An investigation of pulsatile flow in a model cavo-pulmonary vascular system

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SUMMARY

The complexities in the flow pattern in a cavo-pulmonary vascular system—after application of the Fontan procedure in the vicinity of the superior vena cava, inferior vena cava, and the confluence at the T-junction—are analysed. A characteristic-based split (CBS) finite element scheme involving the artificial compressibility approach is employed to compute the resulting flow. Benchmarking of the CBS scheme is carried out using standard problems and with the flow features observed in an experimental model with the help of a dye visualization technique in model scale. The transient flow variations in a total cavo-pulmonary connection (TCPC) under pulsatile conditions are investigated and compared with flow visualization studies. In addition to such qualitative flow investigations, quantitative analysis of energy loss and haemodynamic stresses have also been performed. The comparisons show good agreement between the numerical and experimental flow patterns. The numerically predicted shear stress values indicate that the pulsatile flow condition is likely to be more severe than steady flow, with regard to the long-term health of the surgically corrected TCPC. Copyright © 2008 John Wiley & Sons, Ltd.

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1. INTRODUCTION

The human vascular system comprises the heart and blood vessels, and functions in such a way as to supply each organ with a certain amount of blood, which may vary depending on physiological conditions and organ demands. Haemodynamic factors are hypothesized to be significant in normal adaptive response of blood vessels to chronic changes in physiological demands, and also in maladaptive responses leading to vascular diseases [1]. Specifically, it is postulated that flow
recirculation, high particle residence time and low wall shear stress (WSS) are responsible for the localization of atherosclerotic plaques in the regions of complex flow in arteries [2, 3]. Rathishkumar [4] has analysed the WSS under steady flow in a multiply dilated vessel. The study reveals the variation of WSS along the vessel for different degrees of dilation with two-dimensional (2D) assumption. Considering the strong correlation between the localization of atherosclerosis and arterial WSS, quantifying the recirculation zone size and WSS is important for understanding the development of arterial diseases. The inadvertent introduction of such regions of elevated blood residence time in the design of interventional devices and prostheses (e.g. artificial hearts and valves, stents, bypass grafts) might lead to their ultimate failure. The investigations into WSS and WSS gradients in the femoral bypass graft have been carried out with different geometric parameters [5, 6]. The study indicates the importance of understanding the interaction between blood cells and the vessel wall, for predicting pathological changes in the lumen wall with the near wall blood flow. In the present study, experimental and computational model studies have been carried out to understand the haemodynamics in the cavo-pulmonary vascular system.

The complexities of biofluid dynamic problems in general, and the prediction of WSS in the recirculating and stagnant zones of such flows in particular, have been modelled by several authors using CFD techniques [5–13]. Kim et al. [14] reported that seven out of 1000 babies have congenital heart defects, resulting in improper flow of deoxygenated blood to the lungs. The Fontan procedure has emerged as a surgical means to correct such heart defects. This procedure was initially described by Fontan and Baudet [15] as a non-anatomic correction of tricuspid atresia, where the venous blood is directed to the lungs without passing through the right ventricular chamber. The idea behind Fontan procedure is to reroute the blood flow around the congenital passage blocks; thus, allowing the diversion of systemic venous returns directly to the pulmonary arteries. This enables the blood to be processed by the lungs. The Fontan approach has undergone a series of modifications over the years. The modifications started with the atrio-pulmonary connection and continued to more rational alternatives, where the SVC–IVC (superior vena cava–inferior vena cava) were connected to the pulmonary circulation without passing through the right atrium. This process is often referred to as the total cavo-pulmonary connection (TCPC). The improvement in haemodynamics for the TCPC over the conventional atrio-pulmonary anastomosis has been first demonstrated by de Leval et al. [16] with an in vitro study.

