Springer

KSME

Journal of Mechanical Science and Technology 32 (9) (2018) 4545~4552
www.springerlink.com/content/1738-494x(Print)/1976-3824(Online)
DOI 10.1007/s12206-018-0851-z

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[†]

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(Manuscript Received October 11, 2017; Revised April 10, 2018; Accepted May 17, 2018)

Abstract

Coronary heart disease (CHD) is a fatal condition caused by atherosclerosis or plaque on the arterial walls resulting in its breakage. Curing this disease effectively can be conducted through bypass graft surgery, which helps to restore the blood flow. However, some patients have required repeated surgery because of frequent failures of an implanted bypass graft. In this study, the stenosed coronary artery bypass graft including an analysis of the blood flow phenomena and wall shear stress, based on a three−dimensional computer model, was analyzed and developed to approach a realistic situation, inlet pulse and non−Newtonian behavior. The effects of the anastomotic angles (45°, 60° and 90°), blood characteristics (Newtonian and non−Newtonian) and inlet situations (with and without pulse) were taken into consideration. The results demonstrated that the anastomosis of 45° was the most appropriate for resolving the CHD problem and could act as a guide for medical treatment as well.

Keywords: Bypass graft; 3D computer model; Anastomotic angle; Non-Newtonian; Pulse; Blood flow; Simulation

1. Introduction

Coronary artery disease (CAD) is the most common type of heart disease, which happens when the arteries that supply blood to the cardiac muscle become hardened and narrowed. This is due to the accumulation of cholesterol and other material; namely plaque, building up on the inner walls of the coronary arteries. Therefore, the trend of CAD patients and rates of mortality are still increasing. This trend also includes the rising rates of patients who have had repetitive surgery because of the failure of the treatment. Thus, it is interesting to study CAD treatment by using modern technology; such as, a computer simulation in order to pre-plan before the operation and reduce the repetitive rate of surgery.

Numerous researchers have also investigated the problems of CAD. Lee et al. [1] studied the characteristics of the blood flow; such as, plaque formation, shear stress on the arterial wall and the index of stress oscillation in coronary arteries. The results from a model of an abnormal and normal blood vessel were compared. Owida et al. [2] on the other hand, investigated a numerical model of various coronary artery bypass grafts and updated the trend in the myocardial revascularization field. Alishahi et al. [3] studied the pulsatile flow and arterial wall behavior of a stenosed aorta and iliac arteries. The blood taken was non-Newtonian and the arterial wall was assumed to be rigid and flexible. Vimmr and Jonásová [4] analyzed the blood flow in a three-dimensional bypass of occluded coronary and femoral arteries. The blood flow was also both Newtonian and non-Newtonian in this case. The distribution of the velocity and wall shear stress on the surface at a joint between the plaque and the artery were investigated by Muraca et al. [5]. The relationship between the bifurcated angle and fluid properties was illustrated as well. Sankar et al. [6] studied the effects of the pulsatile blood flow through a stenosed artery with non-Newtonian behavior. The variations of the flow volume with the various parameters of the fluid were analyzed. The bypass grafts with 70 % of an occluded area in the artery with different anastomotic angles of 45°, 60° and 75°, were presented by Ko et al. [7]. The blood flow was considered to be Newtonian fluid in this study. The recirculation structure, secondary flow motion, mass flow rate and wall shear stress on the arterial wall were also analyzed. Furthermore, Hong et al. [8] presented a computer simulation of the pulsatile flow and macromolecular transport in a complex blood vessel. The axial velocity, secondary flow; such as, distribution of the LDL concentration was obtained. The stenosis effects were investigated by Politis et al. [9]. The

