In vitro measurements of velocity and wall shear stress in a novel sequential anastomotic graft design model under pulsatile flow conditions

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Abstract

This study documents the superior hemodynamics of a novel coupled sequential anastomoses (SQA) graft design in comparison with the routine conventional end-to-side (ETS) anastomoses in coronary artery bypass grafts (CABG). The flow fields inside three polydimethylsiloxane (PDMS) models of coronary artery bypass grafts, including the coupled SQA graft design, a conventional ETS anastomosis, and a parallel side-to-side (STS) anastomosis, are investigated under pulsatile flow conditions using particle image velocimetry (PIV). The velocity field and distributions of wall shear stress (WSS) in the models are studied and compared with each other. The measurement results and WSS distributions, computed from the near wall velocity gradients reveal that the novel coupled SQA design provides: (i) a uniform and smooth flow at its ETS anastomosis, without any stagnation point on the artery bed and vortex formation in the heel region of the ETS anastomosis within the coronary artery; (ii) more favorable WSS distribution; and (iii) a spare route for the blood flow to the coronary artery, to avoid re-operation in case of re-stenosis in either of the anastomoses. This in vitro investigation complements the previous computational studies of blood flow in this coupled SQA design, and is another necessary step taken toward the clinical application of this novel design. At this point and prior to the clinical adoption of this novel design, in vivo animal trials are warranted, in order to investigate the biological effects and overall performance of this anastomotic configuration in vivo.

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1. Introduction

Coronary artery bypass grafting (CABG), the primary treatment for coronary artery high-risk patients, has a limited long-term patency mainly due to anastomatic intimal hyperplasia (IH). Hemodynamics is widely believed to be one of the most important factors implicated in the initiation and progression of IH [1–4], predominantly at the downstream anastomosis where IH has been found to be significantly greater [5].

A conventional distal end-to-side (ETS) anastomosis is characterized by abnormal flow conditions, including flow oscillation at the heel, impact on the artery floor, and flow separation at the toe. Typically, there is a low-WSS region at the heel, where a vortex forms due to the interaction of the flow from the graft with the relatively slow flow in the occluded proximal artery. The presence of a slow recirculation flow (i.e., a vortex) increases the blood near-wall residence time and results in platelet activation [6] and fibrin thrombus formation [7,8], which leads to IH development [1,9–11]. Along with the vortex, there is a stagnation point on the artery bed, where the graft flow impinges the arterial floor, whose location oscillates during the cardiac cycle. This moving stagnation point provides a low-magnitude-high–oscillatory WSS condition on the artery bed which has been suggested to enhance atherogenesis [12] and IH formation [13]. In addition, the flow impact on the artery floor is known to be injurious to the endothelium and is believed to be a contributing factor to the graft failure [14]. In a conventional ETS configuration, there is a high spatial gradient of WSS at...
the toe of the anastomosis, which has been postulated as a potential cause of inducing morphological and functional changes in the endothelium that can contribute to elevated wall permeability and consequent atherosclerotic lesions [15–17] and IH development [18].

Several studies have been conducted to improve the hemodynamics at the anastomotic junction [4], in order to attain higher patency rates in bypass grafts. These investigations include studies on the effects of geometrical factors, such as anastomotic angle [19–22], modified configuration of distal anastomosis [23–26], graft-to-host artery diameter ratio [27–30], and out-of-plane graft [31–33], as well as the effects of stenosis severity and proximal artery flow [34,35], irregularities of venous graft wall due to venous valve sinus [36], and distance of grafting (distance between anastomosis and stenosis) [37].

Optimization of the abovementioned geometrical factors in an ETS anastomosis has been shown to somewhat improve the flow field and distribution of the hemodynamic parameters (HPs). However, it has not changed the general problematic flow characteristics, such as the vortex formation at the heel or the impact of the blood on the artery bed. Therefore, in an effort to resolve the dilemma of an optimal downstream anastomotic geometry, a novel coupled sequential anastomoses (SQA) configuration has recently been designed, based on the beneficial flow characteristics observed in the side-to-side (STS) anastomosis of sequential bypass grafts [38]. The computational simulations and fluid–structure interaction (FSI) analyses of this SQA graft design have shown an improved flow field and advantageous distribution of HPs in the coupled SQA design, as compared to the conventional ETS anastomosis [38,39].

