Coupled SPH Model for Prosthetic Valve Induced Thrombus Formation

Laha Sumanta*
Univ. of Manchester
School of Engineering
Manchester, United Kingdom
sumanta.laha@manchester.ac.uk

Fourtakas Georgios
Univ. of Manchester
School of Engineering
Manchester, United Kingdom

Das K. Prasanta
IIT Kharagpur
Dept. of Mechanical Engg.
West Bengal, India

Keshmiri Amir
Univ. of Manchester
School of Engineering
Manchester, United Kingdom

I. INTRODUCTION

Cardiovascular diseases, particularly valvular disorders, remain a leading cause of mortality worldwide, highlighting the urgent need for advanced treatments. While early-stage valvular diseases can be managed with medication, severe cases often require surgical intervention, typically involving prosthetic heart valve implantation. Mechanical heart valve (MHV) transplantation significantly improves life expectancy and quality of life but is often associated with complications like thrombus formation, haemolysis, and pannus growth.

Thrombus formation near the valve is a major concern, primarily caused by improper positioning, tissue damage, and turbulent blood flow. Governed by the coagulation cascade, thrombus formation involves intrinsic and extrinsic pathways that activate platelets and drive fibrin deposition. Prolonged exposure of subendothelial cells to blood flow in damaged regions triggers platelet adhesion and activation, reinforced by the coagulation cascade. MHVs, while effective, disrupt physiological haemodynamics, inducing high shear stresses that damage tissue and elevate thrombosis risk. Additionally, their non-physiological surfaces can trigger platelet activation and aggregation.

Computational fluid dynamics (CFD) is effective and efficient in studying haemodynamics in cardiovascular devices, enabling design optimisation and better post-implantation strategies. However, traditional grid-based methods like Finite Volume Methods (FVM) and Lattice Boltzmann Methods (LBM) face challenges in modelling valve motion, handling large deformations, and capturing dynamic interfaces crucial for thrombosis studies. Smoothed Particle Hydrodynamics (SPH), a fully Lagrangian method, addresses these challenges by representing the fluid domain with particles carrying physical properties like velocity and pressure and biochemical concentrations. This approach excels at capturing sharp interfaces and fluid-structure interactions, making it ideal for studying mechanical valves and thrombus formation.

Building on our earlier SPH-based FSI model for valve motion [1], we have developed a novel SPH thrombus formation model [2] that incorporates wall shear stress effects, advection terms, and fibrin-dependent platelet velocity treatments. This method addresses limitations of earlier models, such as those by Monteleone et al. [3], by employing unique techniques like nearest-neighbour searching for platelet activation a dissipation-based approach for fibrin interactions. The present work proposes a novel coupled SPH model integrating valve dynamics with thrombus formation, offering a comprehensive framework for studying valve-induced thrombosis. Validated against experimental data, including thrombus formation in backward-facing step (Taylor et al.[4]), the initial thrombus model shows accuracy and reliability in predictions.

Finally, the model has been applied to assess thrombus formation across four different valve mounting orientations. By modelling thrombus growth and flow dynamics for each configuration, we identified the optimal implantation site to reduce thrombus-related risks. This integrated approach provides valuable insights for improving valve implantation strategies and reducing post-operative complications, ultimately enhancing patient outcomes.

II. MATERIAL AND METHODS

The present work is structured into three key parts. First, the governing equations of fluid mechanics, including mass and momentum conservation, are utilised using the C++-based open-source SPH solver, DualSPHysics [5]. Fluid-structure interaction (FSI) modelling is modelled via Project Chrono, seamlessly integrated into DualSPHysics, with detailed documentation available in [1][6]. Second, a unique wall shear stress (WSS) model is developed [7], serving as an activation function integral to the thrombus model. This WSS framework is important in the thrombus dynamics throughout the study. Finally, a FSI coupled thrombus model is introduced, employing two distinct approaches—penalty and dissipation—to simulate thrombus growth. The penalty approach utilises a fibrin-dependent velocity penalty term, while the dissipation approach links fibrin concentration with the Einstein equation.

