Journal of Biomechanics 48 (2015) 1789–1795

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Thrombus deflector stent for stroke prevention: A simulation study

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ARTICLE INFO

Article history:
Accepted 6 May 2015

Keywords:
Vascular device
Atrial fibrillation
Aortic flow
Multiphase flow
Embolus
Computational fluid dynamics

ABSTRACT

Atrial fibrillation (AF) is a dysfunction of heart rhythm and represents an increased predisposition to ischemic stroke in AF patients. It has been shown that the AF-induced hemodynamic conditions may contribute to the increased embolic propensity through the carotid arteries. We simulated a stroke-prevention device with a unique strut structure to deflect the trajectory of a blood clot to the carotid artery. We identified the important determinants of functionality in a device design using computational fluid dynamics simulations. Quantitative assessment of deflection efficacy over various clot dimensions was carried out for the device with different strut configurations under AF flow conditions. The simulations demonstrate that the trajectory of a clot destined to the left common carotid artery (LCCA) can be deflected by a strut-structured device at the LCCA inlet with virtually no change in flow resistance. The deflection efficacy of the device is dependent on the clot properties and strut designs of the device. A configuration of 0.75 mm thick and 0.75 mm distant struts with 50% of surface convexity were found to provide maximum deflection efficacy (e.g., 36% greater deflection efficacy than a flat filter) among the strut structures considered. The results suggest that a deflector stent implanted in the aortic branch may be an effective stroke-prevention device. The present simulations motivate pre-clinical animal studies as well as further studies on patient-specific design of the device that maximize the deflection efficacy while minimizing device safety issues.

1. Introduction

Stroke causes high morbidity and mortality in developed countries and is the fifth leading cause of death in the U.S. (Mozzafarian et al., 2015). It has been shown that majority of strokes is ischemic (88% of all strokes) and embolic stroke is the second most common type of stroke at 22% occurrence (Casals et al., 2011). Cerebral embolism is the cause of ischemic stroke in 40–80% of cases (Bogousslavsky et al., 1991; Vukovic-Cvetkovic, 2012). Atrial fibrillation (AF) is a dysfunction of heart rhythm that is a substantial independent risk factor for stroke (Bogousslavsky et al., 1990; Britton and Gustafsson, 1985; Hart and Halperin, 1999; Hinton et al., 1977; Hylek et al., 2003; Manning et al., 1995; Wolf et al., 1991, 1978). Patients with AF have been shown to have an increased risk for ischemic stroke such that those with non-valvular AF have at least a 5-fold increased risk while those with valvular AF have an estimated 17-fold increased risk (Bungard et al., 2000). In this setting, stroke occurs through the formation of a blood clot in left atrial appendage followed by embolization to cerebral vessels (Hart et al., 2003; Hylek et al., 2003). The motion of a blood clot in the blood stream is affected by a number of physical and hemodynamic factors which include clot size and velocity as well as the characteristics of hemodynamic waveforms (e.g., amplitude and frequency) (Choi et al., 2013).

AF has been shown to lead to a broad spectrum of abnormal cardiac hemodynamic parameters (Clark et al., 1997; Greenfield et al., 1968; Nanthakumar and Kay, 2002; Popovic et al., 2002). We have recently shown how the hemodynamic conditions affect the propensity of cerebral embolism and demonstrated that the hemodynamic perturbations caused by AF may play a significant role in the increased embolic propensity to the brain (Choi et al., 2013). Here, we conceptualize a stroke-prevention device with a stent strut structure that deflects emboli. We identify the important determinants of functionality in a virtual stent design and assessed applicability of the device in aortic arch using computational fluid dynamics (CFD) simulations. Under AF flow condition, trajectories of blood clots with diverse physical properties were tracked with and without the device. Deflection efficacy of the device was examined by comparing the clot embolization frequency with and without the device. The present computational platform provides a quantitative analysis for the device feasibility which may be used for future development of patient-specific devices for stroke prevention.

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http://dx.doi.org/10.1016/j.jbiomech.2015.05.006
0021-9290/

2. Methods

2.1. Computational domain

The aortic arch geometry used for computational simulations in the present study is a scale up of a mouse model based on our previous studies (Choi et al., 2013; Huo et al., 2008) (Fig. 1A). Although the dimension of human aortic arch morphology is diverse (Casteleyn et al., 2010; Shin et al., 2008), the morphometric details (diameters) of the aortic arch used are consistent with our previous study (Choi et al., 2013) and commensurate with the human aortic arch (i.e., 14.1 mm of LCCA diameter) (Shin et al., 2008). Specifically, the diameter of LCCA was set at the higher limit of the range of aortic branch dimensions (Shin et al., 2008) for a more aggressive estimate of clot embolization. Fig. 1B and C represents the discretized domain or computational mesh distribution in the aortic arch model for 3D simulations and a schematic drawing of the conceptualized stroke-prevention device examined, respectively. To numerically solve the governing equations for a given computational domain, an optimal fine mesh consisting of 2.4 million tetrahedral elements was adopted with a dense mesh distribution especially near the wall, branching regions, and device to capture the highly dynamic flow patterns and boundary layer (Fig. 1B). This mesh density is in line with the previous studies on aortic arch flow simulations (Cheng et al., 2010; Gallo et al., 2012; Osorio et al., 2011). The mesh distribution was verified to yield mesh-independent solutions; i.e., an increase to 3.1 million mesh elements did not lead to a significant difference ($<2\%$ change) in the temporal profiles of velocity at various positions of aortic arch and branches. Furthermore, the mesh was constructed such that $y+$ values remain <1 over the entire region of wall for the steady flow simulation with a cycle-averaged velocity at the inlet.