In this in vitro study, we have investigated the hemodynamics in fabricated polydimethylsiloxane (PDMS) model of the novel coupled SQA graft design, in comparison with a conventional ETS anastomosis and a parallel STS anastomosis. These three models had been labeled as the base models in the previous computational simulation study [38]. At this stage and prior to the adoption of this novel SQA graft design for in vivo animal trials and clinical applications, we have further examined the efficacy of the coupled SQA graft design compared to the existing conventional ETS anastomosis experimentally, utilizing particle image velocimetry (PIV) technique. The results of the present in vitro investigation can complement the earlier computational studies.

2. Materials and methods

The flow field velocity measurement was conducted by PIV in three life-size PDMS anastomotic models: (i) the novel coupled sequential anastomoses, (ii) a parallel STS anastomosis, and (iii) a conventional ETS anastomosis, shown in Fig. 1(a)–(c), respectively. The latter two were the reference models to be compared, respectively, with the STS and ETS components of the novel SQA graft design.

2.1. Geometric models

The silicone rubber (i.e., PDMS) models were produced, using the method introduced by Chong et al. [40]. In the coupled SQA model, which has been designed based on the beneficial results of sequential bypass grafting, there is initially a STS anastomosis distal to the stenosis, and then the graft end is anastomosed to the same coronary artery further downstream in an ETS fashion, as shown in Fig. 1(a). The dimensions of the models are listed in Table 1, where \( D_G \), \( D_A \), \( L_P \), \( L_{STS} \), \( L_{ETS} \), \( L_D \), and \( d \) stand for the graft diameter, coronary artery diameter, length of the proximal section of the host artery, length of the STS anastomosis, length of the ETS anastomosis, length of the distal section of the host artery, and the distance between the two anastomoses in the novel SQA model, respectively, as shown in Fig. 1. The physical models were bilaterally symmetric with respect to the XZ plane, and the outlet of the proximal segment of the coronary artery was fully occluded during the experiments.

<table>
<thead>
<tr>
<th>Model</th>
<th>( D_G )</th>
<th>( D_A )</th>
<th>( L_P )</th>
<th>( L_{STS} )</th>
<th>( L_{ETS} )</th>
<th>( L_D )</th>
<th>( d )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Coupled SQA</td>
<td>4</td>
<td>2</td>
<td>45</td>
<td>8</td>
<td>7</td>
<td>20</td>
<td>30</td>
</tr>
<tr>
<td>Conventional ETS</td>
<td>4</td>
<td>2</td>
<td>45</td>
<td>N.A.</td>
<td>7</td>
<td>35</td>
<td>N.A.</td>
</tr>
<tr>
<td>Parallel STS</td>
<td>4</td>
<td>2</td>
<td>45</td>
<td>N.A.</td>
<td>30</td>
<td>30</td>
<td>N.A.</td>
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</table>

2.2. Working fluid, flow conditions, and flow circuit

A mixture of 30% glycerin and 70% sodium chloride (NaCl) solution by volume was used as the working fluid in this study (the NaCl solution was made up of 1 part NaCl and 3 parts water by weight). This Newtonian mixture’s density was 1156 kg/m³ and its dynamic viscosity at the working temperature of 25 °C was quantified by a controlled shear rate rheometer (Contraves low rate 40) to be 0.0054 Pa·s (5.4 cP). The refractive index of the mixture was measured by a commercial refractometer (Atago 3T; High-Precision Model) to be 1.409, which matches that of the PDMS model.

Each PDMS model was incorporated into the flow circuit, as shown schematically in Fig. 2, and a computer-controlled gear pump system (Micropump Inc., Vancouver WA USA), including one varying flow pump (Model HG0024-N23 PF15G81) and one steady flow pump (Model HG0024-G050), was used to generate the physiological inlet flow waveform with the time period of 0.9 s, as demonstrated in Fig. 3 [38]. The flow rate was measured and monitored by means of an electromagnetic flow meter (Promag 53, Endress + Hauser, Germany) immediately upstream of the test section.