The modelling process begins with identifying damaged zones where high wall shear stress ($\tau_h > 10Pa$) is activating platelets on a damaged vessel surface, and low wall shear stress ($\tau_l < 0.02Pa$) promotes platelet adhesion and accumulation. Fluid particles within a distance of $2h$ (smoothing radius) from the damaged wall are considered contributors to thrombus formation. Prothrombin concentration reduces with thrombin formation, governed by an advection-diffusion equation (Eq. 1)

Following the methodology proposed by Monteleone et al. [3], a source term has been selected based on the chemical kinetics.

$$\frac{\partial C_k}{\partial \mathrm{t}} = \nabla \cdot (D_k \nabla C_k) - \nabla \cdot (\mathbf{u_k} C_k) + R_k \qquad (1)$$

where C is the concentration of the bio-chemical species, D is the diffusion coefficient, $\mathbf{u}$ is the velocity of the particle and R is the respective source term. Eq. (1) has been solved for prothrombin, thrombin, fibrin and activated platelet respectively. The above equation discretised and computed through SPH approximation. The SPH version of the equation for a particular species is as follows, where i and j represent the interpolating and neighbouring particles, respectively,

$$\frac{\partial C_i}{\partial \mathrm{t}} = \sum_{j \epsilon P} m_j \left(\frac{4 D_i \mathbf{r_{ij}} \cdot \nabla_i W_{ij}}{(\rho_i + \rho_j) + (r_{ij}^2 + 0.01 h^2)} \right) C_{ij} - C_i \sum_{j \epsilon P} \frac{m_j}{\rho_j} (\mathbf{u}_i - \mathbf{u}_j) \cdot \nabla W_{ij} - \mathbf{u}_\mathrm{i} \sum_{j \epsilon P} \frac{m_j}{\rho_\mathrm{j}} (C_i - C_j) \cdot \nabla W_{ij}, \quad + R_i \qquad (2)$$

The source term for the prothrombin and thrombin are as follows,

$$R_{th} = k_{th}^{rp} C_{rp} C_{pt} + k_{th}^{ap} C_{ap} C_{pt} \qquad (3)$$

$$R_{pt} = -R_{th} \qquad (4)$$

where k_{th}^{rp} and k_{th}^{ap} is the kinetic constant of the resting and activated platelets for the thrombin conversation from the prothrombin. C_{rp}, C_{ap}, C_{pt} are the concentration of the resting platelet, activated platelet and prothrombin, respectively.

When thrombin concentration of a particle exceeds a threshold (C_{th_th}), they are considered as activated platelets. Biochemical interactions between prothrombin, thrombin, and fibrin, governed by Michaelis-Menten kinetics, drive this process, while activated platelets are drawn toward the wall or nearby activated platelets, promoting aggregation.

In the final mechanistic step, velocities of the particles must be reduced based on the fibrin concentration. In the penalty approach, a velocity penalty term proposed by Wang et al. [8] has been introduced,

$$u_{penalty} = \beta \left(1 - \left(\frac{C_{fb,i}}{C_{fb,i}, th} \right)^{1.5} \right) \qquad (5)$$

Here $C_{fb,i}$ is the concentration of the fibrin of the interpolating particle and β is a penalty factor considered 1 in the present case. This term gradually reduces particle velocity depending on fibrin concentration, ultimately setting it to rest.

$$\mathbf{u_{i\_new}} = \mathbf{u_i}\, u_{penalty} \qquad (6)$$

It has been demonstrated that reasonable prediction of thrombus formation is achieved based on the above modelling technique. However, the gradual decrease in local velocity during fibrin formation is based on an empirical penalty function. This gradual decrease in velocity is due to the deposition of fibrin being similar to fluid flow with dispersion where the viscosity may be predicted through Einstein constitutive equation. Therefore, in the dissipation approach, an additional viscous term is proposed. In the momentum equation the kinematic viscosity ν has been replaced by ν_c by using the Einstein constitutive equation for the viscosity, thus

$$\nu_c = (1 + C_{fb,i}\varphi)\nu \qquad (7)$$

where φ is an arbitrary constant tuned based on the initial fluid viscosity and has a value of $O(10)^{12}$, sufficiently large to bring the fluid to a complete stop. Therefore, the modified momentum equation based on Eq. (7) can be written as,

$$\frac{\mathrm{D}\mathbf{u}}{\mathrm{Dt}} = -\frac{1}{\rho}\nabla p + \mathbf{f} + \nu \nabla^2 \mathbf{u} + \left(C_{fb,i}\varphi\right)\nabla^2 \mathbf{u} + \frac{1}{\rho}\nabla . \boldsymbol{\tau} \qquad (8)$$

Evidently, while the concentration of fibrin (i.e. the fibrin mesh) increases, the term ν_c rises, leading to the deposition of the particles. This approach aligns closely with the physics of clot formation, as the fibrin mesh captures more platelets and reduces flow velocity, supporting the final clot formation. Thereafter, at the final stage while the concentration of the fibrin exceeds a threshold, velocity of the particles is set to zero.

