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BLOOD FLOW PATTERN IN STENOSED ARTERY WITH COMPLETE BYPASS GRAFT: A CFD APPROACH

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Abstract

In the present study a computational model for the blood flow in stenosed artery with complete bypass graft is proposed. The laminar, incompressible, fully developed, non-Newtonian flow of blood in a stenotic artery having bypass graft is numerically studied. The fluid dynamics simulations are performed by using a control-volume based technique, implanted in the Computational Fluid Dynamics (CFD) code Fluent. The influence of stenosis severity and bypass angle on the various flow features in stenosed arteries with complete bypass graft is discussed with the help of graph. The important flow features, which are correlated with the intimal hyperplasia in arteries, are found to be influenced by the stenosis severity and angle of bypass significantly.

1. Introduction

Cardiovascular disease is one of the major causes of death in all over the world. It is well known that atherosclerosis (narrowing of the artery) would cause partial or total blockage of artery and could result in the heart attack, stroke, or even death.

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Key Words : *Blood flow, Stenosis, Bypass Graft, non-Newtonian, Fluent.*

Atherosclerosis mostly occurs around curvatures, junctions and bifurcation of artery [4]. For the cases of serious blockage in the coronary artery, bypass surgery is the most reliable treatment to restore blood flow. This surgery is not without complications. Restenosis is a major problem occurred one or more years after the surgery leading to the failure of bypass surgery. Initial hyperplasia has been found as a major cause for the developing the restenosis and subsequent failure of bypass surgery. Steele et al. [28] found that the ability to predict changes in blood flow would enable a surgeon to evaluate the efficiency of a treatment strategy after bypass surgery. Numerical models are commonly used in understanding the effects of blood flow and heat transfer in arterial system [14, 33]. Previously, it has been found that Intimal hyperplasia is most frequently seen in the distal part of the bypass anastomosis. Which is thought to be promoted by hemodynamic effects, specially 'abnormal' wall shear stress patterns [7, 11].

Researchers have shown that the local flow dynamics and wall mechanical conditions play a major role in the developments of intimal hyperplasia in bypass graft with subsequent graft failure. The development of intimal hyperplasia is related to different hemodynamic factors including low and high or unidirectional wall shear stress [12,16, 22].

Furthermore geometric factors such as bypass angle [10], curvature of the artery wall [32, 35], Flow rate ratio, and Waveform [7] have been proved to influence the development of stenosis. Previously some authors have investigated the effect of Newtonian and non-Newtonian representation of blood and the wall shear stress distribution in the bypass graft anstomosis [1, 2, 13, 19, 21].

Chua et al. [6] clearly indicated in his study that non-Newtonian effects on any vessel only leads to a maximum of 10% difference in the magnitude of wall shear stress. A lot of studies [1, 7, 12, 15,19, 23] have discussed the flow behaviour in the end to side model of bypass anastomosis. They adopted a simple model to simulate the blood flow in occluded host artery and only a part of bypass graft. Since only the subsection of bypass graft is included in computational domain, the inlet velocity distribution at bypass entrance is usually assumed to be uniform or parabolic. However, as the bypass graft in vivo is with curvature, the velocity of fluids when entering the host artery from bypass graft is accompanied with secondary vortices and therefore the flow fields become neither uniform nor parabolic. Since the inlet condition will influence the resultant flow features in arteries, the resolution and relevant inference from hemodynamic viewpoint to clinical applications based on the resultant flow fields will be affected.

The proximal region is also affected by the diversion of flow in the graft since the entrance and exit junctions of the bypass are the two critical locations, where the flow is complicated [15].

Most of the previous studies using end to side model assumed the host artery as fully occluded and simplified the flow field into flow in bypassed artery with one closed end-side.

Only very few studies [15, 23] considered the cases with partially occluded host artery. However, in the case of bypass surgery, the stenosis in patients arteries is not fully occluded. The residual blood is still allowed to flow through the blocked region of the host artery as well as flow through the bypass graft even after weeks of surgery. As pointed out in the researches of Lu et. al. [19] and Qiao et al. [23], the residual blood flow through the stenosis is not negligible at least until after 2 weeks after grafting, therefore the flow fields in a stenosed artery with 75%-area lumen axisymmetric stenosis was analyzed by them through numerical and experimental methods. Nonetheless, since the End-To-Side model was adopted, the flow rate ratio between the graft and host artery entrance was assumed and imposed as a parameter in their studies.

From the point of view of fluid dynamics, the flow field in stenosed artery with bypass graft can be treated as the flow issuing from the entrance of host artery bifurcates at the junction of host artery and bypass graft, and then develops toward downstream after merging at the anastomosis. Accordingly, the percentage of flow rate partitioning from host artery into bypass graft should be deterministic as soon as the flow configuration is specified and the complete bypass graft is included in the computational domain.