The peak flow rate during systole was \( Q_{max} = 70 \text{ ml/min} \). The maximum Reynolds and Womersley numbers at the peak systole were calculated to be \( Re_{max} = 79 \) and \( \omega_{max} = 2.45 \) respectively, based on the graft diameter of 4 mm. This suggested that the flow was laminar, which was further proven by PIV results, as no turbulence was observed in the flow field.

In the parallel STS model, 15% of the blood flow (on average within each cardiac cycle) was set to pass through the STS anastomosis into the coronary artery, and the remaining 85% was allowed to pass through the graft, by adjusting the resistance at the outlets of the graft and coronary artery of this model. The above flow division was adjusted to make the parallel STS model consistent with the novel coupled SQA configuration, since the flow was bifurcated in the same proportion through the coronary artery and the graft at the STS component of the SQA model (as calculated based on the PIV results).

In this experiment, hollow glass spheres with mean diameter of 10 μm, density of 1.1 g/cm³, and refractive index of 1.52 (HGS-10, Dantec Dynamics, Denmark) were used as tracer particles and seeded into the working fluid.

2.3. Particle image velocimetry (PIV) system

The PIV is an optical measurement technique, allowing the instantaneous acquisition of an entire planar section of a flow field [41]. The PIV system used in this study, consisted of a 15 Hz Q-switched, double cavity pulsed Nd:YAG laser (Minilase-III, New Wave Research, USA) with an energy of 150 mJ at a wavelength of 1064 nm, producing a light sheet which was adjusted to
illuminate the tracer particles in the symmetry plane of the models, as shown schematically in Fig. 2.

The particles’ motion was recorded by means of a synchronized charge coupled device (CCD) Camera (FlowSense, Dantec Dynamics, Denmark), with a spatial resolution of 1600 × 1200 pixel², and a Nikon lens (AF Micro Nikkor, 60/2.8) positioned normal to the laser sheet. The image acquisition in this phase-resolved PIV measurement was triggered by a TTL signal (5 V), generated by the gear pump interface unit, in order to phase-lock the measurements. The flow data were acquired at different phase points during the cardiac cycle, and presented for six distinct points as shown in Fig. 3. For statistical convergence, 100 measurements were found to be sufficient for each time instant, in order to generate a stable velocity vector map.

2.3.1. Post-processing of the data

2.3.1.1. Image processing. After minimum background subtraction, Gaussian smoothing, intensity normalization, and masking [41], the recorded image pairs were analyzed by a two-frame FFT cross-correlation algorithm (DynamicStudio, Version 2.20.18, Dantec Dynamics, Denmark) with interrogation areas of 32 × 32 pixel², overlapped by 50% on each side, to yield the local displacement vector for each interrogation area [42]. This method provided a total of 7326 velocity vectors for an area of 13 mm × 9 mm. Validation of this PIV method is presented in the Appendix.

2.3.1.2. Derivation of WSS. The velocity field obtained by PIV was used to estimate the rate-of-shear (\(\dot{\gamma}\)) along the wall from the near wall velocity gradient, and thereby the WSS, using Eq. (1):

\[
WSS = \mu \frac{du_s}{d\eta}
\]  

where \(\mu\) is the dynamic viscosity of the blood analogue, \(u_s\) is the component of the velocity vector tangential to the wall, and \(\eta\) is the direction normal to the wall. The \(\xi-\eta\) coordinate system was utilized for the ease of calculation and presentation of the results. As shown in Fig. 4, the origins of coordinates \(\xi_1\) to \(\xi_3\) were chosen on the heel, toe, and artery bed of the anastomosis, respectively.

