CFD SIMULATION OF AN UNSTEADY NON-NEWTONIAN BLOOD FLOW AND ITS EFFECTS ON INTEGRITY OF STENT-GRAFTS FOR ABDOMINAL AORTIC ANEURYSM REPAIR

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Abstract: The purpose of this paper is to present some of the characterisation results – obtained using computational based techniques - for the evaluation of hemodynamic forces acting on stent - graft devices, used in the treatment of Abdominal Aortic Aneurysm (AAA). The aims of our investigation are twofold: first, to simulate and analyse a pulsatile, non - Newtonian blood flow through various stent - graft models using the state-of-the art computational fluid dynamics tools; secondly, to export those simulation results into a Finite Element Analysis (FEA) software for assessing the impact of the hemodynamic effects upon the physical integrity of the device. The pulsatile velocity profile related to the blood flow was developed based on experimental data - relevant to the infrarenal abdominal aorta region – available in the literature. The FEA analysis will allow us to achieve an objective design critique of current stent – graft designs, as well as offering the chance of further design improvements (including the possible use of micro-sensors attached to the device, for a real time monitoring of the physical integrity of the stent).

Introduction

Abdominal aortic aneurysm (AAA) is a common vascular disease of many people, typically over 65 years of age [1]. Its characteristic feature is a typical saccular enlargement of the artery in the abdominal region. Presence of the aneurysm in a major blood vessel can cause number of serious health risks. Due to altered blood flow higher pressures on the vessel wall could result in a rupture of the aneurysm and usually a fatal internal bleeding. In other cases thrombosis could develop in regions of low shear rates or the aneurysm itself could increase pressure on other internal organs as it grows in size. Either case shows that an AAA represents serious health risk and has to be treated surgically.

The concept of repairing an abdominal aortic aneurysm (AAA) through remote deployment of an endovascular graft, rather than using an open surgical repair, was developed in 1991 and since then the advantages of this less traumatic procedure have led to an increasing demand to expand the availability of this technique. However, as with any relatively new approach there are still many questions about the design of the graft device. Recently, number of studies has been completed analysing the long-term results of the endovascular repair [2 - 7]. The results revealed that the late failures of the stent-grafts are most commonly due to migration, fracture or kinking. Purpose of this study is to investigate the causes of those late failures in relation to the blood flow dynamics and to suggest means for early detection and monitoring of the graft deformations.

Methods

In the first part of the project, a realistic model of in vivo blood flow conditions was developed for a computational fluid dynamics (CFD) analysis of the blood flow through the stent-grafts employed in the human abdominal aorta. CFD is a very powerful investigation tool either for analysing fluid transport in various systems of the human body, or for designing and optimising the implantable devices to be used in future medical interventions. In both cases, the backup of proper physiological and anatomical facts related to the human body system under investigation, is vital.

For the purpose of the CFD simulations the FLUENT software [8] has been used. The CFD model imitates the real conditions of blood flow in the infrarenal part of the aorta. The blood flow is pulsatile and therefore time dependent flow profile has to be used in the simulations. Even though most of the previous studies considered the blood as a Newtonian fluid, in case of this project the blood has been modelled as a non-Newtonian fluid with a viscosity depending only on a shear rate for this purpose. The blood flow was modelled through stent graft designs based on real devices. The stent device is a bifurcated modular system consisting of self-expanding stainless steel z - stents and woven polyester graft material. The main body of the graft comprises the aortic section with one long iliac (ipsi lateral) limb and one short iliac (contra lateral) limb, which provide the means to attach the extension legs. Both iliac legs are docked into the main body to form two adjacent channels

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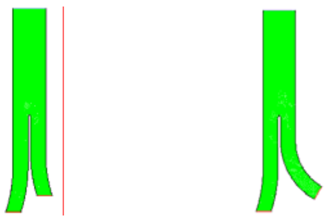
Number of 2- and 3-dimensional models of stent grafts were constructed based on the *Zenith Flex AAA endovascular graft* design from Cook [9]. The models represented the stent graft with following characteristics; the main body is TFFB-26-96; with 26mm main body diameter and 12 mm iliac limb diameter. The iliac legs extensions were chosen to be TFLE-12-37 for extension of the ipsilateral limb and TFLE-12-54 as an extension of the contralateral limb [9]. Models were designed in non - deformed as well as deformed modes. The models were constructed using the *Gambit* and *Pro-Engineer* software packages and meshed using *Gambit* software by Fluent Inc. The un-deformed model represented employed graft under normal circumstances, i.e. the iliac bifurcation angle was approximately 35 degrees according to the T. Shipkowitz et al. model [10]. For simplification the bending of the iliac extension legs is completely radial; see Figure 16 (a) for the model. This model when meshed using the QUAD/TRI paved scheme consisted of 15390 elements with maximum of 0.67 equi angle skew, which characterises a good quality mesh. The distorted model is intended to imitate a typical graft deformation that is likely to occur after prolonged usage. The basic model specification remained the same only the iliac bifurcation angle has increased to approximately 78 degrees due to bending of one of the iliac extension legs. As with the previous model the bend is completely radial for simplification; see Figure 16 (b) for the model. This model was meshed using the QUAD/TRI paved scheme and it contained 15782 elements with maximum equi angle skew of 0.65.



