FLOW DYNAMICS IN THE AORTIC ARCH AFTER A VALVE REPLACEMENT BY A BILEAFLET MECHANICAL VALVE

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INTRODUCTION

The aorta is the main blood vessel transporting blood from the left heart to the systemic circulation. It consists in a complex three-dimensional geometry including curves, branches and tapering. The investigation of blood flow in the human aorta is important because of its potential application to the diagnosis and prevention of disease of the aorta. Many researchers have attempted to reveal the relationship between the flow in the aortic-arch and the apparition of disease such as: aortic aneurysms or aortic dissection. Many studies have been performed with a native valve in aortic position (Yearwood and Chandran [1], Shahcheraghi [2], Kim [3], Mori and Yamaguchi [4], Morris [5], Leuprecht [6]). However, fewer studies are dedicated to the flow in the aorta in the presence of an aortic stenosis or after a valve replacement. Aortic stenosis is usually due to a calcification of aortic valve leaflets leading to an increase in left ventricle after load. If the aortic stenosis is considered as severe, valve replacement by a biological of a mechanical heart valve is the only way to improve patient's life style. However, it is very common that the implanted valve size is not adapted to the body surface area of the patient: this is referred as "patient-prosthesis mismatch" [7].

The objective of this study is to investigate using numerical simulations the impact of valve replacement, in the context of a patient-prosthesis mismatch, on secondary flows and the wall shear stress magnitude in a three-dimensional realistic model of the aorta. As our preliminary results showed that the presence of valve leaflets do not affect significantly the flow pattern in the aorta, we elected to simulate patient-prosthesis mismatch with a simple circular orifice (mimicking the geometrical orifice area of the valve). Under these conditions, there is analogy between non adequate heart valve prosthesis and a residual aortic stenosis.

NUMERICAL METHODS

The typical Reynolds number range for blood flow in the human body varies from 1 in small arteries to approximately 4000 in the largest artery, the aorta [8]. In the absence of a stenosis, the blood flow is usually laminar and does not experience transition to turbulence until the Reynolds number based on the diameter of the aorta and the average flow speed exceeds about 2300. However, the obstruction resulting from a severe stenosis or patient-prosthesis mismatch can lead to disturbed flow region in the ascending aorta. This disturbed flow depends on the conditions and the geometry of the stenosis or the valve. Ghalichi [9] presented numerical results for transitional and turbulent flow through moderate and several arterial stenoses by applying k-ω turbulence model and concluded that this turbulence model is suitable for blood flow studies where both laminar/transitional and turbulent flow regimes coexist. In this study, patient-prosthesis mismatch induced flow disturbance was simulated using k-ω turbulence model by using the commercial CFD software FLUENT 6.3. The results were compared to the flow downstream of a normal aortic valve.

In this study, three-dimensional unsteady blood flow in a realistic model of the human aortic arch (with its three major branches and an out-of-plane descending aorta) was simulated numerically. Two models for both normal and severe stenosis or patient-prosthesis mismatch inlets were investigated.

The unsteady simulations were performed during the systolic phase (with a complete valve opening).

BOUNDARY CONDITION & MODEL PROPERTIES

During the simulations, a steady state solution with the maximum flow rate was obtained first and it was used as an initial condition for the unsteady simulations. The blood was assumed to be incompressible and a Newtonian fluid, which is valid in large arteries [8]. The vessel wall was considered to be rigid and also a no-slip boundary condition was

applied on the wall. As an inlet condition for the normal valve, a uniform velocity corresponding to a pulsatile experimental flow rate (Figure1) was applied. While, as an inlet condition for the mechanical heart valve, the velocity profiles based on experimental data [10] was applied, for the same flow rate. The valve was in fully open position and the cardiac output was 5 l/min with a systolic duration of 300 ms. In the human aorta, approximately five percent of the flow volume is diverted to each of the three aortic branches, Middleman [11]. We imposed this volume flow rate as a boundary condition at the outlet of each of the three aortic arch branches with the use of a zero axial pressure gradient condition at the outlets. We assumed that these flow rate conditions remained constant during the unsteady simulations. As an outlet boundary condition for the descending aorta, we imposed: mass continuity and a zero total pressure condition.

