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Analysis of Hemodynamic on Different Stent Strut Configurations in Femoral Popliteal Artery

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ABSTRACT

Peripheral arterial disease is a narrowing of the peripheral arteries that might result in blockage if not immediately treated. Normally, an invasive technique called stenting is used at the stenosed arterial region to restore normal blood flow. Furthermore, the impact of physiological stresses on the stented arterial wall may worsen the condition and divert blood away from the main direction of flow. This phenomenon may promote the formation of thrombosis in the stented artery due to the presenting flow recirculation. This study aims to investigate the effect of physiological loads with different hemodynamic factors on the strut configuration in Peripheral arterial. The computational fluid dynamics (CFD) method is governed by continuity and Navier-Stokes equations were implemented. Three-dimensional geometries of the different strut configurations inserted into the peripheral artery are modelled. Each of the struts has different characteristics in terms of strut angle and the number of strut configurations. Due to hemodynamic differences, the pulsatile waveform of the velocity and pressure at the peak systole time were investigated. Hemodynamic factors analysed in this study consist of wall shear stress (WSS), velocity, and pressure. From the result, the prediction of thrombosis formation is around 30% higher for the largest number of strut design. The lowest formation of thrombosis was seen with the fewer strut configurations and the reduction of the strut angle with an overall percentage of 15%.

1. Introduction

The Femoral Popliteal arteries are very common sites of involvement in patients with atherosclerotic peripheral arterial disease (PAD). Plaques in the common femoral artery or at its bifurcation are currently treated surgically with endarterectomy, while plaques in the superficial artery are treated with endovascular ballooning and stenting [1]. However, after a few weeks of the stenting procedure, the restenosis or re-blockage of the artery can cause the narrowing of the artery that may reduce the flow of blood supply to the heart, and this may lead to heart attacks [2]. In addition, the availability of blockage within the artery gives enlargement to the complexity of the flow behaviour as well [3]. Thrombosis is the formation of a blood clot, known as a thrombus, within

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a blood vessel. The experts predicted that the stent strut geometry had a major contribution to the restenosis process. Even though stent strut geometry plays such a key role in the restenosis process, it has a relationship with the WSS distribution near the stent strut geometry. High shear stress and blood velocity may create a hemodynamic condition conducive to degenerative vascular injury. This result also suggests the necessity of evaluating flow characteristic and WSS-related indices in addition to the capacity of the device to reduce the flow, the WSS, and the pressure [4]. Based on previous study, lower amount of wall shear stress is found near the inner wall of the artery distal to the plaque region (stenosis), and in both percentages of stenosis, the maximum wall shear stress will accrue in the middle of the stenosis; however, it is much more in the higher rate of stenosis [5]. Besides, hemodynamic variables of velocity and WSS are correlated with intimal thickness, thus proving that atherosclerotic plaques tend to occur in areas of WSS_{low}. Thus, the flow characteristics in the Femoral Popliteal artery require study in order to identify the performance of the strut configuration.

The changes in the flow characteristics due to stent implantation in peripheral arteries have been the subject of numerical investigations through computational fluid dynamics (CFD) analyses. The restenosis development of Femoral popliteal artery implanted with different geometrical stent strut configurations can be invasively analysed and predicted with the current numerical simulation technology via the computational fluid dynamic (CFD) method [6]. On the other hand, computational fluid dynamics (CFD) was to quantify blood flow and determine the hemodynamic factors contributing to the disease from a simplified model [7]. The computational analysis of the stent performance in Femoral popliteal artery via the CFD method was based on hemodynamic parameter distribution that consists of pressure, velocity, and wall shear stress. The model was developed using computer-aided design (CAD) commercial software SOLIDWORKS, which was simulated using CFD commercial software that is ANSYS FLUENT. Physiological loads that act on Femoral popliteal artery, in combination with stenting, can lead to uncharacteristic deformations of the stented vessel. It is imperative to study the different stent strut configurations in Femoral popliteal artery because they are related to the flow characteristics that lead to the arterial disease. However, the flow phenomena in this particular region is not well understood since it involves mechanical load, bending load, and pulsatile load. Thus, this paper is to analyse the hemodynamic effects of different stent strut configurations and predict the growth of thrombosis.

