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Numerical and Physical Models of Distal Anastomosis in Coronary Artery Bypass Grafting

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***Abstract*—The progressive anastomotic intimal hyperplasia has been considered as a major failure reason for coronary artery bypass grafting (CABG), which is associated with the abnormality of the hemodynamic conditions. We designed a set of plexiglass models which can serve as in vitro cells or tissue culture bioreactors, also serve as mechanical flow chambers in a physical perfusion system of CABG. The models of distal anastomosis with different angles were made of plexiglass material. The anastomotic angles α in three models are 60°, 30° and 15°, respectively. The numerical simulation models, reproducing from these physical models, were set up using a computer fluid dynamic (CFD) software. First, the numerical results showed that in the model with α=60° there is a high velocity characteristic on the floor of the host artery opposite the graft orifice, where significant intimal thickening is likely to occur. Decreasing α, the peak velocity will move to the outlet of host artery. Second, the stagnant flow region depends directly on the distance D (The distance from the heel of the graft to the center of stenosed site), the larger is the value of D, and the longer is the stagnant region. We also found the residual flow which is harmful to the orifice region and the distal portion of anastomotic gradually became more slender as severity of the stenosis increased. The results suggest that anastomotic angle and stenosed site are very important factors for preoperative investigation. A smaller α can eliminate the stagnation and decrease the risk of intimal thickening at the anastomotic junction, while a shorter D can decrease stagnant region. The results also showed that our perfusion system and the physical models are feasible for CABG studies.**

Keywords-coronary artery bypass grafting; hemodynamic; numerical model; numerical stimulation; physical model; perfusion system

I. INTRODUCTION

Severe coronary artery disease with narrow lumen is one of the characteristics of atherosclerosis. Coronary artery bypass grafting (CABG) of the partially or fully stenosed arteries, using an autologous saphenous vein or internal mammary artery, is an effective treatment to restore circulation [1]. However, CABG has its complications; the saphenous vein graft only provides palliation of the ongoing process. Approximately 15–20% vein grafts occlude in the first year and 50% occlude within the first 2 years. The pathology in the grafts documented by angiography and histological examinations includes acute thrombosis and intimal hyperplasia (IH) during the first post-operative year and onset of progressive atherosclerosis beyond 3–5 years [2].

Hemodynamics parameters have a profound effect on graft failures due to restenosis that result from acute thrombosis and intimal hyperplasia. Acute thrombosis in the grafts came from the "disturbed flow" at a bypass system. The "disturbed flow" might trigger a cascade of abnormal biological processes leading to thrombi formation [3]. The term IH refers to an abnormal increase in the number of cells in a tissue or an organ. This abnormal thickening of the wall is found to occur principally at the distal anastomosis of a bypass system, especially at the heel and toe, at the junction between the graft and host artery, and on the artery floor opposite to the junction [4].

Many researchers have analyzed the complex flow patterns in the sapheno-coronary anastomotic region, using various numerical simulation models in an attempt to explain the site of preferential intimal hyperplasia based on the flow disturbances and differential wall stress distribution [5-14]. However, only a few of these studies have in vitro physical models to verify their numerical models [15-22]. The physical model of CABG is very valuable to assess the flow fields parameters practically. Moreover, the physical model of CABG would be more bilievable if it contains a segment of blood vessel or cells/tissue culture. In this case, the biological variation under in vitro flow field's environment could be better studied. For the model with a segment of blood vessel, it is neither possible to reproduce the exact shape of the computational model according to the physical model, nor possible to observe the biological variation inside the blood vessel [2]. In our knowledge, there is not such a physical model which has the capability of in vitro cells or tissue culture. Therefore, we designed a set of plexiglass models which can serve as in vitro cells or tissue culture bioreactors, also serve as mechanical flow chambers in a physical perfusion system of CABG.

