Combined MR imaging and numerical simulation of flow in realistic arterial bypass graft models

Y. Papaharilaou a,d, D.J. Doorly a, S.J. Sherwin a, J. Peiro a, C. Griffith a, N. Cheshire b, V. Zervas b, J. Anderson b, B. Sanghera c, N. Watkins d and C.G. Caro d

a Biomedical Flow Group, Aeronautics Department, Imperial College of Science, Technology and Medicine, Prince Consort Road, London, UK
b St. Mary’s Hospital, Hammersmith Hospital, London, UK
c Mechanical Engineering Department, Imperial College of Science, Technology and Medicine, Exhibition Road, London, UK
d Department of Bioengineering, Imperial College of Science, Technology and Medicine, Exhibition Road, London, UK

Abstract. We report methods for (a) transforming a three-dimensional geometry acquired by magnetic resonance angiography (MRA) in vivo, or by imaging a model cast, into a computational surface representation, (b) use of this to construct a three-dimensional numerical grid for computational fluid dynamic (CFD) studies, and (c) use of the surface representation to produce a stereo-lithographic replica of the real detailed geometry, at a scale convenient for detailed magnetic resonance imaging (MRI) flow studies. This is applied to assess the local flow field in realistic geometry arterial bypass grafts. Results from a parallel numerical simulation and MRI measurement of flow in an aorto-coronary bypass graft with various inlet flow conditions demonstrate the strong influence of the graft inlet waveform on the perianastomotic flow field. A sinusoidal and a multi harmonic coronary flow waveform both with a mean Reynolds number (Re) of 100 and a Womersley parameter of 2.7 were applied at the graft inlet. A weak axial flow separation region just distal to the toe was found in sinusoidal flow near end deceleration (Re = 25). At the same location and approximately the same point in the cycle (Re = 30) but in coronary flow, the axial flow separation was stronger and more spatially pronounced. No axial flow separation occurred in steady flow for Re = 100. Numerical predictions indicate a region in the vicinity of the suture line (where there is a local narrowing of the graft) with a wall shear magnitude in excess of five times that associated with fully developed flow at the graft inlet.

Keywords: Coronary arteries, hemodynamics, shear stress, distal anastomosis, unsteady flow

1. Introduction

Bypass graft surgery is a widely applied procedure to relieve the symptoms of arterial occlusion. However, it has been reported [3] that 50% of the grafts implanted fail by 10 years due to restenosis caused primarily by intimal hyperplasia. Although a link between the local flow field and vessel wall biology has been found [4], the exact mechanism of this interaction is not fully understood. However a correlation between regions of low time-averaged or oscillatory shear and long residence times and areas of intimal thickening has been demonstrated [2,9,17]. Furthermore regions of low wall shear stress and flow recirculation have been shown to correlate with locations of atheroma in fixed coronary arteries [1]. Although idealised models are useful in identifying the primary effects of the geometry on the flow field [5,16] current blood flow investigations in models are moving towards more patient specific geometries and flow conditions [8,13].
Several studies have investigated the feasibility of non-invasive in vivo volume flow quantification in coronary arteries [14] or coronary artery bypass grafts [6] by MRI as a means to supplement or replace X-ray angiography. However, detailed in vivo studies of the local flow field in small calibre vessels, which are more prone to failure, are currently not feasible. The aim of this investigation is to assess the effects of the detailed geometry on the local time varying flow field in realistic bypass graft models. For this, bypass graft geometries extracted either in vivo (femoro-tibial) or in vitro (aorto-coronary) by MRI are considered. Preliminary results from numerical flow simulations and in vitro MRI flow measurements in an aorto-coronary distal anastomosis are presented.

2. Materials and methods

2.1. Geometry measurements

Measurements were performed on a General Electric (GE) 1.5 T MRI whole body scanner at St’ Mary’s Hospital and on a GE 1.5 T small (30 cm) bore scanner at Imperial College.

In vivo: The distal anastomosis of a femoro-tibial end-to-side bypass graft was imaged in a patient who had recently undergone bypass surgery, using a three-dimensional time of flight gradient echo sequence. A 1 mm through plane resolution and a 0.23 mm \( \times \) 0.23 mm (interpolated at reconstruction from 0.23 mm \( \times \) 0.46 mm) in-plane resolution was achieved during a 15 minute acquisition.

