Numerical Simulation of Physiological Blood Flow in 2-way Coronary Artery Bypass Grafts

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Abstract. The Coronary Artery Bypass Graft (CABG) yields excellent results and remains the modern standard of care for treatment of occlusive disease in the cardiovascular system. However, the development of anastomotic Intimal Hyperplasia (IH) and restenosis can compromise the medium- and long-term effects of the CABG. This problem can be correlated with the geometric configuration and hemodynamics of the bypass graft. A novel geometric configuration was proposed for the CABG with two symmetrically implanted grafts for the purpose of improving the hemodynamics. Physiological blood flows in two models of bypass grafts were simulated using numerical methods. One model was for the conventional bypass configuration with a single graft (1-way model); the other model was for the proposed bypass configuration with two grafts (2-way model). The temporal and spatial distributions of hemodynamics, such as flow patterns and Wall Shear Stress (WSS) in the vicinity of the distal anastomoses, were analyzed and compared. Calculation results showed that the 2-way model possessed favorable hemodynamics with uniform longitudinal flow patterns and WSS distributions, which could decrease the probability of restenosis and improve the effect of the surgical treatment. Concerning the limitations of the 2-way bypass grafts, it is necessary to perform animal experiments to verify the viability of this novel idea for the CABG.

Key words: anastomosis, CABG, cardiovascular system, geometric configuration, hemodynamics, intimal hyperplasia, restenosis

1. Introduction

Atherosclerosis is a critical cardiovascular disease, characterized by the deposition of atheromatous plaques containing cholesterol and lipids on the innermost layer of the walls of the large and medium-sized arteries, resulting in a reduction in the cross-sectional area of the vessel lumen and an impeded or blocked blood flow. Atherosclerosis is prone to occur in the coronary artery, which is the principal blood supply vessel to the heart. The Coronary Artery Bypass Graft (CABG) yields excellent results and remains the modern standard of care for treatment of occlusive disease in the cardiovascular system. An operative bypass graft using an autologous vein or prosthetic conduit is implanted into the body around the diseased segment. The objective of the bypass is to take blood from just above the diseased segment

and return it to healthy vessels distal to all disease, in order to restore normal blood flow to the myocardia.

Unfortunately 26% of CABG operations may fail within one year [1]. Therefore, a major problem with CABG is its poor long-term patency rate. The causes of failure are varied. Intimal Hyperplasia (IH) and restenosis (i.e. excessive tissue overgrowth and renewed plaque formation) tend to develop in the vicinity of distal anastomosis and result in early graft failure [2–6]. IH has been extensively reviewed, and clinical observations indicate that IH predominantly occurs in four regions of anastomosis [7]: the heel, the toe, the suture-line, and the floor opposite the anastomosis (Figure 1). Many factors influence the initiation and development of IH and restenosis, for example a physiological and biomechanical mismatch between the graft and coronary artery, mural injury, local hemodynamics, etc. [2, 8–10].

Disease formation at the suture-line can be attributed to surgical injury, material mismatch and the abnormal flow patterns around the anastomosis [11–15]. Loth *et al.* suggest that IH is initiated along the suture-line because of mural injury

Figure 1. Geometric models of partially stenosed coronary artery bypass grafts.

[16]. Bassiouny *et al.* [17] and Wieslander *et al.* [18] postulated that suture-line IH formation arises from vascular injury, vascular surgery and compliance mismatch. Evidence has also accrued that hemodynamic factors are associated with the localization and progression of IH [17, 19]. Sottiurai *et al.* suggests that compliance mismatch is not the cause of IH development, but it enhances the pathogenesis, whereas Hemodynamic factors provide the underlying stimulus [4]. The hemodynamic factors can promote reactive mechanisms resulting in growth process and can lead to the activation of thrombocytes, which contribute to the formation of IH in the distal anastomotic region [2, 17]. Characterized by disturbance, eddies, oscillating Wall Shear Stress (WSS), large and sustained temporal-spatial Wall Shear Stress Gradient (WSSG), and increased particle and cellular residence time, the altered flow patterns created by the geometry of the anastomosis have been shown to promote IH [3, 16, 20, 21]. It has been found that IH corresponds to locations of slow particle velocities and low WSS [11, 18]. It has also been suggested that high WSSGs and long residence times might be responsible for the localization of anastomotic IH [4, 11, 17, 23, 24].

