regions commonly experience low velocity and momentum, hence reversing easily with pressure gradient. The core region possessing fluid flow of greater momentum required a longer time to reverse. The traveling vortex which had gathered momentum during the steep deceleration phase had continued its rightward path approaching the outer wall near the toe region. A small recirculating region was observed at the distal entrance although flow along the arterial floor was still dominantly reversed. As backflow accelerated to its peak \((t = 2.675s)\), the vortex had dissipated upon collision with the outer wall. Flow field within the graft, sleeve and host artery was entirely reversing and parallel to the walls with no separation or stagnation points observed. Shear stresses along the walls were negative and low. As fluid decelerated beyond the peak backflow, a small recirculation region was visible near the outer wall above the toe. Flow further decreased to zero velocity where beyond that point, fluid assumed a forward motion and the cycle was repeated.
Figure 6.3 (a) to (h): Flow patterns in Model 1 from $t = 1.825s$ to $t = 2.175s$
Figure 6.3 (i) to (p): Flow patterns in Model 1 from \( t = 2.25s \) to \( t = 2.7s \)
Figure 6.4: Wall Shear Stress Variations along (a) arterial floor, (b) inner wall and (c) outer wall for Model 1.
Chapter 6: Results and Discussion of Geometrical Variations (Pulsatile Flow)

6.2 Flow Characteristics and Wall Shear Stress Variations in Model 2

Model 2 was designed to incorporate a larger sleeve volume by increasing the sleeve height ($H_s$) alone. As observed from Figure 6.5 (a) to (p), flow patterns throughout the pulsatile flow were similar to those in Model 1. Main characteristics included smooth laminar flow with no separation or stagnation points during the acceleration phases, formation of vortex below the heel at the two peaks, distal movement of vortex and stagnation points into the sleeve during deceleration phases and backflow dominance when average velocity fell in the negative range. Two observable differences include the absence of a recirculating zone at the distal artery entrance during the backflow acceleration phase ($t = 2.65s$) and a smaller recirculating region near the outer wall above the toe during backflow deceleration phase ($t = 2.7s$).

At maximum average velocities ($t = 1.95s$ and $2.25s$), peak shear stress values along the outer wall of Model 2 were lower than those in Model 1 and redistributed over a longer distance as shown in Figure 6.6 (c). This was attributed to an increase in outer wall length between the sleeve entrance and the toe region due to height increase. Fluid entered the sleeve of both models at the same angle but reached the outer walls at different positions. High momentum fluid impinged the outer wall of Model 2 at a position further upstream as compared to Model 1 (which was near the toe). The fluid then traveled downstream along the outer wall with decreased velocity (hence the redistributed lower shear stress values) before exiting the sleeve beyond the toe region. Apart from outer wall discrepancies, no significant differences were observed for wall shear stress variations between Models 1 and 2.
Figure 6.5 (a) to (h): Flow patterns in Model 2 from $t = 1.825s$ to $t = 2.175s$
Figure 6.5 (i) to (p): Flow patterns in Model 2 from $t = 2.25s$ to $t = 2.7s$
Figure 6.6: Wall Shear Stress Variations along (a) arterial floor, (b) inner wall and (c) outer wall for Model 2.
6.3 Flow Characteristics and Wall Shear Stress Variations in Model 3

General flow characteristics in Model 3 were very similar to those observed in Model 1 with a few exceptions. During the deceleration phases (from \( t = 2.075s \) to \( 2.125s \) and \( t = 2.575s \) to \( 2.625s \)), a prominent difference was the weaker and smaller recirculating zone at the distal artery entrance of Model 3, as observed by comparing Figure 6.7 (d) to (f) and Figure 6.3 (d) to (f) of Model 1. Time span of the recirculating zone was also shorter in Model 3. At \( t = 2.125s \), flow in the distal artery entrance was entirely forward while at the same time frame, recirculation was still observed in Model 1. In the backflow phase (\( t = 2.65s \)), vortex in the sleeve of Model 3 was observed to be stronger in size and intensity. No recirculation region was observed along the distal artery entrance as compared to Model 1. At maximum backflow (\( t = 2.675s \)), the vortex was still visible at the region above the toe while in Model 1, the vortex had ceased to exist. As backflow decelerated (\( t = 2.7s \)), the recirculating zone enlarged slightly before giving way to incoming forward flow.

Wall shear stress variations (Figure 6.8) were similar for both models although magnitudes of shear stresses were generally lower for all walls (inner, outer and arterial floor) in the sleeve region of Model 3. This was due to the increased cross-sectional area resulting in lower velocity magnitudes and hence lower shear stress values along these walls.
Figure 6.7 (a) to (h): Flow patterns in Model 3 from $t = 1.825s$ to $t = 2.175s$
Chapter 6: Results and Discussion of Geometrical Variations (Pulsatile Flow)

Figure 6.7 (i) to (p): Flow patterns in Model 3 from $t = 2.25s$ to $t = 2.7s$
Chapter 6: Results and Discussion of Geometrical Variations (Pulsatile Flow)

Figure 6.8: Wall Shear Stress Variations along (a) arterial floor, (b) inner wall, and (c) outer wall for Model 3.
6.4 Flow Characteristics and Wall Shear Stress Variations in Model 5

Model 5 represented an increase in sleeve volume with both diameter and height increment from the baseline model, where \( D_S = 1.5 \, D_G \) and \( H_S = 2 \, D_A \). Flow patterns during the first acceleration phase were similar to that of Model 1. At peak flow (\( t = 1.95s \)), a small recirculating region was observed below the sleeve entrance near the inner wall. Flow separation along the arterial floor occurred at a location opposite the heel with reversed flow of a vortex entering the proximal artery section. At the proximal artery entrance, although velocity vectors exhibited a recirculating pattern, they were extremely weak in magnitude (of order \( 10^4 \, \text{m/s} \)) and could thus be considered relatively stagnant. Shear stress variations along the arterial floor, inner wall and outer wall as shown in Figure 6.11 (a), (b) and (c) respectively were similar to Model 1 although the magnitudes were slightly lower due to the enlarged cross sectional area leading to reduced flow velocities. Along the outer wall, a small peak observed at approximately 20 mm downstream was due to the presence of a small notch at that region. Such irregularity was a design flaw which should be avoided in future studies. The notch perturbed the oncoming flow causing rapid spatial acceleration and deceleration as fluid flowed past it.

As flow began to decelerate (\( t = 2.075 \, \text{s} \)), the recirculating zone near the sleeve entrance broke away from the inner wall and moved downwards, forming a source-like structure as described in the steady flow analysis. The small recirculating velocity vectors at the proximal artery entrance had shifted rightwards and increased in magnitude, forming a vortex slightly greater in size and intensity. In the next time frame, two smaller regions of recirculation were observed, one at the top of the sleeve,
the other at the distal artery entrance (which was smaller in size as compared to Model 1). As flow continued to decelerate to the minimum average velocity ($t = 2.125$ s), the source-like structure had occupied almost the entire sleeve volume while the recirculation zone at the distal entrance had disappeared. This phenomenon was also observed in Model 3 where the sleeve diameter was increased in size.

With the onset of acceleration at $t = 2.15$ s, fluid in the sleeve assumed a very complicated flow pattern. The large source-like structure present in the earlier time frame did not have sufficient time to dissipate. It was observed to have shifted proximally left. Two weak and counter rotating vortices were visible near the heel region. Flow patterns for the remaining acceleration were similar to the first acceleration phase. At the second peak, wall shear stress variations were similar to those in the first peak with slightly smaller magnitudes due to a lower flow rate. As fluid decelerated into the backflow phase, a large vortex was observed to occupy approximately 75% of the sleeve volume ($t = 2.65$ s). The large vortex remained throughout the backflow phase only to be washed out when reperfusion in the forward direction occurred again at the beginning of the next cycle.
Chapter 6: Results and Discussion of Geometrical Variations (Pulsatile Flow)

Figure 6.9 (a) to (h): Flow patterns in Model 5 from $t = 1.825s$ to $t = 2.175s$
Chapter 6: Results and Discussion of Geometrical Variations (Pulsatile Flow)

Figure 6.9 (i) to (p): Flow patterns in Model 5 from t = 2.25s to t = 2.7s
Figure 6.10: Wall Shear Stress Variations along (a) arterial floor, (b) inner wall and (c) outer wall for Model 5.
Chapter 6: Results and Discussion of Geometrical Variations (Pulsatile Flow)

6.5 Discussions

Flow characteristics of the cuff-like geometries at different time intervals were presented in the previous sections. In the following sections, similarities and differences between steady and unsteady models would be discussed. Comparisons with published literature would be performed and evaluation of the models would be carried out.