III. Results and Discussions

The initial FSI model of heart valves have been rigorously validated against relevant experimental studies and FVM results. Furthermore, the novel wall shear stress estimation model has been validated to enhance the credibility of the overall framework. Detailed validations are provided in our previous works [1], [7].

Direct validation of any thrombus model remains challenging due to the limited availability of experimental data. Nonetheless, the efficacy of the proposed thrombus model is demonstrated in this work through a case study [2]. As shown in Fig. 1, a three-dimensional backward-facing step has been selected as the validation test case for thrombus formation. This configuration is commonly used in fluid dynamics to study flow recirculation, separation, and reattachment, all of which are conducive to thrombus formation. Moreover, it is a widely recognised benchmark test case for validating thrombus models.

The thrombus formation begins at the step corner and propagates both axially and radially. The top view of the thrombus shape, resembling an expanding wedge, is presented in Fig. 1 and shows good agreement with the experimental data of Taylor et al.[4]. Subsequently, a qualitative validation was conducted to further evaluate the effectiveness of the combined model. Fig. 2a illustrates that the proposed model predicts the periphery of the valve ring to be more susceptible to thrombus formation, consistent with experimental observations by Kreinin et al. [9]. During mechanical valve operation, the standard blood velocity increases through the valve due to its narrow, venturi-like structure. This results in high wall shear stress, increasing the likelihood of thrombus formation.

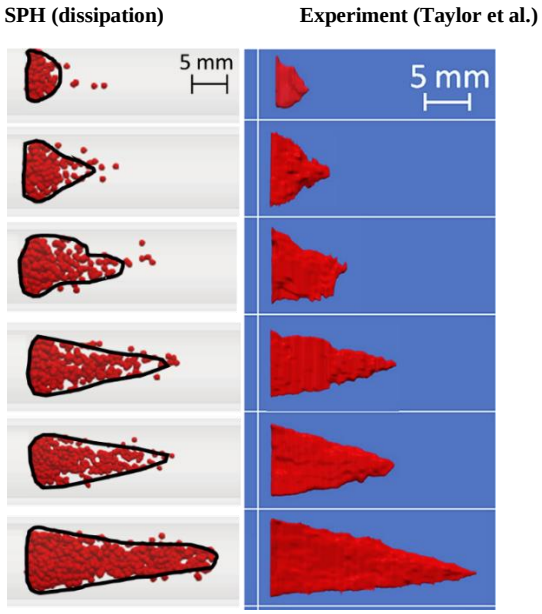

Fig. 1. Comparison of predicted thrombus deposition in the BFS geometry against the experimental observation

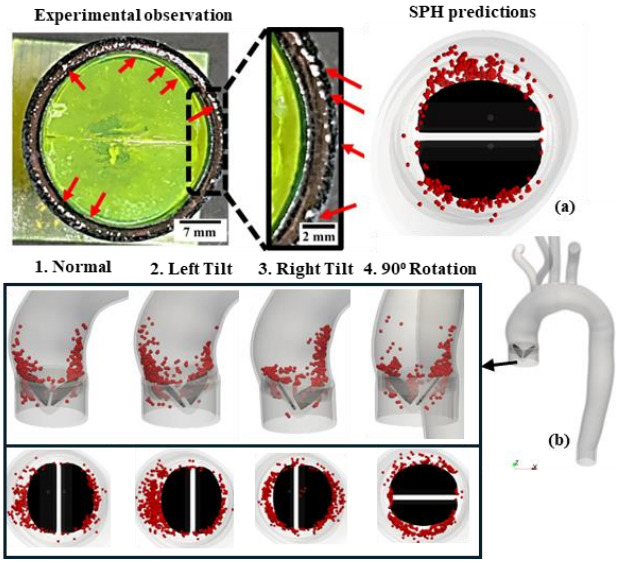

Fig 2 Comparison of the predicted thrombus location near the valve using the present model with experimental results (a). Variation in thrombus formation within the patient-specific aorta for different valve mounting orientations (b).