Some authors [3,6, 8, 9, 15, 25 , 26, 27,28] have investigated the complexity of blood flow in the complete model of arterial bypass. In these studies, the bifurcating flow rate into bypass graft is not a major concern since the fully occluded host artery is assumed in the study and 100% flow issuing from the host artery entrance bifurcates into the bypass graft.

As a matter of fact, the important flow features, including recirculation flow, secondary vortices and shear stress on arterial walls are very sensitive to the severity of stenosis in stenosed arteries. Since these flow features are closely related to intimal thickening which induces the failure of bypass graft surgery, the understanding of the influences of stenosis severity on the flow fields in arteries should be worthwhile for further attacking the exact factors to cause the restenosis in bypass graft.

The anastomotic angle, i.e. the angle between the host artery and implanted bypass graft, is a very important factor responsible for intimal hyperplasia. Various hemodynamic studies [1,8,9, 20, 34, 24) pointed out that the best anastomotic angle for an end-toside or side-to-end reconstruction is between 30^0 and 45^0.

Many authors [5, 29, 30) have studied the problem of blood flow through stenosed tube by considering blood as a Newtonian fluid.

D. Liepsch et al. [18] experimentally investigated the changes in fluid flow through elastic models of arterial bypass under pulsatile flow conditions. They observed that under normal circumstance the flow through the straight section of vessel is high than that of the bypass section, which is attributed to increase in flow resistance and placement of graft in fully occluded arterial segment with minimum angle, minimize the flow disturbances.

Similarly Wang et al. [35] have proved that curvature, in general, lowers the resistance of a tube, but increases the shear stress near the inside wall. Lee et al. 2005 found that substantial area reduction leads to flow recirculation in both upstream and downstream of the stenosis and in the host artery near the toe, while diminishes the recirculation zone in bypass graft near the bifurcation junction.

Recently Qiao et al. [23] proposed a novel geometric configuration with two symmetrically implanted grafts for the purpose of improving the hemodynamics, but the new technique needs a large number of animal experiments to verify its practical viability. Recently Sun et al. [31] suggests that by intentionally introducing swirl flows into the bypassed arteries, the adverse effects along the host artery floor can be suppressed.

However above studies are very important in order to predict the development of intimal hyperplasia but the problem of restenosis and Intimal hyperplasia is still not fully solved. A lot of investigations are still going on to reduce the problem of early graft failure. It has been found that computational fluid dynamics plays a major role in the study of blood flow complexities in different arterial models.

The present paper intends to carry out numerical analysis on the flow fields in a partially stenosed artery with a complete bypass graft implanted on it. The detailed comparison between the cases with different severity stenosis, including 60%, 75% and 100% area occluded and different bypass angles, including 30, 45 and 60 degree cases are considered. The aim of this study is to discuss the influence of stenosis severity and bypass angle on the flow features in stenosed arteries with complete bypass graft.

2. Problem Description

The diameters of the host artery and bypass graft are assumed as the same equals to 1.6 mm for coronary artery. The Reynolds number based on the host artery diameter and the mean velocity at the host artery entrance is 250.

The distance between the heel and toe is 2.24 mm. The axial distance from the narrowest location in the stenosed host artery to heel is 2.88 mm. The axial distance from entrance of host artery and the bifurcating location is 19.2 mm. The host artery extended 40 mm. distally from toe, which is specially set as long enough for setting suitable boundary condition at host artery exit. The location of X=0 is set at the location where the height of stenosis is maximum in the host artery. The arterial wall is assumed to be rigid and solid.

3. Mathematical Model

3.1 Governing Equation

The mathematical description for the current problem applies three-dimensional, steady, laminar, incompressible continuity and Navier-Stokes equations. These equations are expressed as follows,

$$\begin{aligned} \frac{\partial u_i}{\partial x_i} &= 0 \\ \frac{\partial}{\partial x_j}(\rho U_i U_j) &= \frac{\partial p}{\partial x_j} + \frac{\partial}{\partial x_j} + \frac{\partial}{\partial x_j}\lfloor \mu \left(\frac{\partial U_i}{\partial x_j} + \frac{\partial U_j}{\partial x_i} \right) \rfloor \end{aligned}$$

Blood is assumed to be non-Newtonian fluid.

3.2 Initial and Boundary Conditions

(i) At the entrance of host artery, uniform velocity distributions are located.

(ii) The fully developed conditions are applied at arterial outlet.

(iii) On arterial walls, no-slip conditions are applied.

3.3 Numerical Solution

(i) The fluid dynamics simulations are performed by using a control-volume based technique, implanted in the CFD code Fluent. The computation procedure of the commercial code consists of Construction of the geometry using a pre-processor, Gambit

(ii) Meshing the computation domain

(iii) Assigning boundary conditions in terms of velocities and pressure.

(iv) Assigning fluid properties and

(v) The solution algorithm.