While some IH occurrence is expected due to surgical injury and subsequent healing, elevated WSS-based hemodynamic parameters as well as particle-wall interactions appear to provide a potential mechanism responsible for continued advancement of IH occurrence in the suture-line region resulting in late-stage graft failure [11]. Longest *et al.* established a non-dimensional Near-Wall Residence Time (NWRT) model to quantify the severity of particle-wall interactions which can account for particle stasis, elevated near-wall concentration, platelet activation, and surface reactivity [11–15]. This novel model includes a NWRT factor, a Platelet Stimulation History (PSH) factor, as well as a Surface Reactivity (SR) factor. Significant values of the NWRT parameter are indicative of platelet activation and adhesion, which comprise necessary steps in the formation of mural thrombi. The NWRT model provides new severity parameters on the cellular-level basis for the hemodynamic simulation.

It is well known that hemodynamics are sensitive to geometric factors [25–27]. The anastomosis configuration of the CABG is critical, and affects both decision making and outcome of bypass grafting. Thus it is an important factor influencing the medium-and-long term validity of bypass graft [10, 28]. Some researchers have studied the relationship between the junction angle and the local hemodynamics of anastomosis. Inzoli *et al.* [10] and Perktold *et al.* [29] found that a smaller junction angle has more advantages than a larger one. Based on the hypothesis that non-uniform hemodynamics may trigger abnormal biological processes leading to rapid restenosis and hence early graft failure, Lei *et al.* [24] and Kleinstreuer *et al.* [30] were convinced that postoperative disease might be significantly mitigated by finding graft-artery bypass configurations in which the WSSG is approximately zero, thus achieving almost uniform hemodynamics. They optimized the graft-artery bypass configurations with computer aided design techniques to ameliorate the hemodynamics [24]. Others studied different configuration variations such as

patched, cuffed, side-to-side, and end-to-end anastomosis [13, 18, 27, 31–35]. Investigations of improved anastomosis configurations are still ongoing.

In conventional CABG surgery, a single graft was employed to connect with the host artery [36–39]; we have named this configuration the “1-way bypass graft” (Figure 1a). Because of the intrinsic asymmetric geometric characteristics of the 1-way bypass graft, it will inevitably create disturbed flow at the anastomosis. In particular, the blood flow from the graft will strongly strike the floor, resulting in flow stagnation and reversed flow. Disease formation on the bed of the junction is thought to be entirely due to the abnormal flow patterns created by the impinging blood flows from the graft into the artery [40]. Such flow behavior is unphysiological, and it will result in non-uniform and violent hemodynamic changes in the CABG, further bring about the possibility of IH and restenosis. Thus, hemodynamic flow patterns in distal end-to-side anastomoses are widely implicated in the initiation of the disease formation process. In order to improve the hemodynamics induced by the asymmetric geometry of the 1-way CABG and increase the treatment effectiveness of CABG, we propose a novel configuration for CABG employing two symmetrically implanted grafts, termed the “2-way bypass graft” (Figure 1b).

The purpose of the present study is to investigate new approaches to CABGs with regard to anastomosis configuration. The objective of this study is to develop a detailed description of the hemodynamics in the aorto-coronary bypassing models with 1-way and 2-way grafts. The physiologically pulsatile blood flows in the early stage after CABG surgery are investigated by means of numerical simulation using Computational Fluid Dynamics (CFD).

The remainder of the paper is organized as follows. In Section 2, we describe how to construct the geometric models for the 1-way and 2-way CABGs. The simulating conditions and method are described in Section 3. Section 4 demonstrates the simulation results and analysis. Finally, discussion of the present work is given in Section 5, with a particular focus on hemodynamic comparisons between these two models.

2. Geometric Models

For the sake of comparison, the geometric structure of the 2-way model is similar to the 1-way model to the extent that the vessels follow the same centerlines for both models, and the total lumen area of grafts in the 2-way model is equal to that of the 1-way model in order to keep the distal flow rates or perfusion identical for both models (Figure 1). To minimize IH formation, the typically favored conduit of choice for bypass graft is autologous vessels. However, polytetrafluoroethylene (PTFE) grafts are often implemented in many instances. In our models, we employ flexible thin-walled PTFE grafts in order to facilitate comparison between the two models. In addition, the diameter of PTFE graft can be well defined to facilitate numerical simulation. Usually, a graft with a diameter larger than that of the coronary artery

is used in coronary bypass surgery. Hence, the anastomosis cannot be considered as the intersection of two cylinders with equal diameters. The intersection between the graft and the coronary has an elliptical curve, which is due to both the oblique graft cut and the required deformation of the larger-diameter-graft end to allow its surgical suture with the smaller-diameter-coronary side. This fact will be taken into account in this paper for geometric modeling, and the deformation of the bypass graft end is considered in order to reproduce the geometry obtained in the surgical procedure as closely as possible.