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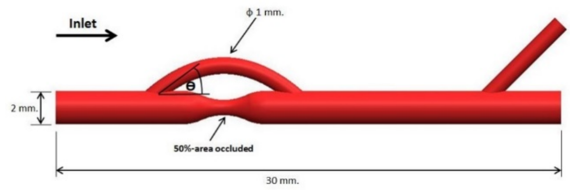
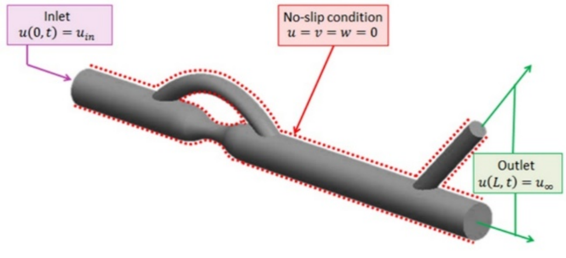


(a)



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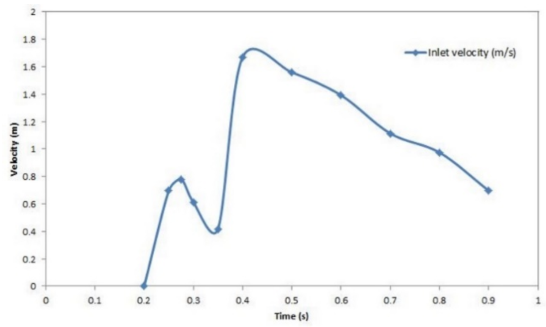
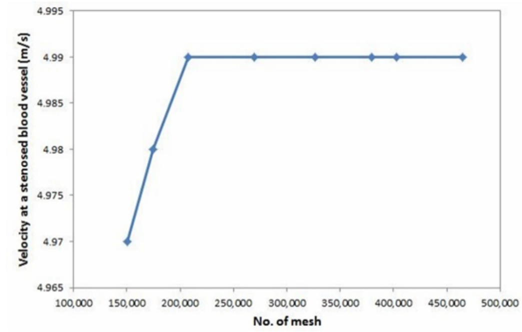


(c)

Fig. 1. (a) Physical model of the problem; (b) the boundary condition; (c) inlet flow situation profile for the pulsatile cases.

analysis of a coronary artery with different grafting distances were shown in this study. The stenosis volume of 25 %, 50 % and 75 %, respectively, as well as without stenosis were modeled with various inflow rate ratios. The velocity and wall shear stress distribution were illustrated. Poltem [10] presented the distribution of the blood flow through the coronary arteries with an unsymmetrical stenosis volume of 25 %, 50 % and 65 %, respectively. The flow distribution, pressure field and wall deformation in the cardiac cycle were carried out. Sakanarayanan et al. [11] demonstrated the graft flow dynamics and wall shear stress distribution at the anastomotic area of the aorta-left and right coronary bypass graft model, which was based on real-life situations. Two different cardiac cycles, at the onset of the ejection and during the mid-diastole, were used for both models. Suh et al. [12] investigated the blood flow behavior in carotid bifurcation and identified the coronary artery related to the hemodynamics; e.g., velocity, pressure and wall shear stress, which played an important role in the mechanism of the plaque's location and formation. Kim and Kim [13] developed a model of a bypass geometric parameter in a coronary artery; i.e., anastomotic angles, curvature radius and length to predict the blood flow phenomenon.

(a)



(b)

Fig. 2. (a) Structured grid used in this computation; (b) the grid independent test in the case of a coronary artery bypass graft with a 45°-anastomotic angle and no pulse situation.

A planar and non-planar model were compared. Other researchers have studied the effects of the various artery bypass graft models on the blood flow phenomena [14-16], as well as some researchers investigated the effect of the non-Newtonian fluid flow on different conditions and models [17-20]. In addition, the pattern of the stenosis artery and mechanical properties; such as, wall shear stress were investigated as well [21-25].