Three physical models of CABG with different anastomosis angle were designed; each of them can change stenosed site and stenosed severity conveniently by fixing two pieces of silicone rubber. Cells or tissue culture can be placed on the trench inside the physical models. The models subsequently were connected with a perfusion system,

respectively. The numerical simulation models, reproducing from these physical models, were set up using computer fluid dynamic (CFD) software. Fluid dynamic parameters obtained from numerical simulation models were benefit to verify the functions of physical models of CABG. Our studies will provide some assistance for surgeons doing preoperative investigation of CABG.

II. METHODS

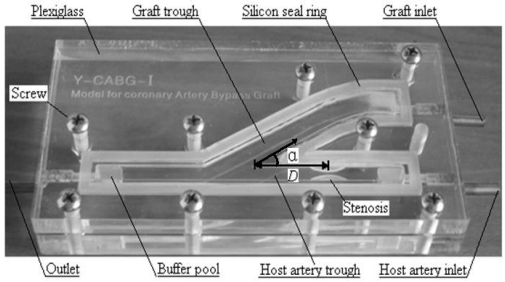
A. Physical Models

The models of distal anastomosis with different angles were made of plexiglass material, are shown in Fig.1. The "host artery" and the "graft" have inner widths of 6mm and 9mm, respectively. We defined α as the mean intersection angle between the axis of the graft and the horizontal axis of the host artery(We defined α as the mean intersection angle between the longitudinal axis of the graft and the longitudinal axis of the host artery), The angles of α in three models are 60 °, 30 ° and 15 °, respectively.

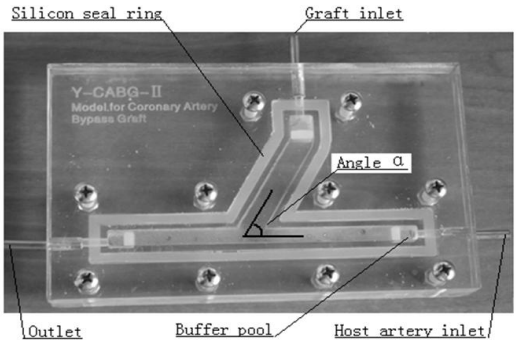
All of the models have two parts, top cover plate and bottom plate. On the surface of the bottom plate, the troughs were carved by an engraving machine under accurate computer control. The depth of all troughs is 2.5mm. The troughs along with the top cover plate consisted of the shapes of distal anastomosis. For preventing leaking from the troughs, a silicone seal ring is placed surrounding the troughs, and several stainless steel screws fix top plate with bottom plate tightly. The inlets and outlets of the models were shaped by drilling holes on the sides of the bottom plates. In order to connect soft tubes easily with the models, copper hollow tubes were inserted into the holes and fixed by adhesive glue. For alleviating turbulent flow near the inlet or outlet, buffer pools were carved with the depth of 7.5mm. The depth of buffer pools is designed by experience.

The stenosed site in host atery trough is made of two pieces of silicone rubber. Each piece is a semicircle plane, sticking to the trough surface by using silicone grease. The distance from the heel of the graft to the center of stenosed site were defined as D. D=1-3cm is the site commonly selected by the surgeon. In our models, the distance D was set at: 1cm, 2cm, and 3cm. Another parameter about the stenosis is stenosed severity (γ), which was calculated according to the cross section area ratio between normal host artery and stenosed site, and γ was chosed as: 70%, 80%, 90%, and 95%. By using two pieces of semicircle planes, which were cut according to the stenosed severity γ, the stenosed site can be flexibly placed on the proximal portion of host artery, see as Fig.1.