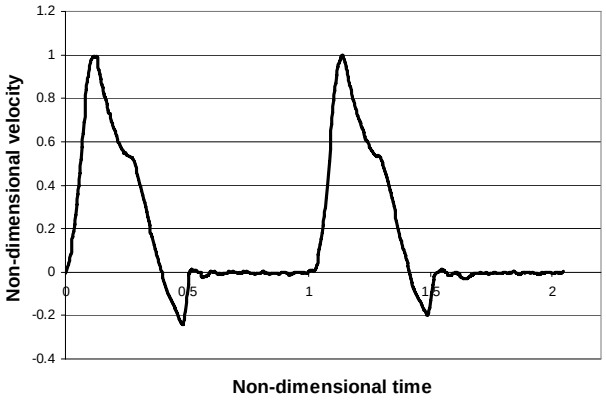


Figure 1: Pulsatile velocity profile according to the experimental data.

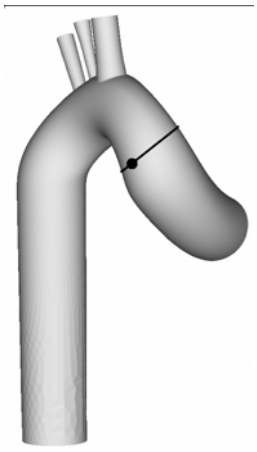


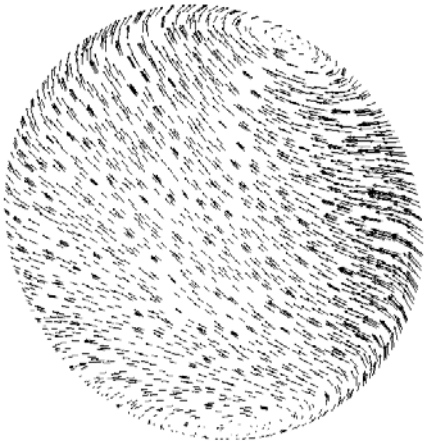
Figure 2: Locations of the cross section used in fig. (3) and fig. (5).

RESULTS

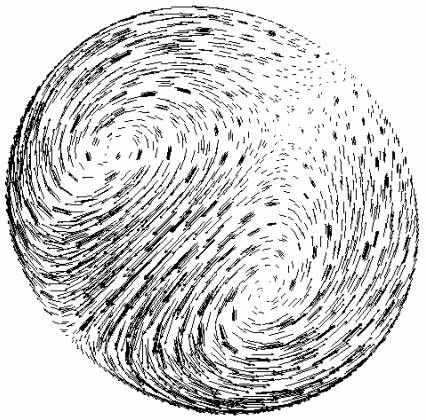
Secondary Velocity

Native valve replacement by a valve which is not adapted to patient's body surface area induces significant modifications in the secondary flow in the aorta.

Figure (3) shows the secondary flow developed in the aorta at the cross section shown in figure (2) for a normal case and in the case of a patient-prosthesis mismatch (or a severe aortic stenosis). It appears that two strong counter rotating vortices are developed due to elevated velocity and centrifugal force. This kind of disturbance prevents flow from going towards its natural direction [12].



[a]



[b]

Figure 3: Secondary flow at the cut cross section (Figure2) for normal case (a) and in the case of patient- prosthesis mismatch (b).

Wall Shear Stress

The wall shear stress is one of the most important parameters from fluid mechanics point of view which is involved in the development of aortic dissection or endothelial cell damage. The wall shear stress distribution is characterized by significantly higher values in the case of patient-prosthesis mismatch in comparison with the healthy configuration (Figures 4) in the ascending aorta. This figure shows the ratio between the wall shear stress induced with a patient-prosthesis mismatch and the normal valve. The maximum ratio occurs in the outer wall of the ascending aorta wall particularly in the vicinity of the arch. Interestingly, this region is very sensitive to aortic dissection.