The research studies mentioned above have analyzed only some effects of the blood flow phenomena in a blood vessel model; such as, the anastomotic angle. Simultaneously, there are a few works that have systematically studied the various effects of the parameters. Therefore, this research investigated the blood phenomena in a three-dimensional model of the coronary artery with a bypass graft by analyzing the effects of the parameters as follows:

- The effects of the bypass graft.
- The effects of bypass grafts' angle (45°, 60° and 90°).
- The effect of blood flow properties (Newtonian and non-Newtonian).
- The effect of cardiac pulse.

The results of the velocity contour and shear stress on the arterial wall were also demonstrated. The Naviar-Stokes equation was used to analyze this problem by applying the finite volume method (FVM).

Therefore, the objective of this study was to propose more realistic conditions of the coronary artery bypass model to achieve an optimum bypassed degree for coronary artery disease treatment. The analysis of the blood flow phenomenon and related parameter effects would prove beneficial in the pre-planning procedures of treatment. Furthermore, this study could act as a database for research in the future. As such, the purpose of the study was to reduce the repetition rate as well as risk for repetitive practical operations. In this study, bypass artery models of 45°, 60° and 90° in situations with a pulse and no pulse were systematically investigated.

2. Problem definition and procedure

2.1 Physical model and boundary condition

Fig. 1(a) shows the three-dimensional computer model with the specified dimensions and geometry for the simulation. This model was modified from Ko et al. [7]. The diameter of the host and bypass arteries were 2 and 1 mm, respectively. The main artery was modeled to be a length of 30 mm and partially constricted (50 %-area). The anastomotic angle of the host artery and bypass graft was denoted by θ .

The assumptions of this work are indicated as follows: The blood flow was assumed to have a laminar flow and incompressible fluid with a density (ρ) of 1060 k/m^3. Dynamic viscosity (μ) of 0.0035 Pa.s was used for a Newtonian blood flow. The blood's viscosity was taken directly from the study of Vimmr and Jonásová [4] in case there was a non-Newtonian blood flow (Carreau-Yasuda model).

The profile of the inlet flow situations with a pulse situation referred to the flow-rate waveform at the left coronary artery, as mentioned by Sakanarayanan et al. [11] in which the heartbeat was 0.9 seconds per time as indicated in Fig. 1(c).

The boundary conditions proposed for the three-dimensional computer model as shown in Fig. 1(b) can be written as:

- Inlet: Blood is assumed to have a uniform velocity profile and the input of the pulsatile case is taken as Fig. 1(c).
- Outlet: Fully developed condition is assumed at the outlet.
- Wall: non-slip conditions are assumed on the entire arterial wall.

2.2 Governing equations

According to the three-dimensional computer model in Fig. 1(a), the Naviar-Stokes and the continuity equations governing the blood flow in a coronary artery with a bypass graft were modeled in a three-dimensional form and written as follows:

Continuity equation:

$$\frac{\partial \rho}{\partial t}+\vec{\nabla}\cdot\left(\rho\vec{V}\right)=0 . \tag{1}$$

Momentum equation:

$$\frac{\partial(\rho u)}{\partial t}+\vec{\nabla}\cdot\left(\rho u\vec{V}\right)=-\frac{\partial P}{\partial x}+\frac{\partial \sigma_x}{\partial x}+\frac{\partial \tau_{xy}}{\partial y}+\frac{\partial \tau_{xz}}{\partial z}+\rho f_x \tag{2}$$

$$\frac{\partial(\rho w)}{\partial t}+\vec{\nabla}\cdot\left(\rho w\vec{V}\right)=-\frac{\partial P}{\partial z}+\frac{\partial \sigma_z}{\partial z}+\frac{\partial \tau_{zx}}{\partial x}+\frac{\partial \tau_{zy}}{\partial y}+\rho f_z \tag{3}$$