2.2. Flow modeling

The flow field in aortic arch was obtained by solving the continuity and momentum equations as follows:

$$\text{Continuity:} \quad \nabla \cdot \vec{V} = 0, \tag{1}$$

$$\text{Momentum:} \quad \rho \frac{\partial \vec{V}}{\partial t} + \rho(\vec{V} \cdot \nabla)\vec{V} = -\nabla p + \nabla \cdot \left(\mu \left(\nabla \vec{V} + (\nabla \vec{V})^T \right) \right), \tag{2}$$

where $\vec{V}$, p, μ, and ρ are velocity, pressure, dynamic viscosity of blood, and blood density, respectively. Blood was assumed as an incompressible Newtonian viscous fluid which is consistent with previous studies (Choi et al., 2013; Gallo et al., 2012; Lantz et al., 2012; Osorio et al., 2011) since the incorporation of non-Newtonian viscosity into the aortic turbulent flow model has been shown to not lead to discernible changes in flow patterns (Cheng et al., 2010).

The hemodynamics in aortic arch is often characterized as complex flow patterns due to the arch curvature and branches. Various computational approaches including turbulence models have been used to simulate the aortic flow dynamics (Cheng et al., 2010; Gallo et al., 2012; Lantz et al., 2012; Niu et al., 2009; Osorio et al., 2011). It is challenging to accurately describe the complex flow patterns in the aortic arch. In our previous study (Choi et al., 2013), the influence of flow modeling on the trajectory of a blood clot in the aortic arch has been examined. To effectively capture the complex flow behavior in the aortic arch, especially in the presence of deflector stents, the k–ω SST (shear stress transport) model was adopted for all simulations.

A variety of velocity profiles has been specified as boundary conditions for CFD simulations at the aortic inlet in previous studies including measured flow rate (Cheng et al., 2010; Gallo et al., 2012; Huo et al., 2008; Lantz et al., 2012; Morbiducci et al., 2013; Niu et al., 2009; Tan et al., 2009). For the present simulations

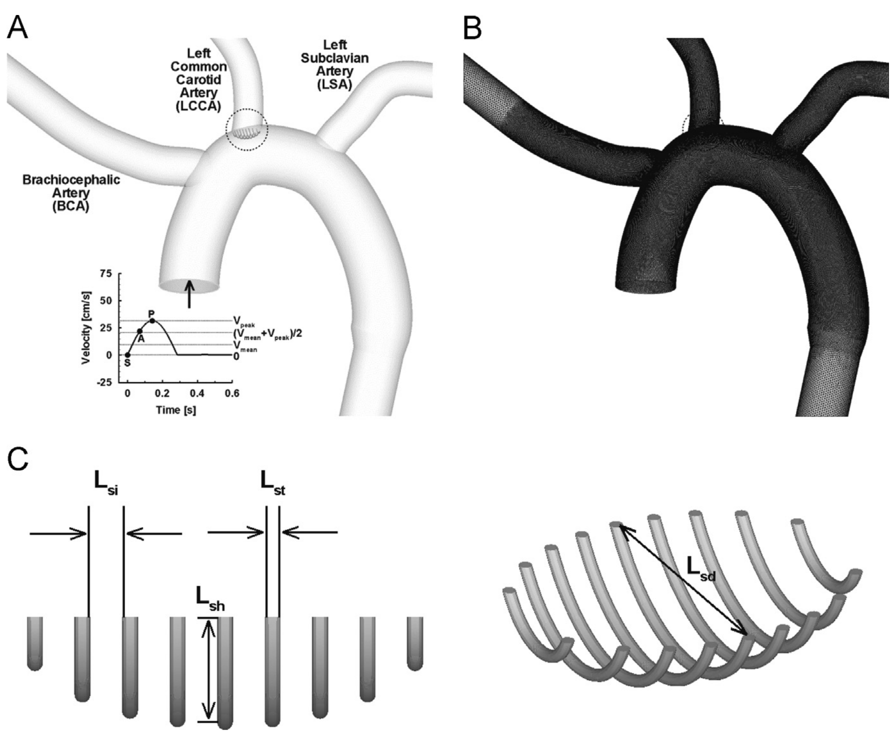

Fig. 1. (A) Aortic arch geometric model and AF flow condition imposed at the inlet. The initial positions of a clot were assumed to be circumferentially distributed at a radial distance of 0–40% of the aortic radius. (B) An optimal mesh distribution that is sufficient to resolve the complex flow patterns near vessel wall, aortic branches, and device struts and to capture the trajectory of a blood clot. (C) Schematic of a deflector implemented at the LCCA inlet (dotted circle in Fig. 1A). L_{st}, L_{si}, L_{sh}, and L_{sd} denote strut thickness, interval, height, and diameter, respectively. S, A, and P denote the beginning of systolic, accelerating, and peak stage of a cardiac cycle, respectively.