The effects of WSS in the blood vessels were studied by many authors [17–20]. It was found that exposure to a high level of WSS causes activation of blood platelets, which if combined with flow recirculation, increases the possibility of thrombosis within the blood vessels. Hedrick et al. and Shirai et al. [20, 21] observed increased levels of thrombus formation for patients (3–20%) with atrio-pulmonary and cavo-pulmonary connections. Shirai et al. [21] have reported a 19% incidence of thrombus formation in a group of patients with TCPC connections. Morgan et al. [22] performed an in vitro study with 3D MRI phase velocity data to calculate shear stresses at the vessel wall. They recorded average WSS values ranging between 5 and 40 dynes/cm$^2$ along the wall over the cardiac cycle and suggested that high values of WSS could affect the longevity of repair. Khunatorn et al. [23] studied the influence of connection geometry and SVC–IVC flow rate ratio on flow structures within the TCPC. The flow was considered as Newtonian and steady. It has been noted that little published data are available on WSS values in the vicinity of the TCPC connection for different flow rate ratios. Liu et al. [24] estimated the energy loss at TCPC based on in vitro experiments (with models similar to the actual TCPC geometry) and computational fluid dynamic studies. The numerical study was carried out with the assumptions of steady, Newtonian flow and rigid vessel walls. In vitro experiments and CFD studies yielded the same energy loss.
trends. Masters et al. [25] studied the power loss in TCPC for rigid and compliant vessel wall assumptions and reported that the power loss is decayed as caval offset increased in the rigid wall case as compared with the case of a compliant wall. Sheu et al. [26] predicted the hydraulic and power losses in the vascular Fontan system with Newtonian and steady flow assumptions. Although some basic studies have been reported on flow through TCPC, the lack of sufficient data for transient flow is apparent. To make a preliminary assessment of pulsatile flow through a TCPC, a geometry similar to the one used by Sheu et al. [26] is considered here. The paper also serves as a means to thoroughly test the characteristic-based split (CBS) method of solution for modelling pulsatile flows. The numerical flow patterns and flow rates obtained have been compared with the pulsatile flow measurements in a model TCPC.

2. THE MODEL PROBLEM

The schematic diagram of a typical TCPC is shown in Figure 1. According to the anatomy of TCPC, blood flows from the SVC and IVC are confluent at the junction. This is followed by a downstream flow into the left pulmonary artery (LPA) and the right pulmonary artery (RPA). The LPA and RPA arteries bifurcate into the left upper and lower pulmonary arteries and, right upper and lower pulmonary arteries, respectively. These branches proceed further downstream to the left and right lungs. For better flow distribution and energy dissipation, surgical reasoning suggests the enlargement of IVC and SVC anastomoses (perpendicular connections of IVC and SVC with pulmonary arteries). However, anastomoses without IVC/SVC offset could lead to some adverse
effects at the junction [26]. Hence, IVC/SVC offset is considered to be a key parameter. In the blood vessels, flow pattern and the direct haemodynamic correlative studies are important for understanding localization of the vascular diseases.

In order to understand the haemodynamic flow patterns in the vicinity of TCPC after the Fontan procedure, an experimental model is also constructed in the present study as shown in Figure 2. In the present study, experimental and numerical investigations have been carried out to analyse the flow pattern for this geometry under steady and various pulsatile flow conditions. The rheological properties of blood normally make it a non-Newtonian fluid, with viscosity depending on vessel diameter and shear rate. However, in large vessels, blood behaviour can be considered as Newtonian [27]. Hence, as a first cut approximation, the flow is assumed to be Newtonian, incompressible and fully developed. For the range of Reynolds numbers ($Re < 600$) studied, the blood flow is also considered to be in the laminar regime. Moreover, the walls of the blood vessels are assumed to be rigid.

3. NUMERICAL PROCEDURE

The non-dimensional form of governing equations for incompressible flow using an artificial compressibility scheme can be written as follows:

**Continuity equation:**

$$\frac{1}{\beta^2} \frac{\partial p}{\partial t} + \frac{\partial (\rho u_i)}{\partial x_i} = 0$$  \hspace{1cm} (1)

**Vector momentum equation:**

$$\frac{\partial u_i}{\partial t} + \frac{\partial}{\partial x_j}(u_j u_k) = -\frac{\partial p}{\partial x_i} + \frac{1}{Re} \frac{\partial \tau_{ij}}{\partial x_j}$$  \hspace{1cm} (2)
where

\[
\tau_{ij} = \left( \frac{\partial u_i}{\partial x_j} + \frac{\partial u_j}{\partial x_i} - \frac{2}{3} \frac{\partial u_k}{\partial x_k} \delta_{ij} \right)
\] (3)

and \( Re \) is the Reynolds number. In the above equations, \( \delta_{ij} \) is the Kronecker delta, \( u_i \) are the velocity components, \( x_i \) are the coordinate axes, \( \tau_{ij} \) are the deviatoric stress components, \( p \) is the pressure, and \( \beta \) is an artificial compressibility (AC) parameter.