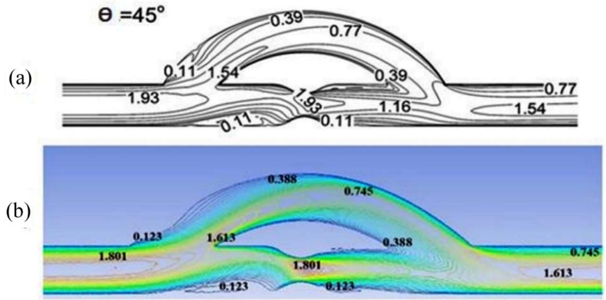


Fig. 3. The comparison of the velocity contour on the symmetric plane in the three-dimensional model: (a) Velocity contour of Ko et al. [7]; (b) velocity contour of the presented study.

$$\frac{\partial(\rho v)}{\partial t}+\vec{\nabla}\cdot\left(\rho v\vec{V}\right)=-\frac{\partial P}{\partial y}+\frac{\partial \sigma_y}{\partial y}+\frac{\partial \tau_{yx}}{\partial x}+\frac{\partial \tau_{yz}}{\partial z}+\rho f_y . \tag{4}$$

When $\vec{\nabla}=\frac{\partial}{\partial x}\bar{i}+\frac{\partial}{\partial y}\bar{j}+\frac{\partial}{\partial z}\bar{k}$

where ρ is blood density, u,v,w is blood velocity in an x, y, z direction, respectively, $\vec{V}$ is the velocity vector, t is time, P is pressure, τ is shear stress, f is body force, σ is normal stress.

The wall shear stress for the Newtonian fluid, where the viscosity is constant, is expressed as

$$\tau_{ij}=-\mu\left[\frac{\partial u_i}{\partial u_j}+\frac{\partial u_j}{\partial u_i}\right]. \tag{5}$$

When μ is the dynamic viscosity.

For the non-Newtonian blood flow, the viscosity of the fluid is shear-dependent and considered to be the molecular viscosity $\eta(\dot{\gamma})$ by using the Carreau-Yasuda model taken from Vimmr and and Jonásová's research [4]. The shear stress term is described as:

$$\eta(\dot{\gamma})=\eta_0+\left(\eta_0-\eta_\infty\right)\left[1+\lambda\left(\dot{\gamma}\right)^a\right]^{(n-1)/a} \tag{6}$$

where η_0 is the zero shear viscosity, η_∞ is the infinite shear viscosity, λ is a characteristic relaxation time, n is a flow index, $\dot{\gamma}$ is shear rate and a is the positive parameter that controls the transition to the lower Newtonian range.

2.3 Numerical procedure

In this study, the Naiver-Stokes equation and related boundary conditions were numerically simulated based on the finite volume method (FVM). All the computational procedures were implanted using the commercial code ANSYS (CFX) to represent the blood flow phenomena within the coronary artery bypass graft. A mesh independent test for all

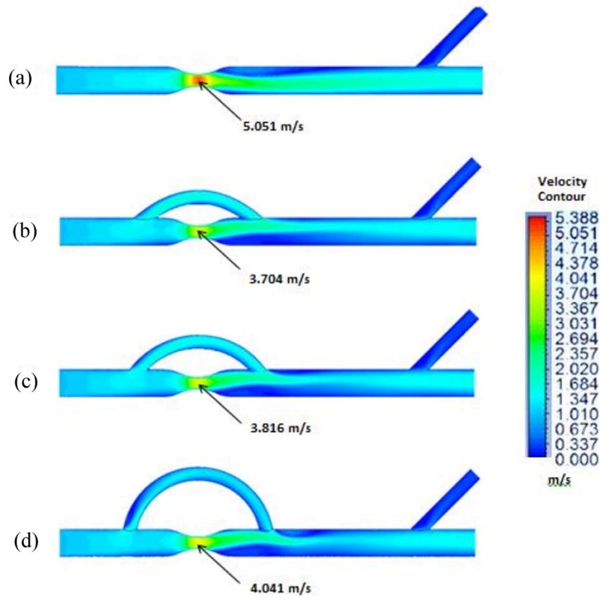


Fig. 4. Resultant velocity contour on the symmetric plane in the case of Newtonian fluid and no pulse situation with the time of 0.4 seconds: (a) Without a bypass; (b) $\theta = 45°$; (c) $\theta = 60°$; (d) $\theta = 90°$.

cases was conducted until the difference in the maximum velocity at the stenosed artery was lower than 0.3 % and the algorithm of the ANSYS ICEM CFD was used to generate the tetrahedral volume and prism layer mesh. The relative tolerance was specified at 1×10^{-3} to ensure that the solutions were accurate.

