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Investigation of hemodynamics in the development of dissecting aneurysm within patient-specific dissecting aneurismal aortas using computational fluid dynamics (CFD) simulations

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ABSTRACT

Aortic dissecting aneurysm is one of the most catastrophic cardiovascular emergencies that carries high mortality. It was pointed out from clinical observations that the aneurysm development is likely to be related to the hemodynamics condition of the dissected aorta. In order to gain more insight on the formation and progression of dissecting aneurysm, hemodynamic parameters including flow pattern, velocity distribution, aortic wall pressure and shear stress, which are difficult to measure in vivo, are evaluated using numerical simulations. Pulsatile blood flow in patient-specific dissecting aneurismal aortas before and after the formation of lumenal aneurysm (pre-aneurysm and post-aneurysm) is investigated by computational fluid dynamics (CFD) simulations. Realistic time-dependent boundary conditions are prescribed at various arteries of the complete aorta models. This study suggests the helical development of false lumen around true lumen may be related to the helical nature of hemodynamic flow in aorta. Narrowing of the aorta is responsible for the massive recirculation in the poststenosis region in the lumenal aneurysm development. High pressure difference of 0.21 kPa between true and false lumens in the pre-aneurismal aorta infers the possible lumenal aneurysm site in the descending aorta. It is also found that relatively high time-averaged wall shear stress (in the range of 4–8 kPa) may be associated with tear initiation and propagation. CFD modeling assists in medical planning by providing blood flow patterns, wall pressure and wall shear stress. This helps to understand various phenomena in the development of dissecting aneurysm.

1. Introduction

Cardiovascular diseases are the leading causes of death in many developed countries and are steadily increasing in the recent years. One of the most catastrophic and non-traumatic cardiovascular diseases is acute dissecting aneurysm of the aorta, which has a mortality rate up to 90% if left untreated (Beckwith et al., 1959; Wheat and Palmer, 1968; Wilbers et al., 1990). Generally, dissecting aneurysm is not an uncommon occurrence (Wheat, 1973) and the most common type is Stanford type B with aneurysm occurring in the descending thoracic aorta (Wheat and Palmer, 1968).

Dissecting aneurysm consists of two stages: aortic dissection followed by false lumenal aneurysm formation. The initial event is a tear of intima. Blood tracks through the tear and separate the intima from underlying media. Thus the initial single lumen of aorta becomes divided into a true lumen and a false lumen (lumen between separated intimal flap and media). Pressure built-up in the false lumen may lead to widening of the false lumen and subsequently aneurysm formation.

Besides family history and genetic factors, there are several mechanical factors that can affect the onset, generation and progression of dissecting aneurysm, including hemodynamics (Herman et al., 1987; Blanco et al., 2006; Feijoo, 2009), vascular geometry (Feijoo, 2009), mechanical properties and composition of the aorta wall (Feijoo, 2009). However, hemodynamics is believed to play a key role in the formation and the progression of the dissecting aneurysm (Herman et al., 1987). Therefore, this study emphasizes particularly on evaluation of hemodynamic parameters such as flow pattern, pressure and wall shear stress

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of a patient-specific pre-aneurismal and post-aneurismal aortas (before and after formation of false lumenal aneurysm).

Computational fluid dynamics (CFD) and fluid-structure interaction (FSI) have been used for simulating the complex cardiovascular system. From the literature, CFD simulations of blood flow in simplified or patient-specific aortic models had been reported (Shahcheraghi et al., 2002; Lam et al., 2008; Yoshiyuki et al., 2008). Although there were some FSI simulations on aneurysm (Borghi et al., 2008; Khanafer and Berguer, 2009), these aortic models were relatively simple and may not be able to represent complex conditions. In our case, FSI was not applicable because the intimal flap in the dissection site was too thin. To our knowledge, there is only one recent simulation work on a patient-specific dissecting aneurismal aorta using CFD (Cheng et al., 2010).