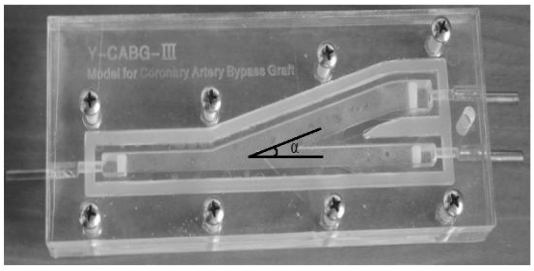
Cells or tissue culture can be placed on any interesting sites of the trough to accept the flow field variation. Usually, the flow field at anastomosis junction has obvious effect on intimal hyperplasia; therefore, it is a target site where the cells or tissue culture were placed. The cells were seeded on a coverslip, and then the coverslip is transferred and adhered to surface of the model trough. For tissue culture, i.e. a segment of blood vessel can adhere to the trough directly by using silicone grease.



(a) Bypass graft with 30 °angle



(b) Bypass graft with 60 °angle



(c) Bypass graft with 15 °angle

Figure 1. Design physical models of coronary artery bypass graft

B. A closed Loop Perfusion System

Fig.2 is a schematic presentation of the experimental arrangements for flow characteristic measurement. The fluid (i.e. cells culture medium) is forced from a tank (1) by a peristaltic pump (2) into the reservoir which has a overflow outlet (3). To maintain a constant static pressure in the model, excess fluid flows back into the tank (1). The fluid flows through the test model (6) and then pressure tank (8) before going back into the tank (1). The two valves (4) and (5) are used to control the inlet flow rate and resistance of the outlet, respectively. The pressure monitor measures the flow pressure of inlet and outlet respectively. When on the cells shear stress experiment, the coverslip seeded with cells was mount on the anasmotomic junction region of test model, showed as the enlarge windows (9) in Fig.2.

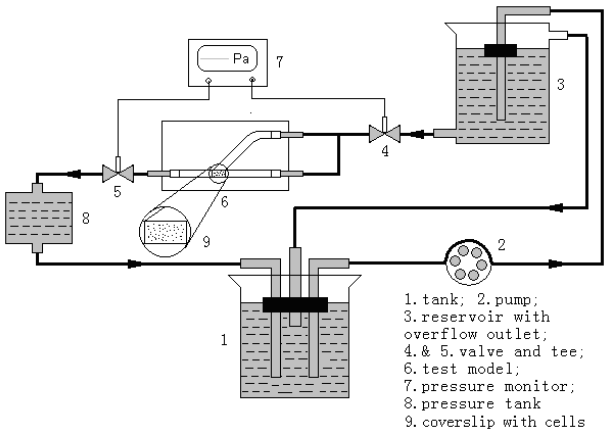


Figure 2. Schematic presentation of in vitro experimental arrangement

C. Numerical Simulation Method

The simulations of fluid dynamics were performed using a control-volume-based technique, implemented in the commercial CFD flow solver, FLUENT (Fluent Inc., USA). The computation procedure consists of (i) construction of the geometry using a pre-processor, Gambit, (ii) meshing the computation domain, (iii) assigning boundary conditions in terms of velocities and flow-rate weightings, (iv) assigning fluid properties, and (v) the solution algorithm.

The geometric dimensions of flow stimulation models were constructed according to the plexiglass test model. The elements employed to mesh the computational domain consisted primarily of regular structured hexahedral elements as well as wedge elements wherever necessary. All stimulation models were meshed with at least 65,936 cells.

In the solution algorithm used by Fluent, the governing equations are solved sequentially. Because the governing equations are non-linear (and coupled), several iterations of the solution loop need to be performed before a converging solution is obtained. By using a semi-implicit simple algorithm, the governing equations were solved. Time stepping employed a fully implicit scheme. A second-order upwind advection scheme was employed to ensure higher accuracy. The solutions of all the flow variables are deemed to have converged once their residuals computed from two successive iterations are below the set desired convergence criteria of 103. The Reynolds number, based on the host artery width and a reference blood viscosity of 0.0035kg m-1s-1, is 230. (The Reynodes number is a non-dimensional number used to characterize the flow.). Flow boundary conditions were arranged as follow: the pressure inlet condition at the host artery inlet and graft inlet with the same pressure value, the pressure outlet condition was stipulated at the host artery outlet. The no-slip condition was applied at all walls.