A computer code, developed in-house, was utilized to interpolate the near-wall velocities within the viscous boundary layer based on the finite element shape functions [43]. Briefly, the discrete pixel wall location for each data set was extracted (no wall movement was detected in this study); by marching on these wall points, the slope of each wall section, located between two
neighboring wall points, was calculated. Then, on the perpendicular bisector of each wall section, two points with an increment of 0.13 mm (16 pixels) between the subsequent points were located within the flow boundary layer, and the tangential components of the velocity vectors (with respect to the corresponding wall section) were interpolated at these points based on the three nearest adjacent velocity vectors, using finite element shape functions. Subsequently, velocity gradients were calculated from a surface fit of the velocity using generalized multiquadratic radial basis functions, optimized to minimize the surface roughness of the resultant fit [44], assuming no-slip condition at the wall. Validation of this WSS estimation method is presented in the Appendix.

3. Results

3.1. Comparison of the velocity profiles in the anastomotic region of the models

The measured velocity fields in the symmetry plane at the anastomotic region of the models were substantially similar to those which had been obtained earlier by our computational simulations [38]; hence, for brevity, the flow fields are not discussed here. However, comprehensive presentation of the velocity vectors (at the six time points shown in Fig. 3) and discussion of the flow fields are provided in online Supplementary Material. Accordingly, the very advantages which had earlier been observed in the flow fields of the SQA graft design were witnessed in the present experimental results too. For instance, the flow at the distal ETS anastomosis of the SQA model during the acceleration phase was smooth, without any vortex formation or impingement of flow on the artery bed (due to the partial flow from the STS anastomosis along the coronary artery changing the velocity components of the ETS anastomosis flow toward the axis of the coronary artery rather than to the artery floor), unlike the conventional ETS model, as shown in Fig. 5 for the peak flow rate (t2). Also, the vortices and stagnation point on the artery bed were eliminated at the STS anastomosis of the SQA model (due to the flow circulation in the vascular loop of the SQA design) during deceleration and reversed flow phases, as demonstrated in Fig. 6 for the time when the net flow rate just became negative (t5). These are positive features of the SQA design, as a stagnation point is associated with low-magnitude-high-oscillatory WSS condition.

![Fig. 2. Schematic of the mock circulation loop and PIV system.](image)

![Fig. 3. The graft velocity waveform; labels indicate the times at which the flow field and WSS are studied, namely, systolic mid-deceleration (t1), peak flow rate (t2), and systolic mid-acceleration (t3), end systole (t4), when the graft average velocity has just become negative (t5), and peak reversed flow (t6).](image)

![Fig. 4. The origin and direction of $\xi$-$\eta$ coordinate system at (1) heel, (2) toe, and (3) artery bed of anastomosis.](image)
3.2. Comparison of the WSS distribution in the anastomotic region of the models

The local WSS was evaluated at the heel, toe, and coronary artery bed of the anastomosis in all the models at the six time instants, as shown in Figs. 7–10, wherein the WSS distribution at the heel, toe, and artery bed of the anastomosis is shown, respectively, in panels (a), (b), and (c).

3.2.1. WSS distribution in the heel region of the ETS anastomoses

Comparison of the WSS distribution in the heel region of the ETS component of the SQA design with that of the conventional ETS model (Figs. 8a and 10a) indicated that the peak value of WSS was slightly reduced on the graft inner wall ($\xi < 0$ mm) of the SQA design as compared to the conventional ETS model (for instance from 3 to 2 Pa at $t_2$). This was due to the lower flow rate passing through the graft toward ETS anastomosis in the novel design compared to the conventional ETS model, since part of the flow was diverted into the coronary artery at the STS anastomosis of the SQA design. However, the WSS at the heel of the ETS anastomosis of the SQA model within the coronary artery ($\xi > 0$ mm) was considerably increased in magnitude compared to the conventional ETS model. This was because in the conventional ETS model, the blood was almost stagnant in the proximal segment of the coronary artery, while in the novel SQA design model the partial flow from the STS component through the coronary artery increased the magnitude of WSS at this location (note that the change in the sign of the WSS is due to the definition of the coordinate system and the location of the origin at the heel region of the models with respect to the flow directions). It is to be noted that a number of studies have indicated a strong relationship between the localized distribution of low-WSS and IT development; as a result, this increase of WSS magnitude at the heel of the ETS anastomosis is a positive feature of the SQA design.