The artery walls are assumed to be rigid with no-slip velocity boundary conditions on them. Fully developed parabolic velocity profiles at SVC and IVC are considered for steady state solutions. For pulsatile motion, the velocity profile is allowed to oscillate at SVC and IVC inlets at an identical frequency. This is achieved by prescribing a sinusoidal variation with respect to time for the maximum velocity value on the axis, in other words, the inlet velocity profile is assumed to be parabolic at each instant with the maximum value on axis varying with time as mentioned above. An AC scheme employing the finite element method, known as the CBS scheme has been adopted [28–31] for solving the governing equations. This is a fully explicit scheme, essentially containing three steps. In the first step, an intermediate velocity field is established; in the second step, the pressure is obtained by solving a modified continuity equation and finally the intermediate velocities are corrected to get the final velocities.

The steps may be written as follows (non-conservation form).

**Step 1:** Intermediate momentum

\[
\Delta \tilde{u}_i = \tilde{u}_i - u^n_i = \Delta t \left[ -(u_j) \frac{\partial u_i}{\partial x_j} + \frac{1}{Re} \frac{\partial \tau_{ij}}{\partial x_j} \right] + \frac{\Delta t^2}{2} u_k \frac{\partial}{\partial x_k} \left[ (u_j) \frac{\partial u_i}{\partial x_j} - \frac{1}{Re} \frac{\partial \tau_{ij}}{\partial x_j} + \frac{\partial p}{\partial x_i} \right]
\] (4)

where \( u^n_i = u_i(t^n) \); \( \Delta t = t^{n+1} - t^n \) and \( \tilde{\cdot} \) indicates an intermediate quantity.

**Step 2:** Pressure

\[
\left( \frac{1}{\beta^2} \right)^n \Delta p = \left( \frac{1}{\beta^2} \right)^n (p^{n+1} - p^n) = -\Delta t \left[ \frac{\partial u_i^n}{\partial x_i} + \theta_1 \frac{\partial \Delta u_i^*}{\partial x_i} - \Delta t \theta_1 \left( \frac{\partial^2 p^n}{\partial x_i \partial x_i} + \theta_2 \frac{\partial^2 \Delta p}{\partial x_i \partial x_i} \right) \right]
\] (5)

Selection of an AC wave speed \( \beta \) for an explicit scheme is discussed in [28].

**Step 3:** Momentum correction

\[
u^{n+1}_i = \tilde{u}_i - \Delta t \frac{\partial p^{n+\theta_2}}{\partial x_i}
\] (6)

where \( 0.5 \leq \theta_1 \leq 1 \) and \( 0 < \theta_2 \leq 1 \). For the fully explicit scheme used here, \( \theta_1 = 0.5 \) and \( \theta_2 = 0 \) are used. In the above equations \( \tilde{u}_i \) is an intermediate quantity. Further details on the CBS method can be found in Zienkiewicz et al. [32]. Although the actual TCPC geometry is three dimensional, for the sake of computational simplicity, a 2D geometry has been considered. It is expected that

the essential features of steady and pulsatile motion within the TCPC can be captured by the 2D simulation, which can be qualitatively compared with the results of experimental flow visualization studies at the same Reynolds number. Unstructured triangular elements have been used for domain discretization with fine grids in high gradient regions such as boundary layers. The time step used in Equations (4), (5) and (6) is locally calculated and is chosen to satisfy the stability conditions [28]. To recover unsteady state, a real time term is added to the third step of the algorithm. In such a dual time stepping arrangement (second order), the pseudo local time step acts as an iterative mechanism to reach a local steady state within each real time step [28]. The real time step is restricted only by the accuracy requirement.

