Estimation and Comparison of T Graft Versus Conventional Graft for Coronary Arteries

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Abstract: Coronary artery disease (CAD) distribution becomes more abundant in the developing world. Coronary Artery Bypass Graft (CABG) is one of the treatments. If there are three blocked coronary arteries, surgeons can use two methods to perform CABG: Conventional graft and T graft. Estimation and comparison of these two methods for LAD, CX & PDA is the aim of this study. First two bypass graft methods were modeled in NX software then simulated with ADINA software. Fluid Solid Interaction (FSI) was considered for simulation with more realistic condition. Pulsatile pressure was used for simulation as well. As will be explained later, it is an appropriate boundary condition for stenosis properties. From the results, coronary arteries flow in T graft bypass is less than that in conventional graft by the factor of 0.2 to 0.5. Then T graft bypass does not provide sufficient flow for PDA. In addition, vortex and wall shear stress (WSS) show more possibility of T graft blockage. Although it is estimated from effective stress that failure in artery may be less in T graft bypass.

Key words: Atherosclerosis • T-shaped Bypass Graft • FSI • Pulsatile flow

INTRODUCTION

Atherosclerosis involves the procedure that leads to most of the diseases distinguished by thickening of artery walls [1]. Atherosclerosis causes many deaths from heart attack and stroke and accounts for nearly three fourths of all deaths from cardiovascular diseases [2]. Atherosclerosis can be the result of plaque deposition in inner wall and then causes blood flow to decrease [3]. When this blockage occurs in coronary arteries, as a result, heart stroke will happen. Coronary artery bypass graft (CABG) is one of the remedies for atherosclerosis. In fact, bypass graft brings blood to the parts beside stenosis. Both artery and vein are used for grafts.

This process can be performed in different ways. If there are many blocked coronary arteries, to increase the quality of bypass, surgeons use two main methods (Figure 1). Suppose three coronary arteries including left anterior descending (LAD), circumflex (CX) and posterior descending artery (PDA) have stenosis. Perfusion can be implemented in two ways. In the first method, conventional bypass graft, direct graft to each coroner, left internal mammary artery (LIMA) is connected to LAD and two pieces of saphenous vein (SV) are connected to CX and PDA. In the second method, T-shaped bypass graft; the graft is between LIMA and LAD. Then right internal mammary artery (RIMA) by a bifurcation from the middle of LIMA, connects to CX and PDA consequently. All of this composes a T-shaped graft. In some cases, the graft is attached to one of the bifurcations of the CX like marginal branches (M2).
Many surgeons believe that the use of T graft for revascularization of the anterior descending and circumflex arteries decreases the risk of death and there will be no need for reoperation or percutaneous angioplastic reintervention when compared with revascularization using conventional bypass with vein grafts [4, 5].

What is more, concerns have arisen for T graft CABG on (a) the potential steal phenomenon of the LIMA by the RIMA, (b) capacity of this arrangement to revascularize the coronary system completely including the right coronary artery (RCA) and (c) potential flow competition with the native coronary artery [6]. Also in [7], it is showed that the function of the RIMA used as a T graft was significantly improved when used on several branches of the circumflex artery or on a severely narrowed first circumflex. Grafting of the intermediate branch of a RCA has a negative prognostic influence on graft function.

In some studies the preference is related to stenosis, like [8] which demonstrates the composite T graft technique of bilateral internal thoracic artery (ITA) grafting should be reserved for patients with severe (70% or more) left anterior descending and circumflex arterial stenosis.

As can be seen, there are disagreements between surgeons about the process of performing open-heart surgery. Some of them believe that the conventional bypass graft is easier whereas some others believe that owing to the complexity of the surgery, using the third graft makes the T-shaped bypass graft simpler. They believe that an increase in connections to the aorta has its own problems as well.

Many of bypass grafts fail because of the formation of intimal hyperplasia. Here an important question is which methods are accompanied by less restenosis or atherosclerosis in graft arteries.

Why atherosclerotic lesions are brought on, cannot be explained by any of the clinical factors; nevertheless, it is shown that the hemodynamic factors can control the location of atherosclerosis within the arteries [9]. Therefore, investigating hemodynamic factors such as wall shear stress (WSS), flow velocity, pressure distribution, blood viscosity and vortical structures is important.

Layers of collagen compose the arterial wall. The strength of the wall is related to the strength of the collagen fiber and its orientation. Wall shear stress regulates the remodeling of endothelial cell via realignment and elongation [10].

Accordingly, wall shear stress is the most important hemodynamic factor. Investigation of [11] and [12] revealed the relation between vessel WSS and the rate of atherosclerosis. Their results showed that there was a significant correlation between low shear stress and an increased rate of atherosclerosis progression.

