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Using Discrete Multi-Physics for studying the dynamics of emboli in flexible venous valves

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ABSTRACT

Emboli, which are parts of blood clots, can be stuck in the vasculature of various organs (most frequently, lungs) and cause their malfunction or even death. In this work, using mathematical modelling, different types of emboli-like structures are studied in a double venous valve system. The model is implemented with a fully Lagrangian Discrete Multi-Physics technique and the flow is governed by flexible walls. The study shows the effect of different diameters and lengths of a free embolus in the flow surrounding the valve. The presence of an embolus strongly affects the dynamics of both the fluid and the leaflets in venous valves and the permanence of the embolus in the valve chamber is narrowly linked with its length.

1. Introduction

An embolus is generally formed when a section of a thrombus detaches and circulates in the cardiovascular system until reaching narrow vessels, most frequently, in the lungs [1–3]. When emboli are trapped, they can obstruct blood flow in the lungs leading to a potentially life-threatening complication known as pulmonary embolism (PE) or deep vein thrombosis (DVT). In the UK alone, around 25,000 deaths are caused by PE or DVT; this number is five times higher than those from breast cancer, AIDS and road accidents combined [4].

While medical research highlights the role of DVT on the hydrodynamics around venous valves [5], the actual physical interaction of the embolus with the valve remains unexplored. The literature provides a wide range of publications about venous and arterial thrombosis, but the majority of these studies focuses on thrombogenesis and clotting [6–14], rather than the dynamics of the embolus.

To circumvent the current limitations of in-vivo and in-vitro models, computer simulations (in-silico modelling) of the venous valve have been carried out but, with a few exceptions [13,15,16], emboli are not accounted for. For DVT however, this represents a serious limitation since the presence of the embolus changes considerably the hydrodynamics around the valve.

Previous studies [15,16] have shown that the diameter, the elasticity and the location of an embolus affect the flow in arterial bifurcations [15] or in Inferior Vena Cava (IVC) [16]. But the interaction of the embolus with more complex settings such as the flexible leaflets of the venous valve has not been investigated. Only Simão et al. [13] consider the presence of solid particles in the venous valve, but these are simple Lagrangian point particles and the flow, therefore, is not fully resolved around them.

By modelling the physical interaction of emboli with different shapes and flexibilities with the soft leaflets of the venous valve and the change of hydrodynamics that this involves, this paper fills a gap in the literature since the valve environment is probably the most critical for DVT and the presence of clots in the vicinity of the valve has been associated with the occurrence of new thrombosis activation sites [5].

2. Methodology

2.1. Modelling

A hybrid approach, based on a particle framework, is implemented to model haemodynamics and solid structure deformation. The technique, called Discrete Multi-Physics (DMP) [14,17,18], associates Smoothed Particle Hydrodynamics (SPH) [19–21] and the Mass and Spring Model (MSM) [22–24] and has been used to model the fluid-structure interactions occurring in deep vein valves [14], cardiac valves [17] and the intestine [18].

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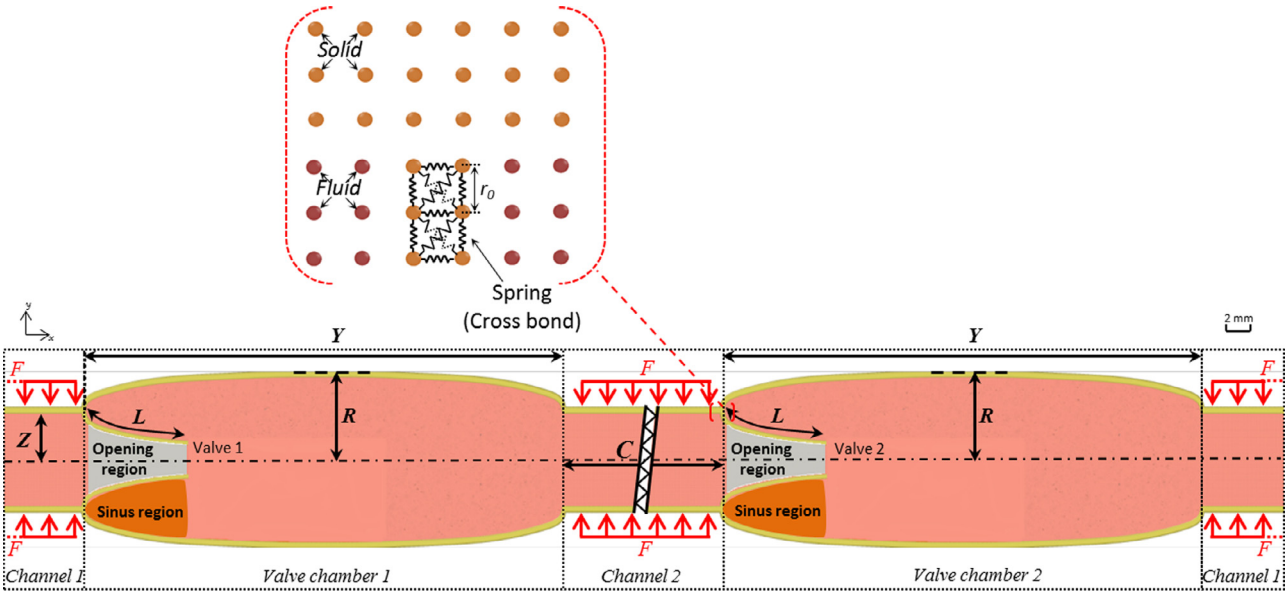

Fig. 1. Illustration of the double venous valve 2D geometry.

