Investigation of flow characteristics of coronary slot stents using computational fluid dynamics

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Abstract. Coronary stents are metallic tubes inserted in coronary arteries to treat cardiovascular diseases. Palmaz-Schatz® Stent represents the original design from which many other slot-stent designs have been derived. In this paper, the impact of the geometry of the stent on its flow characteristics was investigated using the computational fluid dynamics (CFD) analysis. Seven design cases were selected for our research investigation, which were divided into three groups: (a) varying the slot length; (b) varying the slot width; and (c) varying the number of slots. Flow characteristics of stents with various geometries were studied by the CFD analysis using ANSYS CFX software. Results of the analysis revealed significant potential relationships between the geometry of the stent and its flow characteristics.

1. Introduction
Cardiovascular diseases constitute one of the major human health problems of modern times. Recently, the implantation of metallic coronary stents in the coronary artery has become an integral part of most interventional procedures for percutaneous revascularization [1]. The Palmaz-Schatz® stent (Johnson and Johnson Company, Warren, NJ, USA) is the most representative member of the family of slot stents. Much work [2, 3] has been done to investigate different flow behaviours of stents with various geometries. Tseng et al [4] emphasized the importance of computational fluid dynamics to obtain an improved endovascular aneurysm repair treatment outcomes and more accurate diagnosis, prognosis and prediction. Using CFD modelling, Morris et al [5] demonstrated the possibility of acquiring additional data for generating new patient specific insights that would help in better risk prediction and treatment planning. Gao et al [6] discussed the comparison between bioresorbable vascular scaffolds and metallic stents and concluded that bioresorbable vascular scaffolds are non-inferior to certain metallic components. Flow characteristics near stent strut configurations on femoropopliteal artery were studied by Paisal et al [7]. Their investigations show that by varying the design of stent model, different flow characteristics can be obtained and the stent with hexagon cross-section provides the best performance during surgical operations.

Currently, however, an appreciable research gap exists in investigating the flow characteristics of specific coronary stents with various geometries using CFD software. In this paper, a methodology for studying flow characteristics of Palmaz-Schatz type stents with various geometries by the CFD analysis using ANSYS CFX is presented in detail. Analytical results are discussed to reveal significant potential relationships between the geometry of the stent and associated flow characteristics.
2. Computational fluid dynamics model

2.1. Geometry and material properties
To study blood flow characteristics of different geometries, seven cases of varying designs of stents were modelled. It was assumed that all stents were equal in whole length $L$, thickness $T$ and mean diameter $D$ (See figure 1). Stent length, stent width, strut length, strut width, stent thickness, sector angle and numbers of slots vary in seven cases (See table 1).

![Figure 1. Geometric parameters of Palmaz-Schatz type stent.](image)

| Stent design case | Slot length $L_1$ (mm) | Slot width $W_1$ (mm) | Strut length $L_2$ (mm) | Strut width $W_2$ (mm) | Stent thickness | Sector angle $N$ (°) | Number of slots |
|-------------------|------------------------|----------------------|------------------------|------------------------|----------------|---------------------|-----------------|
| 1                 | 3.6200                 | 0.2200               | 0.2200                 | 0.1400                 | 0.1000         | 45                  | 24              |
| 2                 | 3.4390                 | 0.2200               | 0.3407                 | 0.1400                 | 0.1000         | 45                  | 24              |
| 3                 | 3.8010                 | 0.2200               | 0.0993                 | 0.1400                 | 0.1000         | 45                  | 24              |
| 4                 | 3.6200                 | 0.1980               | 0.2200                 | 0.1620                 | 0.1000         | 45                  | 24              |
| 5                 | 3.6200                 | 0.2420               | 0.2200                 | 0.1180                 | 0.1000         | 45                  | 24              |
| 6                 | 3.6200                 | 0.2933               | 0.2200                 | 0.1867                 | 0.1000         | 60                  | 18              |
| 7                 | 3.6200                 | 0.1760               | 0.2200                 | 0.1120                 | 0.1000         | 36                  | 30              |