The geometry is constructed in Gambit, using the dimensions provided in literature (6; 10; 18 and 36). The elements employed to mesh computational domain consisted primarily of regular hexahedral elements. In order to carry out the mesh sensitivity analysis, numerical simulations are carried out by varying the number of mesh elements in the computational domain. Two cases for meshed domain are shown in figure 1.
Several iterations of the solution have been performed for converging solution due to the non-linearity in the governing equations. The governing equations are solved iteratively until convergence of all flow variables is achieved. The solutions of all the flow variables are deemed to have converged once their residuals computed from two successive iterations are below the set desired convergence criteria of 10^{-5}.

4. Result and Discussion

The velocity distribution on the symmetric plane of the three cases with different stenosis severities, including 60%, 75% and 100% occluded cases, are shown in figure 2. It is found that the velocity in bypass graft is larger in the more severely occluded cases. Near the inlet of bypass graft, the velocity in the outer side is slower than that in the inner side. But the velocity distribution changes gradually as the fluid flows along the middle of bypass graft. Near outlet region of the bypass graft, the velocity in the outer becomes faster than that in the inner side. Such changes are found because of the centrifugal force caused by the curvature effect of the bypass graft. The blood is accelerated to be largest in the constrict region for all partially occluded cases in host artery.

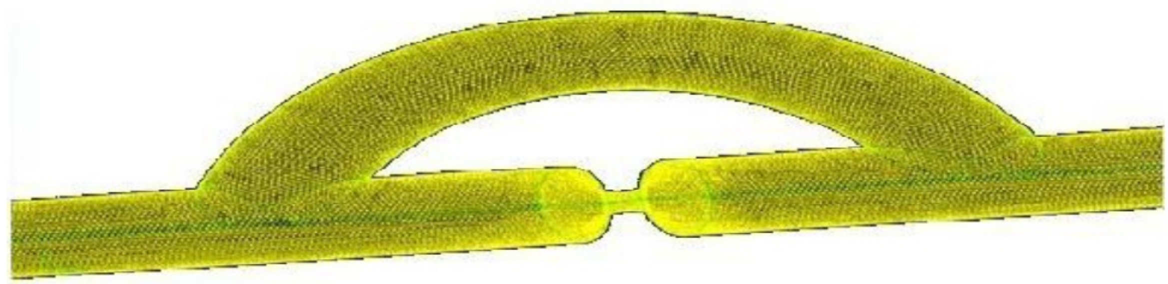


Figure 1 : Grid layout of computational domain

It is found that four low-velocity regions occur in the partially occluded cases. One of them is in the bypass graft and appears in the outer side near the bypass graft entrance. The other three appear in the host artery and distribute in the lower corner upstream of the constrict region and upper and lower corners downstream adjacent to the constrict region, respectively. The low velocity regions of blood in the bypass graft are found to be narrower in the more severely occluded cases. In the 60%-area occluded case, the low velocity zone in the bypass graft extends to the middle of the bypass inlet, but in the 100%-area occluded case, the low velocity region in the bypass graft becomes very minor. The distributing area of the low velocity region increases obviously with increase of the stenosis severity in the host artery.

The axial recirculation flow feature is believed to have close connection with the intimal hyperplasia. In the bypass graft, a large recirculation zone can be seen near the bypass entrance in 60%-occluded case. The recirculation zone reduces slowly as the stenosed degree increases. In the case with fully occluded case, the recirculation in the bypass graft becomes almost disappears. The recirculating velocity in the bypass graft is strongest in 60%-occluded case and also decreases as the stenosis degree increases. In host artery, recirculations can be found for all of the occluded cases. They exist in the three low-velocity regions as mentioned above.