Figs. 2(a) and (b) show the structured grid used in this computation and the convergence grid test at 207500 elements in the case of the coronary artery with 45° bypass graft, respectively.

This calculating program was operated on the computer with Intel Xeon X5660 @2.80 GHz. CPUs and 48 GB ram. The computational time of each case to accomplish the result was less than 15 minutes.

3. Results and discussion

The effect of the anastomotic angles, inlet flow situations (pulse and no pulse) on the blood flow and wall shear stress distributions in the stenosed coronary artery bypass graft model were investigated. The blood flow phenomenon that occurred in the coronary artery was focused on and analyzed as per the following discussion. Navier-Stokes and continuity equations were used for all cases.

3.1 Verification of the model

For this study, the presented simulated results were validated against the numerical result which was directly taken from Ko et al. [7]. The bypassed artery with an anastomotic angle of 45° was selected in this validation case. The blood property was assumed to be Newtonian fluid with a density of 1060 kg/m^3 and viscosity of 0.0035 kg/m/s. The comparisons of the velocity contour on the symmetric plane for both cases are demonstrated in Fig. 3. It was found that the good agreement of the velocity contour on the symmetric plane between Ko et al. [7] and the presented solution was clearly shown. This favorable comparison lends confidence in the accuracy of the present numerical model; furthermore, it is important to note that there were some errors in the simulation, which might have been generated from the input of some properties and differences of the numerical scheme.

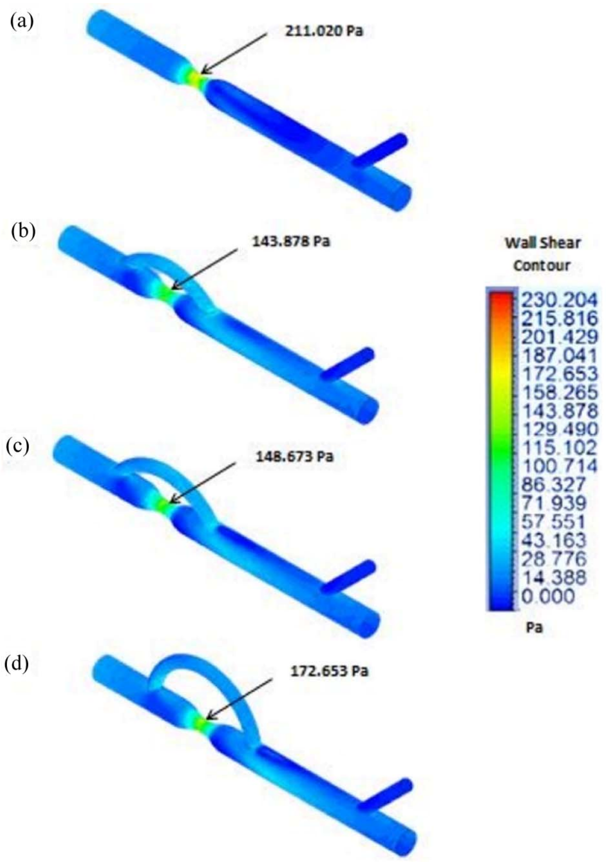


Fig. 5. Wall shear stress distribution in the case of Newtonian fluid and no pulse situation with the time of 0.4 seconds: (a) Without a bypass; (b) $\theta = 45°$; (c) $\theta = 60°$; (d) $\theta = 90°$.