2. Methodology

2.1 Simplified Geometry Model of Stented Femoral Popliteal Artery

The simplified geometry of the stented Femoral Popliteal artery was modelled. This arterial model has a different stent strut configuration. The Type I stent was based on the commercial stent for Femoral Popliteal artery of Absolute Pro. Type II and Type III stents, on the other hand, were modified struts from the Type I stent. Whereas the Type III stent was similar to the Type I but had an additional strut. The differences between Type I and Type II were in terms of strut angle. The strut angle for Type II was greater than that for Type I, which was comparable, as shown in Figure 1. Previous researchers reported that the strut angle has a relationship to the direction of blood flow associated with increases in the area of the vessel exposed to WSS_{low} and high WSSG [8]. Furthermore, Ladisa *et al.,* mentioned that the strut angle has an effect on cellular adhesion in vitro [9]. The variation of strut angle was made in order to analyse the hemodynamic effects of different single strut configurations. In the present study, blood flow was assumed to be Newtonian blood. This is due to the shear rates being greater than $50s^{-1}$ thus validating the widely held Newtonian assumption in large arteries as reported by Long *et al.,* [10].

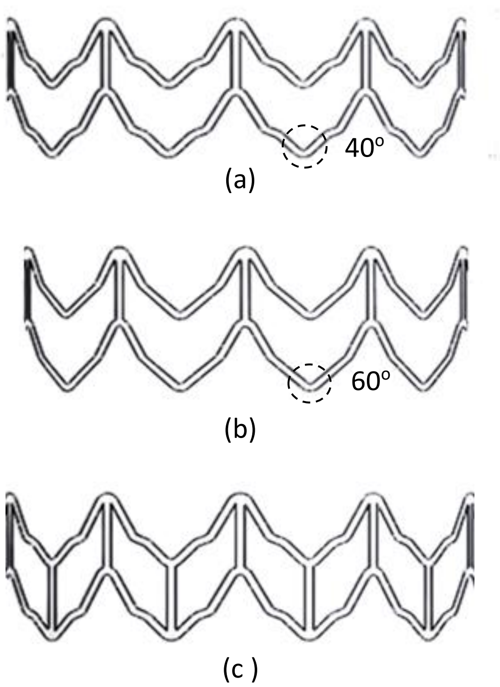

Fig. 1. Simplified model of stented of femoral popliteal artery (a) Type I (b) Type II (c) Type III

2.2 Meshing of Stented Femoral Popliteal Artery Model

A computational domain of a simplified Femoral Poplitel artery model implanted with three types of stents is drawn by using CAD software SOLIDWORKS 2018 (Dassault Systemes Solidworks Corporation, Waltham, MA) and then exported to ANSYS. The computational domain is generated into a tetrahedral mesh as shown in Figure 2. It is necessary for the solution to maintain accuracy and stability in order to get a good quality mesh. There are four processes to generate the mesh, which start with specifying the global mesh setting, inserting a local mesh setting, generating the mesh, and checking mesh quality. The grid independent test (GIT) was required to select the mesh size [11].

Fig. 2. Tetrahedral mesh of the computational domain

2.3 Parameter Assumptions

It has an average density of approximately 1060 kg/m3 and a dynamic viscosity of 0.0035 kg/ms [12]. The blood vessel is assumed to be rigid and there is no slip condition between stents and blood vessels [13]. Figure 2 also depicts the inlet and outlet boundary conditions. In the investigation of the hemodynamic effect on different strut cross-sectional areas, steady flow is taken into account. The velocity inlet and pressure outlet are assigned based on the cardiac phase of maximum systolic time. In this simulation, the velocity input and pressure outlet are set to 0.50 m/s and 21731 Pa, respectively [14].