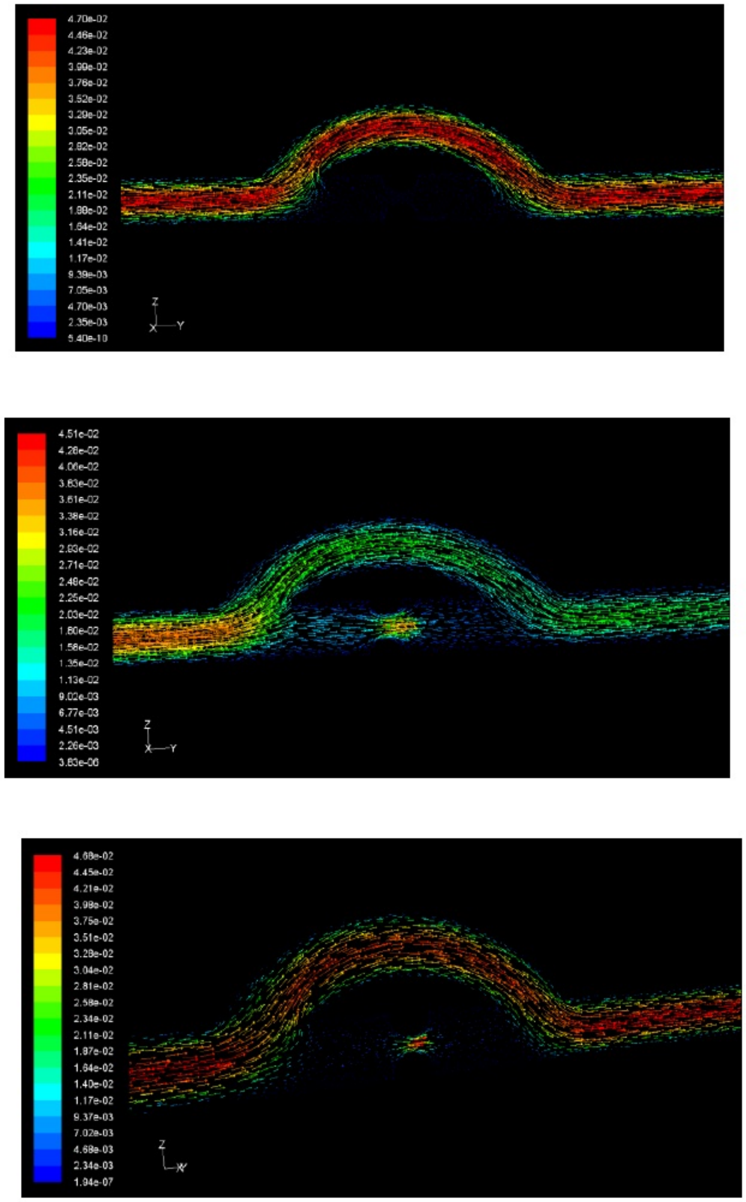
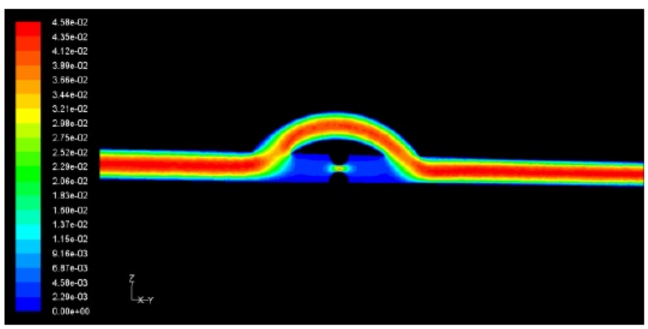
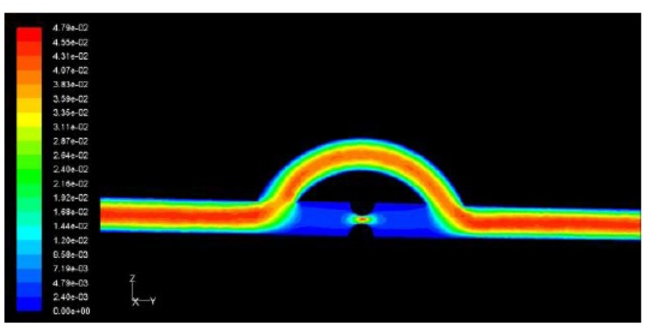


Figure 2 : Velocity Contours in Bypass models with 60, 75 and 100% blocked Artery

The recirculation zone occurring in the lower corner region upstream of the stenosis can be seen to become wider in the more severe stenosed cases. The recirculation zones downstream of the stenosis also exhibit different distributions in fully occluded and partially occluded cases. In the fully occluded case, the recirculation zones are separated into two distinct zones. One locates on the upper corner adjacent to the stenosis and the other is on the impinging bed with some distance to the stenosis. In the partially occluded cases, the recirculation zones downstream of the stenosis distribute on the inner wall and outer wall adjacent to the stenosis. The one distributing on the inner wall even extends to the position of heel, which is very different from the distributions in the fully occluded case. It is noted that there is a very minor recirculation in the region near toe in the fully occluded case, whereas the recirculation does not appear in all partially occluded cases. Through these comparisons, it can be seen that the distributing area of the recirculations in host artery increases as the stenosed severity increases, which shows the opposite trend for the recirculation in bypass graft.





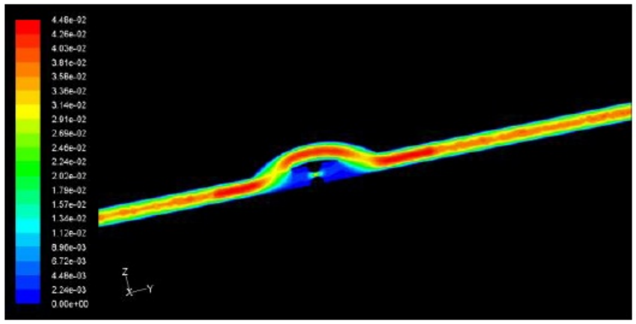


Figure 3 : Velocity contour in bypass models with the angles 30, 45 and 60 degree.