Most previous investigators assumed that the coronary artery was completely occluded when studying the hemodynamics of CABG, and took into account only the blood flow in the graft. In our study, the coronary artery is considered to be a straight tube with a smooth axisymmetric constriction representing a medium stenosis. The stenosis can be described with a cosine function [8]:

$$R(z) = \frac{d}{2} - \frac{\delta}{2} \left[1 + \cos \left(\frac{2\pi z}{L} \right) \right] \quad |z| \leq L \quad (1)$$

where d is the diameter of the coronary artery, δ the thickness of the axial-symmetrical stenosis, and L the length of the lesion. The thickness of stenosis is $\delta = d/4$, i.e. the constriction rate of area at the stenosis is 75%. The length of the stenosis is $L = d$.

The upstream conduit of the graft is parallel to the coronary artery. The downstream segment of the graft follows an arc curve, meeting the coronary artery at an angle of α with end-to-side anastomosis. The anastomosis heel is located at about one diameter of the coronary artery downstream from the stenosis in the 1-way model. The cross-section of the graft alters gradually along its length, changing from a circular to an elliptical shape, to fit the smaller artery at the anastomosis [10]. The cross-sectional ellipse of the circular graft with the diameter of D cut by an oblique plane at an angle of α can be described with the elliptical minor and major axes of

$$\begin{cases} a_1 = D \\ b_1 = D / \sin \alpha \end{cases} \quad (2)$$

where a_1 and b_1 denote the minor and major axes respectively.

Because the graft diameter D is greater than the coronary diameter d , when it is sutured with the coronary, the graft must deform to reduce its minor axis, i.e. the minor axis will become

$$a_2 = k \times d \quad (k \leq 1.0) \quad (3)$$

where a_2 is the reduced minor axis of the intersection ellipse and k the deformation ratio of the graft end. In the present model, $k = 0.8$. Supposing that the deformed

elliptical shape has a perimeter equal to that of the undeformed ellipse, the major axis b_2 of the deformed intersection ellipse should be extended. This deformed graft is quite similar to the Taylor patch model, which is commonly considered to be effective for improving the hemodynamics of bypass graft [27]. The major axis b_2 of the deformed intersection ellipse can be calculated from the following equation:

$$\pi \left[\frac{3}{2}(a_1 + b_1) - \sqrt{a_1 b_1} \right] = \pi \left[\frac{3}{2}(a_2 + b_2) - \sqrt{a_2 b_2} \right]. \quad (4)$$

The length of the coronary conduit upstream from the stenosis and downstream from the toe is 20 times the coronary diameter ($20d$) to allow full development of the velocity profile and minimize the influence of the inlet and outlet boundary conditions. The graft was also extended to obtain fully developed flow patterns. The inlet of the graft is located in the same section as the coronary entrance. The normal mean diameters of the left anterior descending coronary artery and the graft are 2.6×10^{-3} m and 3.8×10^{-3} m respectively [28]. In the present study, the diameter of the coronary artery is $d = 2.6 \times 10^{-3}$ m, and the diameters of the grafts in the 1-way and 2-way models are $D_1 = 3.8 \times 10^{-3}$ m and $D_2 = 2.687 \times 10^{-3}$ m respectively (the total lumen area of the grafts for both models are 1.134×10^{-5} m²). The hemodynamic conditions of the bypass graft depend most critically on the graft-host angle. The larger the angle of incidence at the distal anastomosis, the more favorable these conditions will be for IH. In the present study, the anastomotic angle is a medium one with $\alpha = 30^\circ$.

3. Simulating Conditions and Method

The blood is assumed to be an isotropic, homogeneous, incompressible, Newtonian continuum having a constant density and viscosity. The vessel walls are assumed to be rigid and impermeable. The blood flow is assumed to be unsteady laminar flow. The viscosity and the density of blood are $\mu = 4.24 \times 10^{-3}$ Pa s and $\rho = 1060$ kg/m³ respectively. The period of the cardiac cycle is $T = 0.8$ s assuming a heart rate of 75 beats per minute. The peak entrance velocity of the coronary artery is $v_{\max} = 0.17185$ m/s. The maximum Reynolds number in the whole cardiac cycle is $R_e = \frac{d v_{\max} \rho}{\mu} = 111.7$ based on the inlet velocity and the diameter of coronary artery. The Womersley number is $\alpha = \frac{d}{2} \sqrt{\frac{2\pi\rho}{T\mu}} = 1.82$ based on the host coronary artery diameter.

For the comparative convenience, the boundary conditions in these two models are identical for the simulation. The boundary conditions for the simulation are:

1. At the inlet sections of coronary and graft, longitudinal velocities with blunt or flat profiles are imposed according to the physiological flowrates measured in a human coronary bypass a few days after CABG surgery [7]. The extended graft and coronary are allowed for the flat inlet velocity profiles to fully develop.

Figure 2. Entrance velocity waveforms of coronary and graft. Labels t_1 to t_3 indicate times at which the results are plotted in subsequent figures.

