The Aorta-Heart System Finite Element Modelling with Fluid-Structure Interaction Methods and Validation against Blood Hydrodynamics

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I. INTRODUCTION

The aorta is a vulnerable organ in motor vehicle crash accidents (MVCAs), in which blunt aortic rupture (BAR) is reported as the second common cause of death [1]. Although BAR is common and usually accompanied with serious consequences, the injury mechanism is still uncertain. Several possible mechanisms have been proposed to explain the cause of aortic injury: sudden stretching of the aorta with rupture at isthmus [2]; entrapment of the aorta by the surrounding bony structures [3]; sudden stroke of blood pressure with what is referred to in literature as the water hammer effect [4] and the combination of different mechanisms including these above [5]. To some extent, these mechanisms could also be considered as a combination of two sources of contribution: the inertia of surrounding organs and the hydrodynamics of blood.

Several finite element (FE) models were previously developed to investigate the aortic pathological or traumatic mechanisms. However, most of them could only illustrate either the aortic kinematics [6] or the blood fluid dynamics [7, 8] effects on aortic injuries. Therefore, our research was aimed to simultaneously study cardiovascular kinematics and blood fluid dynamics in MVCAs. The work presented herein is the first step to allow such research by developing a FE model of the aorta-heart system with fluid-structure interaction (FSI) methods to study the blood hydrodynamics, which will then be integrated with a human thoracic model.

II. METHODS

The geometry of the aorta-heart system was acquired from an internet database [9] and was validated by cardiologists for its normal anatomy structures. The geometry was then scaled to make the aortic diameters of different sections (Fig. 1) and the before-systole left ventricle (LV) volume consistent with literature [9]. The FE model (Fig. 1) was meshed and preprocessed with Hypermesh 12.0 (Altair Engineering Inc, MI, USA) and simulated with LS-DYNA 971 R5.1 (LSTC, Livermore, CA, USA). The aorta and heart was assigned with an orthotropic elastic material and elastic material respectively as was done by [6], while the blood was defined as a Newtonian fluid with the same properties as reported in [7]. Three pairs of coupling interfaces were conducted respectively between the fluid and the following three components: aorta, LV and a simplified aortic valve. A standard physiological time-independent pressure (Fig. 2) [8] was correspondingly applied to outlets 1 and 2 (cf. Fig. 1) to simulate the outflow pressure in superior arteries and the distal descending aorta respectively. The blood flow was produced by simulating the contraction and relaxation of the LV, which was loaded with a uniformly distributed force. It should be noted that this model has been idealised to some extent due to some geometry detail elimination for model simplification, the linear fluid property assumption and the boundary constraints for superior arteries and distal aorta.

This model was validated against data in literature on the following responses: the LV volume (Fig. 3), the blood pressure in LV (Fig. 4), the blood flow velocity and rate through different aortic sections as well as aortic section-averaged wall shear stress (WSS) (Fig. 5). Hereinto, the WSS was computed with equation (1):

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where μ is the blood dynamic viscosity, u the blood flow velocity along the lumen direction and x the distance from the wall along its inward normal direction. The flow on the vessel wall was assumed to be zero and the blood axial flow velocity close enough to the vessel was obtained by cubic interpolation based on the fluid velocity nearby during WSS computation.

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III. INITIAL FINDINGS

The simulated before and after-systole LV volume was 123.2 ml and 62.7 ml. As a result, the stroke volume and ejection fraction was 60.5 ml and 49.1%. The simulated LV ejection and systole phase persisted for 290 ms and 350 ms respectively. The peaks of blood mean velocity, flow rate and WSS for different aortic sections were within the range of 0.42 to 0.79 m/s, 178.1 to 380.7 ml/s and 0.47 to 0.89 Pa respectively.