4. EXPERIMENTAL PROCEDURE

The flow pattern in TCPC is investigated using flow visualization by means of a multiple dye technique on an 1:1.6 scale model. Different coloured liquids are sent through the two inlets (i.e. IVC and SVC) at a prescribed flow rate ratio, in order to determine the relative importance of spatial and temporal flow characteristics for steady and unsteady fluid flow through TCPC. Similar methods have been extensively used by earlier researchers for blood flow investigations [33, 34]. The present section describes the details of the model fabrication and set-up, as well as the parameters of flow and experimental procedure employed in the study.

Steady flow experiments by earlier authors [26] show that an optimal offset exists, which results in minimum energy loss. A scaled model with an optimal offset is considered for the present experimental investigation. The diameters corresponding to SVC and IVC streams are 10 mm, and the diameter value for all the pulmonary artery streams is equal to 5 mm. The model has been fabricated using borosil glass with a geometrical configuration as shown in Figure 2. The corresponding flow chart is shown in Figure 3. The offset between SVC and IVC streams at TCPC is 0.4375D (where D is the diameter of SVC/IVC). Here, the offset refers to the shift in the axis of SVC with respect to fixed IVC axis. The IVC and SVC of glass model are connected with two peristaltic pumps, which are capable of delivering pulsating flow at a set flow rate. In order to achieve steady flow, settling tanks have been used in between the pumps and IVC/SVC, whenever desired.

Dynamic similarity is maintained between the physical and numerical investigations by operating at the same Reynolds number. Based on the maximum velocity and diameter of IVC, the Reynolds number has been chosen as 565. This is comparable to the practical Re values as reported by Sheu et al. [26]. The rate of flow through SVC is required to be half of that through IVC. This has been achieved by splitting the flow leading to SVC by means of a T-joint. Such a procedure is advantageous since the same frequency of pulses (and hence rpm of the pumps) needs to be maintained at IVC and SVC inlets during the experiments. The dissimilar inflow between IVC and SVC creates a complex flow pattern at the confluence point of the IVC and SVC streams. By the addition of an appropriate dye to water, the colour of the liquid arriving at each inlet (IVC and SVC) is made different. The experimental study has been carried out for both steady and pulsatile flow conditions.

It is observed that the flow at the confluence region reaches a steady state if steady inflow inlet conditions are forced. The confluence region and the streak lines in the pulmonary artery have been video graphed using a digital camera (Sony DCR-HC90E) at 1024 × 576 pixel resolution and a frame rate of 18 pictures per second. Still pictures of the flow have been obtained using an
image processing software (a combination of windows media player and a MATLAB code) and averaged over frames for eliminating minor flow oscillations that may be present at steady state.

The steady flow investigation provides valuable insight on the impact of various flow and geometric parameters of the TCPC connection. However, steady flow is not physiologically relevant for blood flow circulation. Therefore, pulsatile motion experiments have also been conducted in the present study with the help of peristaltic pumps, without the use of settling tanks (see Figure 3). The frequency of pulsation is set as 0.78 Hz. Averaged pictures at different phases of the cycle have been determined using image processing software from the video clips of the flow pattern.

5. BENCHMARK PROBLEMS

Truskey et al. [35] considered flow over a backward facing step for examining the response of endothelial cells to spatially non-uniform flows. A backward facing step with an expansion ratio of 2.37 was considered for this purpose. In the present study, a grid with 10,822 elements and 5704 nodal points has been used to solve this benchmark problem. Experimental results of recirculation length obtained by Truskey et al. [35] are compared here with the present results. The predicted non-dimensionalized lengths of recirculation zone for different Reynolds numbers agree well with the available experimental results as shown in Figure 4.

As the study presented in this paper is predominantly transient, the next benchmark problem considered is the transient flow past a circular cylinder at a Reynolds number of 100. The problem definition is standard [36] with a total domain length of 16 diameters and width of 8 diameters. The cylinder is placed at the centre of the domain at a distance of 4 times the diameter from the
inlet. The boundary conditions used are: no slip on the cylinder surface and zero normal velocity components along the two horizontal sides. An initial condition of zero pressure and a non-dimensional horizontal velocity component of 1.0 are used. The unstructured mesh used consists of 19650 unstructured elements with fine mesh along the downstream wake of the cylinder.