One study [13] showed WSS had an immediate influence on the endothelial and a 40 Pa shear stress is able to damage endothelial cells. From another study [14] it was revealed that higher WSS (about 100 Pa) could to remove the endothelial cells.

In a research [15] it has been shown that intimal hyperplasia occurs in regions of flow separation.

Another factor influencing the hemodynamic is vascular geometry [16].

From a structural point of view, wall tension is also referred to be among the factors causing graft failure, too [17]. Wall motion, stress and deformation of the vessels are considered to be important factors in atherosclerosis [18].

Hence, the comparison of the bypass method from hemodynamic factors and structural point of view is a major aim to manifest which method is highly suggested.

Although these factors can be obtained in vivo using magnetic resonance imaging (MRI) or color-flow Doppler ultrasound (CDU), there are some limitations to measurements because of the small dimensions of the arteries. Thus, the option to measure the related factors is to use computational fluid dynamics in order to simulate flow in bypass graft geometry then to compute and to compare the factors.

However, the computational simulation for blood flow in arteries confronts difficulties. Blood flow is pulsatile, the pressure and flow rate are dependent on time and the vessel wall is sort of elastic and in constant interaction with the affecting flow field. Because of these difficulties, there are appropriate assumptions in most previous studies according to their exclusive aims. Some proffered the rigid wall [12,19], some assumed flow to be laminar and some focused on steady flow with time dependent pressure conditions [20-22].

Simulations of blood flow in arteries with computational methods have limitations to idealized geometries with stenosis, curvature, bifurcations and graft. But the current trend is to use realistic geometries for better accuracy [23, 24]. In case of atherosclerosis with severely stenosed arteries, fluid–structure interactions (FSI) play an important role. The coupling between fluid (blood) and solid or structure (arterial wall) has significant
implications on atherosclerosis. A fully coupled FSI means that the reaction of the solid is strongly affected by the reaction of the fluid and vice versa.

Computational methods for problems with FSI have been used by some researchers. The ADINA team [25] developed a series of finite element procedures dealing with FSI where the fluid and solid models were solved iteratively for small strain and small deformation. Models with FSI were considered by [23, 24] to quantify pressure and other conditions under which critical flow conditions and artery compression may occur.

One study [26] showed that inlet velocity actually depends on the degree of stenosis and thus for severe constriction in coronary artery, a velocity-based boundary conditions cannot be realistic. They proved that regardless of severity of stenosis in coronary arteries, the upstream pressure and systemic pressure are always constant, thus, they should be used as boundary conditions instead.

The objective of this study is to realize how the flow and structural hallmarks are displayed when one of the two aforementioned bypass methods is proposed which has not been done yet from biomechanics point of view. We use a semi realistic geometry, FSI methods and pulsatile pressure boundary condition.

MATERIALS AND METHODS

Geometry and the Properties of the Model: In order to compare the two bypass methods, we assumed three full blocked coronaries grafted by two different geometry type of bypass. Three coronaries are LAD, CX and PDA. As mentioned in previous section, the depiction of two bypass methods is shown in Figure 1. Using images of [8] and the recommendations of our surgery consultant, the geometry model was generated by CAD software NX (version 7.5); as shown in Figure 2. Table 1 displays detailed geometric information while 0.5 mm was applied to the fluid-structure interfaces are the kinematic condition (or displacement compatibility):

\[ d_s = d_f \]  \hspace{1cm} (5)

and the dynamic condition (or traction equilibrium):

\[ \sigma_s = \tau_f \]  \hspace{1cm} (6)

where \( d_s \) and \( d_f \) are, respectively, the fluid and solid displacements and \( \tau_s \) and \( \tau_f \) are, respectively, the fluid and solid stresses. It is necessary to say that the values are defined only on the fluid-structure interfaces.
Table 1: Dimensional parameter of two models

<table>
<thead>
<tr>
<th>Bypass model type</th>
<th>Vessel</th>
<th>Diameter (mm)</th>
<th>Length (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>T shaped</td>
<td>LIMA</td>
<td>4-2</td>
<td>18.6</td>
</tr>
<tr>
<td></td>
<td>RIMA to CX</td>
<td>4-2</td>
<td>9.6</td>
</tr>
<tr>
<td></td>
<td>RIMA to PDA</td>
<td>3-2</td>
<td>8.0</td>
</tr>
<tr>
<td></td>
<td>Coronary Artery</td>
<td>2</td>
<td>1.5</td>
</tr>
<tr>
<td>Conventional</td>
<td>LIMA</td>
<td>4-2</td>
<td>18.6</td>
</tr>
<tr>
<td></td>
<td>SV to CX</td>
<td>4</td>
<td>10.7</td>
</tr>
<tr>
<td></td>
<td>SV to PDA</td>
<td>4</td>
<td>9.4</td>
</tr>
<tr>
<td></td>
<td>Coronary Artery</td>
<td>2</td>
<td>1.5</td>
</tr>
</tbody>
</table>