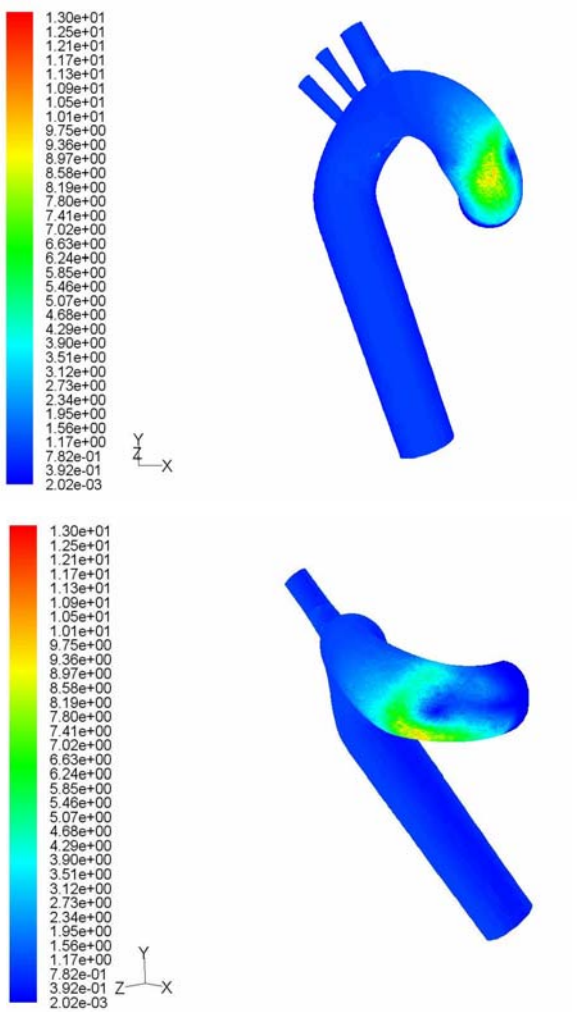


Figure 4: 3-D distribution of the wall shear stress ratio between patient-prosthesis mismatch and a normal case.

Figure (5) shows the time history of the wall shear stress on a point on the cross section indicated in figure (2) for both healthy and patient-prosthesis mismatch cases, during the systolic phase. The instant of maximum and minimum shear stress as well as the overall wall shear distribution patterns are completely different in the case of a patient-prosthesis mismatch. Indeed, in the case of a patient-prosthesis mismatch, the wall shear stress exceeds the cut-off for endothelial cell damage (50 Pa).

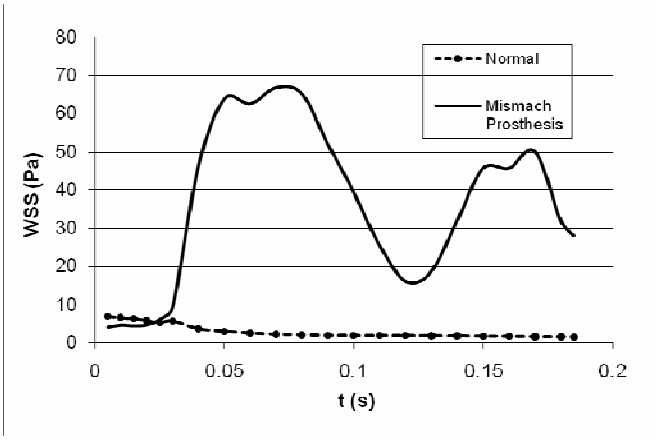


Figure 5: Wall shear history for the point shown in figure (2) during the systolic phase for both normal and patient-prosthesis mismatch.

CONCLUSIONS

Significant differences in wall shear stress distribution, secondary flows and recirculation in ascending of the aorta were observed between a normal case and the patient-prosthesis mismatch case. These differences might lead to an important damage to the aorta.

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