Figure 5 shows the drag and lift coefficient distributions with respect to time. It is evident that vortex shedding is established after some preliminary transient flow. Both non-conservation
and conservation forms of the incompressible flow models are used. In the figure, the average lift coefficient is observed to be zero and the average drag coefficient is 1.512. This is in close agreement with many other accurate schemes such as the orthogonal subscale stabilization schemes of Codina et al. [36].

6. RESULTS FOR CA-VO-PULMONARY SYSTEM

An important parameter used to obtain the influence of the TCPC offset, referred to as the energy loss coefficient, is defined [24] as

\[
E_{\text{loss}} = \sum P_{i, \text{inlet}} Q_{\text{inlet}} - \sum P_{i, \text{outlet}} Q_{\text{outlet}}
\]

(7)

where \( P_i \) is the total pressure and \( Q \) is the flow rate.

The effect of the TCPC offset on the total energy loss coefficient is investigated for \( Re = 565 \), by varying the offset from 0.3125 \( D \) (5 mm) to 0.4375 \( D \) (7 mm) to the right of the axis of IVC and 0.375 \( D \) (6 mm) to the left of IVC, where the reference length scale, \( D \), is the diameter of IVC (16 mm) used in the numerical calculations. The axes of symmetry of both the IVC and SVC are kept normal to the pulmonary artery (see Figure 1). The input velocity at SVC is considered to be half of the corresponding value at IVC inlet. Steady flow predictions are first compared with those of Sheu et al. [26]. Then the numerical model is extended for simulating unsteady flow in TCPC with a periodic motion. Numerical results are also compared with the results obtained using the commercial CFD software Fluent 6.3. Some typical flow patterns and quantitative results are presented and discussed in this section.

6.1. Investigation of steady flow

Before carrying out pulsatile flow simulation, the mesh size needed for the calculation is decided, based upon a grid sensitivity study for the steady flow condition. Table I shows the percentage of energy loss for the no-offset case obtained with three different meshes. As the difference between the results obtained on Mesh-B and Mesh-C is negligibly small, Mesh-B has been considered sufficient for further simulations. The symmetric configuration of SVC/IVC forms a stagnation region at the confluence, which causes a diversion of the flow into the pulmonary arteries. As the SVC–IVC flow rate ratio decreases, the stagnation area moves towards the SVC as a consequence of the increased momentum of the IVC flow stream. More work is therefore needed to maintain the flow through the SVC and connection area, which results in an increased overall pressure drop. In order to avoid greater energy loss and to have a smooth flow, SVC–IVC connection with different offsets is investigated. The optimal offset size has been decided on the basis of minimum

<table>
<thead>
<tr>
<th>Grid</th>
<th>No. elements and nodes</th>
<th>( Re_{\text{element}} )</th>
<th>Energy loss (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>141 170, 72 277</td>
<td>4.744</td>
<td>26.824</td>
</tr>
<tr>
<td>B</td>
<td>187 180, 95 544</td>
<td>3.067</td>
<td>30.363</td>
</tr>
<tr>
<td>C</td>
<td>251 116, 127 918</td>
<td>3.012</td>
<td>30.606</td>
</tr>
</tbody>
</table>

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Figure 6. Steady flow through the TCPC connection. Offset: 0.4375D (7 mm) (SVC moved to right of IVC) S—stagnation point, R—recirculation: (a) (Sheu et al. [26]), reproduced with permission and (b) present numerical solution.

Table II. Energy loss for different offsets.

<table>
<thead>
<tr>
<th>Offsets for SVC–IVC connection</th>
<th>Percentage of energy loss $(E_{\text{loss}}/E_{\text{in}})$ (steady flow)</th>
<th>Percentage of energy loss $(E_{\text{loss}}/E_{\text{in}})$ (pulsatile flow)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0 mm</td>
<td>30.606</td>
<td>41.83</td>
</tr>
<tr>
<td>5 mm</td>
<td>24.499</td>
<td>26.74</td>
</tr>
<tr>
<td>7 mm</td>
<td>23.389</td>
<td>22.62</td>
</tr>
<tr>
<td>−6 mm</td>
<td>24.863</td>
<td>24.18</td>
</tr>
</tbody>
</table>

energy loss. The steady flow patterns in terms of streamlines obtained in the present simulations and those of Sheu et al. [26] for the same offset of 7 mm (0.4375D) are shown in Figure 6. The agreement between the two sets of results is observed to be very good.