Fig. 3: The pressure which is used in inlet boundary

The fluid velocity condition is resulted from the kinematic condition when a no-slip condition is applied:

\[ \vec{v} = \overrightarrow{d_s} \]  \hspace{1cm} (7)

The fluid and solid models are coupled as follows. The fluid nodal positions on the fluid-structure interfaces are determined by the kinematic conditions. The displacements of the other fluid nodes are determined automatically by the program to preserve the initial mesh quality. The governing equations of fluid flow in their ALE formulations are then solved. In steady-state analyses, the meshes velocities are always set to zero even the fluid nodal displacements are updated. Accordingly, the fluid velocities on the fluid-structure interfaces are zero [25].

**Boundary Conditions and Numerical Method:**

Physiological pressure of human coronary arteries were obtained from available literature and used in the model [32]. Details are given as follows. To avoid divergence of the solution, the inlet pressure is raised from zero to aortic pressure within 1 second in 10 intervals. Then the pressure is considered constant for 5 intervals of 0.01 second to reduce transient error. The final solution is done in 95 intervals of 0.01 second and repeated for three periods as shown in Figure 3. At the outlet, flow satisfies zero pressure flow conditions.

First and end of vessel are fully constrained. At the interfaces of the blood flow on the arterial wall, no-slip conditions are imposed. Large displacement and small strain assumption are applied to the model.

The governing equations were solved using finite element commercial computational fluid dynamic software ADINA (version 8.7, ADINA, Watertown, MA) (Adina R&D, Inc., 2010). Direct computing of two-way coupling solution method with applied 0.01 relative relaxing force and displacement/velocity tolerance is used.
Table 2: Number of elements and the density of the meshing

<table>
<thead>
<tr>
<th>Model</th>
<th>Conventional Bypass</th>
<th>T shaped Bypass</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Domain</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Solid (Element/mm)</td>
<td>Fluid</td>
</tr>
<tr>
<td></td>
<td>0.4</td>
<td>0.35</td>
</tr>
<tr>
<td></td>
<td>194478</td>
<td>371298</td>
</tr>
<tr>
<td></td>
<td>Element No.</td>
<td></td>
</tr>
<tr>
<td></td>
<td>168889</td>
<td>292249</td>
</tr>
</tbody>
</table>

Equation solver is sparse. For the iterative algorithm in all simulations, the Newton method is chosen. Maximum iteration repeated 50 times and maximum error was measured to be 0.001.

The field of fluid and solid was meshed by first-order tetrahedral elements. Meshing was composed of free-form meshing algorithm. Free-form mesh generation was based on Delaunay scheme.

Simulation results were supposed to be independent of the computational mesh when the difference between meshes of varying densities was less than 5%. To select the mesh density, the graft to LAD in the conventional model bypass was used and the results were applied to the entire model. The effective stress and maximum velocity parameters were considered to be the convergence scale for the solid and fluid respectively. Table 2 shows the detailed information about meshing.

**RESULTS AND DISCUSSION**

Although the main difference between these two types of bypass is in their geometry and mechanical properties of the wall, in this research, they are compared from the geometry point of view.

To compare the conventional bypass with T-shaped bypass, from the view point of hemodynamic fluid, different parameters can be studied. For the fluid medium, flow, velocity, circulation area, wall shear stress and pressure distribution are important, while for the solid medium, the displacement and the strain of the wall and the effective stress are important.

Since the maximum inlet pressure occurs at T=3.38 second, the figures and graphs presented here, belong to this time.

**Flow:** The main objective of this study is to assess the adequacy of T bypass in comparison to conventional bypass. This could be the most important parameter, as surgeons disagreements are mostly related to flow.

By comparison of the flows, it is clear that the T-shaped bypass has lower flow which is considerable in PDA artery as shown in Figure 4. The flow rates are about 50%, 50% and 25% of normal values for LAD, CX and PDA respectively. Referring to merely these parameters, T-shaped bypass can be considered inadequate.

For the velocities associated with this flow rates, Reynolds number is about 100 to 600. In regards with the above mentioned Reynolds numbers, the assumption of laminar flow for the vessel is reasonable.

**Velocity Field and Recirculation Zone:** It is observed that in the T-shaped bypass graft, average and maximum velocities are smaller than the conventional bypass graft. Therefore, in vortex and rotational flow, the conventional bypass graft is less likely to develop plaques in this area.

Although the velocity is lower in T-shaped bypass graft, the circulation zone is higher and more often has no specific order.

In the junction between LIMA and RIMA in T-shaped graft, a reversal flow occurs which is similar to bifurcations in the cardiovascular system. The velocity magnitudes in this point are zero, but velocity gradient does exist. After the bifurcation, vortices occur and the velocity vectors are negative alongside the wall, as shown in Figure 5.

