Optimal bypass graft design for left anterior descending and diagonal territory in multivessel coronary disease

Sachi Koyama, Keiichi Itatani, Tadashi Yamamoto, Shohei Miyazaki, Tadashi Kitamura, Tuyoshi Taketani, Minoru Ono and Kagami Miyaji

* Department of Cardiovascular Surgery, Kitasato University School of Medicine, Sagamihara, Kanagawa, Japan
b Department of Cardiac Surgery, Graduate School of Medicine, The University of Tokyo, Bunkyo, Tokyo, Japan
c Department of Hemodynamic Analysis, Kitasato University School of Medicine, Sagamihara, Kanagawa, Japan
d Department of Cardiology, Hokkaido Cardiovascular Hospital, Hokkaido, Japan
* Corresponding author. Department of Hemodynamic Analysis, Kitasato University School of Medicine, 1-15-1 Kitasato, Minami, Sagamihara, Kanagawa 252-0374, Japan. Tel: +81-42-7788111; fax: +81-42-7789371; e-mail: keiichitatani@gmail.com (K. Itatani).

Received 20 January 2014; revised in revised form 5 March 2014; accepted 26 March 2014

Abstract

OBJECTIVES: Coronary artery bypass grafting for multivessel disease requires an appropriate graft design to avoid the competition of flow between the graft and the native vessel in order to achieve a sufficient coronary flow and durable graft patency.

METHODS: Three-dimensional computational models of the left coronary artery were created based on the angiographic data. Three stenosis patterns of 75 and 90% combinations were created in the left anterior descending artery (LAD), the diagonal branch (Dx) and the circumflex artery (LCx). The left internal thoracic artery (LITA) was anastomosed to the LAD, and separate saphenous vein grafts (SVGs) were anastomosed to the Dx and the LCx in the ‘Independent’ model. The ‘Sequential’ model included sequential SVG anastomoses to the Dx and the LCx with a left internal thoracic artery–left anterior descending artery bypass, and Y-composite arterial grafts to LAD and Dx were created in the ‘Composite’ model.

RESULTS: The ‘Independent’ model had high reverse flow from the Dx to the LAD in systole, resulting in decreased LITA flow when Dx stenosis was mild. The ‘Sequential’ model also had reverse flow in diastole, resulting in additional LAD flow. The ‘Composite’ model distributed increased flow to the Dx when Dx stenosis was severe, resulting in decreased flow to the LAD.

CONCLUSIONS: Systematic flow evaluation is beneficial for determining the optimal bypass graft arrangement in patients with multivessel disease. Individual SVG anastomoses to the Dx and the LCx are not desirable when Dx stenosis is not severe and a Y-composite arterial graft to LAD and Dx is not desirable when Dx stenosis is severe.

Keywords: Coronary artery bypass graft • Ischaemic heart disease • Revascularization (coronary artery) • Modelling (computer) • Computer applications (simulation)

INTRODUCTION

The increase in life expectancy has confronted cardiac surgery with a rapidly growing population of elderly patients requiring surgical myocardial revascularization [1]. A fundamental aim of coronary surgery is to fashion a perfect anastomosis to deliver blood flow to the ischaemic territories of the myocardium with durable graft patency [2]. However, the ideal designing of coronary revascularization is uncertain, especially in cases with multivessel disease because the competitive flow between the graft and the native vessels, which is closely related to long-term graft patency [3], is difficult to predict. The blood flow distribution from each graft needs to be taken into consideration to provide adequate blood supply to the ischaemic territory.