3.2 The effect of the anastomotic angles

The effects of the anastomotic angles are illustrated in Fig. 4. The resultant velocity contours on a symmetric plane with the time of 0.4 seconds of the Newtonian blood flow with a no pulse situation are shown. The highest velocity in any case occurred at the center of the stenosis coronary artery that was caused by changing the cross-sectional area into a constricted blood vessel. In considering the effects of the anastomotic angles, the simulated results showed that the velocity of the bypassed coronary artery with a 45° angle was equal to 3.704 m/s, which was less when compared with 60° and 90° angles that equaled to 3.816 m/s and 4.041 m/s, respectively.

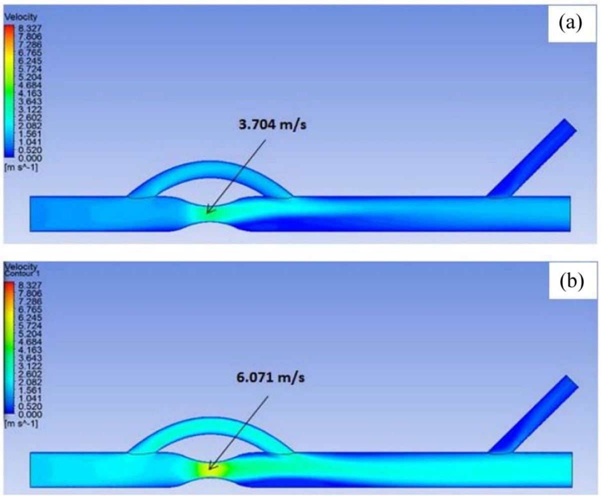


Fig. 6. Resultant velocity contour on a symmetric plane in the case of Newtonian fluid with the time of 0.4 seconds: (a) No pulsatile case; (b) pulsatile case.

This was because there was partial blood flowing into the bypass graft. Moreover, the most convenient blood flow phenomena played an important role at a 45° bypass, and the blood flow from the bypass graft into the host artery showed a smaller region of a low-velocity zone, which caused congested blood, near the outer surface of the host artery than other cases as well.

The result of the shear stress on the arterial wall affected by the gradient of the velocity and viscosity of the fluid with different anastomotic angles are shown in Fig. 5. It was observed that the remarkable point followed the velocity contour like in Fig. 4. Low wall shear stress regions occurred on the connecting point between the host artery and the bypass graft on both the interior and exterior sides and along the floor of the host artery. The lowest wall shear stress in the stenosed cross-sectional area of the host artery was represented in a case of the coronary artery bypass graft with a 45-anastomotic angle.

The velocity contour and streamlines on different cross-sectional planes, A and B, in case of the non-Newtonian and pulsatile blood flow situation, are depicted in Table 1. The researchers chose the result at 0.4 seconds of the flow-rate waveform, as the highest velocity of the inlet flow occurred at this period.

Generally, the coronary artery with a bypass graft at any anastomotic angle cases has a lower velocity at the narrowing of the host artery (at position A) than without a bypass graft case, which could reduce the risk of a vascular wall rupture.

At position A, the center of the stenosed host artery, the high and low-velocity zones occurred near the top side and the exterior edge, respectively. A pair of recirculation zones slightly inclined to the top side and the anastomotic angle values were directly variable to the velocity but had an inverse relationship with the recirculation zone.

At position B, the center of the bypass graft, the high and low-velocity zones slightly tilted to the top side and formed around the exterior edge, respectively, which was similar to position A. The recirculation zone appeared in the central zone of the bypass graft with an asymmetric intensity streamline that was caused by the curve of the artery and nonlinear behavior of the fluid properties. The relationship of the anastomotic angle, the velocity, and the recirculation zone was the same as that happened within position A.