Table II shows the percentage of energy loss for different offsets under steady and unsteady flow conditions. Based on these computations, it is observed that zero offset leads to larger energy loss compared with other offsets this is in-line with the observation found in the literature [22]. The total energy loss coefficient for 7 mm offset is obtained as 1.037, which compares excellently with the value of 1.032 reported by Sheu et al. [26] for steady state flow condition. Masters et al. [25] reported that the decrease in power loss is effective for offsets of up to half the caval diameter. The increase in offset size is not effective for offset beyond half the caval diameter. Hence, the dimensionless offset of 0.4375D is expected to be more suitable for the SVC–IVC connection. Table III shows the sizes of recirculation zones at various offsets. As seen in the table, the agreement between the present results and the published data of Sheu et al. [26] is good. In addition to the energy loss, WSS is also computed in the pulmonary arteries. Figure 7 shows the non-dimensional WSS along the right and left pulmonary arteries for the offset of 0.4375D. The distribution is marked with rapid variation close to the corners and the junction. This is expected as flow separation takes place close to these points. The locations 1, 2, 3 and 4 corresponding to peak (positive or negative) local shear stress values are shown in Figure 6(b). At location 1 three fluid streams are in close proximity, which are: the fluid stream emanating from IVC and flowing
CAVO-PULMONARY VASCULAR SYSTEM

Table III. Size of re-circulation region in the upper wall of LPA.

<table>
<thead>
<tr>
<th>Offsets for SVC–IVC connection (mm)</th>
<th>Sheu et al. [26]</th>
<th>Present study</th>
</tr>
</thead>
<tbody>
<tr>
<td>5</td>
<td>2.38D</td>
<td>2.50D</td>
</tr>
<tr>
<td>7</td>
<td>2.15D</td>
<td>2.30D</td>
</tr>
<tr>
<td>−6</td>
<td>2.20D</td>
<td>2.48D</td>
</tr>
</tbody>
</table>

Figure 7. Steady flow through the TCPC connection. Non-dimensionalized wall shear stress distribution: (a) wall shear stress along upper wall and (b) wall shear stress along lower wall.

Towards RPA, the fluid stream from IVC going towards LPA and the fluid stream flowing from the SVC to LPA. As location 1 borders these three fluid streams, the local shear stress reaches a maximum value at this location. Location 4 corresponds to the corner situated very close to the stagnation point of the strong recirculation zone in the offset region. Therefore, the shear stress at this corner location is also very high. Points 2 and 3 correspond to the corners where flow separation occurs. It is evident that the three streams, viz. SVC to LPA, IVC to LPA and IVC to RPA, are also identifiable in the experimentally visualized image (Figure 8(b)), as predicted by the simulations (Figure 8(a)). The size and shape of the recirculation zone at the junction also qualitatively matches with the experimental results.

A still image obtained through the processing of video recording during the steady state experiment is presented in Figure 8 and it is also compared with the corresponding numerical results. The figure shows flow details near the confluence of TCPC. The size of recirculation zone obtained numerically in the LPA is 2.30D and the corresponding size obtained from measurement is 2.33D, where D is the diameter of IVC/SVC (experimental model). The numerical and experimental results are indeed quite close. More comparisons for the recirculation zone sizes between the present predictions and the results of Sheu et al. [26] are presented in Table III. In Figure 8(b), at the confluence of TCPC junction, one can clearly see the mixing of coloured liquids, implying that the streams entering through SVC and IVC partially mix with each other and then redistribute in the form of streams going towards RPA and LPA. These features from visualization are in good

Figure 8. Steady flow through the TCPC connection: (a) numerical simulation \((R = 2.30D)\) and (b) experimental flow pattern \((R1 = 2.33D)\).

agreement with the patterns observed in numerical simulation. In general, the flow patterns from the experimental investigation correlate well with those of the numerical results.