Given that the 2-way model has the same total graft cross-sectional area and flow rate as the 1-way model, the longitudinal entrance velocities at the coronary and grafts in the 2-way model are identical with those of the 1-way model. The radial and tangential components of the velocity are zero at the entrance. Figure 2 plots the longitudinal velocities in which t_1 , t_2 and t_3 are the three points for which the result will be shown. Here, t_1 represents the acceleration phase, t_2 the peak value of graft flow rate, and t_3 the deceleration phase.

2. Zero velocity is assigned on all walls corresponding to no-slip flow conditions.
3. At the outlet section of the coronary conduit, a constant reference pressure is given as $P = 0$.
4. Because of the planar symmetry of the bypass graft model, half of the 1-way model and a quarter of the 2-way model are taken as the computational domain in order to reduce the computational consumption. On the plane of symmetry, normal velocity components are zero.

There is a problem that must be considered when meshing the geometric model: the mesh size must be much finer in the boundary layer near the vessel wall in order to improve the precision of the calculations. For this purpose, the meshing procedure should be considered when constructing the geometric model. Several volume domains are glued together to compose the whole model. Different meshing schemes are employed in different directions or domains. Non-uniform grids are employed in the radial direction near the wall (Figure 3). Almost uniform grids are employed in the radial direction near the centerline and in the longitudinal and circumferential directions.

The physiological blood flows through the 1-way and 2-way CABGs are simulated using the CFD module of the commercial package ANSYS6.0. ANSYS6.0 is capable of computing the transient blood flow patterns through the vessels and the shear stresses on the vessel wall. The time step for transient analysis is 8×10^{-3} s. The convergence precision of the velocity field and the pressure field at each time

Figure 3. Meshes in the cross-section of vessels.

step is limited at 10^{-6} during the iteration. Periodic steady state solutions are reached at the third cardiac cycle.

In our series of studies, we have tested several cases with different grid densities to assess mesh independence. We found that the grid density near the vessel wall has great influence on the precision and convergence of the simulation. With the special strategy of applying gradually finer grids near the wall, we can improve the calculation efficiency significantly in our studies. We have also verified our calculations based on the above-mentioned mesh generation and numerical method using previous works [19, 41–43], and have confirmed that the numerical approach used is valid for the simulation of hemodynamics in the CABGs.

4. Results and Analysis

The numerical simulations of physiological blood flows in the 1-way and 2-way models are performed primarily to analyze the difference between the physical variables of the flow field in these two models. The demonstrated results were obtained from the third cycle of calculation. Numerical results shown here include flow patterns and WSS distributions. We will focus especially on the hemodynamics in the vicinity of distal anastomosis because disease is prone to develop in this region and its aftereffects are more critical.

4.1. FLOW PATTERNS

Velocity vector distributions in the plane of symmetry at different times are shown in Figure 4. The flow patterns do not vary extremely during the whole cardiac cycle. From the viewpoint of time, the flow disturbance is weak during the acceleration phase, and the flow patterns in the downstream area of the toe are quite similar in both models (Figure 4a). At the time of peak graft flow rate and during the deceleration

Figure 4. Distributions of velocity vectors in the plane of symmetry at different times. Upper – 1-way model; Lower – 2-way model.

phase, the flow disturbances in the two models are different (Figure 4b and c). The blood flow from the graft in the 1-way model skews to the floor and affects the flow patterns both near the toe and on the floor approaching the stenosis. There is a large stagnation region on the floor of 1-way model, which will create high pressure on the floor near the stenosis. Thus the floor downstream from the stenosis has a high potential for the stagnation of flow.

In both models, the areas downstream from the junction are occupied by positive longitudinal velocities throughout the whole cycle. From the viewpoint of space, the blood flow velocity in the coronary artery increases at the toe section because of the join of the inflow graft, and the velocities near the coronary wall are much larger, which consequently weakens the reversed flow at the distal junction. The residual flow issued from the stenosed artery creates a confined jet flow. The jet flow accelerates the blood flow at the stenosis, shortens the residence time of blood in the area between stenosis and anastomosis, and alleviates the development of restenosis.

There is a reversed flow between the stenosis and the heel, whereas no reversed flow exists at the toe and its downstream region. Just as the situation in the Taylor patch model [27], since the cross-sectional area of the bypass graft gradually becomes larger at the toe section, the momentum of blood flow in the graft becomes smaller and the impact of the graft blood flow on the coronary wall opposite the anastomosis is decreased. In particular, the blood flows from the 2-way bypass graft are concentrated on the center of the coronary conduit, which weakens reversed flow near the stenosis and the strong strike on the arterial wall, and strengthens the jet flow from the host coronary artery. This also has positive effects on the hemodynamics of the CABG.