The velocity distribution on the symmetric planes of the three cases with $\alpha = 30^0, 45^0$ and 60^0 respectively, are shown in figure 3. In the host artery, it can be seen that the velocity magnitude generally increases aα increases, indicating the larger bypass bifurcating degree with larger α impedes the fluid to flow into the bypass graft and pushes fluids to flow toward the stenosed host artery. In all the three cases with different bypass angles, the flow is accelerated to be largest in the constrict region. Three low-velocity regions can be seen in the host artery. They distribute at the upper and lower corners near arterial walls upstream and downstream adjacent to the constrict region. In the bypass graft, the velocities in the smaller case are larger, revealing that the smaller anastomotic angle is beneficial for conducting the fluid flow through the bypass graft. In addition, some common flow features can be found in different angle cases. Near the bypass-graft entrance, the velocity is larger in the inner region, whereas a very low-velocity region appears in the outer region. Along the curved bypass graft, the curvature effect makes the larger-velocity fluid transfer toward the outer region. As the flow develops to the outlet of bypass graft, the velocity in the inner side becomes slower than that in the outer side, which is just opposite to the distribution at the bypass-graft inlet. These flow patterns are reliable with the previous discussion for stenosis severity, indicating the principal flow patterns in the bypass graft are not influenced by the variation of anastomotic angles. In the host artery, three recirculation structures can be seen in the three low-velocity zones which were mentioned previously. At the lower corner upstream of the constrict region, both of the strength and distribution area of the recirculation are larger in the smaller angle case, whereas the recirculation occurring at the upper and lower corners downstream of the constrict area can be found as stronger and wider in the larger angle cases. It is noted the recirculation at the upper corner downstream of the constrict region even extends almost to the heel where the flow structures are very influential to cause intimal hyperplasia. In the bypass graft, the recirculation also occurs in the low-velocity region locating adjacent to the outer wall near the bypass-graft entrance as mentioned previously.

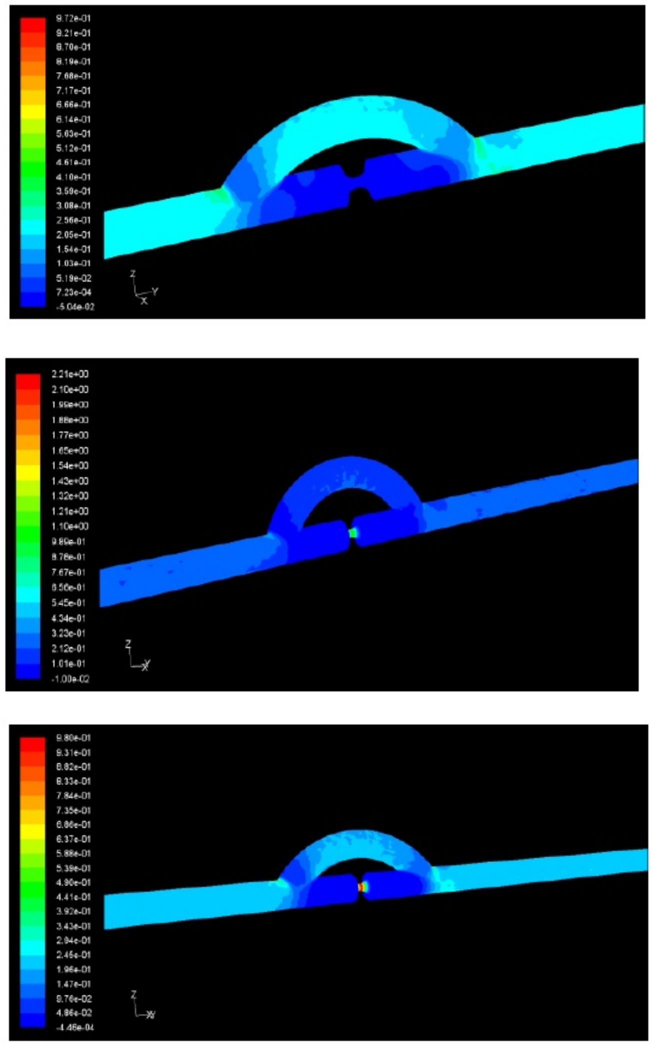


Figure 4 : WSS in Bypass models with 60 ,75 and 100% blocked Artery

The different angles do not change the distribution zone of recirculation in the bypass graft significantly, but it is found that the recirculation strength increases as bypass angle increases.