3.2.2. WSS distribution in the toe region of the ETS anastomoses

At the toe of the ETS anastomosis of the SQA model and the conventional ETS model, the qualitative distribution of WSS was similar, as shown in Figs. 8b and 10b. There was a sharp increase in the WSS magnitude to its peak at the toe, and then a rapid drop to form an apex, continued by a slow increase to an asymptotic value further downstream along the toe within the coronary artery mainly during the forward flow phases. The peak value of the WSS at the toe during the peak flow rate ($t_2$) was higher in the conventional ETS model (44 Pa) than that in the ETS component of the SQA
model (34 Pa). This represents a more moderate distribution of WSS (with lower spatial and temporal gradients) at the toe of the ETS component of the SQA model, which may reduce the probability of atherosclerotic lesion development, since it has been postulated that the progress of atherosclerotic lesions and growth of plaque formation occurs mainly in regions of extreme (either maxima or minima) WSS [45,46].

3.2.3. WSS distribution on the coronary artery bed of the ETS anastomoses

On the coronary artery bed opposite to the heel of the ETS anastomosis \((-2 \leq \xi_3 \leq 3\) mm), the WSS was slightly higher in magnitude in the SQA model than that in the conventional ETS model, as shown in Figs. 8c and 10c, respectively. This is due to the fluid flowing through the coronary artery segment between the two anastomoses of the novel SQA model (contrasted with the flow circulation in the conventional ETS model). This moderate increase in the WSS magnitude, conjugated by the absence of the stagnation point on the artery bed in the ETS anastomosis of the SQA model (as mentioned earlier) can reduce the risk of IT and atherosclerosis development at this critical location. On moving downstream along the artery bed, the WSS magnitude was increased, reaching its peak value opposite to the toe of the ETS anastomosis, as shown in Figs. 8c and 10c, and then, was reduced to an asymptotic value in both models (not shown here). These values of WSS at the downstream of the artery bed in the ETS anastomosis of the conventional ETS model \((\xi_3 \geq 6.5\) mm) were higher than those along the toe of the anastomosis within the coronary artery \((\xi_2 \geq 1\) mm in Figs. 8b and 10b) during the forward flow phases, indicating that the velocity profile was skewed toward the artery floor, inducing larger velocity gradients on the floor side. However, this phenomenon was less pronounced in the SQA model (Fig. 8) than in the conventional ETS model (Fig. 10), due to the partial flow from the coronary artery changing the direction of the graft flow toward the coronary artery axis and reducing the velocity profile skew toward the artery bed at the ETS component (as shown earlier in Fig. 5).

3.2.4. WSS distribution in the heel region of the STS anastomoses

Upon comparing the distribution of WSS in the STS anastomosis of the SQA model and the parallel STS model, shown in Figs. 7a and 9a respectively, the qualitative distribution of the WSS was mostly similar in the heel region of the STS anastomosis of the two models. However, at time \(t_5\) the WSS in the graft over the heel region \((\xi_1 \leq 0\) was positive (with respect to \(\xi_1\) coordinate at the heel, shown in Fig. 4) in the parallel STS model, while it had negative values in the STS component of the SQA model. This was consistent with the opposing flow directions over the heel region in the two models at this time instant (as shown in Fig. 6).

3.2.5. WSS distribution in the toe region of the STS anastomoses

At the toe of the STS anastomosis of the two models, the distribution of WSS was about the same. However, at times \(t_4\) and \(t_5\), the magnitude of WSS on the graft and the coronary artery walls had different distributions in the two models, as demonstrated in Figs. 7b and 9b. In the STS component of the novel SQA model, due to the strong backflow through the coronary artery at these time instants, the WSS magnitude was higher on the coronary artery wall at the toe of the anastomosis \((\xi_2 \geq 0\) than on the graft wall
at the toe ($\xi_2 \leq 0$), where a small recirculation zone resulted in a low WSS magnitude at time $t_5$ (as shown in Fig. 6). However, in the parallel STS model, owing to the large vortices within the coronary artery, the WSS magnitude was lower on the artery wall at the toe ($\xi_2 \geq 0$) than on the graft wall ($\xi_2 \leq 0$) on account of the higher flow rate in the graft (than in the coronary artery).