In vitro: Bypass grafts of the left anterior descending (LAD) coronary artery were constructed by an experienced heart surgeon using excised porcine hearts and human saphenous vein surplus to requirements for coronary artery bypass grafting. The grafts were cast using an epoxy resin (Bi-resin G49, Cika Chemie) at a pressure equivalent to 74 mm Hg and extracted using hydrochloric acid. They were then imaged immersed in a CuSO\(_4\) doped water solution, using a standard three-dimensional gradient echo sequence with 0.19 mm in plane and 0.7 mm through plane resolution.

2.2. Surface reconstruction

The raw MRI images are segmented and single pixel edges defining the blood vessel boundaries are extracted. A closed spline is fitted through each edge and evaluated at a suitable number of sites to obtain a cloud of points representing the surface of the blood vessel. An implicit function which has minimum curvature and fits through these points and the centres of all closed splines is calculated. The zero value contour of the implicit surface provides a good interpolation of the surface of the blood vessel (Fig. 1(a)). A geometry definition in terms of bicubic spline surfaces is then passed to the FELISA mesh generator [11] that produces the computational mesh. Details on the surface reconstruction have been discussed in [12].

2.3. Model fabrication

A 1650 FDM rapid prototyping (RP) fused deposition modeller (Stratasys Inc., USA) using Acrylonitrile-Butadine-Styrene (ABS) thermoplastic applied in thin layers was used to produce the bypass graft replica. For the purposes of ensuring appropriate flow conditions for the MRI flow measurements and boundary conditions for the computations circular cross section extensions were blended at the inlet and outlet of the bypass graft geometry definition that was then scaled by a factor of 1.5. This
information was then input in stereolithography (STL) format to the RP manufacturer’s processing software Quickslice™ to generate an output file defining the position of the RP head for physically creating the model, slice by slice every 0.25 mm. Each model was constructed in approximately 90 min (Fig. 1(b), Fig. 3(a)). A negative flexible transparent silicone (Elastosil, Wacker-Chemie) model of the stereolithographic replica was fabricated and used for MRI flow measurements. The physical model was assumed rigid for the range of pressures applied in the flow experiments.

2.4. Flow measurements

A two-dimensional phase contrast sequence was used to measure the velocity vector by exciting a 2 mm slice with a 0.23 mm in-plane resolution. To improve the signal-to-noise ratio of the measurement a contrast agent (Gd-DTPA) was introduced into the circulating fluid. The MRI flow acquisition method has been discussed in [10]. Two flow waveforms were applied at the graft inlet: a simple sinusoidal waveform with a superimposed steady flow component and a modified coronary bypass graft waveform measured in vivo by Doppler ultrasound during surgery [7]. The mean Reynolds (Re) number was 100 and the Womersley parameter was equal to 2.7 in both cases. For the sinusoidal flow the normalised flow rate ($Q/Q_{\text{mean}}$) varies in the range of 0.05 to 1.75 whereas for the coronary flow it varies between $-0.1$ and 2.25. Measurements with steady flow ($Re = 100$) were also acquired. A computer controlled flow simulator (Shelley Medical Imaging Inc., Ontario, Canada) was used to generate the flow waveforms.

2.5. Computations

The computations were performed using a spectral/hp element algorithm [15] to solve the three-dimensional incompressible Navier–Stokes equations for steady flow. The code has been extensively validated both experimentally and analytically [10,15,16]. At the inflow boundary (located approximately three diameters upstream of the junction) a fully developed laminar pipe flow was imposed with the conditions of constant pressure and zero normal derivatives of velocity enforced at the outflow (located approximately 7 diameters downstream of the junction). The wall shear stress vector was computed from the steady flow solution at $Re = 100$. The computational domain was subdivided into 2469 tetrahedral elemental subdomains. Simulations were performed at polynomial orders of $P = 2, 4, 6$ which correspond to approximate local degrees of freedom of 30,000, 116,000 and 291,000, respectively. A time step of $\Delta t = 0.001$, normalized by mean inlet velocity per vessel diameter, was applied.