6.5.1 Steady Flow vs. Unsteady Flow

Steady flow models are widely used as preliminary studies for hemodynamics in blood vessels for two main reasons. Firstly, simulations concerning steady flow are easier to set up and secondly, time required for computation of steady flow results is shorter. Therefore, more flow conditions or geometries can be investigated and the results can be analyzed quickly and more efficiently. However, analysis can only be performed on a qualitative basis and not a quantitative basis since blood flow is pulsatile in nature. Assuming Reynolds number in steady flow to be denoted by $Re_{\text{steady}}$ and mean Reynolds number based on average velocity values in pulsatile flow to be denoted by $Re_{\text{mean}}$, it is to be emphasized that flow patterns at $Re_{\text{steady}} = Re_{\text{mean}}$ do not coincide due to presence of inertial effects in pulsatile flow. Using Model 3 as an example, the phenomenon of a “source-like” structure in the steady sleeve model at $Re_{\text{steady}} = 250$ coincided with the unsteady model at approximately $Re_{\text{mean}} = 23$ when $t = 2.1 \text{s}$ as demonstrated in Figure 6.11. Velocity magnitudes and wall shear stresses were significantly different. Nevertheless, steady flow analysis provided knowledge about the existence of such a flow structure. Varying the Reynolds numbers in steady conditions would thus reveal the different flow characteristics found in a particular sleeve model.
6.5.2 Comparison with Published Literature

Good agreement was found in flow characteristics between the sleeve models and published literature on Miller cuff models. Rowe in 1999 suggested that the geometry due to presence of interposition vein cuffs may alter local hemodynamics in a predictable and advantageous way. Using a mean flow rate $Q_{\text{mean}} = 180 \text{ ml/min}$ corresponding to $Re_{\text{mean}} = 193$, with a waveform of frequency $f = 1.25 \text{ Hz}$ (Figure 6.12), Womersley parameter $\alpha = 3.6$ and period $T = 0.8 \text{ s}$, studies were conducted to compare hemodynamics between conventional bypass models and the Miller cuff models (Figure 6.13).

Figure 6.11: Comparison of flow structures in (a) steady and (b) unsteady flow.

Figure 6.12: Inlet waveform used by Rowe et al. (1999).
Chapter 6: Results and Discussion of Geometrical Variations (Pulsatile Flow)

Figure 6.13: Hemodynamics in (a) conventional bypass model and (b) Miller cuff model (Rowe et al. 1999).

Common characteristics found in the conventional model include laminar flow with stagnation point along the artery floor opposite the heel during acceleration, low velocity with secondary flow components inside the graft hood during peak systole and flow separation at the graft hood and reattachment at the graft toe during deceleration. Fluid motion in the Miller cuff was more dynamic. Flow patterns during
the acceleration phase resembled that of the conventional model. At peak systole, a small recirculation zone was formed at the graft toe, which expanded to become a large vortex occupying most of the cuff region during early deceleration phase. In late diastole the main vortex in the cuff became larger while the rotational velocity appeared to increase in magnitude. As observed by Rowe et al. (1999), the vortex appeared to be coherent with no sign of random particle motions that would suggest turbulent flow. The vortex remained in the cuff during the remaining cycle, only to be completely removed with the arrival of the next cycle, where particle pathlines were observed to be essentially parallel to the cuff walls.

Observations made in the present study were highly agreeable with results published by Rowe et al. (1999), especially on the formation and dissipation times of the vortex. As observed in the sleeve models, the vortex formation began at peak flow and dissipation occurred at the end of the cycle. However, two differences were noted in the present study. Firstly, within a cycle, vortex formation and dissipation occurred twice. This was due to biphasic nature of the coronary bypass graft waveform. If only a single peak of the biphasic waveform was considered, the results would be in better agreement. Secondly, according to results of Rowe et al. (1999), the vortex was formed initially at the graft toe, which eventually expanded to occupy most of the cuff region. As observed from the present study, at peak flow, the vortex originated below the heel region in all sleeve models. The difference in site of origination was due mainly to inherent geometrical differences between the Miller cuff and the sleeve models. As observed from Figure 6.14, due to incorporation of the vein cuff, flow exiting the graft would be highly skewed towards the outer wall, resulting in direct flow separation along the arterial floor opposite the toe. The vortex was partially
formed by fluid traveling back into the proximal artery. On the other hand, the sleeve models were designed to incorporate a smoother transition from the graft to the cuff region, hence flow separation (as observed from all sleeve models) occurred along the arterial floor nearer to the heel. Therefore, vortices formed in the sleeve models generally originate from the heel instead of the toe. Although the main function of the vortex was to enhance mixing, the site of origination could play an important role. In a fully occluded model, fluid velocity below the heel is generally low in nature. A vortex originating from the heel would thus ensure better prevention of flow stasis which may reduce chances of thrombus formation or deposition.

Figure 6.14: Geometrical differences between (a) Miller cuff and (b) sleeve model.

How et al. (2000) published a follow up study on Miller cuffs. The investigation was aimed at verifying if flow patterns in the cuffed anastomoses were able to minimize areas of low wall shear stresses, hence inhibiting development of intimal hyperplasia. Flow patterns and WSS variations in a conventional end-to-side (ETS) model and an interposition vein cuff (IVC) model as shown in Figure 6.15 were investigated. Experimental parameters used included a mean flow rate $Q_{\text{mean}} = 180$ ml/min
Chapter 6: Results and Discussion of Geometrical Variations (Pulsatile Flow)

corresponding to $Re_{mean} = 193$, with a waveform of frequency $f = 1.25$ Hz as demonstrated in Figure 6.16 and Womersley parameter $\alpha = 4.6$.

Figure 6.15: Flow studies using (a) ETS and (b) IVC models (How et al., 2000).

Figure 6.16: Waveform used in experiments by Wow et al. (2000).

Figure 6.17: Flow characteristics of (a) ETS and (b) IVC models (How et al., 2000).
Chapter 6: Results and Discussion of Geometrical Variations (Pulsatile Flow)

Flow characteristics using a 50:50 flow split are as shown in Figure 6.17. Flow patterns were similar to that reported by Rowe et al. in 1999. As observed from Figure 6.17 (b), the interposition vein cuff altered the incoming flow such that fluid highly skewed towards the outer wall resulted in flow separation along the arterial floor just below the toe. Subsequent fluid motions led to formation of the vortex near the graft toe. In the ETS model however, recirculation was not evident.

How et al. (2000) further compared mean wall shear stress distributions along vessel walls in the ETS and IVC models. An earlier study by Salam et al. (1996) had shown that mean WSS below 0.5 Pa was associated with the greatest increase in neointimal thickness. Using 0.5 Pa as a critical WSS value (i.e. $WSS_{\text{crit}}$) such that regions with WSS below $WSS_{\text{crit}}$ are considered low WSS, How et al. (2000) went on to demonstrate that while the ETS model showed two regions of low WSS, one at the heel and the other along the arterial floor as shown in Figure 6.18 (a), the IVC model only had one low WSS region (along the arterial floor) (Figure 6.18b). They also pointed out that the extent of low WSS was reduced from 9 mm in the ETS model to 5 mm in the IVC model.

Due to differences in geometries and outflow conditions, direct quantitative comparisons between results from How et al. (2000) and the current study would be impossible. On the other hand, since both the sleeve and IVC models exhibit the same characteristics, it can be postulated that the sleeve model may perform better than a conventional bypass having the same geometry and outflow conditions (i.e. same diameter ratio and fully occluded) in the same way the IVC model outperformed the ETS mode. However, this postulation can only be validated in future studies where a
conventional model of similar geometry and outflow conditions is created and compared with

Figure 6.18: Mean wall shear stress distribution for (a) ETS model and (b) IVC model (How et al., 2000).
Evaluation of the sleeve models was performed earlier in Chapter 5. Under steady flow conditions, Model 1, having a volume larger than conventional bypass anastomoses had exhibited improved hemodynamics which may inhibit development of intimal hyperplasia or thrombus formation. It was also recommended that a further volume increase through height instead of diameter increment was more desirable.

