be incompressible, homogeneous and Newtonian. No-slip boundary condition was applied at the aortic dissection wall. An arbitrary Lagrangian-Eulerian formulation was employed to describe the fluid motion in the FSI model. The governing equations for the fluid domain are described as:

Continuity:

$$\nabla \cdot \mathbf{u} = 0 \tag{1}$$

Momentum:

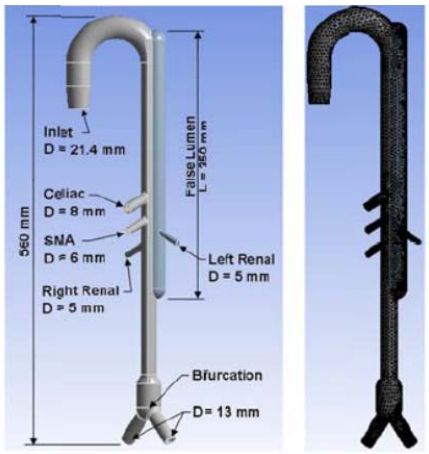
$$\rho_f \frac{\partial \mathbf{u}}{\partial t} + \rho_f (\mathbf{u} - \mathbf{w}) \cdot \nabla \mathbf{u} = \nabla \cdot \boldsymbol{\sigma}_f + \mathbf{f}_f^B \tag{2}$$

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where **u** is the fluid velocity vector, ρ_f the fluid density, **w** the moving coordinate velocity, $\mathbf{f}_f^B$ the body force per unit volume, $(\mathbf{u}-\mathbf{w})$ the relative velocity of the fluid with respect to the moving coordinate velocity, and $\boldsymbol{\sigma}_f$ the fluid stress tensor. The governing equation for the solid domain of the FSI model can be described by the following elastodynamics equation:

$$\rho_s \ddot{\mathbf{d}}_S = \nabla \cdot \boldsymbol{\sigma}_s^{\text{total}} + \mathbf{f}_S^B \tag{3}$$

where $\ddot{\mathbf{d}}_s$ represents the local acceleration of the solid region ($\ddot{\mathbf{d}}_s = \dot{\mathbf{w}}$), $\mathbf{f}_S^B$ the externally applied body force vector at time t, ρ_s the density of the arterial wall, and $\boldsymbol{\sigma}_s$ the solid stress tensor.



(a) (b)

Fig. 1 (a) Schematic diagram of the dissection model, (b) Mesh

Time-dependent pressure waveforms and velocities obtained from our experimental setup (Figs. 2 and 3) were applied at the inlet and outlet branches of the fluid domain (Fig. 1).

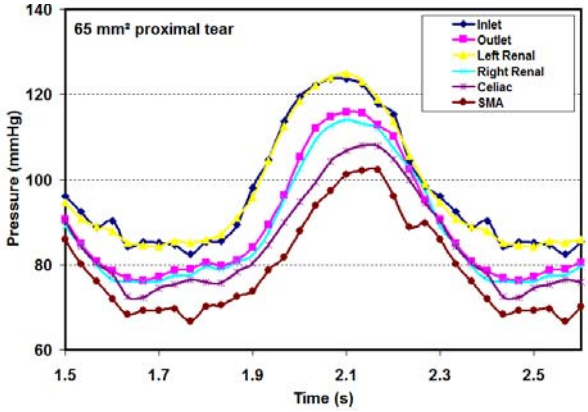


Fig. 2 Boundary conditions of pressure waveforms for 65 mm^2 proximal tear and 0 mm^2 distal tear model

III. Numerical Scheme

Finite element method (ANSYS FLUENT 15) was used to solve the governing equations for flow pattern and wall stresses in a flexible wall of an aortic dissection model. Variable mesh size was used in the vicinity of the tears and the walls of the model to capture the large variations in the dependent variable. Both Newton-Raphson method and 4-Node tetrahedral element were selected in this investigation. For better accuracy and convergence, second-order scheme was used for spatial discretization and transient formulation. The solution was considered converged when the maximum relative error between two following time steps in the dependent variable was less than 10^{-3}. The time step in the transient analysis was assumed constant and was equal to 5×10^{-3} s.