6.2. Pulsatile flow results

In this section, pulsatile flow conditions in the TCPC junction are thoroughly studied. Flow rate is fixed depending on the desired time-averaged Reynolds number of the flow. The experiments for pulsatile flow condition are conducted for a frequency of 0.78 Hz (corresponding to a non-dimensional pulse period, \(T = 1\)). Quantitative estimates of the flow rate are made using the measured outflow rate in the experiments for both the left and right branches and these are compared with the corresponding numerical predictions. In the 2D numerical simulation, the volume flow rate ratio of left to RPA is obtained as 1.28. The measured experimental volume flow rate ratio is 1.347, which is in reasonable agreement with the numerical results.

Figure 9 shows the numerically predicted flow patterns (streamlines) at different phases of the cycle. Based on instantaneous pictures, movement of vortices is visualized over a pulsation cycle. The corresponding experimentally visualized dye patterns (streaklines) at approximately similar instances within a cycle are shown in Figure 10. It is evident from these numerical and experimental patterns that the recirculation zones at the junction as well as at the upper part of LPA and the lower part of RPA, also grow and diminish within a pulsation cycle. It is observed that the stream from IVC to RPA moves continuously with respect to time, although the instantaneous flow rate of this stream may fluctuate within the cycle. On the other hand, the streams going from SVC to LPA or IVC to LPA are intermittent. When the SVC stream is in motion, the IVC–LPA stream rolls up as a recirculation vortex at the junction; the corresponding recirculation zone on the upper part of LPA has maximum strength. On the other hand, when the IVC–LPA stream is flowing with full strength (see Figures 9(b) and (c) corresponding to \(t = T/4\) or \(T/2\)), the SVC–LPA stream almost stops. Owing to such pulsating variation of flow in the junction region, the associated shear stresses also fluctuate significantly. Intermittent movement of the IVC–LPA and SVC–LPA streams and the corresponding growth/decay of the strong recirculation vortex at the junction are also seen very clearly in the experimental streak line patterns of Figure 10.
Figure 9. Unsteady flow through the TCPC connection for $T = 1$. (Flow pattern at different instances.)
Figure 10. Unsteady flow through the TCPC connection, Dye visualization technique at different instances over the cycle.
A detailed study is now carried out focusing on the quantities of interest in the vicinity of the TCPC connection. For conducting a pulsatile flow simulation, a steady state solution is used as an initial condition. The explicit scheme is iterated with a local pseudo time step to a maximum of 800 steps to satisfy the mass balance as accurately as possible. Standard problems such as flow past a circular cylinder need only about one hundred pseudo time iterations. As the variation here is more intense and rapid, a higher number of pseudo time steps is used. The pseudo steady state is assumed to have been achieved if the $L^2$ norm of pressure error is of the order of $10^{-6}$ or less. This convergence to a pseudo steady state helps to achieve accurate mass balance at every real time step for the pulsatile flow simulation. The non-dimensional sinusoidal inlet velocity variation for a non-dimensional pulse period of $T = 1$ with an amplitude of 1 is used at the IVC inlet. This corresponds to an actual frequency of 1.086 Hz. The time-dependent variation of the velocity at the IVC inlet is given as $0.5 \times V_{\text{max}}[1 - \cos(2\pi t/T)]$, where $t$ is the real time. A similar pulse with half the amplitude is prescribed at the SVC inlet, which is in phase with the IVC inlet pulse.

Figure 11 shows the time history of pressure at the mid-point of the inlets. As seen in the figure, the pressure value changes rapidly at the beginning before reaching an approximate periodic state. The initial rapid change may be attributed to the effects of initial flow development within the TCPC geometry.

In Figures 9 and 10, it was seen that in pulsatile flow, periodic growth and decay of recirculation regions occur in the TCPC junction region. As a consequence of such flow variations, WSS also undergoes periodic changes in direction. Therefore, the blood vessel walls are exposed to temporal and spatial gradients of WSS, which could be construed as a more adverse situation for the health of endothelial cells than the steady flow case. In fact, prolonged exposure to such alternating stresses may lead to damage in the artery walls. Figure 12 shows the non-dimensional WSS averaged over a cycle along the upper (Figure 12(a)) as well as the lower walls (Figure 12(b)) of the arteries. The time-averaged shear stress variations are consistent with the steady state results. The values are fairly constant near the exit and they rapidly change close to the junction. The shear stress also changes sign close to the junction, as a recirculation zone appears in both RUPA and LUPA near the junction. Similar explanation applies to the shear stress distribution along RPA and LPA. It is
noted that the magnitude of average WSS is generally lower for the pulsatile flow than that for the steady state. This is an important point as the reliability of steady state solutions for representing the time-averaged flow features, may be questionable. As mentioned before, lower WSS values in the long term can lead to arteriosclerosis and other complications. Thus, the transient features observed in the present study appear to be extremely important for the long-term health of the repaired blood vessels.