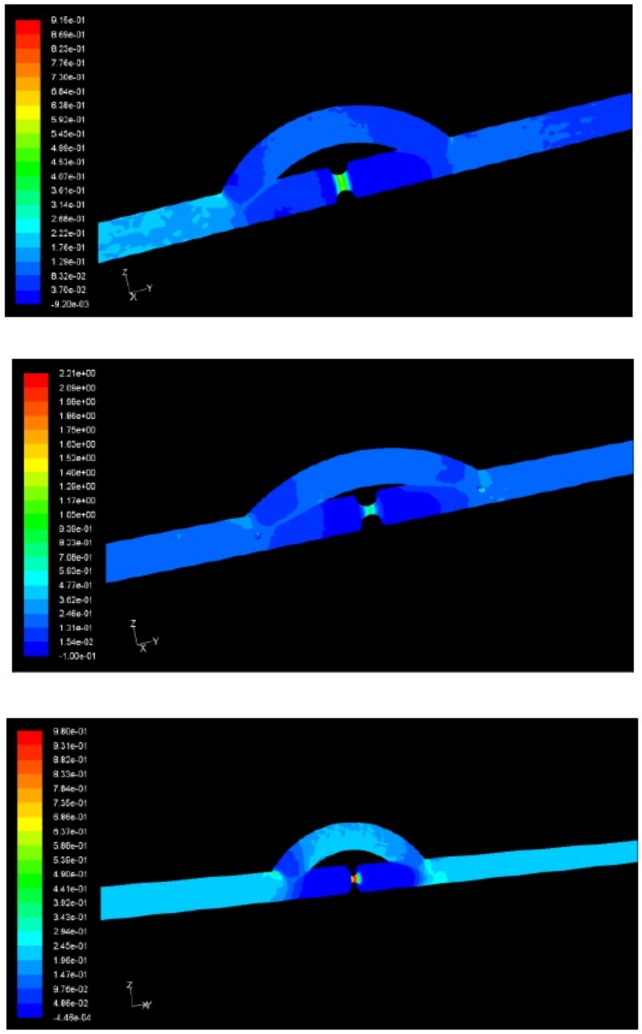


Figure 5 : WSS in Bypass models with the angles 30, 45 and 60 degree

Figure 4 and 6 shows the distributions of magnitudes of shear stress on arterial walls for the cases with different stenosis degrees. For all cases, the shear stresses in the junctions of the host artery and bypass graft are relatively larger and the magnitudes are affected by the stenosis. The highest shear stress for all of the partially occluded cases occurs in the constricted region of the host artery and its value can be seen to increase as the stenosis becomes more severe. It is noted that the shear stress near the occlusion is almost zero in the fully occluded case, which is very different from that in the partially occluded cases.

The wall shear stresses on the arterial wall for the three cases with different anastomotic angles are shown in figure 5. In the figures, it can be seen that the wall shear stress distributions for the three cases are very similar. The most evident influences of bypass angle on the shear stress distributions occur near the inlet of bypass, the narrowest region and the region between heel and toe. The largest shear stress concentrates in the constrict region, whilst near the junction between the bypass graft and host artery the shear stresses are also relatively high.