Four models were studied in pulsatile flow conditions. General flow characteristics included: laminar flow during early acceleration phase where fluid traveled parallel to vessel and sleeve walls, vortex formation below heel region towards peak flow, movement of vortex rightwards to occupy most of sleeve volume during deceleration phase and dissipation of vortex at the beginning of the acceleration phase.

Hemodynamics throughout the flow cycle in Model 1 was shown in Figure 6.3. As compared to other sleeve models with increased volume, flow characteristics in Model 1 during pulsatile flow conditions were more predictable and less complicated. In a fully occluded host artery, sites of low velocity were located at the distal artery entrance and region below the heel. Recirculation in the distal artery entrance was reported by some researchers (Einav et al., 1985, Pietrabissa et al., 1990 and Fei et al., 1994). Pietrabissa et al. (1990) suggested the use of lower bypass angles to eliminate recirculation in that region. Minimization of bypass angles was however limited by surgical constraints. In fact, a $0^\circ$ bypass angle would require a total change in bypass configuration from end-to-side to side-to-side anastomosis. Even as bypass angles were lowered, fluid at the distal artery entrance would still experience low velocity due to diagonal entrance of flow into the distal artery section. Slow moving fluid
would thus elevate blood residence time leading to increased possibilities of platelet activation or thrombus deposition. Therefore, a temporal reverse flow as observed at times $t = 2.1s$, $2.125s$ and $2.625s$ in Figure 6.3 would attempt to wash out trapped particles near the outer wall of the distal artery entrance.

Slow fluid motion at the heel was unavoidable as a result of physiological constraints in a fully occluded host artery. As explained earlier, due to inherent geometrical differences between the sleeve models and Miller cuffs, vortex in the former originated at the heel while the latter originated at the toe. In the particular case of a full occlusion, a vortex originating from the heel would ensure better prevention of flow stasis which may reduce blood exposure time and chances of thrombus formation or deposition. The dynamic vortex helped enhance fluid mixing by entraining blood into areas of low velocity at the heel region. Perhaps the most important aspect of the vortex was its spatial movement and temporal existence. As observed from Figure 6.3, the vortex was observed to travel rightwards with increasing size and intensity. The movement was important as it implicated spatial mixing of fluid instead of flow recirculation at a specific region. The temporal existence was another essential factor of consideration. The vortex was formed near peak flow and was fully suppressed by the arrival of the next acceleration phase. Suppression of the vortex implied that fluid particles may not remain in the sleeve for extended periods of time as they were replaced by new fluid. The biphasic nature of flow in a bypass graft implied that two vortices were formed and dissipated per cycle of period, $T = 0.9s$. As compared to results at lower limb bypasses by Rowe et al. (1999) and How et al. (2000) which reported only one vortex per cycle having approximately the same period ($T = 0.9s$ for present study and $T = 0.8s$ for studies by
Rowe et al. (1999) and How et al. (2000), the washout mechanism in this case could be considered twice as efficient.

For a height increment in the case of Model 2, the only significant difference was the modulated peak shear stress values near the toe region along the outer wall. This modulation reduced peak WSS by approximately 20%. Flow structures throughout the cycle were similar to that of Model 1 with comparable velocity magnitudes. Such resemblance implied that a height increment had the ability to increase sleeve volume yet replicating the desired flow characteristics of Model 1.

Models 3 and 5 represented individual diameter increment as well as combined diameter and height increment respectively. As discussed in Section 5.5.3, an increase in sleeve diameter would result in an overall decrease in fluid velocity within the sleeve. Geometrical changes in cross-section, coupled with curvature at the sleeve entrance would lead to an increase in secondary flow components. The result of this combined effect was a highly three-dimensional flow with decreased fluid velocity. On one hand, a highly three-dimensional flow would implicate an enhanced mixing effect; however, a reduced fluid velocity on the other hand would imply a decreased rate of mixing and an increase in particle exposure time. Also, fluid motion in a large volume such as Model 5 exhibited complex flow structures which would result in energy losses and diminished perfusion capability. Therefore, considering hemodynamic factors alone, Models 3 and 5 were deemed inappropriate as designs for the sleeve model.
Based on steady and pulsatile flow studies, Model 1 had shown highly desirable qualities as a sleeve design. It provided temporary washing to the low flow regions, mainly the heel and distal artery sections twice per cycle. In addition, the increased anastomotic volume due to the cuff geometry would minimize flow restrictions and delay occlusion time hence improving patency in the long run. Model 2 was cited as an appropriate alternative design with an increased sleeve volume (due to height increment) capable of maintaining flow characteristics in Model 1 while Models 3 and 5 with increased sleeve diameters were considered as less suitable designs.
Chapter 7: Results and Discussion of Out-Of-Plane Angles and Bypass Angles

In coronary artery bypass surgery, the vein graft connects the aorta (proximal anastomosis) to the coronary artery (distal anastomosis) of the heart. Due to general curvature of the human heart, both out-of-plane and bypass angles were required to counter the three dimensionality of CABG. A sleeve designed to give a variation of the two angles would offer full flexibility to the surgical process.

7.1 Out-Of-Plane Angles

The distal anastomosis involved grafting of the saphenous vein to the coronary artery. Due to three dimensionality of the bypass surgery, the axis of the saphenous vein and the coronary artery were often non-planar in nature. A sleeve designed with out-of-plane feature had to be incorporated to facilitate attachment of the vein graft to the sleeve. Without such a feature, kinking would likely occur which could result in poor hemodynamics. The out-of-plane angle ($\theta_{OP}$) was defined as the angle between the incoming vein graft axis and the coronary artery axis as shown earlier in Figure 4.5 and repeated in Figure 7.1 for ease of reference.

In the present study, the primary objective was to determine the required entrance length, $L_E$ for flow to be fully developed with minimal secondary fluid motion. Steady inlet conditions was applied using $Re = 80$. The Reynolds number was obtained from data provided by Galjee et al. (1996) as mentioned in Section 4.4.2. Since the inlet length required for oscillatory flow was shorter than that for steady flow as indicated
in Section 3.1.5, it was thus safe to assume that $L_E$ obtained in steady flow analysis would also satisfy the inlet length requirement for fully developed flow in pulsatile conditions. Three models were designed with varying out-of-plane angles of $\theta_{OP} = 30^\circ, 60^\circ$ and $90^\circ$. The desired entrance length was determined by:

i) Analyzing wall shear stress values where a constant wall shear stress on both the inner and outer walls indicated a fully developed profile.

ii) Examining secondary flow velocity such that secondary motion as a result of fluid passing through the bend was minimized.

Figure 7.1: (a) Out-of-plane angle ($\theta_{OP}$) and entrance length $L_E$ of sleeve design and (b) definition of inner wall and outer wall for out of plane models.
7.1.1 Results and Discussion of Out-Of-Plane Angles

A fully developed profile was determined by observing the convergence of wall shear stress towards a constant value on both sides of the (inner and outer) wall. Shear stresses along the inner and outer walls of varying out of plane angle models were plotted as shown in Figure 7.2. The initial variation in shear stresses was fluid motion after moving through the bend. After some distance downstream, a fully developed profile started to develop characterize by the convergence of the shear stresses to about 1.1 Pa. From Figure 7.2, it was determined that an entrance length of approximately $L_g = 20$ mm would ensure the development of a parabolic profile for all angles of $\theta_{OP}$.

![Variation of WSS (Pa) Along Inner and Outer Walls of Varying Out of Plane Angle Models](image)

Figure 7.2: Variation of WSS (Pa) along inner and outer walls of varying out of plane angle models.
Secondary fluid motion arose due to flow traveling past the bend. In the present study, the sleeve should be designed with entrance length sufficient to minimize the magnitude of secondary flow components. As observed in Figure 7.3, secondary velocity vectors in the $\theta_{OP} = 90^\circ$ model were quite prominent about 5 mm after passing the bend. As fluid traveled further downstream, the secondary motion had minimized to almost negligible magnitudes by about 15 mm from the bend. This distance also corresponded to the desired entrance length as mentioned earlier.