III. RESULTS AND DISCUSSION

A. The Effect of Anastomotic Angle

The numerical results showed that in the model with α=60° there is a high velocity characteristic on the floor of the host artery opposite the graft orifice, where significant intimal thickening is likely to occur. Decreasing α to 30°, the peak velocity appeared at the outlet of host artery, while the graft orifice did not have a very high velocity. Further decreasing α to 15°, however resulted in uniform velocity in the proximal segment of the host artery and the length of the entire graft, are shown in Fig. 3(a-f).

The α=60° model showed that a stagnation point happened near the toe of the graft orifice. The stagnation point was also thought to play a role in the disease progress. Hence, it was plausible that a smaller α could eliminate the stagnation and decrease the risk of intimal thickening at the stagnation site, which would threaten blood supply to the vasculatures downstream. In all of the models, the anastomotic junction area walls are suddenly subjected to high WSS; These WSS can dramatically change by varying the anastomotic angle of the graft. In α=30°, there is a peak WSS at the toe up to 11Pa, while in α=60° and α=15°, the peak WSS are 7Pa and 7.5Pa, respectively. It is valuable to study why the peak WSS in α=30° is higher than others, and one reason may be the shape of the anastomosis.

B. The Effect of The Distance from The Stenosis to Distal Anastomosis Junction

Fig. 3(g-h) and Fig.4 showed that the flow in the proximal portion of the anastomosis could be perfused from the graft in a retrograde fashion. Consequently, an extensive region of relatively stagnant flow was introduced when retrograde flows interacted with the postoperative residual flow through the stenosis. The stagnant flow region depends directly on the distance D, The more larger is the value of D, the more longer is the stagnant region. Although all of the boundary conditions are the same in the three numerical models in Fig. 3(g-h) and Fig. 4, we observed that the residual flows through the stenosis had different ways to shear the host artery wall, In Fig. 3(g), the residual flow sheared the floor of the host artery wall, while in Fig. 3(h) it sheared the ceiling. The residual flow and the flow through the graft finally combine at the anastomotic junction.

For the surgeon, the stenosed severity is one criteria in evaluating the necessity for coronary bypass surgery. Generally, the cut off value of the stenosed severity is taken at 80%. According to this knowledge, we also investigated the flow fashion in different severities of the stenosis as 70%, 80%, 90% and 95% (data are not shown in here). We found the residual flow gradually became more slender as severity of the stenosis increased. Nevertheless, the residual flow still plays a competing role with the graft flow. The flow competition is harmful to the orifice region and the distal portion of anastomotic [18].

C. Cells or Tissue Culture Inside The Physical Models

From the analysis of numerical simulation models, the cells coverslip and tissue culture have better to place on the anastomotic junction or near the stenosed site. Because of the higher WSS at anastomotic junction, the cells on the coverslip can not afford long time perfusion. In this case, we can close one of the valves in perfusion system (see as Fig. 2) and let the cells to rest. The medium in the perfusion

system should be changed every 2-3 days. As the physical models were made of plexiglass, it is transparent totally, thus the cells can be observed conveniently under a phase contrast microscope.