(a) (b)

Figure 1 2D meshed models of the endovascular graft used in the simulations (a) un-deformed (b) deformed model

The problem had to be treated as unsteady (time-dependent) simulation, because, as already explained above, the blood flow is unsteady due to the pumping action of the heart. Looking at this concept from practical point of view the pressure gradient varies depending on the time position in the cardiac cycle. Varying pressure gradient also implies varying flow velocity; most of the information about abdominal aortic blood flow was in terms of varying flow rate and therefore the transient flow velocity was chosen to be implemented as the inlet condition for the CFD model. The transient flow profile had to be modelled mathematically so it could be used in the FLUENT software. The largest problem arose while determining the outflow boundary conditions, the information about the in vivo conditions is already very limited and to implement different boundary conditions for the outflow is almost impossible. Faced with the problem of knowing only conditions at the inlet, the FLUENT support team had to be consulted for solution. As advised the pressure outlet boundary condition was used on both outlets discharging at 0 gauge pressure. The operating pressure was set to 13300 Pa, which corresponds approximately to the mean abdominal blood pressure of 100 mmHg.

The blood was treated as a non-Newtonian fluid with a viscosity following a power law (1) implemented in the *Fluent 6* software.

$$\eta_{\min} < \eta = k\dot{\gamma}^{n-1}e^{T_0/T} < \eta_{\max} \qquad (1)$$

The viscosity varies with the shear rate between maximum and minimum limit. The input parameters were obtained from [11] and are shown in Figure 2 as a window from FLUENT. The blood was as a fluid with constant density of 1060 kg/m^3. The simulation allowed only for laminar flow as the Reynolds number based on the peak flow velocity was far from the turbulent transition range. The selected time step size was 0.005 seconds with maximum of 20 iterations per step; the solution always converged to the specified convergence criteria of 0.001 in less than 20 iterations. Reporting interval for all the results was chosen to be 10 time steps, i.e. 0.05 seconds.

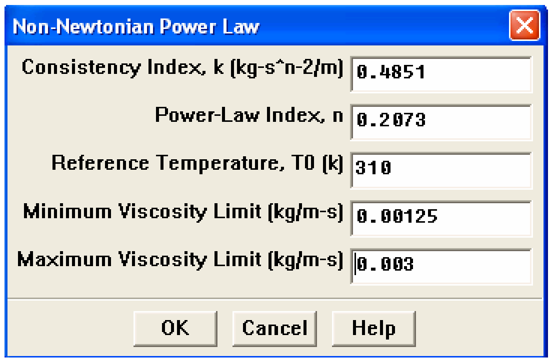


Figure 2 Non-Newtonian power law parameters

To minimise the influence of initial flow conditions, all simulations were carried out over 6 cardiac cycles. And results over the 5th cycle were used for the results analysis. For purpose of this study, non-turbulent pulsatile blood flow was assumed as the Reynolds number based on the peak flow velocity was far from the turbulence transition range.

The inlet velocity profile was developed to simulate the realistic infrarenal abdominal aortic flow profile obtained from MRI data under resting conditions [4]. The profile was developed with help of MATLAB and ORIGIN programmes by combining a decaying sinusoidal function with a Lorentz function to obtain the right shape with a right spacing in time.

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Blood Inlet velocity (m/s) =

$$0.031\times(0.7+2.9\times e^{((-t)\times t^{0.2}/0.39)}\times\sin(\pi\times t/0.25))\times(1-(0.5/\pi0\times(0.25/(4\times(t-0.7)^2+0.18^2)))$$

,where *t* is time (seconds).

The flow velocity profile is shown in Figure 3 as two complete pulses; the pulse period is 0.9 s.