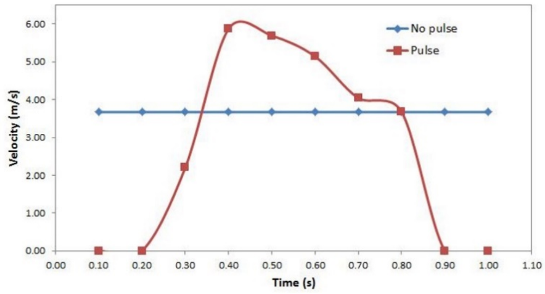


Fig. 7. Comparison of the velocity with the elapses in time in the case of the stenosed coronary artery with the bypassed 45°-anastomotic angle and Newtonian fluid at the center of the host artery between no pulse and pulse situations.

3.3 The effect of the inlet flow situations

The inlet flow situation affected the characteristics of the blood flow as shown in Figs. 6(a) and (b).

Figs. 6(a) and (b) demonstrate the resultant velocity contour with a flow time of 0.4 seconds in the case without a pulse and with a pulse, respectively.

The figures show that the maximum velocity of blood at the center of the stenosed coronary artery in the case of bypassed and no pulse was 3.704 m/s while the artery with a pulse was 6.071 m/s at the same time. It was found that the blood velocity at the stenosis area with no pulse was slower and the stagnation of the blood flow in both the host artery and the bypass graft was more than the pulse situation case due to the highest systole, which was occurring at a time of 0.4 seconds. Furthermore, the inlet pulse flow situation was similarly an actual systolic than that of the bypassed coronary artery with a no pulse situation as shown in Fig. 7.

3.4 The relationship between the velocity and length for three anastomotic angles

Fig. 8 illustrates the relationship between the velocity at the center line and the length of the host artery with three different anastomotic angles, 45°, 60° and 90° at a time of 0.4 seconds in the case of Newtonian blood flow and pulse input situation. It can be considered remarkable that all testing cases had the highest velocity at the length of 0.01 m. This was because this length was the narrowest cross-section area. Furthermore, the result of the 45° bypassed angle had the least velocity at 0.01 m of length and reached the smoothest flowing more than other angles due to the convenient blood flow within the artery.

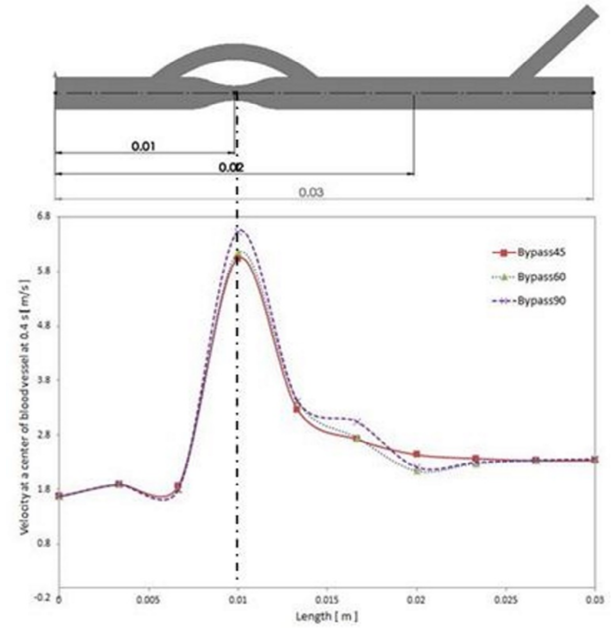


Fig. 8. The relationship between the velocity at a center line and the length of the artery with different anastomotic angles in the case of a pulse input situation at time of 0.4 seconds.

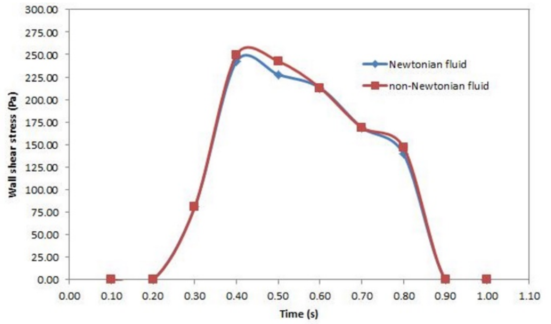


Fig. 9. Comparison of wall shear stress at the stenosed area with different fluid properties (in the case of a coronary artery bypass graft with a 45°-anastomotic angle and pulse situation).