2.4 Computational Model and Governing Equation

The Navier-Stokes equation and continuity equation are solved by the computing velocity inlet and pressure outlet. Under the assumptions of incompressibility, homogeneity and Newtonian blood, the equations are defined as illustrated these equation [15].

$$\rho\frac{\partial u}{\partial t} + \rho(u.\nabla)u = -\nabla P + \mu\nabla^2 u \tag{1}$$

$$\nabla.u = 0 \tag{2}$$

Where u is the primary vector u= [u, v, w], and P is the pressure that vary in space x, y, z and time, t. Wall shear stress estimation is common analysis in predicting performance of medical devices towards vascular wall. Wall shear stress (WSS) can be defined as shown in Eq. (3).

$$WSS_\tau = \frac{\partial v}{\partial r} \tag{3}$$

Where u is the fluid viscosity, v is the velocity along the vessel axis, and r is the perpendicular to and away from the wall.

3. Result

3.1 Grid Independence Test

Grid independence is a term used to describe the improvement of results by using successively smaller cell sizes for calculations. Figure 3 shows the result of the velocity profile for stented femoral popliteal artery models with different numbers of nodes, which are 380K, 191K, and 88K. The selection of a node is based on an unchanging result of the parameters with respect to the increasing number of nodes. Thus, the selection of the sample is based on the nodes. However, the higher the number of nodes, the higher the time consumed. Therefore, the sufficient number of nodes obtained using the present study mesh setting was 191K nodes

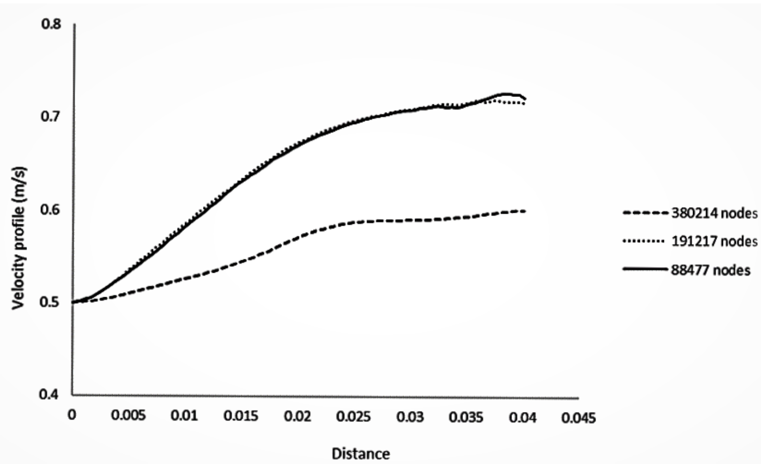

Fig. 3. Velocity distributions at the centre of femoral popliteal artery for different number of nodes

3.2 Verification

In order to extract the data, it is necessary to consider the same region for both the current and previous studies in order to make a comparison. The comparison of simulation results was analysed based on the previous study conducted by Ali A. Al-Allaq [16]. This finding was used as a reference for verifying the present study by calculating the relative error [17]. From the calculation, the relative error is approximately 7.8% and is acceptable due to the fact that the pattern of the result is quite similar as illustrated in Figure 4.