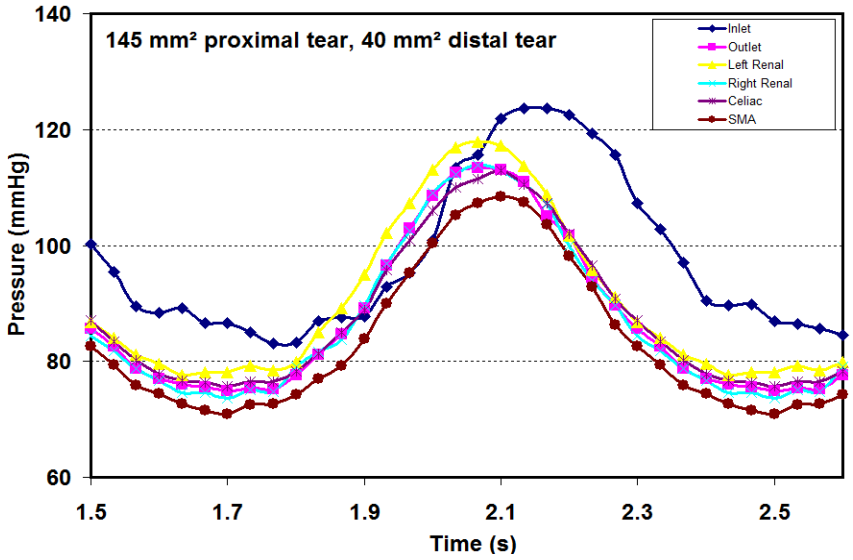


Fig. 3 Boundary conditions of pressure waveforms for 145 mm^2 proximal tear and 40 mm^2 distal tear model

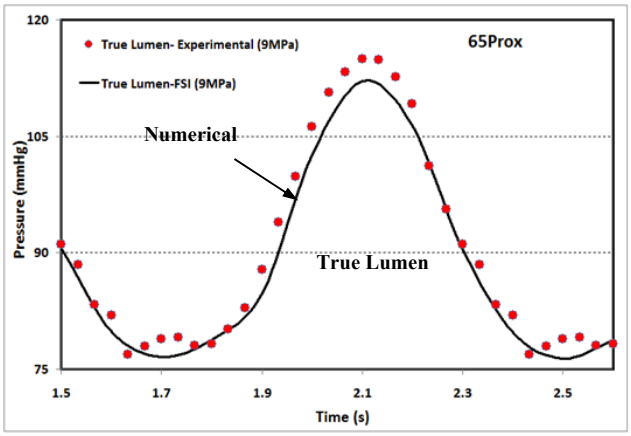
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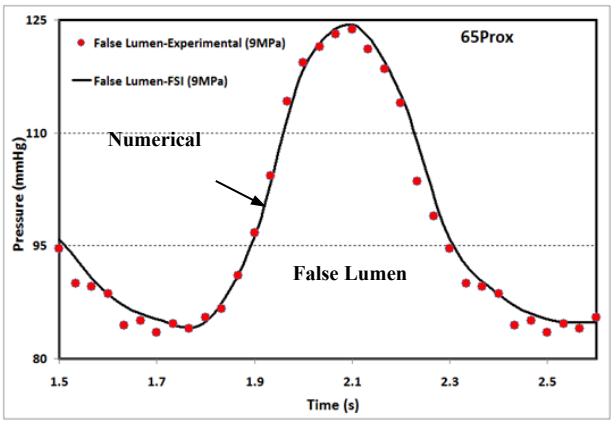
IV. Results and Discussion

The present FSI simulation of an aortic dissection model was validated in different bench-top models of aortic dissection representing different scenarios of proximal and distal tears. A 65 mm^2 proximal tear only model as well as 145 mm^2 proximal tear and 40 mm^2 distal tear of an aortic dissection model were used for validating the CFD tool.

Fig. 4 shows a comparison of the temporal variation of the pressure in the true and false lumina between experimental and numerical results for 65 mm^2 proximal tear and 0 mm^2 distal tear. A very good correlation is found between the FSI numerical and experimental results.