### 3.2.6. WSS distribution on the coronary artery bed of the STS anastomoses

On the coronary artery bed, during the acceleration phase and the peak flow, the magnitude of WSS was higher in the parallel STS model than that in the STS anastomosis of the SQA model (e.g., 5 versus 2 Pa at $\xi_3 = 8.8$ mm at time $t_2$), as shown in Figs. 9c and 7c, respectively. During the deceleration phase ($t_3$), in the parallel STS model (Fig. 9c), the WSS magnitude was generally low, with a region of negative values at the middle of the bed opposite to the anastomosis junction ($−0.9 \leq \xi_3 \leq 6.2$ mm), indicating that the shear was directed proximally. This was due to the large vortex, formed in the coronary artery at this time instant (see the online Supplementary Material). The zero value of WSS at point $\xi_3 = 6.2$ mm corresponded to the stagnation point, created just distal to the vortex. However, in the STS anastomosis of the SQA model
(Fig. 7c) at this time ($t_3$), the results showed fluctuations in the WSS values on the artery bed. This could possibly be due to the inaccuracy induced into the PIV results, caused by scattering of the laser light sheet due to the unevenness of the inner surfaces of this fabricated model at this particular region. Another possible reason for this inaccuracy can be the accumulation of some seeding particles in this low-velocity region, which would induce errors into the velocity measurement, and consequently, the WSS calculation. These oscillations also resulted in fluctuations in the sign of WSS, due to the very small (nearly zero) velocity magnitude at this time ($t_3$). Some fluctuations in the magnitude of WSS could also be observed in this model at other time instants. At time $t_5$, the WSS magnitude on the artery bed of the parallel STS model (Fig. 9c) was much lower than that in the STS anastomosis of the novel SQA model (Fig. 7c), due to the recirculation zones and low velocity regions created within the coronary artery in the parallel STS model (as shown in Fig. 6c). However, during the backflow phase ($t_6$), both models showed similar WSS distributions.

FIG. 8. WSS distribution in the ETS component of the SQA model, along the (a) heel, (b) toe, and (c) artery bed.
4. Discussion

A novel CABG coupled SQA model has been designed and fabricated, and its flow fields and WSS distributions were studied and compared to those of the conventional ETS and parallel STS anastomoses. PIV measurements were utilized in this investigation to complement the previous computational modeling of the blood flow through this novel CABG design [38,39], from which more complex HPs had been derived to quantify hemodynamics in the novel CABG coupled SQA design models with a variety of the design parameters, including the anastomotic angle at the ETS component of the SQA design, the distance between the two (STS and ETS) anastomoses, and the out-of-plane grafting.

Fig. 9. WSS distribution in the parallel STS model, along the (a) heel, (b) toe, and (c) artery bed.
4.1. Features of the coupled SQA graft design

(i) During the acceleration and early deceleration phases the flow at the ETS anastomosis is smooth as part of the graft flow (which is diverted into the coronary artery at the STS anastomosis of the SQA model) lifts up the flow coming from the graft at the ETS anastomosis and directs it smoothly into the coronary artery. By this means, there is no impact on the artery bed and no flow recirculation at the heel. It is to be noted that the flow impact on the artery floor is believed to be injurious to the endothelium and a contributing factor to graft failure [14], and a stagnation point is always associated with a low WSS and high spatial WSS gradient region, which may contribute to IT and atherosclerosis development, and may increase the risk of aggregation of red blood cells [12,13].
(ii) During the late deceleration phase at the distal anastomosis, part of the flow goes back to the STS anastomosis through the coronary artery. There, it lifts up the flow coming into the STS anastomosis junction from the graft, and directs the flow smoothly back into the graft, thereby, avoiding the formation of stagnation point and vortex in the STS anastomosis region, as demonstrated in Fig. 6. It is to be kept in mind that a vortex is always associated with low WSS and high particle residence time which result in platelet activation and IT [10,11,47,48].