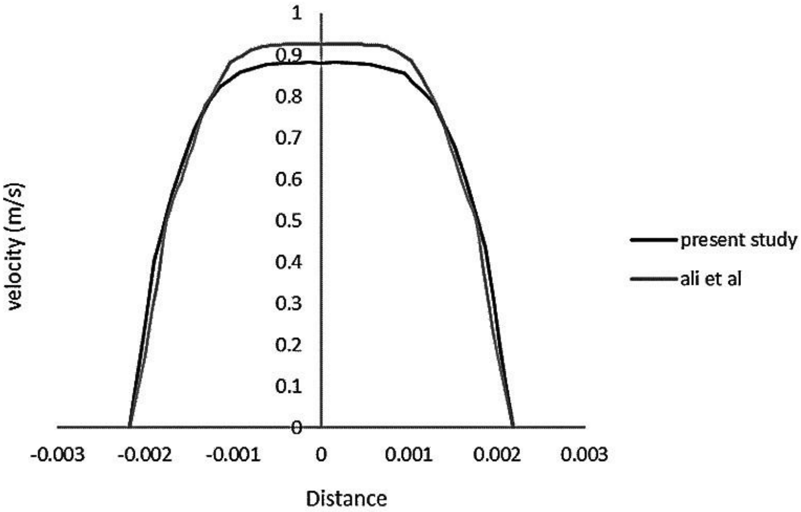

Fig. 4. Distribution of velocity between previous study and present study

3.3 Distributions of Velocity and Pressure at the Femoral Popliteal Artery

Despite the use of stents to improve blood flow, the physiological process can produce restenosis, a phenomenon that causes tissue growth, again closing the vessels. The hemodynamic parameters, such as velocity distribution and pressure distribution, were susceptible to re-stenosis [18]. Currently, direct calculation of flow velocity and pressure at the luminal surface is not possible in vivo. However, technological and scientific advances in mathematical and numerical models allow the mathematical simulation of blood flow and the precise determination of local hemodynamic effects [19]. High pressure may occur in the blood vessel, causing damage such as arteriolar medial hyperthrophy, intimal proliferation and fibrosis, and eventual obliteration of pulmonary arterioles and capillaries, thus increasing pulmonary resistance [20, 21]. In addition, blood damage can occur when there are large pressure drops in the artery. This is due to the high shear rate and outflow velocity from the inlet of the artery.

Based on Figure 5, the velocity distributions for different stent strut configurations are shown. The velocity has slightly reduced from the proximal to the distal region, which represents the vortices after the flow passes through the strut. The results also show that the highest velocity was observed at the distal region of the Type III stent as compared to Type II and Type I. Instead of velocity, pressure is an alternative parameter used to predict arterial stenosis. The contour of pressure for different stent strut configurations in Figure 6 indicates the pressure distribution for each type of stent. Overall, there was an increment of pressure from the proximal region to the distal region. Type III shows the highest reading value of pressure at the distal region. Compared with Figure 5 and Figure 6, the velocity decreases and the pressure increases after the blood flows through the stent. The velocity of blood flow increased due to the formation of plaques that narrowed the lumen size. The lesion will cause atherosclerosis, and over time, the plaque will grow to be massive, producing a mass of plaques and rupturing the artery [22]. The thrombogenic process is accelerated due to the effect of misalignment of blood flow direction near the stent strut that causes tiny injuries to the arterial wall. The hemodynamic parameters generate the formation of plaques that will stop the circulation of blood and cause backflow in severe cases [23].