Recently, several attempts have been made to evaluate coronary artery diseases using computational fluid dynamics (CFD), which is considered to be a novel method that can predict blood flow and pressure from the cardiovascular system [4]. The CFD method enables the quantitative evaluation of the severity of coronary ischaemia based both on the anatomical features of the disease site and on the myocardial reserve of the perfused lesion in each branch. Taylor et al. developed the fractional flow reserve derived from coronary computed tomography (CT) angiography, a noninvasive method for identifying ischaemia-causing stenosis [5], and Samady et al. calculated the wall shear stress in the disease site to evaluate the risk of plaque rupture and progression [6]. This method has the potential to provide a systematic evaluation of complicated cardiovascular diseases, and can facilitate graft design definition for treating multivessel disease.

The objective of this study was to establish the optimal graft arrangement for the left anterior descending and the diagonal territory in multivessel disease. To fulfil this objective, we developed idealized 3D CFD models that enable the systematic evaluation of blood flow and pressure distribution of each native...
branch and grafted vessel in various types of coronary artery bypass grafting (CABG).

METHODS

The present study was based on CFD models with basic idealized geometry to examine the details of the haemodynamics in patients with coronary artery disease before and after bypass graft procedures. The analysis involved the following three steps: firstly, 3D coronary arterial models were constructed using the computer-assisted design (CAD) technique. Models of three types of coronary artery disease and three types of bypass grafts were created. Secondly, numerical simulations of the physiological blood flow were performed with boundary conditions reflecting the cardiac muscle impedances and arterial pressure waves. Finally, from the calculated results, the severity of ischaemia and effects of revascularization after bypass grafting were systematically

Figure 1: (A) The methods used to create the models using computer-assisted design techniques. The 3D left ventricular shape was assumed to be a cone with a mitral annular base. The angiographic data of 10 patients with normal coronary vessels were used to create idealized 3D geometry of the left coronary artery. (B) The stenotic lesions in the coronary artery were created as follows: the edges of the punched-out lesions were made smooth in order to represent natural plaque attachment. (C) The three types of bypass models. Left, the left internal thoracic artery (LITA) graft to the left anterior descending artery (LAD) and separate saphenous vein grafts (SVGs) to the diagonal branch (Dx) and the left circumflex artery (LCx) (‘Independent’ model). Middle, the LITA to the LAD and a sequential SVG to the Dx and the obtuse marginal (‘Sequential’ model). Right, a Y-composite LITA and radial artery (RA) graft to the LAD and the Dx (‘Composite’ model).
evaluated with the flow rate recovery and pressure recovery for each coronary branch.

Creation of coronary artery models and bypass graft models

We created 3D left coronary artery geometry using a commercial CAD software program, SolidWorks (SolidWorks Japan, Tokyo, Japan). The geometrical features of the coronary arterial branches, including the vessel diameter and angles, were determined using the angiographic data of 10 normal cases. The average values were substituted for the geometric values. We obtained the diameter and length of the left main trunk (LMT), left anterior descending (LAD), diagonal branch (Dx), obtuse marginal (OM) and posterior lateral (PL), as well as the following angles between branches: LMT and LAD, LAD and Dx, and LAD and LCx. The virtual left ventricle (LV) was assumed to be a cone, the elliptical base of which had a mitral annular size of 36 and 24 mm for the long and short axes, respectively, and the height of which was set to be 80 mm as a representative value of the LV long axis. Then, all coronary branches were directed towards the apex, and the circumflex branch was designed to pass through the mitral annulus. The details of these features are illustrated in Fig. 1A.

Stenotic lesions in the coronary artery were created as follows. Hemicircles with a radius of 75 and 90% of the vessel diameter were punched out in the native vessels to create 75 and 90% stenosis models with eccentric plaque, respectively. In order to imitate natural plaque attachment, we smoothed the edges of the punched-out lesions (Fig. 1B). As a result, the plaque area of the 75% stenosis models occupied 86.3% of the vessel luminal area and that of the 90% stenosis models had 99.8% of the vessel luminal area. Finally, three coronary disease models were created: LAD 75%, Dx 75% and LCx 75% stenosis; LAD 90%, Dx 90% and LCx 90% stenosis; and LAD 90%, Dx 75% and LCx 90% stenosis.

