Impacts on Hemodynamics of Different Bifurcation Coronary Stent Configurations


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Abstract

Percutaneous coronary intervention such as implantation of a stent (stenting) is the modern gold standard for coronary revascularisation. Stenting for a simple lesion has become a common procedure and is highly successful, but stenting for bifurcation lesions is more difficult and requires multiple stent implantations. The presence of the stent struts partially occludes blood flow from the main vessel to the side branch and leads to chaotic flow. As a result, bifurcation coronary stenting is often associated with dreaded complications like in-stent restenosis and thrombosis. Computational fluid dynamics (CFD) simulations were conducted on the classical T-stenting (TS) and reverse T-stenting (R-TS) techniques under pulsatile physiological conditions. The effects of TS and R-TS techniques on the changes in hemodynamics were systematically analysed. Removing the stent struts at the side branch (SB) ostium in the R-TS technique enlarges the recirculation bubble at the entrance of the SB, but reduces the area of non-physiological mean wall shear stress (WSS < 0.5 Pa) at baseline flow condition. However, the larger recirculation bubble results in a greater area of low WSS at induced hyperemia. The presence of the stent struts in the TS technique also reduces blood flow into the SB and creates low velocity regions in the vicinity of the stent struts. The differences in blood flow rate between the TS and R-TS techniques diminishes at induced hyperemia.

Introduction

Percutaneous coronary intervention (PCI) such as implantation of a stent (a metal wired mesh device) into the coronary artery remains one of the most effective treatments of coronary lesions [1]. The use of bare metal stents (BMS) and drug eluting stents (DES) has proven to be highly successful for simple coronary lesions with minimal occurrence of restenosis [6, 7, 10, 13]. However, 15–20% lesions are found in the vicinity of a coronary artery bifurcation, e.g., the diameters of MV and SB; angulation of the SB and the lesions location near the bifurcation [7]. As a result, the optimal stenting strategy remains debated among practitioners with many suggesting a provisional stenting approach (stenting of the MV only) with small diameter SB (< 2.75 mm) except for SB with visually suboptimal angiographic [1, 7]. However, SB closure is not uncommon for a provisional stenting approach; whereas complex two-stenting techniques may lead to other adverse clinical events such as in-stent restenosis (ISR) and thrombosis [7].

The aim of this study is to employ high-fidelity computational fluid dynamics (CFD) research tools to study the changes in hemodynamics in a coronary artery bifurcation with the classical T-stenting (TS) and reverse T-stenting (R-TS) techniques. Fundamental research in hemodynamics will lead to an unprecedented understanding of the linkages between specific bifurcation coronary stent configuration and post-stenting complications such as ISR and thrombosis. This advanced knowledge in hemodynamics can help customise two-stenting’s geometry and placement that will minimise the likelihood of fatal post-stent complications.

Figure 1. Schematic diagram of the T-stenting (TS) technique in an idealised coronary artery bifurcation. The main vessel (MV) of the bifurcation artery has a diameter of 3.0 mm whereas the side branch (SB) is of 2.5 mm and at an angle of 70° to the MV. A time-dependent parabolic velocity profile is imposed at the proximal end of the MV and zero pressure at the distal ends of both MV and SB. The top inset illustrates the tetrahedron elements density around the stent struts and lumen wall. The presence of stent struts (green arrow) near the SB ostium for TS technique is demonstrated in the bottom left inset. The reverse T-stenting (R-TS) technique, where the stent struts are artificially removed (red arrow), is shown in the bottom right inset.
Methodology

Computational Fluid Dynamics Methodology

The hemodynamics around bifurcation stents inside a coronary artery bifurcation were numerically simulated using CFD. We assumed blood to be Newtonian with a density of $\rho = 1.000 \text{ kg/m}^3$ and dynamic viscosity of $\mu = 2.504 \times 10^{-3} \text{ Pa} \cdot \text{s}$. Similar assumption on the blood rheology has been utilised by Taylor et al. [17] in CFD studies of coronary arteries. The lumen wall was assumed to be rigid as stent deployment has been shown to reduce arterial compliance [2] and blood flow inside the coronary stent follow the Medtronic Driver stent [16, Medtronic, Minneapolis, MN, USA] and a small cell area of 1 mm$^2$.

For the incompressible Navier–Stokes equations, the pressure and blood velocity, are $\rho \frac{\partial \mathbf{u}}{\partial t} + \mathbf{u} \cdot \nabla \mathbf{u} = -\nabla P + \mu \nabla^2 \mathbf{u}$, (1) and $\nabla \cdot \mathbf{u} = 0$, (2)

where the primitive variables $\mathbf{u}$ and $P$ in equations (1) and (2) were discretised using an open source, finite-volume solver OpenFOAM (OpenCFD Ltd.) with the convective term being approximated by the Gauss linear corrected scheme; Gauss self-filter central differencing scheme and Crank–Nicolson scheme for the diffusive and unsteady terms, respectively. Relaxation factors for $\mathbf{u}$ and $P$ are 0.5 and 0.7, respectively. Pressure and velocity coupling are through a standard SIMPLE algorithm. Convergence criteria are set at $10^{-3}$ for both $\mathbf{u}$ and $P$.