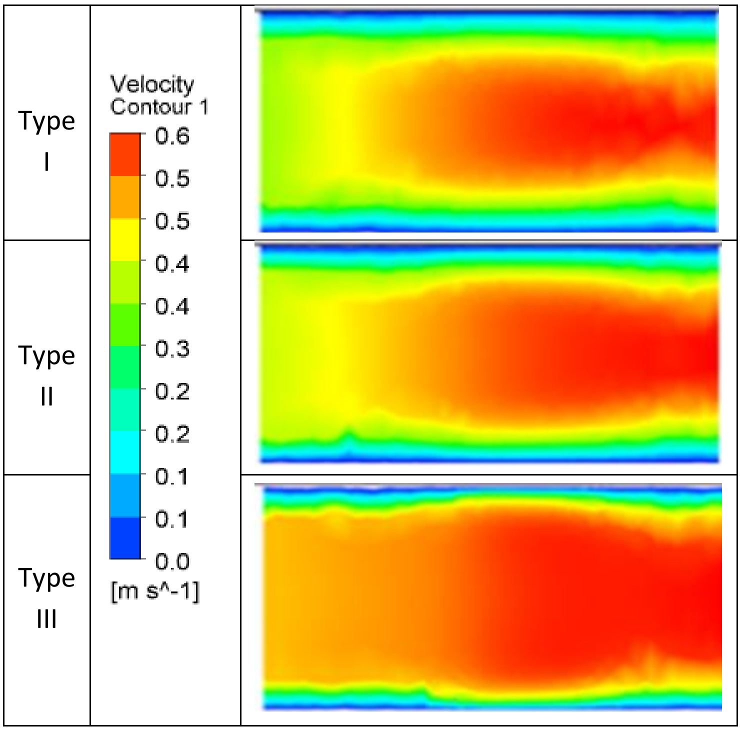

Fig. 5. The velocity distribution for different stent strut configuration

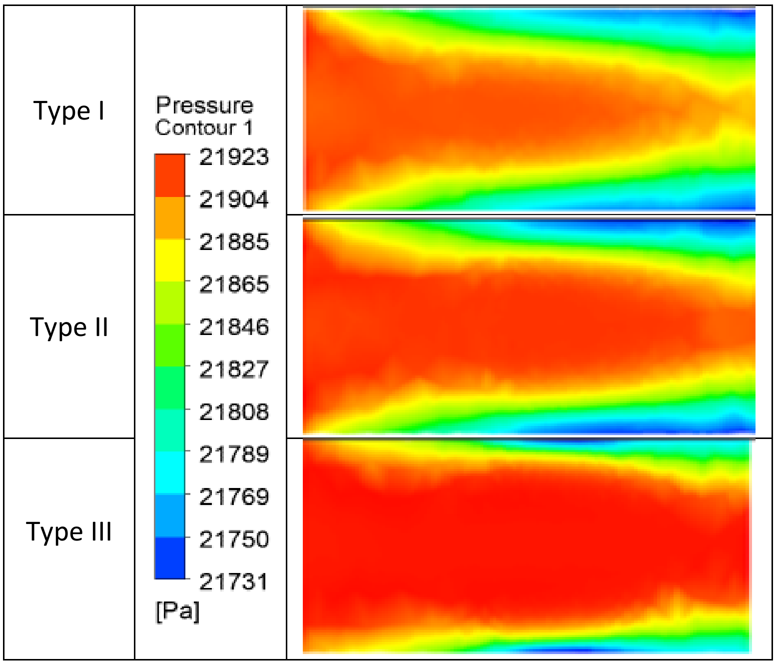

Fig 6. The pressure distribution for different stent strut configuration

3.4 Distributions of Wall Shear Stress at Strut Configurations

The vascular endothelium is exposed to a hemodynamic stress generated by the blood flow known as the wall shear stress. Wall shear stress is defined as the force per unit area exerted on the vessel wall by the blood flow, and it depends on blood viscosity and the blood flow velocity profile. The presence of local hemodynamic factors such as wall shear stress plays a major role in the generation, progression, and destabilization of atherosclerotic plaques. The wall shear stress increases as the atherosclerotic plaque progresses to create stenosis, particularly in the proximal part of the plaque at the stenotic region's entrance. It has been suggested that this increase in the wall shear stress could give rise to a reduction in the fibrous cap of the plaque through a process of apoptosis and, therefore, would favour destabilization and rupture of vulnerable plaques [24]. Local hemodynamic factors influence the evolution of atherosclerotic disease and may contribute to explaining the differences in distribution and progression of different atherosclerotic plaques. The presence of low wall shear stress promotes plaque progression, whereas physiologic shear stress protects the vascular endothelium. These local hemodynamic parameters, and in particular, an increase in the wall shear stress, have been suggested to play a role in the generation and destabilization of vulnerable plaques [25].