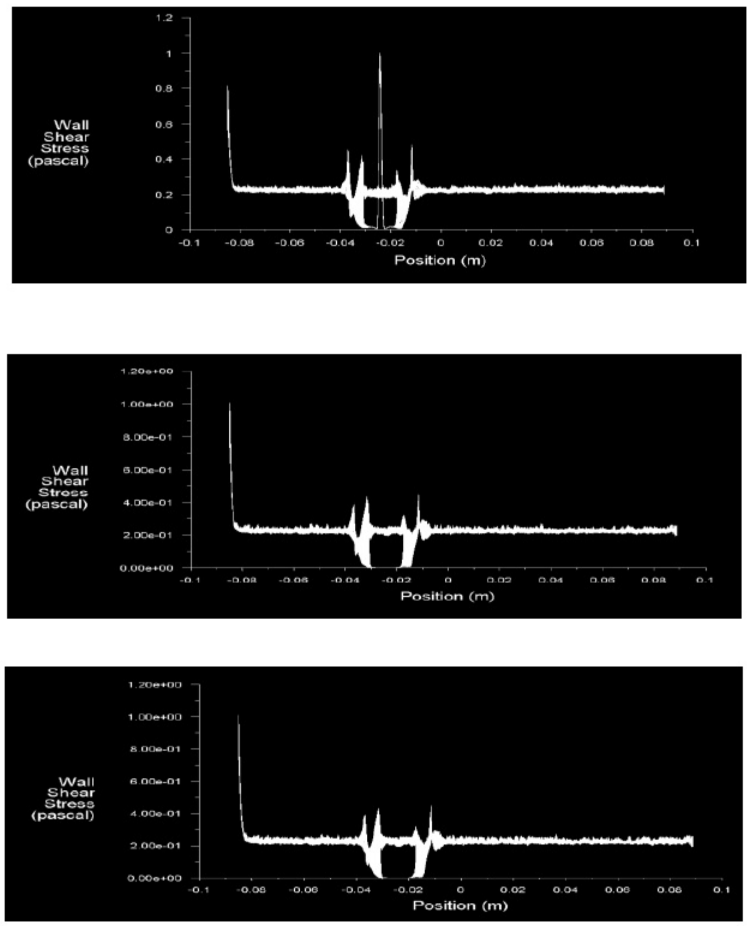


Figure 6 : WSS in Bypass models with 60 ,75 and 100% blocked Artery

5. Conclusion

The effects of stenosis degree on the flow fields in a partially occluded artery with complete bypass graft have been investigated in detail by numerical methods. The important flow features, which are suspicious to relate with the intimal hyperplasia in arteries, including the recirculation and the shear stress distributions in host artery as well as bypass graft, are verified to be influenced by the stenosis severity and angle of bypass significantly. It is concluded that the axial recirculation takes place in less severely occluded cases in bypass model. In all the three anastomotic-angle cases, there are four low-velocity regions in the artery. One of the regions is in the bypass-graft entrance and the other three are in the lower corner upstream of the constricted area and the upper and lower corners downstream of the constricted area. The bypass angle only slightly influences the distribution area of the recirculation. In the bypass-graft inlet region, the velocity is larger in the inner region and as the flow develops toward downstream, the higher velocity region transfers to the outer region. It is concluded that the velocity in the bypass graft becomes smaller as the anastomotic angle becomes larger. The influences of bypass angle on the shear stress distributions is found to occurs near the bypass-graft inlet, the narrowest region of the stenosed host artery and the region between heel and toe.

This study will provide a better understanding of the blood flow in the artery after bypass surgery. The information provided in the present study will be useful for further investigation of the exact factors to cause the restenosis in the artery with bypass graft. The data provided by this study would allow healthcare professionals to make more accurate prognoses for their patients after bypass surgery and reduce the problem of restenosis.

References

[1] Ballyk P. D., Steinman D.A. and Ethier C.R., Simulation of Non-Newtonian blood flow in an end-to-side anastomosis. Biorheology, 31(5) (1994), 565-586.

[2] Basavarajappa K. S., A study on Mathematical modeling on human blood in reference to the variations of Hematocrit. Journal of analysis and computation, 4(1) (2008), 1-12.

[3] Borhan Alhosseini Hamedani and Mahdi Navidbakhsh, Blood flow pattern in bending coronaries in coronary artery, Bypass Grafts, J. Forensic Biomed, 6(2) (2015), 1-6.

[4] Chakravarty S. and Mandal P. K., An analysis of pulsatile flow in a model aortic bifurcation. Int. J. Engng science 35(4) (1997), 409-422.

[5] Chaturani P. and Palanisamy V., Pulsatile flow of blood with periodic body acceleration. Int. J. Engng. Sci. 29(1) (1991), 113-121.

[6] Chua L. P., Zhang J. and Zhou T., Numerical study of a complete anastomosis model for the coronary artery bypass. Int. Com. in Heat and Mass Transfer, 32 (2005), 473-482.

[7] Ethier C. R., Steinman D.A., Zhang X., Karpik S. R. and Ojha M., Flow waveform effects on end-to-side anastomosis flow patterns, Journal of Biomechanics 31 (1998), 609-617.

[8] Ghista D. N., Sankaranarayanan M., Chua L. P. and Tan S. T., Computational model of blood flow in the aorto-coronary bypass graft, Biomedical-Engineering-Online, (2005), 4-14.

[9] Ghista D.N., Sankaranarayanan M., Chua L.P. and Tan S. T., Flow studies in three dimensional aroto-right coronary bypass graft system, Journal of Medical Eng. And Technol. 30(5) (2006), 259-282.

[10] Jackson Z. S., Ishibashi H., Gotleib.A. I. and Langille B. L., Effects of anastomotic angle on vascular tissue responces at end-to-side arterial grafts. J. Vasc. Surg. 34(2001), 300-307.

[11] Katiyar V. K., Agarwal R., Pradhan P. and Liepsch D., Finite Element Simulation Of Cardiac Output And Pressure Waveform in Carotid Artery, Canadian Journal of pure and applied Sciences, 2(1) (2008), 117-203.

[12] Keynton R. S., Evancho M. M., Sims R. L., Rodway N. V., Gobin A. and Rittgers S. E., Intimal hyperplasia and wall shear in arterial bypass graft distal anastomosis: an in vivo model study Journal of biomechanical engineering 123 (2001), 464-473.

[13] Ku J. P., Draney M. T., Arko F. R., Lee W. A., Chan F. P., Pelc N. J., Zarins C. K. and Taylor C. A.,In Vivo Validation of Numerical Prediction of Blood Flow in Arterial Bypass Grafts. Annals of Biomedical engineering 30 (2002), 743-752.

[14] Kumar S., Mathematical modeling of freezing and thawing process in tissues: a porous media approach, International Journal of Applied Mechanics, 2(3) (2000), 617-633.