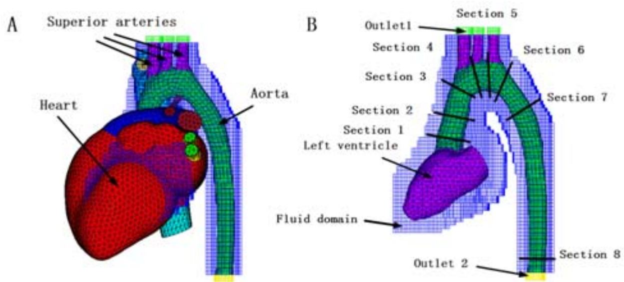

Fig. 1. FE model depicting different components and the location of each aortic section

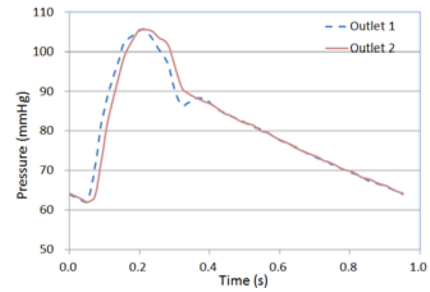

Fig. 2. Boundary conditions of time-dependent pressure for outlet 1 and 2

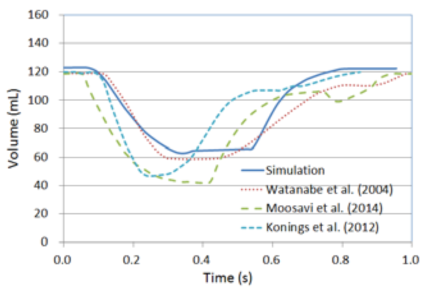

Fig.3. LV volume time-history

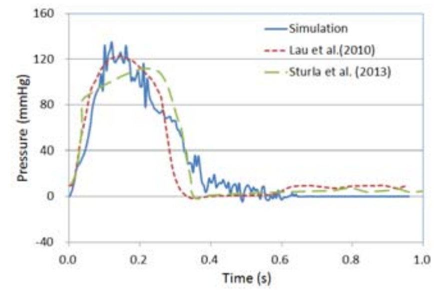

Fig.4. Left ventricular pressure time-history

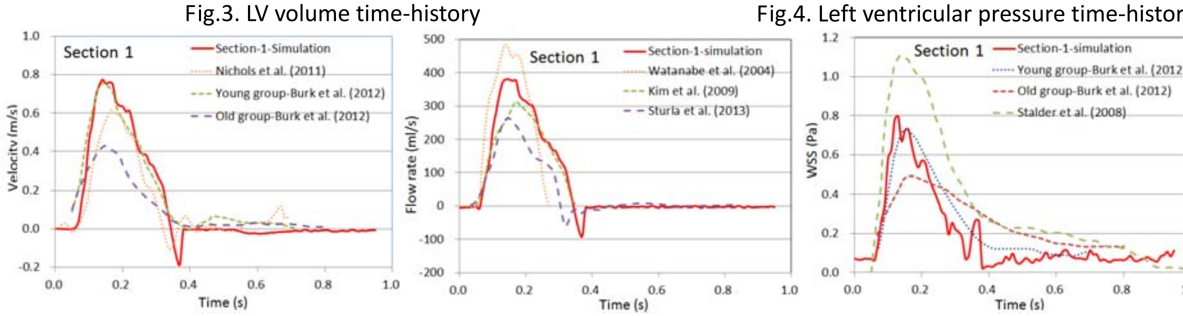

Fig. 5. Example of blood mean velocity, flow rate and WSS on section 1 during the cardiac cycle

IV. DISCUSSION

This research was the first successful attempt to reproduce the blood flow from the beating heart to the descending aorta. The simulation results of LV volume, blood pressure and blood hydrodynamics were generally consistent with the data in literature. The validation against the LV volume change and the blood pressure ensured the ability of this model to offer enough blood flowing through the aorta within an appropriate period. The consistency of blood flow velocity and rates confirmed the feasibility of studying the fluid dynamics in different aortic sections with FSI methods. The comparison of WSS indicated the possibility to study the effects of WSS on aortic pathologies and injuries. The integration of this aorta-heart FE model with the human thoracic model is already in progress, based on which further research can be expected to be conducted on the effects of aortic kinematics and blood hydromechanics on aortic injuries in MVCAs.

V. REFERENCES

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