![Diagram showing secondary fluid motion](image)

**Figure 7.3: Secondary fluid motion in $\theta_{OP} = 90^\circ$ model**
The initial design of the out-of-plane feature is as shown in Figure 7.4. The non-planar feature was accomplished by twisting the sleeve inlet at an angle $\theta_{op}$ to the sleeve outlet. Fluid thus entered the sleeve at an angle to the coronary artery axis. Some disadvantages arising from such a configuration included:

(i) complicated fluid motion due to swirling of flow,

(ii) energy wasted in the complicated swirling motion instead of propulsion of flow in the axial direction to perfuse the coronary artery,

(iii) flow would require a longer distance to stabilize hence resulting in the need for a longer sleeve length to contain the unstable flow.

Figure 7.4: Initial out-of-plane design. Figure 7.5: Final out-of-plane design.
Chapter 7: Results and Discussion of Out-Of-Plane Angles and Bypass Angles

The design was thus improvised to integrate an extended length (entrance length, $L_E$) prior to the sleeve entrance as shown in Figure 7.5. Such a configuration yielded several advantages, which included:

(i) flow was able to fully develop and stabilize with minimal secondary motion before entrance into the sleeve,

(ii) fluid entering the sleeve was parallel to coronary artery axis hence maximizing perfusion capabilities (therefore conserving energy), as opposed to swirling in the previous design,

(iii) a shorter sleeve length would be required since swirling motion was absent and flow in the coronary artery could stabilize within a shorter distance,

(iv) simulation work would be simplified since flow became fully developed at the end of the entrance length which would satisfy inlet conditions used in computational studies carried out in Chapters 5 and 6.

From the current investigation, a desired entrance length of $L_E = 20$ mm for all out of plane angles was obtained. It was thus suggested that such an entrance length be incorporated for sleeves with varying out of plane angles to offer greater flexibility to the surgical process.
7.2 Bypass Angles

Due to general curvature of the human heart, the saphenous vein is often grafted at an angle to the coronary artery at the distal anastomosis. The bypass angle ($\theta_B$) was defined as the angle between the graft axis and the coronary artery axis in the same plane as shown in Figure 7.6. As explained in Section 6.5.1, steady flow analysis could be utilized in place of physiological flow analysis where only qualitative comparisons of results were required. The objective of this study was to identify flow patterns in sleeve designs with various bypass angles. The Hagen-Poiseuille equation was used to compute velocity values which act as inlet boundary conditions for the steady model. Two flow rates were modeled, using Reynolds numbers 80 and 250. The cuff geometry was designed similar to the baseline model (Model 1) as described in Chapters 4, 5 and 6. Model 1 in this case had a bypass angle of $\theta_B = 0^\circ$. In addition, four other models were designed with bypass angles of $\theta_B = 15^\circ$, $30^\circ$, $45^\circ$ and $60^\circ$ as shown in Figure 7.7 (b), (c), (d) and (e) respectively.

The inner and outer walls were defined from the sleeve inlet to the proximal and distal outlet of the sleeve respectively while the floor region stretched from the proximal outlet to the distal outlet of the sleeve. Positive wall shear stress is represented by flow in positive $x$ direction while negative wall shear stress represented flow in negative $x$ direction. Terminologies used in discussing flow at various sites are also shown in Figure 7.8. In this study, flow patterns within the graft as well as the proximal and distal artery sections will not be discussed as they were similar to and already discussed in detail in Chapters 5 and 6.
Chapter 7: Results and Discussion of Out-Of-Plane Angles and Bypass Angles

Figure 7.6: Bypass angle ($\theta_B$) of sleeve design.

Figure 7.7: Bypass angles of $\theta_B$ = (a) 0°, (b) 15°, (c) 30°, (d) 45° and (e) 60°.

Figure 7.8: Terminology in bypass angle study.
Chapter 7: Results and Discussion of Out-Of-Plane Angles and Bypass Angles

7.2.1 Results

Velocity vectors of the bypass angle models at $Re = 80$ are as shown in Figure 7.9. General flow characteristics were similar for all models with varying bypass angles. They include strong skewing of particles towards the outer wall and arterial floor, location of stagnation point along the arterial floor approximately near the mid-point opposite the graft and presence of a weak recirculation region below the heel.

![Figure 7.9: Velocity vector plots for bypass angle $\theta_R = (a) 0^\circ$ and (b) $15^\circ$ at $Re = 80$.](image)

Figure 7.9: Velocity vector plots for bypass angle $\theta_R = (a) 0^\circ$ and (b) $15^\circ$ at $Re = 80$. 
Figure 7.9: Velocity vector plots for bypass angle $\theta_B$ = (c) 30°, (d) 45° and (e) 60° at $\text{Re} = 80$. 
Along the inner wall, shear stress variations were similar for models $\theta_2 = 0^\circ$, $15^\circ$, $30^\circ$, $45^\circ$ and $60^\circ$ as shown in Figure 7.10. They were characterized by a sudden increase in wall shear stress values to a peak followed by an immediate decrease to zero and subsequently negative values thereafter. The relatively steep increase initially in wall shear stress values was due to alteration of the flow profile as the fluid traveled along the curvature into the cuff region of the sleeve, resulting in sudden increase in velocity magnitudes near the inner wall as shown in Figures 7.9 (a) to (e). Bulk flow then entered the cuff region of the sleeve at an angle, leaving an area of low flow. This resulted in wall shear stresses plummeting to zero value, characterized by a stagnation point as shown in Figure 7.10. Further downstream, flow reversal due to the vortex formation was observed, represented by the negative wall shear stress values.

![Variation of WSS (Pa) Along Inner Wall (mm) for Re = 80](image)

**Figure 7.10:** Wall shear stress variation along inner wall at Re = 80.
In comparison, as bypass angle was increased from 0° to 60°, peak wall shear stress values along the inner wall decreased. In addition, slight redistribution of the peak shear stresses was also observed with increasing bypass angle. This was due to reduction in curvature in which fluid had to overcome, hence resulting in smoother flow transition. When computed, it was found that an increase of $\theta_b$ from 0° to 60° yielded an approximate 40% decrease in wall shear stress values.

As flow entered the cuff region, shear stresses along the outer wall increased drastically to reach peak values as shown in Figure 7.11. The drastic increase was a result of impingement of high velocity fluid traveling almost parallel to the outer wall. As observed in Figure 7.9, when the high momentum fluid exited the sleeve and entered the distal artery diagonally, a low flow region along the top outer wall was formed, resulting in rapid decrease of shear stresses beyond the peak to a minimum point. Flow beyond this point then progressively became aligned with the cylindrical geometry before attempting to re-establish the fully developed profile as wall shear stress values increased gradually to approach a constant Hagen-Poiseuille value. At low flow rate of $Re = 80$, no flow reversal was observed along distal artery entrance, therefore wall shear stresses were generally positive in values.

When compared, an increase in bypass angle $\theta_b$ resulted in a decrease and redistribution in wall shear stress peak values as shown in Figure 7.11. Both phenomenon were due to smoother flow transition resulting from reduction in curvature as the bypass angle was increased. An increase of $\theta_b$ from 0° to 60° yielded an approximate 27% decrease in wall shear stress values. In addition, since the curvature length decreased as the bypass angle increased, it was thus logical that the
Chapter 7: Results and Discussion of Out-Of-Plane Angles and Bypass Angles

Wall shear stress peak occurred at different locations. As a matter of fact, the larger the bypass angle, the shorter the distance from the sleeve inlet to the point of maximum wall shear stress.

![Variation of WSS (Pa) Along Outer Wall (mm) for Re = 80](image)

Figure 7.11: Wall shear stress variation along outer wall at Re = 80.