The distribution of the main flow will affect the magnitude and profile of the WSS at locations where intimal thickening occurs. High velocity fluctuations indicate

Figure 5. Secondary flow in the cross-section of coronary artery $2d$ downstream from the toe at different times. Left – 1-way model; Right – 2-way model.

flow disturbances that should be avoided. This will also be confirmed by the analysis of WSS distributions.

The distributions of secondary flow in the cross-section of the coronary artery $2d$ downstream from the toe at different times are depicted in Figure 5. From the spatial point of view, the secondary flow induced by the skewed graft flow is still strong in the region downstream from the toe. From the temporal point of view, the strength of secondary flow varies proportionally with the entrance flow rate. The secondary flow in the acceleration phase is weaker than that in the deceleration phases, and reaches its maximum at the time of peak flow rate. The vortex center moves upward during the acceleration phase and downward during the deceleration phase (upper half lumen for 2-way model). Although the 2-way model exhibits stronger secondary flows than the 1-way model, there is no significant difference of eddies and vortices between both models. The 2-way configuration makes the flow pattern in a cross-section symmetrical and uniform, which may also affect the distribution of pressure and WSS acting on vessel wall in the circumferential direction.

4.2. WALL SHEAR STRESS

WSS is one of the most crucial hemodynamic variables associated with intimal thickening. Endothelial cells in blood vessels are exposed to blood flow and thus fluid shear stress. The adhesion of blood cells, the hyperplasia of endothelium cells, the thickening of the blood vessel wall, and the restenosis of the conduit are closely related to local hemodynamic WSS [11, 16–18].

It is obvious that there are stress concentrations at the toe in the both models (Figure 6). Without taking the stress concentration at the toe into account, the WSSs in the 2-way model are larger and more uniform than those in the 1-way model. Especially downstream from the toe, there is little variation in WSS magnitudes, so the hemodynamics in that area is stable. The WSSs along line b in the 1-way model vary more violently than those along line a in the 1-way model and line c in

Figure 6. Distributions of WSS along lines a, b and c at different times.

the 2-way model. Thus, the interaction between the blood flow and vessel lumen changes much more on the floor in the 1-way model.

The distribution of WSS can reflect the flow patterns in the boundary layer. The variation of WSS along the floor of the 1-way model illustrates flow oscillation in that region. For example, the flow stagnation in the 1-way model can be distinguished in Figure 6 by the changes of WSSs from negative to positive. The uniform values of WSS in the 2-way model indicate that there are fewer fluctuations of flow. This is in accordance with the aforementioned flow patterns.

Figures 7 and 8 show respectively the distributions of longitudinal and circumferential WSSs along the vessel wall downstream from the toe at different times. The magnitudes of longitudinal and circumferential WSSs in both models vary with the entrance velocities. Although the boundary conditions at the entrance are identical for both models, the WSS distributions at the distal anastomosis are quite different between these two models. The most obvious variations of WSSs occur in the vicinity of the toe, while the far downstream of the toe features uniform WSS distributions. This phenomenon indicates that the anastomosis configuration can influence the hemodynamics only in the vicinity of anastomosis, but not in the far downstream area. The symmetrically implanted 2-way bypass configuration significantly influences the hemodynamics in the distal anastomosis. Because the blood

Figure 7. Distributions of longitudinal WSS along the vessel wall downstream from the toe at different times. Upper: 1-way model, $\theta = 0^\circ$ denotes the toe, $\theta = 180^\circ$ denotes the floor; Lower: 2-way model, $\theta = 0^\circ$ and $\theta = 180^\circ$ denote the two toes respectively. $z = 0$ denotes the toe section.

flows in the 2-way model are concentrated in the horizontal plane, the enhanced flows in this plane generate significantly increased longitudinal and circumferential WSSs. These increased WSSs can scour the IH and thrombus from the distal anastomosis, and may improve the long patency rates of the CABGs. However, the

Figure 8. Distributions of circumferential WSS along the vessel wall downstream from the toe at different times. Upper: 1-way model, $\theta = 0^\circ$ denotes the toe, $\theta = 180^\circ$ denotes the floor; Lower: 2-way model, $\theta = 0^\circ$ and $\theta = 180^\circ$ denote the two toes respectively. $z = 0$ denotes the toe section.

longitudinal and circumferential WSSs vary on a larger scale in the 2-way model than in the 1-way model, so the WSSGs in the longitudinal and circumferential directions in the 2-way model are larger than those in the 1-way model. This feature compromises the advantages of 2-way model to some extent if WSSGs have a significant influence on the patency rate.