(a)

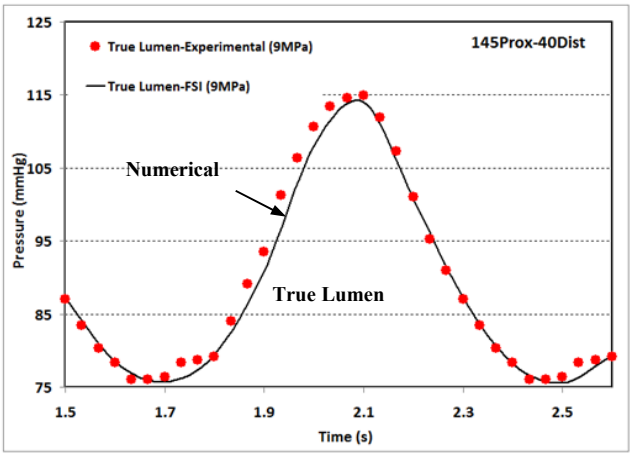


(b)

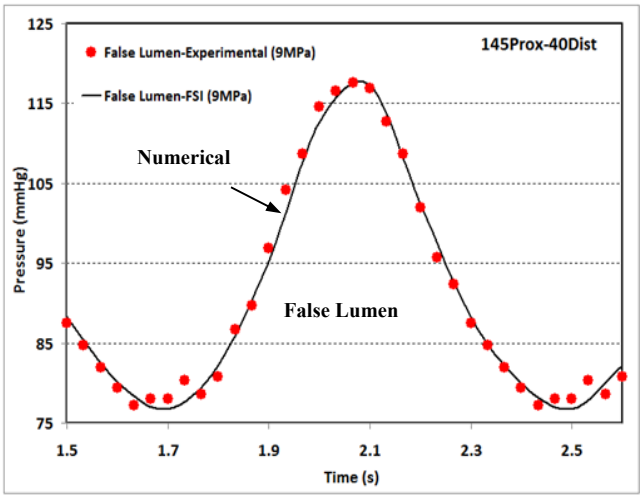
Fig. 4 Comparison of the temporal variation of the pressure in (a) true lumen and (b) false lumen between experimental and numerical results for 65 mm^2 proximal tear and 0 mm^2 distal tear model

Similarly, Fig. 5 illustrates an excellent agreement between experimental and the FSI numerical results of the temporal variation of the pressure in true and false lumena for 145 mm^2 proximal tear and 40 mm^2 distal tear aortic dissection model.

Fig. 6 illustrates the effect of adding the areas of proximal and distal tears on the pressure difference between false lumen and true lumen at both peak systole and end diastole. Fig. 5 shows that as the total area of both tears increases, the pressure difference between both lumena decreases. The total area of 250 mm^2 created by septectomy may equalize the pressure in both lumena.



(a)



(b)

Fig. 5 Comparison of the temporal variation of the pressure in (a) true lumen and (b) false lumen between experimental and numerical results for 145 mm^2 proximal tear and 40 mm^2 distal tear model

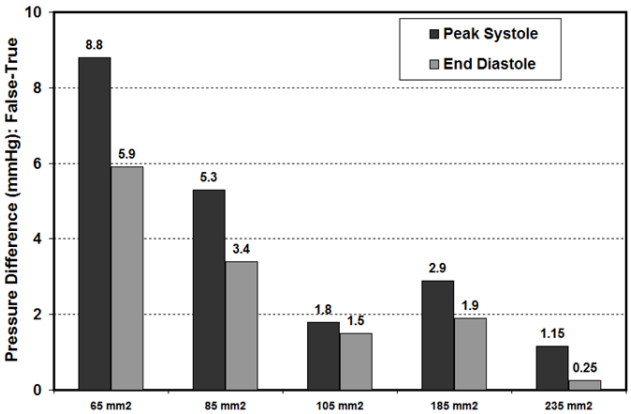


Fig. 6 Pressure difference between false and true lumina vs. total area of the tear sizes

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V. Conclusion

The results of the present numerical simulations were validated against our experimental data for different tear sizes and location. The results indicated that, as the total area of proximal and distal tears increases, the pressure difference decreases. This study shows that CFD can be used to study correctly hemodynamics in vascular diseases and may help clinicians to develop new treatment methods and medical devices.

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