In general, small differences were observed in wall shear stress variations along the arterial floor corresponding to different bypass angles $\theta_B$ as shown in Figure 7.12. Incoming flow from the cuff region impinged the arterial floor almost perpendicularly, resulting in bifurcation of flow and a stagnation point. Stagnation point along the arterial floor was found to be shifted toward downstream of the coronary artery as bypass angle increased, implicating a spatial effect on location of the induced vortex. Comparing among the wall shear stress plot obtained an 8% increase from a 0° to 60° change in the bypass angle.
Chapter 7: Results and Discussion of Out-Of-Plane Angles and Bypass Angles

Wall shear stresses to the left of the stagnation point were negative, as a result of flow reversal caused by vortex formation below the heel as shown in Figures 7.9 (a) to (e). As $\theta_h$ was increased from $0^\circ$ to $60^\circ$, the magnitude of the negative wall shear stress peak was observed to increase correspondingly by approximately four times. It can be observed in Figure 7.12 that after the rapid increase in wall shear stress, there was a region of fluctuating wall shear stresses. Such phenomenon was explained earlier in Section 5.1, caused by interaction between upward moving fluid reflected from the floor and downward incoming flow. Bypass angle however did not have a significant effect on the magnitude of this fluctuating zone. It was thus concluded that at low flow rates, variation in bypass angle did not have a significant impact on wall shear stress variations along the arterial floor.

![Variation of WSS (Pa) Along Arterial Floor (mm) for Re = 80](image)

Figure 7.12: Wall shear stress variation along arterial floor at Re = 80.
Chapter 7: Results and Discussion of Out-Of-Plane Angles and Bypass Angles

The main purpose of a cuff-like structure in the sleeve design was to improve wash-out via induction of a vortex that varied temporally and spatially. As reported in Chapter 6, through a detailed analysis of unsteady flow patterns, the presence of temporal variation of the vortex was validated. In addition, it was concluded that due to biphasic nature of flow in the bypass graft, two vortices were formed and dissipated per flow cycle, hence improving wash-out efficiency as compared to the lower limb bypasses. The main objective of this study was to assess influence of bypass angle on the spatial variation of vortex movement. As explained in Section 6.5.1, steady flow analysis could be utilized in place of physiological flow analysis where only qualitative comparisons of results were required. In the present study, spatial movement of the vortex was analyzed by subjecting the fluid to higher flow rates ($Re = 250$) and to compare the effects of bypass angle on vortex movement. Velocity vectors of the bypass angle models are as shown in Figure 7.13.

![Velocity vector plots for bypass angle $\theta_B = (a) 0^\circ$ and (b) $15^\circ$ at $Re = 250$.](image)

**Figure 7.13:** Velocity vector plots for bypass angle $\theta_B = (a) 0^\circ$ and (b) $15^\circ$ at $Re = 250$. 
Figure 7.13: Velocity vector plots for bypass angle $\theta_b = (c) 30^\circ, (d) 45^\circ$ and (e) $60^\circ$ at $Re = 250$. 
Intensity of the vortex formed below the heel region was observed to increase with flow rate for all bypass angles. As shown in Figure 7.14, shear stress values along the inner wall were observed to be larger at Re = 250 as compared to Re = 80 (Figure 7.10) as a result of increased velocity magnitudes. Apart from that, variation patterns of shear stresses along the inner wall were similar for both flow rates (decreasing peak shear stresses value and increasing shear stress redistribution with increasing bypass angle). When computed, a change in bypass angle from $\theta_b = 0^\circ$ to $60^\circ$ yielded a decrease in wall shear stress values by approximately 50%, a value comparable to that at Re = 80.

![Variation of WSS (Pa) Along Inner Wall (mm) for Re = 250](image)

**Figure 7.14:** Wall shear stress variation along inner wall at Re = 250.
Chapter 7: Results and Discussion of Out-Of-Plane Angles and Bypass Angles

At higher flow rate of Re = 250 near the sleeve entrance, a lower angle design resulted in fluid impingement on the outer wall. For smaller bypass angles of $\theta_B = 0^\circ$ and $15^\circ$, a small recirculation region was observed characterized by negative wall shear stresses. The recirculation ceased to exist as bypass angle increased to $\theta_B = 30^\circ$ and $45^\circ$ but impingement of fluid resulted in wall shear stresses close to zero. At $\theta_B = 60^\circ$, fluid was able to travel almost parallel to the outer wall as shown in Figure 7.13 (e), hence resulting in larger shear stresses. Near the outer wall at the distal artery entrance, a region of recirculation was observed for all models as shown correspondingly in Figures 7.13 (a) to (e). Such phenomenon was caused by departure of high momentum flow from the outer wall, creating a region of low pressure hence resulting in some fluid particles to take a reversed path. Flow reversal along the outer wall was represented by negative wall shear stresses as shown in Figure 7.15. The magnitude and intensity of the recirculation as a result of the reversed flow was observed to vary inversely with bypass angle. As bypass angle was increased, the recirculation was observed to decrease in magnitude and intensity. When compared, an increase in bypass angle from $0^\circ$ to $60^\circ$ yielded a decrease of approximately 32% in wall shear stress values.

![Figure 7.15: Wall stress variation along outer wall at Re = 250.](image)
Chapter 7: Results and Discussion of Out-Of-Plane Angles and Bypass Angles

At a low bypass angle, fluid was highly skewed towards the outer wall as it had to overcome a large curvature when traveling from the graft to the sleeve. Fluid entering the sleeve was therefore traveling at a high velocity and had a larger tendency to overshoot and bypass the area distal to the toe, thereby creating a small recirculation in that region as shown in Figure 7.16 (a). As the bypass angle was increased to $\theta_B = 60^\circ$, flow transition from the graft to the sleeve was smoother, as the effects of skewing were lesser, leading to reduced flow velocity. This was further evidenced by reduced wall shear stress values along the outer wall in Figure 7.15. As shown in Figure 7.16 (b), fluid traveling at a lower velocity tended to follow the curvature along the toe before entering the distal artery segment. As the fluid motion conformed to the geometry well, tendency to overshoot was reduced and thus resulting in vortex formation was lower. In the case of $\theta_B = 60^\circ$, no flow reversal was observed.

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{Fig716.png}
\caption{Comparison of hemodynamics between (a) $\theta_B = 15^\circ$ and (b) $\theta_B = 60^\circ$.}
\end{figure}
As flow rate was increased from $\text{Re} = 80$ to $\text{Re} = 250$, the reversed flow as well as fluctuating fluid motions along the arterial floor became more pronounced. This was evidenced by an increase in wall shear stress magnitudes as shown in Figure 7.17 when compared to Figure 7.12. Also, stagnation points along the arterial floor were observed to have moved distally downstream when the flow rate was increased. The model with bypass angle $\theta_B = 60^\circ$ exhibited the largest percentage change (12%) as a result of increase in flow rate. The effects of bypass angle on hemodynamics were similar to that discussed earlier for $\text{Re} = 80$. The magnitude of negative wall shear stress peak was observed to increase by approximately two times when $\theta_B$ was increased from $0^\circ$ to $60^\circ$. Similarly, stagnation points were found to be located more towards distal direction with larger bypass angles. The difference between the $0^\circ$ and $60^\circ$ models was found to be 10%.

![Variation of WSS (Pa) Along Arterial Floor (mm) for Re = 250](image)

**Figure 7.17:** Wall shear stress variation along arterial floor at Re = 250.
7.2.2 Discussion

The influence of bypass angles on sleeve designs with a cuff-like geometry had been discussed earlier. Uniqueness of this research stemmed from the fact that no studies had been made on the combined effects of both bypass angles and cuff geometries, although both parameters had been investigated independently by other researchers. The effects of bypass angles had been studied extensively by Crawshaw et al. (1980), Einav et al. (1985), Pietrabissa et al. (1990), Fei et al. (1994) and Ojha et al. (1994). Several papers had also been published on the effects of interposition vein cuffs on hemodynamics (Rowe et al., 1999, How et al., 2000, and Leuprecht et al., 2002). Incorporation of the sleeve as a mechanical connector offered greater freedom to study designs that were previously impossible to create during bypass surgeries, such as a cuff-like sleeve with different bypass angles.