Table I. Mean WSS at different positions

1-way model	A ₁	B ₁	C ₁	D ₁	E ₁	F ₁	G ₁	H ₁
Mean WSS	0.985	1.045	2.685	1.789	9.443	7.613	3.710	5.815
2-way model	A ₂	B ₂	C ₂	D ₂	E ₂	F ₂	G ₂	H ₂
Mean WSS	1.384	1.435	3.145	5.46	10.474	10.474	4.207	4.207

It has been shown that a low mean WSS tends to promote IH while a high WSS inhibits the process [3, 16, 19, 23]. WSS on the graft wall is probably the mechanical factor and an important trigger that induces the adhesion of circulating cells onto the luminal wall. The mean values of WSSs at different positions are listed in Table I. Generally speaking, the 2-way model has greater mean WSSs than the 1-way model not only at the toe but also in the graft conduits. Higher mean WSSs in the 2-way model can decrease the residence time of blood particles in the lumen and weaken the adhesion of blood particles onto the vessel intima. Thus, the 2-way model is advantageous over the 1-way model with larger mean WSSs in the bypass graft and in the host artery.

The variation of WSS over the cardiac cycle can be elucidated with the temporal WSSG. Figure 9 depicts the temporal distributions of WSS and WSSG at points G₁, H₁ and G₂(H₂). From the viewpoint of time, WSS is influenced predominantly by the flow rate of grafts, and increases or decreases with the magnitude of entrance velocity of grafts during the whole cardiac cycle (Figure 9a). The most violent changes of WSSs occur during the systolic phase, and the WSSGs are large in this period (Figure 9b).

In order to assess the variation of oscillating hemodynamic WSSs with time, here we introduce an additional index of cyclic shear stress ratio from structure mechanics. The most commonly used shear stress ratio is the ratio of minimum stress to maximum stress

$$\gamma = \pm \frac{\tau_{\min}}{\tau_{\max}} \quad (-1 \leq \gamma \leq 1) \quad (5)$$

Figure 9. Temporal distributions of WSS and WSSG at points G₁, H₁ and G₂(H₂).

Table II. Shear stress ratios (γ) at different positions

1-way model	A ₁	B ₁	C ₁	D ₁	E ₁	F ₁	G ₁	H ₁
γ	0.2356	0.2305	0.2165	0.1526	0.1975	0.1919	0.2761	0.2158
2-way model	A ₂	B ₂	C ₂	D ₂	E ₂	F ₂	G ₂	H ₂
γ	0.2619	0.2575	0.2337	0.0675	0.1839	0.1839	0.2710	0.2710

where γ is the shear stress ratio, $\tau_{\min}$ and $\tau_{\max}$ denote the minimum and maximum absolute values of WSS respectively. The sign of γ is determined by the original signs of $\tau_{\min}$ and $\tau_{\max}$. For partially reversed stress, γ is a negative quantity between 0 and -1 . For fully reversed stress, $\gamma = -1$. Fully reversed stress means the type of stress cycle has an amplitude symmetric about the abscissa of time; the maximum and minimum stresses are equal, but opposite in sign. $\gamma = 1$ means the shear stress is steady, which is the most preferable case.

Table II lists shear stress ratios at different positions. In the present study, there are no negative values of γ at the selected positions since all the WSSs are positive. The magnitudes of γ in the grafts are larger in the 2-way model than in the 1-way model. Meanwhile, the magnitudes of γ in the artery are smaller in the 2-way model than in the 1-way model. With regard to γ , the 2-way model also has better hemodynamic performance in the grafts.

5. Discussion

We are convinced that geometric modification can improve the medium-and-long term curative effects of coronary bypassing procedures. We present here a novel concept of 2-way CABG. The concept stems from the fact that the asymmetric geometry of the conventional 1-way bypass graft creates disturbed flow in the vicinity of the anastomosis. We suppose that a symmetrically implanted 2-way bypass graft may improve the hemodynamics of CABG and eventually increase the medium-and-long term patency rates by symmetrizing the flow patterns and increasing the flow velocities and hence the WSSs. Additionally, the 2-way bypass model can provide one more graft. If one of the grafts is occluded and another remains patent, then the post-operative effects for the patients can be improved. As a first step, we simulated the physiological blood flows in both the conventional 1-way model and the novel 2-way model of CABG, to investigate how the symmetric 2-way bypass grafts affect the flow patterns and WSS distributions.

The symmetric and relatively uniform flow patterns in the 2-way bypass graft showed that a symmetrical geometric anastomosis configuration can improve the hemodynamics in CABGs. It is widely recognized that reversed flow and large flow separation zones may promote the development of atherosclerosis [44–47] because the particles in blood are allowed to adhere to endothelial cells within the recirculation zone. At the anastomosis, disturbed flow causes

zones of recirculation and stagnation in the 1-way model. However, the 2-way model features larger and more uniform longitudinal velocity and less reversed flow. The strong sustained longitudinal velocity at the distal downstream is favorable for accelerating the blood flow, weakening the reversed flow at the distal junction and shortening the residence time of blood particles at the junction region.