Coronary Artery Bifurcation and Different TS Techniques

Figure 1 presents a schematic diagram of the coronary artery bifurcation, the stent models and the boundary conditions applied in this study. The main vessel (MV) of the coronary artery bifurcation has a diameter of 3 mm whereas the side branch (SB) is of 2.5 mm at an angle of 70° to the MV. The dimensions of the coronary stent follow the Medtronic Driver stent (Medtronic Inc. Minneapolis, MN, USA), which has a strut thickness of 9 micron and a small cell area of 1 mm$^2$. The geometry of the bifurcation artery and the Driver stents were created using SolidWorks (SolidWorks Corp., Concord, MA, USA) and were subsequently imported to ICEM-CFD (ANSYS INC., Canonsburg, PA, USA) for computational mesh generation. The computational meshes consist of approximately 8 million tetrahedral elements. A time-dependent parabolic velocity profile was prescribed at the proximal end of the MV. The physiological waveform (shown in figure 1) was taken from the study of Kim et al. [9] and was reconstructed using the first four Fourier coefficients. Mean blood flows ($Q_{TM}$) at baseline and induced hyperemia (with a coronary flow reserve (CFR) = 1.5) are 1.5 and 3.1 mL/s, respectively. Reynolds numbers, based on MV’s diameter and blood velocity, are $Re = \rho Ud/\mu \approx 169$ and 349, and the Womersley number is $\alpha = 0.5d/\sqrt{2\pi \mu} \approx 2.74$. $\mathbf{U}$ denotes the mean blood velocity into the MV; $d$ stands for the diameter of the MV and CC is period of the cardiac cycle at 75 BPM. No-slip and no-penetration boundary conditions were applied to the lumen wall and stent struts. Zero pressure was employed at both distal ends of MV and SB. Although zero pressure outlets may induce unrealistic physiological transient pressure, it is offset by the prescribed Dirichlet velocity boundary condition [15] and lengthen outlet boundary from the stent struts [5]. It also allows the finite volume solver to adjust the blood flow out ($Q_{SB}$) of the SB according to the chosen bifurcation stent configurations.

Results

Changes in Hemodynamics over a Cardiac Cycle

Figure 2 presents the 3-dimensional streamline patterns over a CC for the baseline case with TS technique. At the beginning of the CC (far left in figure 2), a small recirculation bubble (yellow arrow) is observed at the entrance of the SB. The blood flow decreases in the next snapshot and the recirculation bubble decreases in size. The recirculation bubble is completely suppressed when the blood flow rate passes the minimum point of the CC. As the blood flow rate increases again, the recirculation bubble begins to build-up. However, the presence of the stent
struts partially occludes the blood flow through the SB ostium and leads to a "zig-zag" streamline pattern in the vicinity of the stent struts (red arrow). At maximum blood flow rate, the recirculation bubble is more pronounced and it retains its size after the maximum flow rate. The recirculation bubble reduces its size as the blood flow rate is further decreased, but it slightly increases in size before the CC is completed.

Employing the R-TS technique and inducing hyperemia result in qualitatively similar flow patterns. As a result, the effects of R-TS on the changes in hemodynamics are only discussed at maximum blood flow of the CC. The streamline patterns for both TS and R-TS techniques at this time instant are shown in the inset of figure 2. The recirculation bubble (yellow arrows) is at the most predominant stage for both techniques. However, the R-TS technique removes the stent struts near the SB ostium. As a result, blood flows freely into the SB and no "zig-zag" streamline pattern (red arrow) is observed for the R-TS technique. The absence of the stent struts near the SB ostium also leads to an increase in streamline velocity and thus a significant larger recirculation bubble in the SB.

Changes in Mean Hemodynamics Behaviour

![Figure 3](image)

Figure 3. Contours of mean wall shear stress (WSS) patterns. Contours are only shown for WSS < 0.5 Pa to clearly demonstrate areas of the lumen wall that are subjected to non-physiological WSS. Green arrows show the area of WSS < 0.5 Pa in the MV.

While instantaneous streamline patterns demonstrate the levels of chaotic flow using different TS techniques, adverse clinical events are better predicted by locating the non-physiological mean wall shear stress (WSS < 0.5 Pa) regions. Low WSS is shown to have significant correlations with excessive neointimal hyperplasia (NH), which is the main reason for ISR [3, 4, 8, 14]. Low WSS (< 0.5 Pa) affects the shape and alignment of the endothelial cells that increase the permeability of the endothelial layer and thus excessive NH [12]. Figure 3 shows the contours of WSS patterns for TS and R-TS techniques at baseline and induced hyperemia. To clearly demonstrate the regions of low WSS, only lumen areas with WSS < 0.5 Pa are displayed. At baseline, the R-TS technique reduces the area of low WSS at the distal end of the SB ostium as compared to the TS technique. It also shows slightly higher WSS values at the proximal end of the SB ostium. However, the R-TS technique has resulted in a larger area of low WSS in the MV distal to the bifurcation (green arrow).