Ojha et al. in 1994 studied the effects of bypass angles (20°, 30°, 45° and 60°) on hemodynamics in distal anastomoses. Using photochromic tracer technique, Ojha et al. (1994) reported low shear stresses at the heel and on the bed opposite the heel for all angles. Computational results obtained in the present study were in good qualitative agreement with the experimental results reported by them. The low shear stresses were believed to be due to complete occlusion of the proximal end of the host vessel. In addition, they observed increased flow separation near the toe and increased shear stresses on the bed (arterial floor) across the toe when bypass angle was increased. Such observations were reported earlier by Crawshaw et al. (1980), Einav et al. (1985) and Pietrabissa et al. (1990).
Chapter 7: Results and Discussion of Out-of plane Angles and Bypass Angles

One of the most prominent findings by the researchers on the effect of bypass angle was recirculation at the distal artery entrance. Using dye injection technique, Crawshaw et al. (1980) reported that a low insertion or bypass angle would reduce flow separation at the distal artery entrance. Einav et al. (1985) further illustrated using a coronary artery bypass model as shown in Figure 7.18 the possibility of delaying turbulence at the graft-coronary intersection by reducing graft-artery diameter ratio and bypass angle. Pietrabissa et al. (1990) established a direct correlation between the magnitude of recirculation with bypass angle. They concluded that the magnitude of recirculation increased with bypass angle as shown in Figure 7.19. Using computational models as shown in Figure 7.20, Fei et al. (1994) showed that for a flow rate of $Re = 100$, recirculation at the distal artery entrance occurred only with bypass angles greater than $60^\circ$. When the flow rate was increased to $Re = 205$, recirculation occurred for angles larger than $45^\circ$. Findings by the above mentioned researchers were in good agreement with one another, but contrary to the present results.

![Figure 7.18: Model of coronary bypass by Einav et al. (1985).](image-url)
Figure 7.19: Effect of bypass angle on recirculation at distal artery entrance (Pietrabissa et al., 1990).

Figure 7.20: Computational model by Fei et al. (1994).
Chapter 7: Results and Discussion of Out-Of-Plane Angles and Bypass Angles

Models used by Crawshaw et al. (1980), Einav et al. (1985), Pietrabissa et al. (1990), and Fei et al. (1994) employed two straight tubes connected at an angle. Relationship between bypass angle and recirculation at the distal artery entrance was predictable for such end-to-side models as the fluid had a tendency to continue its motion in a direction parallel to the graft axis. Since bypass angle was defined as the angle between the graft and coronary artery axis, a larger bypass angle would result in a greater tendency for fluid to overshoot the region distal to the toe, thereby creating recirculation. A lower bypass angle on the other hand would facilitate change in direction as fluid enters the distal artery section.

Hemodynamics in the current study was altered as a result of the presence of curvature due to the cuff-like geometry. As explained earlier, enhanced skewing and high velocity fluid motion in low bypass angle models resulted in greater tendency to overshoot and create recirculation as compared to larger bypass angles with lesser skewing and reduced fluid velocity. It could thus be concluded that incorporation of the cuff-like geometry had inversed the relationship between bypass angle and magnitude of recirculation at the distal artery entrance found in normal end-to-side models.

Detailed analysis of effects of bypass angle on hemodynamics was discussed earlier. In general, as the bypass angle was increased, shear stresses along the inner and outer walls decreased in value. The peak wall shear stresses were also observed to be modulated with increment in bypass angle. For low flow rate of Re = 80, no recirculation was observed for all angles but as flow rate was increased to Re = 250, recirculation near the distal artery entrance was also found to diminish in magnitude.
for increasing bypass angle. Vortex in the sleeve was observed to increase in intensity and size, characterized by stagnation points that were located more distally as bypass angle increased.

In conclusion, to reduce possibility of thrombus deposition due to flow stasis, a smaller bypass angle (resulting in larger wall shear stresses) was preferred. To minimize recirculation near the distal artery entrance in small bypass angle models, the sleeve diameter could be increased as mentioned in Chapter 5. It should however be mentioned that the choice of bypass angle should not be based solely on hemodynamic considerations, but also on facilitating the demanding anatomical constraints of bypass surgery.
Chapter 8: Conclusions and Future Works

8.1 Conclusions

The study was undertaken with the aim of designing and evaluating sleeve models with cuff-like geometries of varying volumes as well as out-of-plane and bypass angles using CFD. The sleeve models were designed to accommodate a systematic variation of volume by splitting it into the “diameter” and “height” components so that their individual or combined effects on hemodynamics could be evaluated efficiently under steady and unsteady flow conditions. In addition, models with out-of-plane and bypass angle features were designed to study the effects of such angle variations on flow patterns. Hemodynamics in all sleeve models were simulated using a commercial solver FLUENT. Detailed validation procedures were carried out prior to numerical computations to ensure high accuracy of the solutions.

The initial study involved investigation of hemodynamics in varying volumes under steady flow conditions. General flow characteristics included strong skewing of particles towards outer wall and arterial floor, helical particle trajectories due to secondary motions, increased strength of secondary velocity components with greater flow rates, location of stagnation point along the arterial floor approximately at the mid-point opposite the graft and presence of recirculation at the distal artery entrance. All of the above characteristics were found in the baseline model (Model 1) while Model 2 with a height increment showed similar trends with the exception of modulated shear stresses at the toe region along the outer wall. Fluid motion in Model 3 (diameter increment) was generally slower, exhibiting a source-like structure within
Chapter 8: Conclusions and Future Work

the sleeve volume at higher flow rate of Re = 250, due possibly to influence of secondary flow components. Flow in models with combined height and diameter increment (Models 4 and 5) revealed characteristics observed separately in Models 2 and 3, namely modulated wall shear stresses around toe region and source-like structure in sleeve volume respectively.

When subject to pulsatile flow conditions, typical flow patterns observed included laminar flow during early acceleration phase where fluid traveled parallel to vessel and sleeve walls, vortex formation below heel region near peak flow, movement of vortex rightwards to occupy most of the sleeve volume during deceleration phase and suppression of vortex at the beginning of the acceleration phase. Comparison of flow characteristics between steady and pulsatile conditions showed qualitative similarities. Varying Reynolds numbers in steady conditions would reveal the different flow characteristics found in a particular sleeve model. Therefore, steady flow analysis is a practical and efficient preliminary approach in understanding effects of different sleeve geometries on flow characteristics.

As compared to other models, hemodynamics throughout the flow cycle in Models 1 and 2 were more predictable and less complicated. The dynamic vortex formed near peak flow provided improved mixing by entraining blood into areas of low velocity at the heel region. Movement of the vortex during deceleration was crucial as it implicated spatial mixing of fluid instead of flow recirculation at a specific region. At the beginning of the next acceleration phase, the vortex was completely suppressed, implying that fluid particles did not remain in the sleeve for extended periods of time as they were replaced by new fluid. Such was an excellent washout mechanism provided by the cuff-like geometry. Due to the biphasic nature of flow in coronary
Chapter 8: Conclusions and Future Work

artery bypass grafts, two vortices were formed and suppressed per cycle as compared to one vortex per cycle in the lower limb bypasses, hence indicating a more efficient washing capability. Flow in Models 3 and 5 were characterized by the source-like structure due to influence of the secondary flow components. The overall decreased fluid velocity would be detrimental as it increased exposure time of blood to the prosthetic material, hence increasing possibilities of thrombus formation or deposition at the same time. From the above studies, Model 1 was more desirable as a sleeve design due to improved hemodynamics which may inhibit or reduce the possibilities of reocclusion. For a further increase in volume, a height increment was recommended.

Further studies were conducted by subjecting the sleeve model (Model 1) to varying out-of-plane ($\theta_{OP}$) and bypass angles ($\theta_b$). Models with out-of-plane angles of $\theta_{OP} = 30^\circ$, $60^\circ$ and $90^\circ$ were subject to steady flow conditions of Re = 80 to determine the desired entrance length $L_E$. Visual inspection on the velocity contour plots were used to identify distance required for fully developed profile after passing through a bend. Secondary flow conditions were also checked to ensure minimal secondary fluid motions beyond the entrance length. For each of the out-of-plane angles, $\theta_{OP} = 30^\circ$, $60^\circ$ and $90^\circ$, the corresponding entrance length was found to be $L_E = 9\text{mm}$, $12\text{mm}$ and $14\text{mm}$ respectively.