Three models of coronary bypass to the LAD, Dx and LCx were created. In the 'Independent' model, the left internal thoracic artery (LITA) was connected to the LAD and separate SVGs were grafted to the Dx and the OM. In the 'Sequential' model, the SVG was sequentially anastomosed to the Dx and the OM along with the left internal thoracic artery–left anterior descending artery (LITA–LAD) bypass. In the 'Composite' model, a Y-composite arterial graft was anastomosed to the LAD and the Dx and a separate saphenous vein graft (SVG) was anastomosed to the OM.

Figure 2: (A) The pressure waves of the aorta and the LITA. We substituted these pulsatile waves for the inlet boundary conditions for all models. (B) The peripheral resistances of the LAD, Dx, OM and PL. We substituted these impedances for the outlet boundary conditions for all models. (C) The LITA flow wave of the three bypass graft designs in one cardiac cycle under LAD 75%, Dx 75% and LCx 75% stenosis. I: The isovolumetric contraction period (t = 0.10); II: aortic pressure peak in the early systolic phase (t = 0.29); III: LITA pressure peak (t = 0.40); IV: coronary resistance peak (t = 0.49); V: phase of dicrotic notch in the middle diastolic phase (t = 0.65). (D) The aortic pressure wave and LITA pressure wave with LAD resistance in one cardiac cycle. Dx: diagonal branch; LAD: left anterior descending artery; LITA: left internal thoracic artery; LCx: left circumflex artery; OM: obtuse marginal; PL: posterior lateral; ECG: electrocardiogram; Ao: aorta.
diameters of the LITA and arterial graft were assumed to be 2.0 mm, and that of the SVG was assumed to be 3.0 mm (Fig. 1C).

**Flow simulation and boundary conditions**

The finite volume method was used to solve the mass and momentum conservation equation (Navier–Stokes equation) using a commercial CFD software program, ANSYS-CFX 13.0 (ANSYS Japan, Tokyo, Japan). Blood was assumed to be an incompressible Newtonian fluid with a density of 1060 kg m$^{-3}$ and a viscosity of 0.004 kg m$^{-1}$ s$^{-1}$ [6]. A commercial grid generation code, ICEM-CFD (ANSYS Japan, Tokyo Japan), was used to divide the numerical domains into tetrahedral unstructured meshes with three boundary-fitted prism layers. The number of nodes was set around 5.0 × 10$^6$, and there were around 1.2 × 10$^6$ elements for all models. The time step was set to be fine (<10$^{-5}$) enough to achieve a sufficiently low Courant number, and the transient calculations for two cardiac cycles were performed.

Pulsatile calculations were adopted to realize the physiological pulsatile flow. We measured the pressure and flow velocity of the aorta, left subclavian artery and distal coronary artery branches using simultaneous measurement of the pressure and velocity using a Combo Wire catheter (Volcano Corp., Rancho Cordova, CA, USA) on normal coronary arteries. The blood pressure and velocity profile were assessed with a 200-Hz sampling frequency and were smoothed by a Gaussian filter with a 25-ms standard deviation. The pressure and velocity profiles of one cardiac cycle were taken as the averaged value of 10 cardiac cycles. We set one cardiac cycle as 1 s. We set the aortic and left subclavian arterial pressure wave as the inlet boundary conditions (the orifices of the LMT, SVG and LITA) (Fig. 2A). The peripheral coronary impedances were assessed with a 200-Hz sampling frequency and were thus set to be 0.13 times the resistance value at rest in order to attain the total LAD flow between three and five times that without a vasodilator [7]. A rigid non-slip wall condition was set for all vessel walls.

**Evaluation of ischaemia and revascularization based on the calculated results**

From the calculated results, the following parameters were examined: the flow rate, pressure distribution and streamlines for each vessel branch and bypass grafts. In the pulsatile simulation, we calculated the reverse and competitive flow from the results. The LITA flow in the composite model was measured distal to the Y-anastomosis site.