In contrast, inducing hyperemia reduces the area of low WSS in the SB as compared to the MV. The reduction in low WSS area is mainly observed at the distal end of the SB stent and can be explained by the relatively smaller recirculation bubble in the TS technique. Although the area of low WSS is comparatively smaller in the TS technique, the WSS values at the proximal end of the SB ostium in the TS technique remain lower than the R-TS technique. It is also clearly observed that the absolute minimum WSS in the TS technique is closer to the entrance of the SB as compared to the R-TS technique due to a smaller recirculation bubble.

![Figure 4](image)

Figure 4. Contours of mean velocity magnitude (U) patterns. Slow velocity regions near the SB ostium (red arrows) are presented in the TS technique. These slow velocity regions are not observed in the R-TS technique.

Adverse clinical events can also be predicted by examining the mean velocity magnitude (U) as shown in figure 4. Besides the low U regions at the proximal end of the SB ostium, the presence of the stent struts in the TS technique has resulted in low U regions on top of the stent struts (red arrows) and near the distal end of the SB ostium. These low U regions are prone to platelets, lipids and white blood cells deposition and thus lead to NH [15]. The absence of the stent struts in the R-TS technique significantly improves the blood flow to the SB. However, low U region at the proximal end of the SB ostium remains which may be attributed to the large angulation of the SB that generates a recirculation bubble at the entrance of the SB.

The increase in blood flow to the SB is quantified by presenting the mean blood flow rates out of the MV (Q_MV) and SB (Q_SB) for different bifurcation stenting techniques at baseline and induced hyperemia in table 1. In both TS and R-TS techniques, majority of the blood flow out of the MV and less than 20% of the blood flows into the SB. One of the reasons for the high volume of blood flows out of the MV may be attributed to the large angulation of the SB (at 70°). At both baseline and induced hyperemia, the R-TS technique results in higher blood flows into the SB as compared to the TS technique. However, the advantage of the R-TS technique diminishes (from 19% to 16%) with induced hyperemia. This reduction is significant as hyperemia is considered at only CFR = 1.5, but it is unclear whether further increase in CFR will keep reducing the percentage increase in blood flow into the SB using the R-TS technique.
Table 1. Mean blood flow (mL/s) out of the MV ($Q_{MV}$) and SB ($Q_{SB}$); percentage of blood flow out of SB relative to the blood flow into MB; and percentage increases in blood flow out of the SB using the R-TS technique as compared to the TS technique.

<table>
<thead>
<tr>
<th>study(s)</th>
<th>technique(s)</th>
<th>$Q_{MV}$ (mL/s)</th>
<th>$Q_{SB}$ (mL/s)</th>
<th>$Q_{SB}/Q_{IN} \times 100%$</th>
<th>$Q_{SB, R-TS} - Q_{SB, TS}/Q_{SB, TS} \times 100%$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Baseline</td>
<td>T-stenting (TS)</td>
<td>1.29</td>
<td>0.21</td>
<td>14</td>
<td>19</td>
</tr>
<tr>
<td>Hyperemia</td>
<td>reverse T-stenting (R-TS)</td>
<td>1.25</td>
<td>0.25</td>
<td>17</td>
<td>19</td>
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<tr>
<td>Hyperemia</td>
<td>T-stenting (TS)</td>
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<td>0.31</td>
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<td>16</td>
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<tr>
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<td>2.75</td>
<td>0.36</td>
<td>12</td>
<td>16</td>
</tr>
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</table>

Discussions and Conclusion

The changes in hemodynamics in the presence of a bifurcation coronary stent were studied using computational fluid dynamics (CFD) research tools. The T-stenting (TS) and reverse T-stenting R-TS techniques were employed into an idealized coronary artery bifurcation with side branch (SB) at large angulation (70°) to the main vessel. Two different physiological flow conditions were considered: baseline and hyperemia (simulating a coronary flow reserve of 1.5). For both stenting techniques and flow conditions, the instantaneous streamline patterns are qualitatively similar to each other with the recirculation bubble at the entrance of the SB builds-up and suppresses throughout the cardiac cycle. In the R-TS, the recirculation bubble is relatively larger as compared to the TS technique. The TS technique shows larger area of low mean wall shear stress (WSS < 0.5 Pa) at baseline but has smaller low WSS region at induced hyperemia. The larger recirculation bubble in the R-TS technique is the reason for the increased area of low WSS at induced hyperemia. For both baseline and hyperemia flow conditions, the presence of the stent struts at the SB ostium with the TS technique creates low velocity regions near the stent struts and reduces the blood flows into the SB. These reductions in flow velocity and blood flows into the SB with the TS technique are more pronounced at baseline condition.

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References