Bypass angles of $\theta_b = 0^\circ$, $15^\circ$, $30^\circ$, $45^\circ$ and $60^\circ$ were subject to steady flow conditions of Re = 80 and Re = 250. General flow characteristics included decrease and redistribution of peak shear stresses along the inner and outer walls as bypass angle was increased. Along the arterial floor, stagnation point was found to be more distally
located with the increment in bypass angle. As flow rate was increased from $Re = 80$ to $Re = 250$, stagnation points along the arterial floor was found to move distally downstream, implying spatial movement of the vortex. At low flow rate ($Re = 80$), no recirculation was observed for all bypass angles at the distal artery entrance. However, at a higher flow rate of $Re = 250$, an inverse relationship existed between $\theta_B$ and the magnitude of recirculation, a phenomenon opposite to that of the common end-to-side anastomosis model. To minimize flow stasis which may lead to thrombus deposition resulting in reocclusion of the sleeve, a lower bypass angle was preferred. Recirculation at the distal artery entrance caused by small $\theta_B$ values could be minimized by increasing the sleeve diameter.

### 8.2 Future Works

As mentioned earlier, the objectives of this study was to evaluate effects of geometry and configuration variations on flow patterns. The main design considerations in this study involved sleeve diameter and height as well as out-of-plane and bypass angles. Although the present work provided a comprehensive evaluation of the major parameters, other design factors were suggested for future studies. These included the sleeve thickness as well as inlet and outlet rim designs. In the case of a fully inserted sleeve, thickness played an important role in determining flow patterns since an increase in sleeve thickness could possibly reduce the cross sectional area available for perfusion of blood. In addition, both inlet and outlet rims of the sleeve could be tapered to minimize flow disturbances as a result of change in cross-sectional area as shown in Figure 8.1.
Another design consideration would be the proximal outlet condition. Presently, all computations were carried out on the assumption that the proximal outlet was fully occluded. Such assumption was based on clinical observations that full occlusion occurred at the stenosis within three to six months after bypass surgery. Prior to full occlusion, incoming flow past the stenosis would interact with outgoing flow from the sleeve, resulting in complicated flow structures as shown in Figure 8.2. To prevent unnecessary flow complications, it was suggested that the sleeve be designed with only one outlet at the distal artery segment as shown in Figure 8.3. However, such design would only be possible if the anastomosis was constructed near the stenosis. In reality, a number of bypasses were constructed at a distance downstream from the stenosis. It is not uncommon that in such cases, side vessels exists which required perfusion from the proximal outlet as shown in Figure 8.4. In the above mentioned situation, the proximal outlet was required and in-depth study of flow patterns corresponding to two outflows would be necessary.
The sleeve could be designed for full or partial insertion into the coronary artery as shown in Figures 8.5 (a) and (b) respectively. A full insertion would subject blood in the anastomosis to a more controlled environment since all flow instabilities were contained within the sleeve itself. This would eliminate intimal hyperplasia (IH) as a causing factor for reocclusion since the common sites of IH formation (the toe, heel...
and arterial floor) were "protected" by a synthetic material not capable of thickening through cell proliferation. However, possibilities of thrombus formation and deposition still exist and had to be investigated. A full insertion on the other hand would result in a reduced cross-sectional area for perfusion due to thickness of the sleeve. In addition, interface between the sleeve and coronary artery would pose additional problems if mismatch in diameters occurred. Lack of perfusion to the vessel walls covered by the sleeve may also have undesirable physiological effects, in the worst case leading to tissue necrosis in that area. The disadvantages of a full insertion could be relieved by a partial insertion design as shown in Figure 8.5 (b). However, in a partial insertion, the anastomosis would be subjected to two possible reoccluding mechanisms – thrombus formation and deposition as well as IH formation along the arterial floor due to exposure to flow instabilities. Hence, the true pros and cons of both designs could only be evaluated in an animal trial once the material and procedure for inserting the sleeve is finalized.

Figure 8.5: Sleeve design for (a) full and (b) partial insertion.
Last but not least, experiments should be performed to validate all results obtained in the present study. Hemodynamic patterns and wall shear stress values from the present computations could be compared with experimental results using PIV to obtain necessary validation. In addition, in-vitro experiments using blood as a medium could be performed on the varying sleeve models (different geometries as well as out-of-plane and bypass angles). By comparing the amount and location of thrombus formation and deposition over a period of time, the true efficiency of wash-out mechanism corresponding to each model can then be validated.
References


References


205


References


References


References


References


References


Appendix A: Sleeve Model Designs

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<thead>
<tr>
<th>Model</th>
<th>$D_A$</th>
<th>$D_G$</th>
<th>$D_S$</th>
<th>$H_S$</th>
<th>Remarks</th>
<th>Approximate Volume</th>
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<td>3 mm</td>
<td>$D_G$</td>
<td>$D_A$</td>
<td>Baseline Model</td>
<td>88 mm$^3$</td>
</tr>
<tr>
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<td>2.6 mm</td>
<td>3 mm</td>
<td>$D_G$</td>
<td>1.5 $D_A$</td>
<td>Increased Height</td>
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</tr>
<tr>
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<td>2.6 mm</td>
<td>3 mm</td>
<td>1.25 $D_G$</td>
<td>$D_A$</td>
<td>Increased Diameter</td>
<td>104 mm$^3$</td>
</tr>
<tr>
<td>4</td>
<td>2.6 mm</td>
<td>3 mm</td>
<td>1.25 $D_G$</td>
<td>1.5 $D_A$</td>
<td>Increased Diameter and Height</td>
<td>120 mm$^3$</td>
</tr>
<tr>
<td>5</td>
<td>2.6 mm</td>
<td>3 mm</td>
<td>1.5 $D_G$</td>
<td>2 $D_A$</td>
<td>Increased Diameter and Height</td>
<td>160 mm$^3$</td>
</tr>
</tbody>
</table>

Table A.1: Summary of sleeve dimensions.

Figure A.1: Illustration of physical model dimensions.

Figure A.2: Illustration of sleeve model 1.
Appendix A: Sleeve Model Designs

Figure A.3: Illustration of sleeve model 2.

Figure A.4: Illustration of sleeve model 3.
Figure A.5: Illustration of sleeve model 4.

Figure A.6: Illustration of sleeve model 5.
Appendix B: User Defined Functions for Steady and Unsteady Flow

User-Defined Functions (UDF) are used to compute inlet velocity profiles which act as boundary conditions for the numerical methods.

UDF for the steady flow condition is as shown below. The velocity profile was calculated based on the equation \( v_z = 2v_z \left[ 1 - \left( \frac{r}{R} \right)^2 \right] \). In the following example, at \( \text{Re} = 80 \), the mean velocity was \( \bar{v}_z = 0.103127962 \) m/s, radius of the graft was \( R = 0.0015 \) m and the radial coordinate was defined as \( r = \sqrt{y^2 + z^2} \). The function name in this case is “inlet_x_velocity”.

```c
#include "udf.h"

DEFINE_PROFILE(inlet_x_velocity,thread,position)
{
    real x[ND_ND];
    real y;
    real z;
    face_t f;

    begin_f_loop(f,thread)
    {
        F_CENTROID(x,f,thread);
        y=x[1];
        z=x[2];
        F_PROFILE(f,thread,position)=2*0.103127962-
            (y*y+z*z)/(0.0015*0.0015)*2*0.103127962;
    }

    end_f_loop(f,thread)
}
```
Appendix B: User Defined Functions for Steady and Unsteady Flow

For pulsatile flow conditions, the graft inlet velocity profile was calculated based on the
Womersley solution as shown in Section 4.5.2:

\[
v_z(r,t) = 2B_0 \left[ 1 - \left( \frac{r}{R} \right)^2 \right] + 2 \times \text{Real} \sum_{n=1}^{n=n/4} B_n \left( \frac{\alpha_n r}{R} \right) e^{i \omega t}
\]