**RESULTS**

**Flow and pressure distributions of the control model and disease models**

The table (Table 1) shows the mean flow splits of all branches on our control model and the three coronary disease models under the maximum vasodilatation conditions. The total mean flow of the normal left coronary artery was 567.6 ml/min, which was an acceptable value under the maximum vasodilatation condition [7]. The branch flows of the three coronary disease models were decreased according to the increase of the degree of stenosis: under the LAD 75%, Dx 75% and LCx 75% condition, the LAD flow was 135.3 ml/min; and under the LAD 90%, Dx 75% and LCx 90% condition, the LAD flow was 69.1 ml/min. The Dx flow was also decreased with an increase in the stenosis ratio.

<table>
<thead>
<tr>
<th>Disease</th>
<th>LAD (ml/min)</th>
<th>Dx (ml/min)</th>
<th>OM (ml/min)</th>
<th>LITA–LAD (ml/min)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normal coronary in maximum vasodilatation</td>
<td>205.0</td>
<td>187.2</td>
<td>101.8</td>
<td>None</td>
</tr>
<tr>
<td>LAD 75% Dx 75% LCx 75%</td>
<td>135.3</td>
<td>64.1</td>
<td>80</td>
<td>None</td>
</tr>
<tr>
<td>LAD 90% Dx 90% LCx 90%</td>
<td>82.1</td>
<td>18.1</td>
<td>39.6</td>
<td>None</td>
</tr>
<tr>
<td>LAD 90% Dx 75% LCx 90%</td>
<td>69.1</td>
<td>34.6</td>
<td>39.1</td>
<td>None</td>
</tr>
<tr>
<td>After CABG</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>LAD 75% Dx 75% LCx 75%</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Independent</td>
<td>240.5</td>
<td>235.0</td>
<td>168.5</td>
<td>55.3</td>
</tr>
<tr>
<td>Sequential</td>
<td>239.2</td>
<td>218.0</td>
<td>153.7</td>
<td>63.0</td>
</tr>
<tr>
<td>Composites</td>
<td>213.3</td>
<td>153.1</td>
<td>169.7</td>
<td>73.7</td>
</tr>
<tr>
<td>LAD 90% Dx 90% LCx 90%</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Independent</td>
<td>208.4</td>
<td>235.8</td>
<td>167.5</td>
<td>111.7</td>
</tr>
<tr>
<td>Sequential</td>
<td>213.1</td>
<td>211.3</td>
<td>147.9</td>
<td>121.5</td>
</tr>
<tr>
<td>Composites</td>
<td>195.4</td>
<td>137.9</td>
<td>166.8</td>
<td>110.5</td>
</tr>
<tr>
<td>LAD 90% Dx 75% LCx 90%</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Independent</td>
<td>215.6</td>
<td>233.3</td>
<td>167.1</td>
<td>96.1</td>
</tr>
<tr>
<td>Sequential</td>
<td>219.0</td>
<td>210.9</td>
<td>148.0</td>
<td>110.1</td>
</tr>
<tr>
<td>Composites</td>
<td>189.1</td>
<td>140.0</td>
<td>166.6</td>
<td>127.5</td>
</tr>
</tbody>
</table>

CABG: coronary artery bypass graft; Dx: diagonal branch; LAD: left anterior descending artery; LCx: left circumflex artery; left internal thoracic artery–left anterior descending artery; OM: obtuse marginal.
Left internal thoracic artery–left anterior descending artery bypass flow in one cardiac cycle

Figure 2C and D shows LITA–LAD flow in each bypass model on the LAD 75%, Dx 75%, LCx 75% stenosis model and its boundary conditions in one cardiac cycle, respectively. LITA–LAD flow stayed at almost the same level during the isovolumetric contraction period (Fig. 2C, labelled I). LITA–LAD flow decreased in early systole with an increase in the aortic pressure (Fig. 2C, labelled II), and then it increased with an increase in LITA pressure (Fig. 2C, labelled III). LITA–LAD flow decreased again in late systole with increase of the coronary impedance (Fig. 2C, labelled IV). Without the phase of the dicrotic notch in mid-diastole (Fig. 2C, labelled V), LITA–LAD flow was maintained during diastole.