To study the blood flow behaviour of the stent, a three-dimensional model of the blood flow domain was created. It was assumed that the stent had expanded against the stenosed artery to 4 mm, which was the diameter of the assumed healthy artery. Additional length was added to the two ends of the artery, respectively, to ensure the fully developed blood flow when it passes the domain of the stent. The minimum addition length ($L$) was determined by the following formulas [2]:

\[
Re = \frac{\rho u D}{\mu}
\]

\[
L = 0.06 \times Re \times D
\]

Blood behaves as a Newtonian fluid for shear rates great than 100 s$^{-1}$ [8]. Shear rates in large arteries are in general considerably larger than this value [9, 10]. In this model, blood was assumed to behave like a Newtonian fluid due to the relatively large size of the artery and high speed of the blood flow. According to the previous assumption, the diameter of the flow domain $D$ was 4 mm. The mass density $\rho$, speed of the blood flow $u$ and dynamic viscosity of the blood $\mu$ were defined as 1060 kg/m$^3$, 0.105 m/s, and 3.7 $\times$ 10$^3$ N s/m$^2$, respectively [2]. Thus, the Reynolds number $Re$ was 120.32, and an additional length $L$ should be larger than 28.88 mm. Due to different longitudinal recoils of stents with various geometry, lengths of various deployed stents were about 7.9 – 8.0 mm. Thus, the whole length of the model for each case was selected as 67.9 mm, which means the value of an additional length $L$ for each case was approximately 30 mm.

The model is axisymmetric with respect to its central axis in all aspects, including the geometry, materials, loadings and boundary conditions. To take advantage of these symmetric characteristics and to reduce the size, scope and processing time of the model, a basic section of it was used in the analysis using CFX (see figure 2). Three-dimensional solid triangular elements were generated by
CFX-Mesh depending on the geometry of the model. Minimum and maximum edge lengths were determined as 0.05 mm and 1 mm, respectively. The inlet boundary condition and loading specifies inlet flow parameters. A steady-state velocity of 0.105 m/s corresponding to the average blood flow velocity during one cardiac cycle was imposed on the inlet region for which the direction was along the positive direction of the z axis. The outlet boundary condition and loading specifies outlet flow parameters. An average static pressure 0 Pa was applied on the outlet region, which ensured the outlet pressure would not influence the blood flow pattern. It was assumed that there is no slip between the blood and luminal surface. To ensure accuracy, the advection scheme was set as High Resolution, the maximum number of iterations was defined as 100, and the residual target was chosen as 0.00001. Other parameters remained default values.

3. Results and discussion

3.1. Velocity distribution in stents

CFD results indicated that the velocity of blood flow changed suddenly when the blood flow made contact with struts of a stent, which resulted in large pressures imposed on the lumen of the artery. In the analysis, velocity patterns of the blood flow with different geometries of stents were investigated. The line which was located at the centre of the slot was selected as the sample (see figure 3).

![Figure 2. Discretization of model.](image_url)

![Figure 3. Line selected in CFD.](image_url)
The effect of varying the slot length upon the velocity distribution in stents was investigated in Cases 1, 2 and 3. Figure 4 illustrates the velocity along the flow direction at the centre of the slot of Cases 1, 2 and 3. In Case 2, the velocity of blood flow was the lowest in the three cases when it passed through the first strut. But its velocity became highest of all when it made contact with the second strut. When the blood flow passed through the last strut, the velocity was larger than that in Case 1. In Case 3, the velocity of blood flow was the highest of all when it was at the first and last strut, while the velocity was least when it made contact with the second strut. This revealed that shorter slot length, i.e. larger strut length, results in the reduction and increase of velocity in first and second struts, respectively. The velocity of blood flow in the third strut seems to have no direct relation to the slot length.

The effect of varying the slot width upon the velocity distribution in stents was investigated in Cases 1, 4 and 5. Figure 5 illustrates the velocity along the flow direction at the centre of the slot of Cases 1, 4 and 5. The three velocity curves nearly overlap at the positions of three struts. The slot width was a circumferential dimension. The slot length, i.e. the distance between struts, of these stents after full expansion was similar, so the impact of the slot width on the velocity of axial flow was weak.