In our study, CFD was adopted for the simulation of hemodynamics in a whole patient-specific dissecting aneurismal aorta because of its capability in obtaining quantitatively the three-dimensional (3D) velocity field, pressure and wall shear stress (Papathanasopoulou et al., 2003; Marshall et al., 2004), which provide important information on hemodynamic in the development of dissecting aneurysm. In addition, aortic branches, which were reported to have significant effect on the flow field (Shahcheraghi et al., 2002), were included in our patient-specific dissecting aneurismal aorta models.

2. Method and material

2.1. 3D reconstruction from computed tomography (CT) imaging

The representative case of a dissecting aneurysm was obtained from the medical database—a 38-year-old female patient, who had been diagnosed as suffering from a Stanford type B aortic dissection followed by lumenal aneurysm formation. Aortic valve replacement together with replacement of the proximal ascending aorta (Dacron graft) was performed immediately after the first CT scan to correct the leaking valve. Geometrical information of a patient-specific aorta, before and after the formation of lumenal aneurysm in the thoracic aorta, was extracted as axial images (in-plane resolution of 512 by 512 pixels with a pixel size of 0.68 mm and slice thickness of 0.6 mm) from two separate sets of the contrast enhanced CT scan data. These CT images were imported into Mimics v13.0 (Materialise, Leuven, Belgium)—a medical image processing software, to reconstruct 3D models of the pre-aneurismal and post-aneurismal patient-specific aorta before and after the lumenal aneurismal formation.

2.2. Meshing and elements

A semi-automatic adaptive meshing technique was employed in HyperMesh v10.0 (Altair HyperWorks, Troy, MI, USA) to optimize between computational efficiency and element quality. 4-noded tetrahedral elements were assigned to both models, with 114,518 elements for the pre-aneurismal model and 84,209 elements for the post-aneurismal model (Fig. 1).

2.3. Boundary conditions and flow models

For the CFD simulation, a commercial finite element package—ADINA v8.6.0 (ADINA R & D, Inc., Watertown, MA, USA) was chosen. Transient analysis was adopted to investigate the pulsatile nature of blood flow. Time-dependent pulsatile waveforms of velocity and pressure at the ascending aorta inlet located just above the coronary artery and at the common iliac artery outlets, respectively, were obtained from Olufsen et al.'s (2000) work and used as boundary conditions in both models (Fig. 2A and C). These boundary conditions had been verified by various in vivo experimental data (Fung, 1997; Pedley, 2003) and were used extensively for both healthy and unhealthy aortas in various numerical simulation works (Li and Kleinstreuer, 2006, 2007; Cheng et al., 2008; Lam et al., 2008; Wolters et al., 2010). Based on the literature, each of the three supra-aortic branches, namely the brachiocephalic artery, left common carotid artery and left subclavian artery, was prescribed with 5% of the flow volume (Shahcheraghi et al., 2002) while each renal artery was prescribed with 10% of the thoracic flow (Shipkowitz et al., 2000) (Fig. 2D).

According to Fung (1997), shear rates in large arteries are typically large enough to assume that the blood flowing through it behaves as a Newtonian fluid. For the purposes of this study, blood flowing through the aorta was modeled as a homogeneous, incompressible Newtonian fluid (Pedley, 1980; Fung, 1997), with the dynamic viscosity of blood being taken to be 0.00371 Pa s and the mass