The orientation of the mechanical valve is observed to significantly affect thrombus formation. Since the shape of the human aorta varies, the mounting position of the valve has a pronounced impact on flow dynamics. In the present study, four different valve orientations, depicted in Fig. 2b, were analysed. Results indicate that while thrombus formation occurs predominantly at the valve periphery, the extent of thrombus propagation varies downstream of the valve. In the second and third orientations, significant thrombus formation is observed along the aortic wall adjacent to the valve mount, in the direction to the high-velocity jets induced by the mounting orientation. These jets generate high shear stress, further promoting thrombus formation. Conversely, in the first and fourth orientations, thrombus formation is more symmetrical. Notably, the fourth orientation, rotated 90 degrees relative to the first, results in moderately less thrombus formation.

IV. CONCLUSION

The novel SPH-based model successfully integrates mechanical valve dynamics and thrombus formation processes, addressing critical limitations of traditional CFD methods. Based on the findings, the mechanical valve should ideally be implanted without any tilt for this specific patient geometry to minimise excessive thrombus formation in the ascending aorta. However, it is important to note that real-world scenarios typically involve the use of anticoagulant medications, which were not considered in this study. All test cases were conducted without anticoagulants to provide an unbiased understanding of thrombus formation variations based solely on valve orientation.

In summary, the coupled smoothed particle hydrodynamics based FSI and thrombus model provides valuable insights for optimising valve implantation tailored to patient-specific geometries, ultimately enhancing patient outcomes and reducing post-operative complications.

REFERENCES

[1] S. Laha, G. Fourtakas, P. K. Das, and A. Keshmiri, “Fluid-Structure Interaction Modelling of Bi-Leaflet Mechanical Heart Valves Using Smoothed Particle Hydrodynamics,” *Phys. Fluids*, pp. 1–24, 2023.

[2] S. Laha, G. Fourtakas, P. K. Das, and A. Keshmiri, “Graphics processing unit accelerated modelling of thrombus formation in cardiovascular systems using smoothed particle hydrodynamics,” *Phys. Fluids*, vol. In Press, pp. 1–35, 2025.

[3] A. Monteleone, A. Viola, E. Napoli, and G. Burriesci, “Modelling of thrombus formation using smoothed particle hydrodynamics method,” *PLoS One*, vol. 18, no. 2 February, pp. 1–24, 2023.

[4] J. O. Taylor, K. P. Witmer, T. Neuberger, B. A. Craven, R. S. Meyer, S. Deutsch, and K. B. Manning, “In vitro quantification of time dependent thrombus size using magnetic resonance imaging and computational simulations of thrombus surface shear stresses,” *J. Biomech. Eng.*, vol. 136, no. 7, pp. 1–11, 2014.

[5] J. M. Domínguez, G. Fourtakas, C. Altomare, R. B. Canelas, A. Tafuni, O. García-Feal, I. Martínez-Estévez, A. Mokos, R. Vacondio, A. J. C. Crespo, B. D. Rogers, P. K. Stansby, and M. Gómez-Gesteira, “DualSPHysics: from fluid dynamics to multiphysics problems,” *Comput. Part. Mech.*, vol. 9, no. 5, pp. 867–895, 2022.

[6] I. Martínez-Estévez, J. M. Domínguez, B. Tagliafierro, R. B. Canelas, O. García-Feal, A. J. C. Crespo, and M. Gómez-Gesteira, “Coupling of an SPH-based solver with a multiphysics library,” *Comput. Phys. Commun.*, vol. 283, p. 108581, 2023.

[7] S. Laha, G. Fourtakas, P. K. Das, and A. Keshmiri, “Computational Particle Mechanics Towards the Estimation of Wall Shear Stress in Smoothed Particle Hydrodynamics,” *Comput. Part. Mech. Towar.*, 2024.

[8] F. Wang, S. Xu, D. Jiang, B. Zhao, X. Dong, T. Zhou, and X. Luo, “Particle hydrodynamic simulation of thrombus formation using velocity decay factor,” *Comput. Methods Programs Biomed.*, vol. 207, p. 106173, 2021.

[9] Y. Kreinin, Y. Talmon, M. Levi, M. Khoury, I. Or, M. Raad, G. Bolotin, J. Sznitman, and N. Korin, “A Fibrin-Thrombin Based In Vitro Perfusion System to Study Flow-Related Prosthetic Heart Valves Thrombosis,” *Ann. Biomed. Eng.*, vol. 52, no. 6, pp. 1665–1677, 2024..