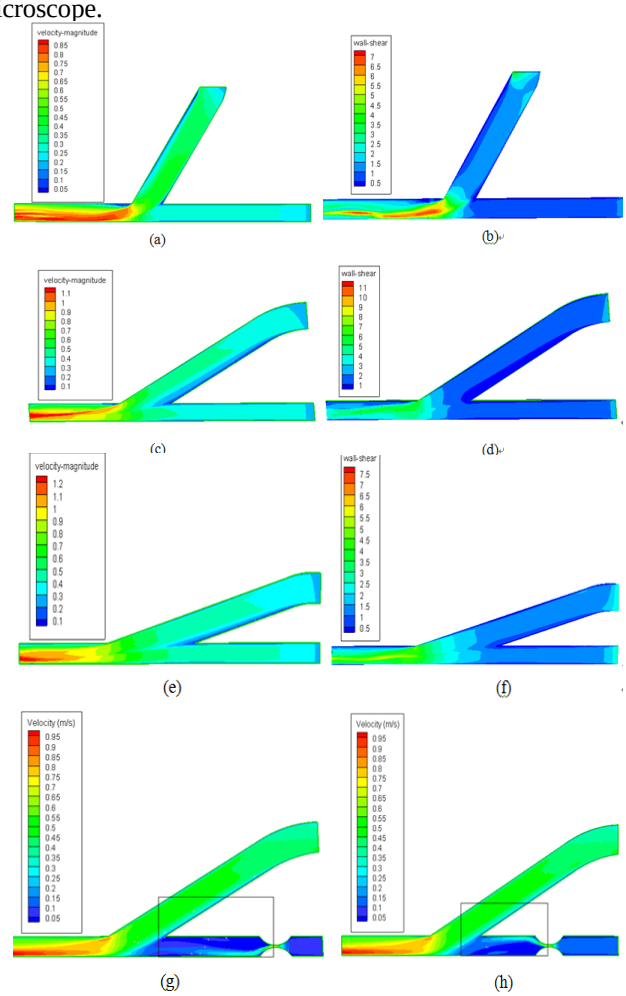


Figure 3. Velocity distributions for the model with ⊿P=4 Pa and (a) α=60°, (c) α=30°, and (e) α=15°. WSS distributions for the model with (b) α=60°, (d) α=30°, and (f) α=15°. Velocity distributions for the model with (g) D=3 cm, (h) D=2 cm, and (i) D=1 cm. Here, ⊿P denotes the pressure difference between the inlet and outlet, α denotes the graft-artery angle. D denotes the distance from stenosed site to the heel. The rectangle frame highlight the stagnant flow region. Velocity unit: (m/s); WSS unit:(Pa)

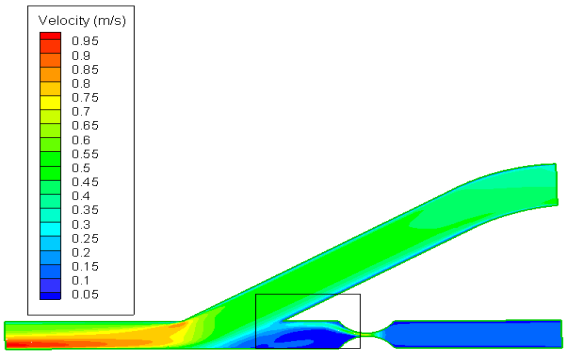


Figure 4. Velocity distributions for the model with D=1 cm.

IV. CONCLUSION

We report a set of models designs of CABG; these models include three kinds of anastomotic angle: 15 °, 30 ° and 60 °. The flow fields in physical models were analyzed by numerical simulation models. From the results of numerical simulation, we found that flow in a bypass graft is greatly dependent on the anastomotic angle. The velocity and WSS distributions show that the low WSS and stagnation regions occur near the toe, where the intimal thickening is apt to happen. A smaller α would eliminate the stagnation and decrease the risk of intimal thickening at the anastomotic junction. The effects of stenosed site have also been investigated in detail by numerical methods. The important flow features, which could be relative to the intimal hyperplasia in arteries, including the recirculation, and the shear stress distributions in host artery as well as bypass graft, are verified to be influenced significantly by the stenosed site and stenosed severity. The stagnant flow region depends directly on the distance D. The larger is the value of D, the longer is the stagnant region. We found the residual flow which is harmful to the orifice region and the distal portion of anastomotic gradually became more slender as severity of the stenosis increased. The residual flow plays a competing role with the graft flow. From the analysis of numerical simulation models, we think that these physical models and the arrangement of the perfusion system are feasible for CABG studies.

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