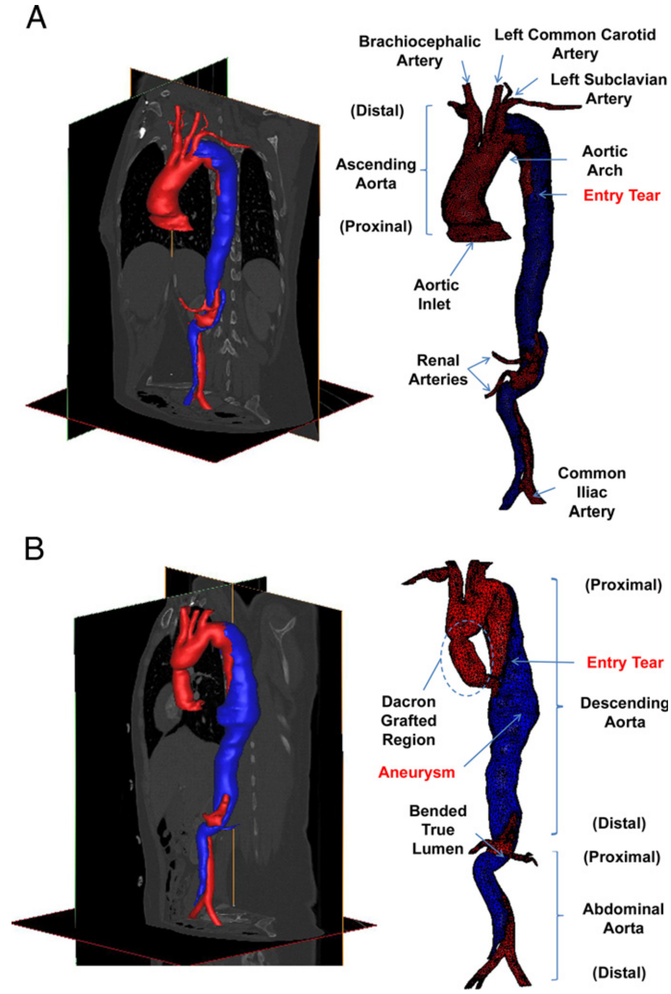

Fig. 1. Patient-specific pre-aneurismal (A) and post-aneurismal (B). On the left, aorta models generated from CT data by Mimics. The red colored portion in the model denotes the true lumen while the blue colored portion denotes the false lumen whereby blood from true lumen flows in via the intimal tear (aortic dissection). The right side shows meshed models for the pre-aneurismal aorta (A) and the post-aneurismal aorta (B), show the anatomy of the aorta as well as the sites of dissection and aneurysm of the false lumen around the true lumen. It shall be noted that the dilation of proximal ascending aorta was caused by a Dacron graft placement after the first set of CT scan had been taken. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

density of 1060 kg m^{-3} (Paul et al., 2009). Aortic wall was assumed to be rigid and therefore no-slip condition was applied at the aortic wall.

The blood flow is usually assumed to be laminar in large vessels since the mean flow velocity is predicted low enough to result in relatively low Reynolds number (Fung, 1997; Morris et al., 2004; Cheng et al., 2008). In our study, the average Reynolds numbers (Re_{ave}) based on the average flow velocity (V_{ave}) and average hydraulic diameter ($D_{h,ave}$) at peak systole are 1804 for the pre-aneurismal model and 1410 for post-aneurismal model. The maximum Reynolds numbers (Re_{max}) in our models are 2866 and 2643, respectively. The Womersley numbers (α) based on the $D_{h,ave}$ are 15.08 and 14.27, respectively. For pulsatile unsteady flow, turbulence occurs at a Reynolds number much larger than expected for steady flow due to the fact of a more stable accelerating flow and a more unstable decelerating flow (Fung, 1997; Hart, 1997). From the experimental study on canine aortas by Nerem et al. (1972), critical Reynolds number (Re_c) for unsteady flow takes the form of $Re_c = k\alpha$, with k being the constant of proportionality ranging from 250 to 1000. For our models, Re_c ranges from 3600 to 15,100. Since the Re_{max} for our models are lower than this threshold range of Re_c, the flow in our model was assumed to be laminar (Morris et al., 2005).

According to Shahcheraghi et al. (2002) and Kim et al.'s (2004) works, a steady state solution at the maximum flow rate is obtained first before it is used as the input for the CFD solver as the initial condition for the transient equations. Several cardiac cycles are required for obtaining a time-periodic solution (Shahcheraghi et al., 2002; Kim et al., 2004). In this study, five cardiac cycles of these pulsatile