Haemodynamics and bypass graft design

The peripheral perfusion pressure and branch flow in all bypass models improved when compared with those in the disease models. The table describes the mean LAD, Dx, OM, PL and LITA flows of all three bypass models in all three stenosis patterns. When the LAD, Dx and LCx had 75, 75 and 75% stenosis, respectively, LAD flows of all bypass models recovered to almost the same level of the normal control one, and LITA–LAD flow stayed relatively lower compared with that under more severe stenosis patterns (Table 2). LAD flows of the ‘Composite’ models were 10% lower than other bypass models and the normal model in all situations under the maximum vasodilatation condition.

When the LAD, Dx and LCx had 90%, 75% and 90% stenosis, respectively, LITA–LAD flow was higher in the ‘Sequential’ and ‘Composite’ models than in the ‘Independent’ model. In the ‘Independent’ model, prominent decreases in LITA flow were observed during systole with an increase in aortic pressure, and with reverse flow from the Dx to the LAD. In the ‘Sequential’ model, LITA flow during systole was higher than that in the ‘Independent’ model, and the reverse flow was lower than that of the ‘Independent’ model. In the ‘Composite’ model, the peripheral LAD pressure was lower than that of the other models, and no reverse flow was detected (Fig. 3). When the LAD, Dx and LCx had 90, 90 and 90% stenosis, respectively, LITA–LAD flow in the ‘Composite’ model decreased prominently due to the flow split to the Dx branch. During diastole, LITA–LAD flow of the ‘Sequential’ models was higher than that of the ‘Independent’ model (Fig. 4).

**DISCUSSION**

The optimal bypass graft design in CABG has been discussed for decades, and yet there has been no consensus on the ideal graft design. Several studies have reported an incremental survival benefit by increasing the number of arterial grafts [8-11], and this has increased interest in avoiding vein grafts altogether in favour of an all-arterial CABG for multivessel coronary disease [12]. Such all-arterial revascularization is usually accomplished through varying combinations of multiple arterial conduits and grafting methods (e.g. T or Y grafts) [13-15]. Zacharias et al. reported that all-arterial revascularization was associated with a significantly better 12-year survival rate compared with the standard single internal thoracic artery (ITA) with SVG [12]. Nakajima et al. reported that a composite graft using in situ and free grafts is necessary for complete revascularization in patients with multivessel disease, and the arterial graft is commonly used because of its beneficial characteristics in terms of both graft patency and improved late outcomes [3]. However, Sakaguchi et al. reported that the composite Y graft demonstrated improved myocardial blood flow at rest, but it was not as effective as independent grafts with SVG for improving the coronary flow reserve under exercise [16]. These clinical reports did not perform a quantitative evaluation of the blood flow of each coronary branch and the reverse flow of grafted vessels, resulting in difficulty to compare several graft designs quantitatively.

In the present study, the composite Y grafting did not reveal reverse flow from the Dx to the LAD, but the blood flow to the LAD was relatively low under the maximum vasodilatation condition when compared with the other models. These results are consistent with the conclusion of Sakaguchi’s report indicating that the composite Y graft was not as effective as independent SVGs for improving the coronary flow reserve under the maximum vasodilatation condition [16]. In addition, when the Dx stenosis became more severe, LITA flow of the ‘Composite’ model was distributed more to the Dx, resulting in relatively low LAD perfusion compared with other models.