Regarding the distributions of WSS, the 2-way model exhibits more favorable hemodynamics. It is shown that large sustained WSSG indicates injury to the endothelium and rapid recurrence of IH plus atheroma [48]. If the WSSG can be decreased considerably, the non-uniformity of the flow field can be alleviated and the probability of IH and restenosis can be reduced. At the anastomosis, disturbed flow induces both spatial and temporal gradients of WSS. Such gradients have been linked to the generation of atherosclerotic plaques. In the 1-way bypass graft, WSSs vary violently along the floor, generating large WSSGs. In the 2-way bypass graft, the floor does not exist any more, so there are no large WSSGs in that area, and consequently IH in that “danger zone” is eliminated.

Violently changing WSS and large WSSG exist along the suture-line, so it is still the dominant zone for IH. The area downstream from the toe exhibits uniform hemodynamics with uniform flow patterns and WSS distributions, especially in the 2-way model, which is quite beneficial for eliminating IH and restenosis. The “sharp” edges and corners along the suture-line cause stress concentrations and large WSSGs. We can eliminate or reduce these stress raisers by streamlining the models [46]. In actual *in vivo* CABG surgery, the “sharp” suture-line may become modest because of the compliance of vessels. Therefore the WSS along the suture-line in the *in vivo* CABG may not vary as violently as that of the numerical simulation, and the actual WSS concentrations and WSSGs may also be much smaller. Nevertheless, it is better to use a “rounded toe” model which more correctly approximates physiological reality. In general, the 2-way model demonstrates larger and more uniform WSS, and hence significantly reduced local WSSGs, which should consequently improve the hemodynamics.

For the first time, we introduced shear stress ratio as an assessment criterion for oscillating WSSs in CABGs. We put forward the shear stress ratio as an important coefficient for physiological blood flow in CABGs, to quantify the variations of WSSs over a cardiac cycle and study their correlation with hemodynamics. A small shear stress ratio means violent WSS variations which may have negative effects on CABGs. The present simulation results indicate that the 2-way model has obvious advantages over the 1-way model in the distributions of shear stress ratio in the grafts. Meanwhile, the large mean WSSs in the 2-way grafts have positive effects for the hemodynamics within the graft conduits.

The graft material properties significantly affect the WSSs along the suture-line. Synthetic and autologous vessels are commonly used for grafts. The most common prosthetic substitutes are Dacron or PTFE, which have the advantage of convenience and guaranteed availability. The compliance of prosthetic material

with the coronary artery is much worse than that of autologous vessels [27, 31, 49]. Use of graft material with mechanical properties more closely matched to those of the normal artery wall may result in a reduction of negative flow effects. A compliant autologous (arterial or venous) graft material is preferred in CABG surgery if available.

It is well known that the initiation and advancement of IH and restenosis result from complex biological and mechanical multi-factors. Factors associated with surgical injury, compliance mismatch and the associated intramural stresses, as well as NWRT parameters have not been rigorously addressed in this study. We presented herein with the CFD simulation only the hypothetical hemodynamic parameters responsible for the development of IH and graft failure. As evident in this study, geometric variations of the graft-to-artery junction area significantly influence the hemodynamics in the vicinity of distal anastomoses. The enhanced flow velocities and WSSs in the 2-way model can improve the hemodynamics of CABG. On the basis of these phenomena, we infer that our results coincide with those of Longest *et al.* [11–15]. The achievements obtained by Longest *et al.* make it possible to assess the particle hemodynamics in the 2-way CABGs. Ongoing studies of the 2-way CABG should include the analysis of NWRT-related parameters to realize a comprehensive evaluation.

The smaller grafts used in the 2-way model might be more prone to rapid thrombosis, clotting, and short-term graft failure. Binns *et al.* concluded that smaller grafts typically present an increased risk of early graft failure due to thrombosis [50]. Abbott *et al.* found that larger caliber grafts were typically associated with higher graft survival rates for above-the-knee femoropopliteal bypass applications [51]. Large graft exhibits better and more regular hemodynamic phenomena and may be effective in decreasing the probability of the initiation and development of postoperative IH and restenosis [52]. However, beyond a certain limit, graft diameter has no effect on improving CABG hemodynamics. Results of Longest *et al.*'s study indicated a significant increase in the potential for IH occurrence for longer grafts (graft-artery diameter ratio 2:1) in comparison to smaller graft (graft-artery diameter ratio 1.5:1), particularly in regions critical to graft function [11]. Thus, appropriately large grafts are applicable in the clinical practice of 2-way CABG operations. Sanders *et al.* [53] studied the influence of graft caliber on patency rate in a canine anastomotic model and concluded that graft diameter should be approximately equal to or a little greater than the recipient vessel (graft-artery diameter ratio 1.5:1). In this study, the graft-artery diameter ratio for the 1-way model is 1.46:1. The 2-way model reserved the identical cross-section area as the 1-way model and increased the WSS in the grafts with the graft-artery diameter ratio of 1.03:1. According to the previous work [52], 2-way bypass grafts with slightly larger diameter versus the host artery are preferable for CABG surgery. The controversial features between small and large grafts raise the question of the optimal graft diameter in clinical operations. Further investigations on the influence of graft diameter reduction might be necessary.