3.5 The effect of fluid properties

The comparison of the wall shear stress at the stenosed area in the host artery with different fluid properties, Newtonian and non-Newtonian fluid, are displayed in Fig. 9. The results of both fluid properties were flowing in the same direction but were slightly different in magnitude, which conformed to Vimmr and Jonásová's research [4]. Therefore, we can consider the blood flow properties within the coronary artery to be Newtonian fluid for convenience and utilized less time for the calculation.

4. Conclusions

This research studied the flow behaviors in a three-dimensional computer model of a stenosed coronary artery bypass graft with varying anastomotic angles, inlet flow situations (pulse and no pulse situation), and fluid properties in the case of Newtonian and non-Newtonian blood. The effects of these different parameters on the velocity and wall shear stress during the blood flow in the coronary artery bypass graft were systematically investigated.

The main results can be summarized as follows:

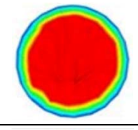
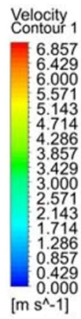
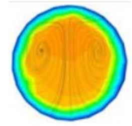
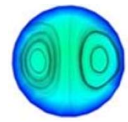
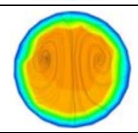
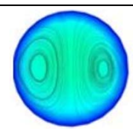
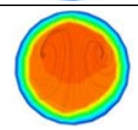
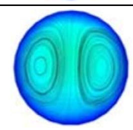
The anastomotic angle that connected the host artery and bypass graft was 45°, 60° and 90°, respectively. The result showed that the velocity at the stenosed coronary artery bypass graft of 60° and 90° was more than the 45° bypassed artery.

The fluid properties, Newtonian and non-Newtonian fluid, showed a similar result; therefore, the blood flow properties were considered as Newtonian fluid for convenience according to Vimmr and Jonásová [4]. The inlet flow velocity situations, no pulse and pulse situation, and the inlet pulse flow situations showed an appropriate result more than others because of the similarity to a realistic systolic.

In this study, the case of the coronary artery bypass graft with a 45°-anastomotic angle was found to be the best proper model for implementing in a coronary heart disease (CHD) problem because the velocity at the stenosis and wall shear stress on the arterial wall was less than other angles. Moreover, the case of the considered blood flow model, as a non-Newtonian fluid and used pulse input situation, was the closest to a reality situation as well.

The results presented here can be used as guidance for medical treatment and help to predict the outcome of the treatment.

Table 1. The resultant velocity contour and streamline pattern within the cross-sectional planes at position A and B in the case of pulsatile and non-Newtonian blood flow at 0.4 seconds.

	At position A	At position B	Range
No bypass		N/A	
Bypass 45°			
Bypass 60°			
Bypass 90°			

Acknowledgments

This research was supported by Thailand Research Fund (Contract No. RTA598009) and Faculty of Engineering Scholarship, Thammasat University, Thailand.

Nomenclature

a	: The positive parameter of Carreau-Yasuda model
f_i	: Body force [N/m^3]
n	: Flow index of Carreau-Yasuda model
P	: Pressure [Pa]
t	: Time [s]
u, v, w	: Velocity in x, y, z direction
$\vec{V}$	: Effective work
η_0	: Zero shear viscosity [Pa·s]
η_∞	: Infinite shear viscosity [Pa·s]
$\dot{\gamma}$	: Shear rate [1/s]
λ	: Characteristic relaxation time [s]
μ	: Dynamic viscosity [Pa·s]
ρ	: Density [kg/m^3]
σ	: Normal stress [N/m^3]
τ	: Shear stress [N/m^3]
θ	: Anastomotic angles of bypass graft

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