The vortex after stenosis viewpoint puts the sufficiency of "T-shaped bypass graft" under the question, which again shows a weak point of this graft.

**Wall Shear Stress:** Comparing the figures and their results shows that the average shear stress in conventional bypass graft is more than that in T-shaped bypass graft.

The maximum shear stress in conventional bypass graft is up to 46 Pa (Figure 6), which is slightly higher than damage threshold but does not happen in many points, while the minimum shear stress in T-shaped bypass graft, happens in more points.

The investigation of T-shaped bypass graft shows that when the maximum shear stress is at least 17 Pa (when the pressure inlet is at its minimum amount); more points with shear stress less than 0.5 Pa could be observed. However, there are the same points in conventional bypass graft model too, especially after the stenosis, but as there are plenty of such points in the T-shaped bypass graft models, the possibility of having higher restenosis is much more.

The proximity of obtained range with the results of [33] confirms that the answer is close to the actual amount.
Fig. 4: Outflow of LAD in both methods

Fig. 5: The circulation zone in the T-shaped graft after bifurcation

Fig. 6: Wall shear stress in CX artery from conventional graft
Pressure: The difference in values of LAD arterial pressure is visible. For example, maximum pressure for LAD in conventional and T-shaped graft are 17324 Pa and 4093 Pa, respectively. For graft vessel with bigger diameter, the pressure gradient in coronary graft decreases and in long-term creates a severe wall shear stress in coronary and results hyperplasia. Due to 4 mm saphenous diameter, this effect appears more likely in conventional bypass graft. As the pressure wave propagation velocity is depended on the strength and diameter of a vessel, using a narrower and more strength vessel, could lead to better results.

These are the main advantages of T-shaped bypass graft. In addition to further strengthening of arteries, artery to artery graft is more useful than artery to vein graft.

Wall Displacement and Strain: The maximum displacement to vessel length ratio in each model is between 1 to 5 percents. The maximum value belongs to T-shaped bypass graft. Thus, the assumption of large displacements in this problem is necessary and appropriate. In T-shaped bypass graft model due to more freedom, the vessel is displaced about 12 mm, from its original position.

The higher wall strain causes the reaction of living tissue and thickening of the wall which brings about new problems. The fluctuating pressure is the main factor that destroys a transplant and using a high strength vessel is necessary for success in a long-term transplant.

In normal situation, the major arteries expand about 5% in a cycle and for the veins due to their lower compliance, this amount exceeds to about 10 to 15 percent. In this bypass model, 10% strain and pulsation flow of coronary shows the necessity of problem solving with regard to fluid-solid interactions. As the pressure at the inlet is considered to be aortic pressure and the displacement of vessel is dependent on the elastic modulus, the elastic modulus of coronary artery and saphenous vein are appropriate.
At the junction of two bypass grafts in conventional model, namely CX and PDA, there is a great strain as can be seen in Figure 7. When the cycles are numerous, this problem can lead to rupture and inefficiency of the graft. As in conventional bypass model more pressure is applied on the vessel wall, it is followed by additional strain.

Effective Stress: One of the most important parameters is the stress influencing vessel wall or effective stress. Tip and heel of the graft are among the points that their tension must be investigated. In this point the grafted vessel is sutured to the host vessel and has high tension concentration then it is necessary to have enough strength.

The major causes of effective stress are the pressure applied to the wall and thickness of the wall area. Thus, because of higher pressure in conventional model, higher effective stress is reasonable. For example maximum effective stress for LAD in conventional model is 42808 Pa (Figure 8), while for T-shaped model is 21637 Pa. There is stress concentration at the tip and heel of the graft. The existence of sutures leads to higher stress concentration in the region and increases the possibility of rupture in this part of the bypass graft.

The results of the study [34] from stress point of view are in good agreement.

CONCLUSION

The summary of the study is presented in Table 3. The results and the proposals are from engineering point of view and it is necessary to add the physiological and physician point of view to it. Of course, this is surgeons who decide about the optimum model according to the characteristics of each patient's surgeon or surgical conditions.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Conventional Bypass</th>
<th>T shaped Bypass</th>
</tr>
</thead>
<tbody>
<tr>
<td>Flow</td>
<td>*</td>
<td></td>
</tr>
<tr>
<td>Vortex</td>
<td>*</td>
<td></td>
</tr>
<tr>
<td>Wall shear stress</td>
<td>*</td>
<td></td>
</tr>
<tr>
<td>Strain</td>
<td></td>
<td>*</td>
</tr>
<tr>
<td>Effective stress</td>
<td>*</td>
<td></td>
</tr>
<tr>
<td>Strength</td>
<td>*</td>
<td></td>
</tr>
</tbody>
</table>

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REFERENCES