Figure 7 shows the distribution of wall shear stress for different stent strut configurations. From the observations, there is the formation of vortices due to the flow recirculation, especially at the distal region. This figure also shows the flow recirculation was slightly higher for the Type III stent as compared to other stents. This is due to the diverging of flow directions after being attached to the blunt shape of the strut configuration. However, the Type I stent shows the lowest activity of the vortices presented at the distal region. This phenomenon occurred because the diverging flow was aligned with the direction of the flow after passing through the strut configuration. Thus, the Type I stent shows the desired stent configuration due to the flow characteristics, which might reduce the formation of the thrombosis. A high re-circulation of the blood flow could increase the possibility of thrombosis being formed. Regions with flow recirculation may appear, causing the biological molecules to accumulate considering the change in near-wall flow behaviour and wall shear stress distribution. The commonly studied hemodynamic characteristic in the cardiovascular system is the parameter of wall shear stress (WSS), which is able to predict both the restenosis phenomenon of atherosclerotic and thrombotic patients [26]. A higher value of WSS than 70 dyne/cm^2 indicates a high shear thrombosis of the arterial wall, while WSS between -4 and 4 dyne/cm^2 shows a region prone to atherosclerosis formation [27]. From the calculation, the total wall surface area in a range greater than 70 dyne/cm^2 which was approximately 30% for Type III, 23% for Type II, and 15% for Type I, respectively.

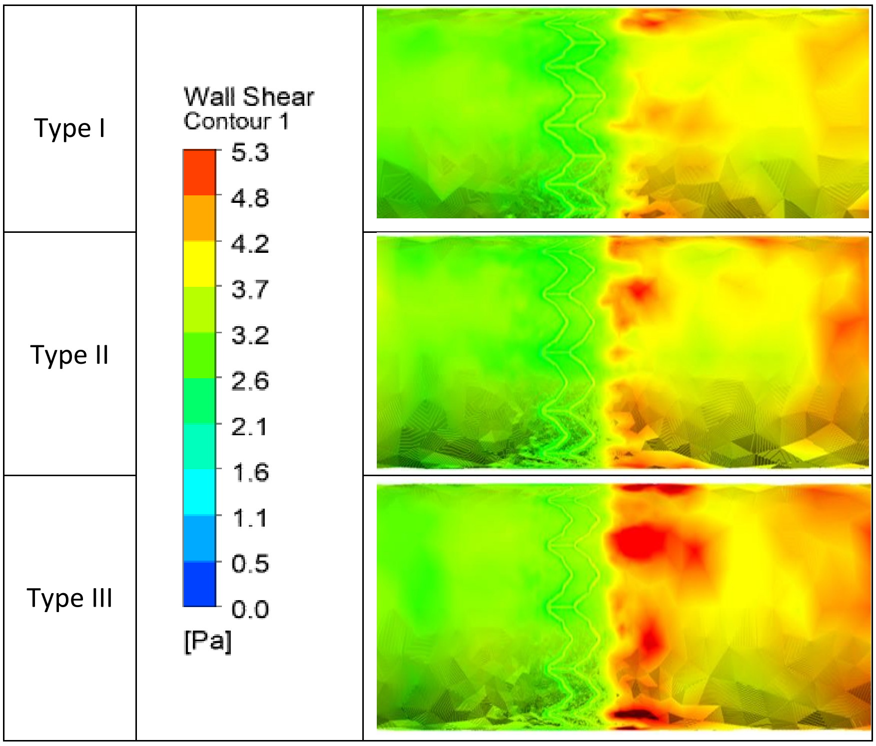

Fig. 7. Contour of wall shear stress for different stent strut configuration

4. Conclusion

In conclusion, the blood movement should be aligned with the direction of flow in order to reduce the chaos, especially at the distal region. The formation of the thrombosis was seen to be highest near the strut configuration at the distal region of FPA. However, the highest void area in the stented region reduced approximately 15% of the thrombosis growth. Thus, the fewer linkages of the strut and the reduction of strut angle manage to reduce the risk of thrombosis growth.

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