On the other hand, the reverse flow from the Dx to the LAD was detected both in the ‘Independent’ and ‘Sequential’ models during one cardiac cycle. The reverse flow of the ‘Independent’ model during systole was higher than that during diastole, resulting in a decrease in LITA flow, especially when the Dx stenosis was mild. These results indicated that the reverse flow of the ‘Independent’ model from the Dx to LAD was competitive with the LITA–LAD graft flow, which would be considered to cause reduced LITA graft patency as Kawamura et al. revealed the relationship between reverse flow and graft patency [17]. On the other hand, the reverse flow of the ‘Sequential’ model from the Dx to LAD was also detected during one cardiac cycle, but LITA–LAD flow increased especially during diastole compared with the other two models (Fig. 2C), and the reverse flow of this model during diastole could act as an additional flow to the LAD. The pulsatile fluctuation of the pressure and flow of the LITA in the ‘Sequential’

---

**Table 2:** The branch averaged flow split of the left coronary arteries after coronary artery bypass graft in diastole

<table>
<thead>
<tr>
<th></th>
<th>LAD (ml/min)</th>
<th>Dx (ml/min)</th>
<th>OM (ml/min)</th>
<th>LITA–LAD (ml/min)</th>
</tr>
</thead>
<tbody>
<tr>
<td>After CABG</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>LAD 75% Dx 75% LCx 75%</td>
<td>218</td>
<td>197.0</td>
<td>142.0</td>
<td>88.6</td>
</tr>
<tr>
<td>Independent</td>
<td>218</td>
<td>197.0</td>
<td>142.0</td>
<td>88.6</td>
</tr>
<tr>
<td>Sequential</td>
<td>215.8</td>
<td>184.2</td>
<td>130.3</td>
<td>99.6</td>
</tr>
<tr>
<td>Composite</td>
<td>208.0</td>
<td>143.8</td>
<td>142.6</td>
<td>92.4</td>
</tr>
<tr>
<td>LAD 90% Dx 90% LCx 90%</td>
<td>196.3</td>
<td>196.1</td>
<td>150.0</td>
<td>124.4</td>
</tr>
<tr>
<td>Independent</td>
<td>196.3</td>
<td>196.1</td>
<td>150.0</td>
<td>124.4</td>
</tr>
<tr>
<td>Sequential</td>
<td>200.7</td>
<td>177.7</td>
<td>125.5</td>
<td>137.1</td>
</tr>
<tr>
<td>Composite</td>
<td>185.5</td>
<td>131.6</td>
<td>140.9</td>
<td>122.3</td>
</tr>
<tr>
<td>LAD 90% Dx 75% LCx 90%</td>
<td>200.8</td>
<td>194.3</td>
<td>125.4</td>
<td>113.5</td>
</tr>
<tr>
<td>Independent</td>
<td>200.8</td>
<td>194.3</td>
<td>125.4</td>
<td>113.5</td>
</tr>
<tr>
<td>Sequential</td>
<td>203.8</td>
<td>178.3</td>
<td>132.4</td>
<td>132.1</td>
</tr>
<tr>
<td>Composite</td>
<td>181.7</td>
<td>133.8</td>
<td>140.7</td>
<td>134.3</td>
</tr>
</tbody>
</table>

CABG: coronary artery bypass graft; Dx: diagonal branch; LAD: left anterior descending artery; LCx: circumflex artery; LITA–LAD: left internal thoracic artery–left anterior descending artery; OM: obtuse marginal; PL: posterior lateral.
model was higher than that of the other models, and this fluctuation might contribute to the long-term graft patency.