[15] Lee D., Su C. M., Tran-Son-Tay R. and Shyy W., Fluid flow structure in arterial bypass anastomosis, J. Biomechanical engineering, 127 (2005), 611-618.

[16] Lei M., Giddens D. P., Jones S. A., Loth F. and Bassiouny H., Pulsatile flow in an end-to-side vascular graft model: Comparison of Computations with Experimental Data, Trns. Of ASME 123 (2001), 80-87.

[17] Lie M., Archie J. P. and Klinstreuer C.,Computational design of a bypass graft that minimizes wall shear stress gradients in the region of the distal anastomosis, J. of Vascular Surgery 25 (1997), 637-646.

[18] Liepsch D. and Singh M., Localization of bypass-induced changes in flow in coronary artery models, Indian J. of Experimental Biology 41(11) (2003), 1249-1252.

[19] Lu X. Y., Chen J. and Wang W.,Non-Newtonian effects of blood flow on hemodynamics in distal vascular graft anastomoses, J. Biomechanics, 39(11) (2006), 1983-95.

[20] Migliavacca F. and Dubini G., Computational modeling of vascular anastomosis, Biomechan Model Mechanobiol. 3 (2005), 235-250.

[21] O'Callaghan S., Walsh M. and McGloughlin T., Numerical modeling of Newtonian and non-Newtonian representation of blood in a distal end-to-side vascular bypass graft anastomosis. Medical engineering and physics 28 (2006), 70-74.

[22] Parktold K., Leuprecht A., Prosi M., Berk T., Trubel W. and Schima H.,Numerical Study of hemodynamics and wall mechanics in distal end-to-side anastomoses of bypass grafts, J. Biomechanics 35(2) (2002), 225-236.

[23] Qiao A., Liu Y., Li S. and Zhao H .,Numerical simulation of physiological blood flow in 2-way coronary artery bypass graft, J. of biological physics, 31 (2005), 161-182.

[24] Razavi S. E., Zanbouri R., Arjmandi-Tash O.,Simulation of blood flow coronary artery with consecutive stenosis and coronary-coronary bypass, BioImpacts 1(2) (2011), 99-104.

[25] Sankaran S. and Marsden A. L., The impact of uncertainty on shape optimization of idealized bypass graft models in unsteady flow. Physics of Fluids, 22(12) (2010), 121902.

[26] Sankaran S., Esmaily Moghadam M., Kahn A. M., Tseng E. E., Guccione J. M., Marsden A. L., Patient-specific multiscale modeling of blood flow for coronary artery bypass graft surgery. Ann Biomed Eng., 40(10) (2012), 2228-42.

[27] Satiraporn K., Phadungsak R. and Patcharaporn W., 3D numerical model of blood flow in the coronary artery bypass graft during no pulse and pulse situations: Effects of an anastomotic angle and characteristics of fluid, Journal of Mechanical Science and Technology, (September 2018), 32(9), 4545-4552.

[28] Steele B. N., Jing W. K., Joy P. H., Thomas J. R. and Taylor C. A., In Vivo Validation of a One-Dimensional Finite Element Method for Predicting Blood Flow in Cardiovascular Bypass Graft. IEEE Tran. Of Biomedical Engineering, 50(6) (2006), 649-655.

[29] Sud V. K. and Sekhon G. S., Arterial flow under periodic body acceleration. Bulletin of Mathematical Biology 47(1) (1985), 35-52.

[30] Sud V. K., Von Gierke H. E., Kaleps I. and Oestreicher H. L., Analysis of blood flow under time-dependent acceleration. Med. & Biol. Engi. & Comput. 23 (1985), 69-73.

[31] Sun A., Fan Y. and Deng X., Numerical investigation of blood flow in the distal end of an axis-deviated arterial bypass model. Biorheology 46(2) (2009), 83-92.

[32] Tandon P. N., Kawahara M. and Rana U.V., A mathematical model for blood flow through an arterial bifurcation. Int J Biomed Comput. 35(4) (1995), 309-325.

[33] Varshney G. and Katiyar V. K., Computational study of steady blood flow simulation in a complete coronary artery bypass anastomosis model. Canadian Journal of Pure and applied Sciences, 1(1) (2007), 103-109.

[34] Varshney G., Katiyar V. K. and Kumar S., Effect of magnetic field on the blood flow in artery having multiple stenosis: a numerical study. International Journal of Engineering, Science and Technology, 2(2) (2010), 67-82.

[35] Wang C. Y. and Bassingthwaighte J. B., Blood flow in small curved tubes, Trans. Of the ASME 125(6) (2006), 910-913.

[36] Wiwatanapataphee B., Poltem D., Wu Y.H. and Lenbury Y.,Simulation of pulsatile flow of blood in stenosed coronary artery bypass with graft. Mathematical Biosciences and Engineering, 3(2) (2006), 371-383.