Figure 4. Velocity distributions in Cases 1, 2 and 3 (Varying slot length).

Figure 5. Velocity distributions in Case 1, 4 and 5 (Varying slot width).

The effect of varying the number of slots upon the velocity distribution in stents was investigated in Cases 1, 6 and 7. Figure 6 illustrates the velocity along the flow direction at the centre of the slot of Cases 1, 6 and 7. The number of slots has a significant influence on the velocity of the blood flow at the first strut, while the impact of the number of slots on the velocity at the last strut was weak. In Case 6, as the number of slots decreased, the velocity reduced at the first strut and increased at the second strut. To expand to the same diameters, Case 6 required more axial foreshortening, which reduced the actual slot length after full expansion. So the velocity pattern of Case 6 was similar to that of Case 2 to some extent. On the contrary, as the number of slots increased, the velocity increased at the first strut and decreased at the second strut. To expand to the same diameters, Case 7 required the
least axial foreshortening, which enlarged the actual slot length after full expansion. So the velocity pattern of Case 7 was similar to that of Case 3 to some extent.

Figure 6. Velocity distributions in Cases 1, 6 and 7 (Varying number of slots).

3.2. Wall Shear Stress (WSS) distribution in stents

Stent geometry and expansion during deployment produce alterations in vascular anatomy that may adversely affect wall shear stress (WSS) [2]. The luminal surface of an artery with lower WSS is prone to form atherosclerosis [11]. Atherosclerosis is a disease characterised by the degeneration of the arteries because of the build-up of fatty deposits. Thus, a well-designed stent should have higher WSS distribution of flow in the stented artery. In the analysis, WSS patterns of the blood flow with different geometries were studied. The line which was located at the centre of the slot was selected as the sample, i.e. the line selected in the previous analysis. WSS was determined as the product of dynamic viscosity $\mu$ and shear rate $\dot{\gamma}$. Shear rate was calculated using the second invariant of the strain rate tensor, where $u$, $v$ and $w$ are the $x$, $y$ and $z$ components of velocity vector $\mathbf{u}$, respectively. [2]

$$\dot{\gamma} = \sqrt{2 \left( \frac{\partial u}{\partial x} \right)^2 + \left( \frac{\partial v}{\partial y} \right)^2 + \left( \frac{\partial w}{\partial z} \right)^2 + \left( \frac{\partial u}{\partial y} + \frac{\partial v}{\partial x} \right)^2 + \left( \frac{\partial u}{\partial z} + \frac{\partial w}{\partial x} \right)^2 + \left( \frac{\partial v}{\partial z} + \frac{\partial w}{\partial y} \right)^2}$$  \hspace{1cm} (3)

$$WSS = \mu \dot{\gamma}$$  \hspace{1cm} (4)

The effect of varying the slot length upon the WSS distribution in stents was investigated in Cases 1, 2 and 3. Figure 7 illustrates WSS distributions along the flow direction at the centre of slot of Cases 1, 2 and 3. The WSS of Case 2 was the highest of all three cases at the first strut. The WSS of Case 2 was slightly less at the second strut and more at the third strut than that in Case 1, respectively. The WSS of Case 3 was the lowest of all three cases at all positions of strut. Thus, this revealed that increasing the slot length enabled a reduction in the WSS.

Figure 7. WSS distributions in Cases 1, 2 and 3 (Varying slot length).
4. Summary
In this paper, a parametric model of a certain Palmaz-Schatz® Stent has been developed by varying the slot length, slot width and number of slots. The CFD simulation for studying flow behaviours of stents with various geometries has been conducted by CFX software. A good design stent requires lower velocity and higher WSS of the blood flow in the stented artery. To achieve these goals, the following guidelines can be considered. Decreasing the slot length and number of slots can decrease the velocity at the first strut, while increasing the slot length and number of slots can decrease the velocity at the second strut. Decreasing the slot length can increase WSSs at the first and third strut. However, it revealed that the slot width has no significant effect on flow behaviours, because it is a circumferential dimension.

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