In evaluating coronary artery disease, especially multivessel disease, angiography has been reported to be unable to assess the geometrically complicated lesions or to estimate the severity of myocardial ischaemia [18, 19], and is also unable to detect quantitatively the flow in each vessel. On the other hand, positron emission tomography can provide a quantitative evaluation of myocardial ischaemia, but is reported to underestimate the extent of multivessel disease [20], and provides little anatomical information about the diseased vessels. Shimizu et al. used a Doppler ultrasound-tipped guidewire and reported that the grade of the native coronary artery stenosis obviously affected the ITA graft flow [21]. These catheter flow guidewires were suitable for the quantitative evaluation of the severity of coronary ischaemia in each branch, but were unsuitable for assessing the total balance of other branch stenosis and the coronary flow volume. Therefore, in order to perform a quantitative evaluation of the flow of the whole branch systematically, we developed 3D CFD models based on idealized left coronary artery geometry created by patients’ averaged data using a CAD program. Taylor et al. reported that the CFD method may enable the noninvasive assessment of lesion-specific ischaemia and the prediction of changes in the coronary flow and pressure resulting from therapeutic intervention [5].

Figure 3: The pressure distributions and streamlines after CABG: the independent model, sequential model and composite bypass model with the stenosis pattern: LAD 90%, Dx 75% and LCx 90%. Dx: diagonal branch; LAD: left anterior descending artery; LCx: left circumflex artery; CABG: coronary artery bypass graft.
Regarding the boundary condition setting, Taylor et al. adopted the ‘lumped parameter method’, which is an analogy of an electrical circuit to imitate microvascular characteristics. In our system, we adopted directly measured coronary impedance to achieve realization of pulse and flow wave patterns especially including their accurate characteristics with phase in one cardiac cycle. Our measurement-based impedance boundary conditions can realize the peripheral vascular beds reflecting cardiac muscular viabilities, and have the potential to realize an altered condition with ischaemic myocardial changes.

On the other hand, regarding the 3D geometry of the present models, the present study was not based on a patient-specific model [22–25] but on an idealized and rather simplified one. CFD studies, by their nature, enable detailed and systematic evaluations that are not available in usual clinical examinations, even though they do not provide directly measured data, but only virtually simulated calculation results. Recent CFD models have reconstructed patient-specific 3D geometry from cardiovascular imaging data including CT, magnetic resonance imaging and/or intravascular ultrasound [6, 25], and one study constructed statistical evidence from these calculation results [5]. However, because the purpose of the simulation study is to reveal the generalized abstract knowledge that is essential to the surgical strategies independent of the characteristics of each patient [6], an idealized model created based on the averaged patient data would be necessary. Future studies will include comparison and verification of these idealized numerical studies with accumulated results from patient-specific numerical models of many cases, and now we work in progress towards a future study that will include comparison and verification of these idealized numerical studies with accumulated results from patient-specific numerical models of many cases.

In numerical models, several assumptions may oversimplify the multifactorial human body system and become study limitations. In this study, we assumed the vessel walls to be rigid and dissociated from cardiac muscle movement [25], and we could not include the valve structures in the SVG or spasm in the arterial graft. Studies using more physiological models with impedance reflecting the muscle viability of each vessel perfusion region or the effects from the collateral vessels are required in order to
cover all possible cases or situations. Another limitation was that our idealized models did not reflect patient-specific anatomical characteristics such as kinking of the graft in the sequential anastomosis in clinical practice. Further studies will be needed encompassing varieties of coronary disease patterns, reflecting vessel compliance and myocardial viability, and revealing the impact of these characteristics on both the numerical and clinical results. In this study, only three patterns of coronary grafting with only three stenosis patterns were evaluated, and future studies should include more grafting patterns known in the literature.

CONCLUSIONS

The simulation of quantitative and systematic haemodynamics based on computational models was feasible and useful for evaluation of the competitive flow and reverse flow of coronary arteries and bypass grafts. In the left coronary multivessel disease models, when Dx stenosis was not severe, individual SVG anastomoses to the Dx and the LCx were not desirable because of high reverse flow from the Dx to the LAD in systole. When Dx stenosis was severe, a Y-composite arterial graft to the LAD and the Dx was not desirable because of the low flow split of the LITA graft.

Conflicts of interest: none declared.

REFERENCES