The UDF is as shown below:

```
#include "udf.h"

DEFINE_PROFILE(unsteady_velocity, thread, position)
{
    face f;
    double x[ND_ND];
    double x1,y;
    double z;
    float t;
    int timeno;
    double ratio;
    int n,k,j,coefficient;
    double R;
    double w;
    double kv;
    double power,even,rank,rank2,am,sumreal,sumimg;
    double rf,a,b,c,d,e,g,j0r,j0m;
    double j1r,j1m,jr,jm,Er,wr;
    double FC[90][2];

    R = 0.0015; /* where R is the graft radius */
    w = 2*3.14159265359/0.9; /* where w is the angular frequency, \omega */
    kv = 3.867298e-6; /* where kv is the kinematic viscosity, \nu */
```
Appendix B: User Defined Functions for Steady and Unsteady Flow

/* The following are the Fourier Coefficients */

FC[0][0]  =  7.9941;  FC[0][1]  =  0.0000;
FC[1][0]  = -1.1795;  FC[1][1]  = -0.1165;
FC[2][0]  = -1.0894;  FC[2][1]  = -1.5468;
FC[3][0]  = -1.5739;  FC[3][1]  = -0.1086;
FC[4][0]  =  0.0280;  FC[4][1]  = -0.1891;
FC[5][0]  = -0.4326;  FC[5][1]  = -0.6029;
FC[6][0]  = -0.1919;  FC[6][1]  =  0.1111;
FC[7][0]  =  0.1603;  FC[7][1]  = -0.1399;
FC[8][0]  = -0.0841;  FC[8][1]  = -0.0752;
FC[9][0]  =  0.0127;  FC[9][1]  =  0.0545;
FC[10][0] =  0.0247;  FC[10][1] = -0.0084;
FC[11][0] = -0.0160;  FC[11][1] = -0.0230;
FC[12][0] = -0.0498;  FC[12][1] =  0.0048;
FC[13][0] =  0.0053;  FC[13][1] =  0.0331;
FC[14][0] = -0.0093;  FC[14][1] = -0.0244;
FC[15][0] = -0.0444;  FC[15][1] =  0.0218;
FC[16][0] = -0.0108;  FC[16][1] =  0.0225;
FC[17][0] = -0.0248;  FC[17][1] = -0.0057;
FC[18][0] = -0.0343;  FC[18][1] = -0.0080;
FC[19][0] = -0.0217;  FC[19][1] =  0.0015;
FC[20][0] =  0.0036;  FC[20][1] = -0.0181;
FC[21][0] = -0.0108;  FC[21][1] = -0.0137;
FC[22][0] =  0.0036;  FC[22][1] =  0.0038;
FC[23][0] =  0.0055;  FC[23][1] = -0.0065;
FC[24][0] = -0.0072;  FC[24][1] =  0.0012;
FC[25][0] = -0.0010;  FC[25][1] =  0.0034;
FC[26][0] =  0.0031;  FC[26][1] =  0.0005;
FC[27][0] = -0.0019;  FC[27][1] = -0.0059;
FC[28][0] = -0.0070;  FC[28][1] =  0.0023;
FC[29][0] =  0.0006;  FC[29][1] =  0.0048;
FC[30][0] = -0.0097;  FC[30][1] = -0.0080;
FC[31][0] = -0.0141;  FC[31][1] =  0.0057;
FC[32][0] = -0.0085;  FC[32][1] = -0.0040;
FC[33][0] = -0.0151;  FC[33][1] = -0.0029;
FC[34][0] = -0.0052;  FC[34][1] = -0.0014;
FC[35][0] = -0.0083;  FC[35][1] = -0.0086;
FC[36][0] = -0.0047;  FC[36][1] =  0.0012;
FC[37][0] = -0.0010;  FC[37][1] = -0.0018;
FC[38][0] = -0.0036;  FC[38][1] = -0.0035;
FC[39][0] = -0.0098;  FC[39][1] = -0.0016;
FC[40][0] = -0.0086;  FC[40][1] =  0.0019;
begin_f_loop(f, thread)
    x1=x[0];
    y=x[1];
    z=x[2];
    ratio=sqrt((y*y+z*z)/R/R);  /* where ratio = \frac{r^2}{R^2} */
    t = RP_Get_Real("flow-time");
    wr=2*FC[0][0]*(1-ratio*ratio);  /* where wr = 2B_0\left(1-\frac{r^2}{R^2}\right) */
    for(n=1;n<=40;n++)
        rf=R*sqrt(n*w/kv);  /* where rf is \alpha_n = R\sqrt{\frac{n\omega}{\nu}} */
/* This section calculates the term \int_0^1 \left(\frac{\alpha_n^2 r}{R}\right) i^j and stores it as a + ib */
    sumreal=1.0;
    sumimg=0.0;
    for (k=1;k<=20;k++)
        {power=1.0;
         rank=1.0;
         coefficient=pow(-1,k);
         power=pow(rf*ratio*rf*ratio,k);
         even=pow(4,k);
         for(j=1;j<=k;j++)
             {rank=rank*j;
              am=coefficient*power/even;
              am=am/rank;
              am=am/rank;
              if(k%2==0)
                  {if(k%4==0) sumreal=sumreal+am;
                   else sumreal=sumreal-am;
                  }
              else
                  {if((k+1)%4==0) sumimg=sumimg+am;
                   else sumimg=sumimg-am;
                  }
          }
    a=sumreal;
    b=sumimg;
/* This section calculates the term \( \int_0^\infty \left( \frac{3}{i^2} \alpha_n \right) \) and stores as \( c + id \) */

    sumreal=1.0;
    sumimg=0.0;

    for (k=1;k<=20;k++)
    {
        power=1.0;
        rank=1.0;
        coefficient=pow(-1,k);
        power=pow(rf*rf,k);
        even=pow(4,k);
        for(j=1;j<=k;j++)
        {
            rank=rank*j;
        }
        am=coefficient*power/even;
        am=am/rank;
        am=am/rank;
        if(k%2==0)
        {
            if (k%4==0) sumreal=sumreal+am;
            else sumreal=sumreal-am;
        }
        else
        {
            if((k+1)%4==0) sumimg=sumimg+am;
            else sumimg=sumimg-am;
        }
    }
    c=sumreal;
    d=sumimg;

/* This section calculates the term \( \int_0^\infty \left( \frac{3}{i^2} \frac{\alpha_n}{R} \right) \) and stores it as \( j0r + ij0m \) */

    j0r=(c*c-c*a+d*d-d*b)/(c*c+d*d);
    j0m=(d*a-b*c)/(c*c+d*d);

/* This section calculates the term \( 2\int_0^\infty \left( \frac{3}{i^2} \alpha_n \right) \) and stores it as \( e + if \) */

    sumreal=0.5;
    sumimg=0.0;
for (k=1;k<=20;k++)
{power=1.0;
 rank=1.0;
 rank2=1.0;
 coefficient=pow(-1,k);
 power=pow(rf,2*k);
 even=pow(4,k+1);
 for(j=1;j<=k;j++)
 {rank=rank*j;
 rank2=rank2*(j+1);}
am=coefficient*power/even;
am=am/rank;
am=am/rank2;
if(k%2==0)
 {if(k%4==0) sumreal=sumreal+am;
 else sumreal=sumreal-am;
 }
else
 {if((k+1)%4==0) sumimg=sumimg+am;
 else sumimg=sumimg-am;
 }
}
e=2*sumreal;
g=2*sumimg;

/* This section calculates the term \( \frac{2J_{\frac{3}{2}}}{J_{\frac{3}{2}}(i\hat{\alpha}_{n})} \) and stores it as \( jlr + jlm \) */

jlr=(c*c-\( e+d*d\times d\times d\times g\))/(c*c+d*d); 
jlm=(g*a-b*e)/(c*e+g*g);

/* This section calculates the term \( \frac{1-2J_{\frac{3}{2}}}{1-\frac{3}{2}(i\hat{\alpha}_{n})} \) and stores it as \( jlr + jlm \) */

jr=(j0*r*j1r+j0*m*j1m)/(j1r*j1r+j1m*j1m);
jm=(j1r*j0m-j1m*j0r)/(j1r*j1r+j1m*j1m);
/* This section calculates the final Womersley solution */

Er=cos(n*w*t)*(FC[n][0]*jr-FC[n][1]*jm)
-\sin(n*w*t)*(FC[n][0]*jm+FC[n][1]*jr);
wr=wr+2*Er;
}

F_PROFILE(f, thread, position) = 0.01*wr;
end_f_loop(f, thread)