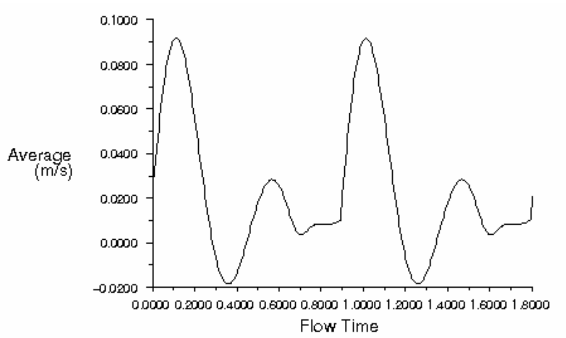

Figure 3 Simulated velocity profile over first two cardiac cycles (cardiac cycle period of 0.9s)

The transient blood flow velocity, in mathematical form, had to be implemented into the FLUENT software. This was done by using User Defined Functions (UDF), for which an additional function library (UDF is written in C language) is available; these functions are able to enhance many standard FLUENT simulation features, like boundary condition types in this case. UDF is the only tool that is able to implement such a transient behaviour as an inlet boundary condition.

Results

Results were recorded at equal time periods along the cardiac cycle. The reporting interval was set to 10 time steps, so every 0. 05 seconds the flow conditions were recorded and saved. The results were obtained in all cases from the third pulse that proved to ensure a fully developed pulsatile flow. Therefore along the pulse there were 18 equally spaced points in time at which the flow conditions were recorded. Figure 4 shows the equally spaced points along the flow velocity profile at which the relevant quantities were recorded (the figure shows a second pulse but the actual readings were taken from the third one).

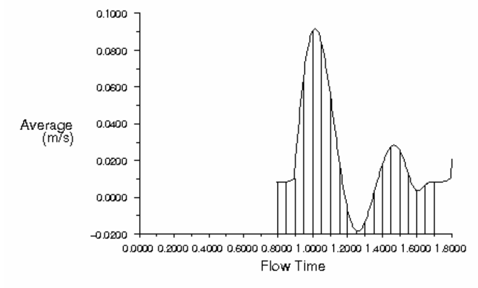

Figure 4 Monitoring intervals along the pulse

Figure 5 shows the velocity profile across a plane that is located 60 mm down flow from the inlet for the un-deformed graft. This velocity profile is similar to other pulsatile flow simulations presented in research papers, thus the CFD model used could be justified. Due to the non-slip condition the velocity is always zero at the wall. Looking at (b), corresponding to the peak flow velocity, the profile is trying to reach the usual parabolic shape typical for pipe flow, but never reaches it completely. As the flow start to reverse (c) in a late systole, the velocity decreases producing two peaks in the profile. At the lowest velocity flow in the diastole (d) the flow direction is already reversed. Due to the initial momentum of the flow, the flow profile does not have enough time to fully develop into an opposite flow; i.e. the middle of the flow is very close to zero (stationary).

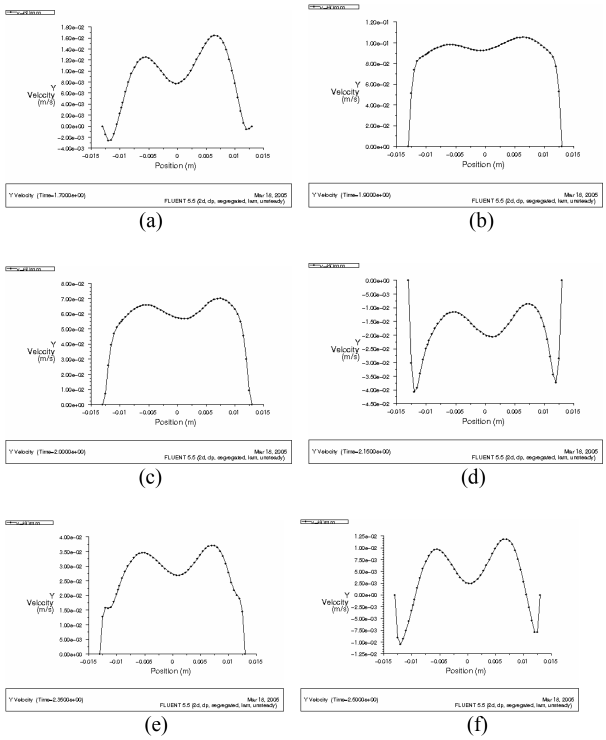

Figure 5 Flow velocities across a plane 60mm down from the inlet – undeformed (a) t/T = 0, (b) t/T=0.22, (c) t/T=0.33, (d) t/T=0.5, e) t/T=0.72, (f) t/T=0.88

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Contours of velocity magnitudes in the whole stent-graft model are shown in Figure 6 (velocity in m/s). The patterns in the un-deformed model are very similar to the deformed model. It could have been expected that this relatively small deformation would not cause any significant changes in the flow. For future simulations larger deformations would be preferable. Generally the flow in the longer (ipsi - lateral) extension leg was faster than in the other one, but in the trough of minimal flow velocity, the flow in the ipsi - lateral leg was almost stationary. In both models the permanent stagnation point was at the tip of the bifurcation, which could possibly lead to a development of a small-scale blood thrombosis.