Figure 13. Unsteady flow through the TCPC connection, $T=1$. Instantaneous non-dimensionalized wall shear stress along the wall over a cycle (RUPA—left and LUPA—right).

To visualize the temporal variation of shear stress, instantaneous values over a cycle are plotted along the walls of right and left upper pulmonary arteries, in Figure 13. Spatial gradients of WSS are the largest in the immediate vicinity of the flow separation point (close to the junction) and reattachment point next to the recirculation zone. The shear stress changes from negative to positive.
values close to the recirculation region. It is evident that at any spatial location, the shear stress also undergoes temporal oscillations, due to the pulsatile nature of flow through the TCPC.

Figures 14–16 show the results over a pulsatile cycle, for a dimensionless cycle time of $T = 0.5$. This situation corresponds to a pulse with double the frequency to that of the previous case. This
Figure 17. Location used for FFT analysis.

case is studied here to examine the effects of the frequency on shear stress and flow patterns as the frequency can be double the normal value during exercise (Pedersen et al. [37]). The present results indicate that even at the higher pulsation frequency (corresponding to $T=0.5$) flow features are quite similar to the case of $T=1.0$. The flow pattern shown in Figure 14 exhibits very similar characteristics as the $T=1$ case; however, the sizes of recirculation zones are smaller than those observed for $T=1$. WSS patterns shown in Figures 15 and 16 also show variations, which are somewhat similar to the patterns for $T=1$. However, the magnitudes are different. In general, pulsation occurring at higher frequency will be more detrimental to the health of the repaired vessels.

The FFT plots of pressure and velocity components at point 1 (see Figure 17) in the recirculation zone of the TCPC junction are shown for the dimensionless cycle times of $T=1$ and 0.5, in Figures 18(a)–(f). It is clear that higher harmonics of the pulsation frequency appear (particularly in the pressure and $v$-velocity fluctuation), although the maximum amplitude component in the spectra corresponds to the pulsation frequency itself. It is also observed that the amplitudes of all the flow variables decrease for the case with $T=0.5$, as compared with the case of $T=1.0$.

6.3. Comparison between FLUENT predictions and present solutions

The general purpose CFD software, Fluent 6.3, has been used to simulate the blood flow for 7 mm ($0.4375D$) offset, with the same flow parameters and assumptions as discussed above. This exercise is to ensure that the accuracy of the present method is comparable to those of standard software. The computed WSS using the commercial code is compared with the present numerical results in Figure 19 and the agreement between the results is seen to be quite good. The small deviations observed between the two sets of results can be attributed to the differences in the implementation of the exit boundary conditions in the FLUENT software and the present code.
Although the results obtained are similar, due to various controls and flags, FLUENT was slower than CBS in obtaining the final solution.

7. CONCLUSIONS

A suitable offset value of SVC-IVC connection for reducing energy losses has been obtained by numerically analysing TCPC junction, using the CBS scheme. Extension of the model for estimating the quantitative haemodynamic characteristics such as wall shear stress (WSS) and velocity field in the recirculation zone under the pulsating flow condition reveals interesting information about alternating shear stress values close to reattachment point of the flow. It is also identified that the position and size of the recirculation zone is a function of pulsation frequency. Experimental
investigation of flow patterns has been carried out using a dye visualization technique. The flow features identified are in qualitative agreement with numerical results, thereby ascertaining the validity of the CBS method used. The notable difference between the steady and unsteady state results is observed in the WSS values. The time-averaged WSS value obtained using the unsteady state simulation is, in general, lower than the corresponding steady flow value. These differences in the features of the steady and unsteady blood flow in the vicinity of the TCPC are likely to have a strong influence on the long-term health of the repaired blood vessel junction.

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