In this approach, the liquid is represented by SPH particles that interact with each other by viscous and pressure forces and the tissues by MSM particles inter-connected by means of computational springs (to model the elastic modulus) and dashpots (to model viscoelasticity). The essential ideas behind the DMP method are summarized in Appendix A; the reader can refer to [24] for a more extensive explanation of the DMP theory and to [22] for applications in different fields such as lava flows, cell dynamics and solid-liquid flow.

2.2. Geometry

In this study, we use a 2D schematic representation of a double leg venous valve system. The geometry is similar to the short valve model [14] but here it is used with two antagonist valves (Fig. 1). The channel radius is $Z = 0.004$ m, the membrane length is $L = 0.01$ m, the radius of the valve chamber is $R = 0.007$ m and its length is $Y = 0.04$ m [14]. The two valve systems are interconnected and the total length between the two chambers is $C = 0.048$ m (0.046 m in [25]). The external walls are divided into four parts: two flexible sections where an external force is applied (Channel 1 and Channel 2 in Fig. 1), and two valve chambers (Valve chamber 1 and Valve chamber 2 in Fig. 1) that contain the leaflets. Since periodic boundary conditions are enforced, the fluid exiting from the channel opening on the right is reinserted to the channel opening on the left and vice versa. For the same reason, Channel 1 in Fig. 1 appears to be divided into two sections (one on the left and another on the right), but, computationally, the two ends are joined together by the periodic boundary conditions. A net fluid flow is achieved by means of external forces acting alternatively on Channel 1 and Channel 2 (Fig. 1). When a 'squeezing' force F (see Fig. 1) is applied to Channel 1, Valve 1 opens, Valve 2 closes, and the fluid flows from the left to the right. When F is applied to Channel 2, Valve 2 opens, Valve 1 closes, and the fluid maintains the same direction from the left to the right. This approach mimics the actual motion of blood in the legs' veins induced by the contraction of the surrounding muscles.

In the rest of the paper, we refer to the regions between the leaflets (in both Valve 1 and Valve 2) as the 'opening regions' and to the regions between the walls and the leaflets (in both Valve chamber 1 and Valve chamber 2) as 'sinus regions' (Fig. 1).

Table 1
Model parameters used in the simulations.

Parameter	Value
SPH (Eqs. (A.5)–(A.7))	
Number of SPH wall particles (3 layers)	10,722
Number of SPH valve particles (2 layers)	584 (146 particles/leaflet)
Number of SPH fluid particles	157,370
Mass of each particle (fluid)	1.05×10^{-5} kg
Mass of each particle (solid)	2×10^{-5} kg
Initial distance among particles Δr	1×10^{-4} m
Smoothing length h	2.5×10^{-4} m
Artificial sound speed c_0	10 m s^{-1}
Density ρ_0	1056 kg m^{-3}
Viscosity μ_0	0.0035 Pa s
Time step Δt	10^{-6} s
Force F	0.008 N
MSM (Eq. (A.10))	
Hookian coefficient k_b (Wall)	1×10^5 J m^{-2}
Hookian coefficient k_b (membrane)	5×10^6 J m^{-2}
Viscous damping coefficient k_v (Wall)	1 kg s^{-1}
Viscous damping coefficient k_v (membrane)	0.1 kg s^{-1}
Equilibrium distance r_0	1×10^{-4} m
BOUNDARIES (Eq. (A.14))	
Constant K	4×10^{-4} J
Repulsive radius r^*	1×10^{-4} m

According to the DMP approach (see Appendix A) both the fluid and the solid are represented by discrete computational entities, which we call 'fluid particles' and 'solid particles'. The different behaviour of fluid and solid particles depends on the type of forces the DMP algorithm applies to each computational particle. If these forces model the viscous and pressure forces commonly acting on fluids, the computational particle behaves like a fluid; if they model the viscoelastic forces acting on solid, the particle behaves like a solid. Computationally, the fluid forces are calculated with the SPH method while the solid forces particles by means of springs (MSM model) as detailed in Appendix A. The wall delimiting the valve chamber is considered stationary and, therefore, no forces are applied to the computational particle representing this part of the domain. Solid-liquid boundary conditions are also modelled by means of inter-particle forces that model no-penetration and no-slip conditions as explained in Appendix A.