(iii) WSS distribution is more moderate in the novel SQA model, especially at the ETS anastomosis, as compared with the conventional ETS model; this can reduce the chance of IH development.

(iv) Moreover, this novel coupled STS-ETS SQA design provides a spare route for the blood flow to the coronary artery, in order to avoid re-operation in case of re-stenosis in either of the anastomoses.

Although this novel design does involve one additional anastomosis (and prolonged operation time), the distinct advantages observed in the flow field and WSS distribution in this coupled SQA design may motivate the use of this novel design over conventional ETS anastomosis.

Also, a recent study on topology optimization with application to bypass graft geometries [49] has shown that among the different investigated geometric models, the optimized bypass graft has two outlet channels, resembling our proposed novel sequential anastomoses design. This optimized model has shown the best performance among the studied models in terms of WSS, with a relatively uniform WSS distribution and the smallest maximum value of WSS. The study has referred to our proposed novel sequential anastomoses design and concluded that their topology optimization method leads to this sequential anastomotic design too.

5. Conclusion

A novel CABG anastomotic configuration has been designed, and its flow fields and WSS distributions were investigated by utilizing PIV method. This in vitro investigation complements the previous computational studies, and constitutes another necessary step toward the clinical application of this novel coupled SQA design. So far, an innovative CABG design has been provided and its hemodynamic benefits have been validated (by this in vitro experiment and the earlier computational studies [38,39], as well as by some topology optimizations [49]). At this point and prior to the clinical adoption of this novel design, in vivo animal trials are warranted, in order to investigate the biological effects and overall performance of this anastomotic configuration in vivo.

Funding

None.

Ethical approval

None.

Conflict of interest

The authors affirm that there are no financial and personal relationships or involvement with any commercial organization that could inappropriately influence or bias the present manuscript.

Appendix

A.1. Validation of PIV method

In order to validate the accuracy and precision of the PIV measurements, a comparison was conducted between the PIV results for a fully developed sinusoidal flow ($Q = 140(1 + \sin(4.2t))$ ml/min)
in a straight tube (with an inner diameter of 3 mm) and the corresponding Womersley solution [50].

As shown in Fig. 11, a good qualitative match was observed between the experimentally obtained and analytical velocity profiles at different phases of the flow cycle. The quantitative comparison of the results revealed an average relative difference of 8.6% in the velocity magnitude between the two sets of data. This reasonable discrepancy confirmed that the accuracy of the PIV results was acceptable.

A.2. Validation of the WSS estimation method

In order to validate the accuracy of the WSS estimation method, a comparison was conducted between the WSS values estimated from PIV results for the abovementioned fully-developed sinusoidal flow in a straight tube (with an inner diameter of 3 mm) and the corresponding analytically calculated WSS values obtained from the Womersley solution at the very time instants for which the PIV results were validated (i.e., \( \tau/T = 0.2, 0.75, 0.856, \) and 1, as shown in Fig. 11).

Moreover, the WSS calculation method was applied to data points obtained from the analytical solution (rather than the PIV data) at the abovementioned time points, and compared with those obtained from the analytical solution and from the PIV data (Table 2). The WSS values obtained from the analytical data points do not include (and are not influenced by) the PIV measurement error; hence, they can indicate the accuracy/error of the WSS calculation method. Whereas the WSS values obtained from the PIV results are highly influenced by the near-wall error of the PIV measurement results, as shown in Table 2, the WSS values estimated based on the analytical data points showed a good agreement with the analytically calculated WSS values with an error of <13% (except for \( \tau/T = 0.75 \) with an error of <33%), which was mainly due to the small magnitude of the WSS at this time point, resulting in higher percentage of error even for small differences between the two sets of data). However, the WSS values estimated based on the PIV data (which include the near-wall PIV measurement errors) had relative errors of up to about 55% with respect to the analytically calculated WSS values. Nevertheless, the trend of the WSS variations was identical among the different sets of data which made the comparison of the WSS variation among different geometric models valid in the present study.

Appendix B. Supplementary data

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.medengphy.2014.06.024.

References