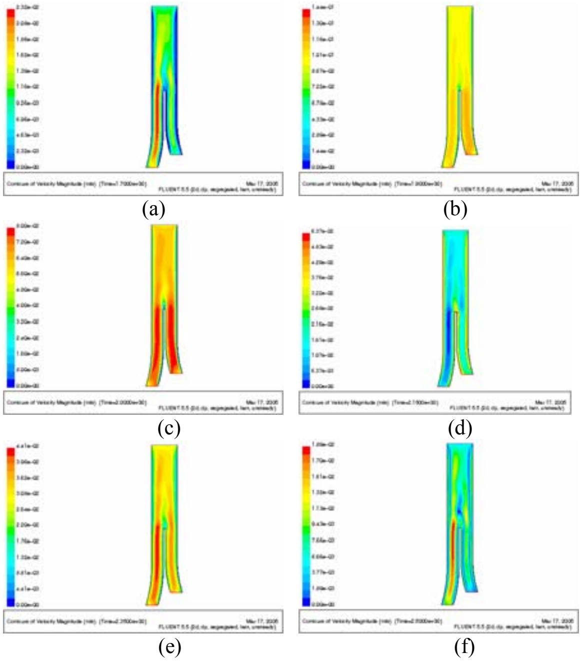
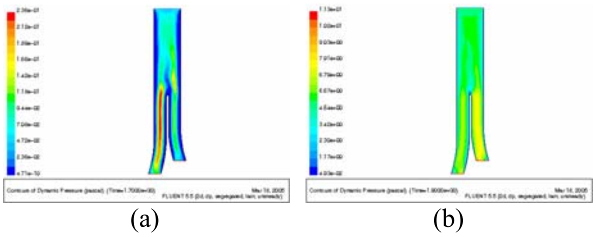


Figure 6 Velocity magnitudes at various times along the pulse (T=0.9 s) – undeformed (a) t/T = 0, (b) t/T=0.22, (c) t/T=0.33, (d) t/T=0.5, (e) t/T=0.72, (f) t/T=0.88

Figure 7 shows the time variation for the dynamic pressure in the device; the pressure reaches maximum (b) in both legs at the same level. The maximum is of the order of 3 Pa, in contrast to the minimum that occurs at (d) and approximately of two orders of magnitude smaller. The main dynamic pressure variations occur in the both iliac legs.



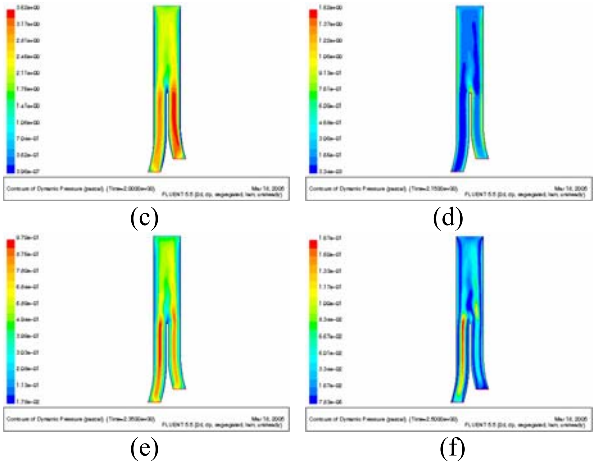


Figure 7 Dynamic pressure at various times along the pulse (T=0.9 s) – undeformed, (a) t/T = 0, (b) t/T=0.22, (c) t/T=0.33, (d) t/T=0.5, (e) t/T=0.72, (f) t/T=0.88

Obtained pressure and force distributions on the graft walls under various flow conditions are going to be extracted to FEA software *Pro-Mechanica* in order analyse and quantify stresses and strains within the device due to the pulsatile blood flow effects.

Having information about the stress distributions within the structure, the present graft design could be analysed and potential improvements suggested. The main focus is on development of real-time in vivo monitoring and sensing techniques that could provide an immediate warning of any changes in the graft integrity.

Conclusion

In the present work, we developed a working pulsatile non-Newtonian blood flow model – used in the CFD analysis - that simulates conditions in the infrarenal region of abdominal aorta. The modelled pulse profile is based on an MRI data from number of patients. The CFD simulations were done on realistic stent-graft models with varying degree of deformation. Purpose of the simulations is to use the obtained results in a FEA analysis that will help us to investigate the stress/strain behaviour of the stent-graft model due to the hemodynamics effects. This study will allow the examination of long-term integrity changes of the graft, which are crucial for further design improvements of the grafts.

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