There are some simplified assumptions in the present study. First, we have used idealized blood vessel models. Idealized models are advantageous in analyzing the influence of some special geometric parameters on the hemodynamics. The present study focuses on the comparisons between 1-way and 2-way bypass models with similar geometries. Further endeavors should consider realistic patient-specific geometry models based on anatomical data. Second, the blood is assumed to obey Newtonian rheology and the vessel walls are simplified as rigid tubes without considering the deformation of vessels. Buchanan *et al.* [9] and Longest *et al.* [11, 13] suggest that non-Newtonian blood viscosity is an important consideration which can significantly affect hemodynamic conditions within critical low shear regions and during periods of flow reversal, and thus should not be neglected. In contrast, Ballyk *et al.* [54] have shown that non-Newtonian blood rheology has no significant effect on unsteady flow WSS. Future models should include non-Newtonian properties to illustrate more realistic hemodynamics. Previous research confirms that the distensibility of vessel walls does not greatly affect the hemodynamics. These assumptions cause errors in the calculation of WSS. WSS is smaller in models of compliant grafts and compliant arteries than in rigid wall models, but the differences are generally less than 25% in magnitude [33, 44], so the rigid vessel models are still widely used. The numerical simulation of the present study still has meaning for guiding research work and medical treatment. Further studies will take into account all the above-mentioned factors so as to carry out more precise and realistic numerical simulations.

Naturally, any innovative surgical technique needs a large number of animal experiments to verify its practical viability. This is certainly the case for the 2-way CABG operation. It is necessary to investigate the technical details of 2-way bypass grafts with the animal experiments henceforth, and to observe the long-term hemodynamic and biophysiologic variations in the grafts and the bypassed coronary artery as well as the variations of cardiac functions. The research significance of animal experiments comparing the 1-way and 2-way bypass grafts is to obtain deeper insights into the role which the hemodynamic abnormality plays in the development of IH and restenosis. Besides feasibility verification with animal experiments, the main problem encountered for the 2-way bypass graft is the operation technique: the operation difficulty is enhanced by suturing two hemodynamically satisfied grafts on one piece of coronary artery. Furthermore, bypassing with 2 grafts means that more graft materials are needed and more injury is done to the coronary artery. A normal vein (or artery) is not always available because of previous use (previous lower extremity or coronary bypass), thrombosis, or biological variability. The connection of the graft to the vessel needs space and the connection of two grafts needs twice as much space. However, the coronary vessel is attached to the myocardium and only a part of the vessel circumference may be used in surgery. There are only a few locations in the coronary tree which may be used for the 2-way bypass procedure. This situation compromises the application of the 2-way CABG. Problems with insufficient graft length, prolonged operative time

and further incisions may also make each of these options less desirable. Whether the improved hemodynamics achieved by 2-way CABG can guarantee reduced IH formations is unclear, thus clinical evaluation is warranted for this new approach. In any event, the assumption of the novel 2-way bypass grafts provides a new potential solution for improving the hemodynamics of the CABG. The data presented here provide an initial basis for comparison with other experiments, and may be used as initial test simulations for more physiological models.

In summary, we have presented a novel configuration for CABG with a symmetrically implanted 2-way bypass graft which features uniform longitudinal flow patterns and WSS distributions. The troublesome floor in the 1-way model is eliminated in the 2-way model, and WSSG in the 2-way model can be decreased significantly. It can be concluded that CABGs with 2-way bypass grafts should demonstrate more reasonable hemodynamics which would, at least in theory, reduce the risk of IH development and restenosis, and consequently generate better surgical effectiveness for the CABG operation. Despite the advanced hemodynamic features of the approach presented here, uncertainties about anastomatic wall behavior limit its application in clinics because anastomatic IH and restenosis are not only hemodynamic issues, but also complex biological events. Continuing studies on the development of IH and the roles of WSS, WSSG, particle-wall interactions, and other parameters can be later used to interpret the present results. Before such CFD analyses can be accepted at face value for use in basic or clinical studies, issues related to their validity and the underlying modeling assumptions must be considered. Concerning the limitations of 2-way bypass grafts, it is necessary to perform animal experiments to verify the viability of this novel idea for the CABG.

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