3. Results and discussion

The agreement between the numerical flow simulations and the experiment was very good with respect to the axial velocity component but less so for the in-plane velocity components (Fig. 2). This is
Fig. 2. (a) Surface representation of the coronary artery bypass graft distal anastomosis in side view with distal host section extended. Dimensions are normalised by the graft inlet diameter (D). (b) Comparison of the numerically computed and MRI measured axial velocity contour plots for steady flow (Re = 100) extracted from the plane indicated by the dashed line in (a). Cross flow streamlines are superimposed.

evident from the differences in the cross flow streamlines between the numerical predictions and the measurements. However both experiments and computations indicate the presence of a dominant vortex and are in agreement with respect to the location of its core. The wall lumen boundary applied in the experimental results is calculated by fitting a closed spline through the single pixel edge extracted from the respective modulus MRI images. Differences in the shape of the computational and experimental boundary can be attributed to the finite slice thickness of the measurements and the additional errors introduced in the transformation of the digital representation of the geometry to a physical model.

A comparison of the rapid prototype replica and the cast aorto-coronary distal anastomosis is shown in Fig. 3(a). The replica model dimensions are increased by a factor of 1.5 to allow more precise flow experiments. The spatial distribution of the numerically computed wall shear stress magnitude normalised by the inlet Poiseuille wall shear is shown in Fig. 3(b). The peak wall shear region (arrowed in Fig. 3(b)) is located in the vicinity of the graft suture line and is associated with a sudden contraction of the graft as it joins the host artery. The cross sectional area of the graft lumen is reduced by approximately 60% in the locality of the suture line as compared to the graft inlet lumen area. This contraction occurs primarily within a short axial distance equivalent to less than 0.5 graft inlet diameters. A second elevated shear region is located on the bed of the host artery with a peak shear magnitude 50% less than that found in the vicinity of the graft suture line. In the occluded section of the anastomosis, shear levels are low due to the reduced mass flux through the heel of the anastomosis.

By fabricating a physical model of the distal aorto-coronary bypass graft it was possible to investigate the effects of flow unsteadiness on the perianastomotic flow field. Three flow regimes were examined: (a) steady, (b) single harmonic and (c) multi harmonic flow. The strong influence of the flow waveform
Fig. 3. (a) Close-up view of the distal anastomosis and the scaled (1.5 × original) stereolithographic replica corresponding to the outlined cast region. (b) Views of the bed or myocardial (0°), and lateral (120° and 240°) epicardial surface distribution of the numerically computed normalised shear magnitude for steady flow (Re = 100). Dotted line indicates extraction location of slice shown in Figs 2 and 4. Arrows indicate elevated shear regions in the vicinity of the suture line.

Fig. 4. Contours of the MRI measured normalised axial velocity and cross flow streamlines just distal to the toe of the distal coronary bypass graft anastomosis near end deceleration applying sinusoidal flow (a) and a multi harmonic physiological flow waveform (b) at the graft inlet. The instantaneous Reynolds numbers are approximately 30 and 25, respectively. Although cross flow is similar, axial velocity distribution is markedly different.

applied at the inlet of the anastomosis is apparent in Fig. 4. Although the instantaneous Reynolds number is not significantly different between the multi and single harmonic flow results presented (30 and 25, respectively), there is a significant change in the flow field. In the coronary waveform case just distal to the toe and at near end deceleration the flow reversal region is more extensive and its magnitude is on average 20% higher than that found at the same location in the sinusoidal flow field. This can be attributed
to the steeper flow deceleration associated with the higher harmonics of the coronary waveform. It should be noted that there is no axial flow separation in steady flow. The cross flow streamlines indicate that although a primary vortex dominates the flow, a weak counter rotating vortex is present at the bottom of the cross section. Finally the similarity in cross flow streamlines between flow regimes suggests a lesser effect of the unsteadiness on the in-plane velocity distribution.

4. Conclusion

A procedure whereby the effects of geometry on the local flow field in bypass grafts can be investigated in more realistic configurations was developed. The good agreement shown between results from numerical computations and MRI measurements of the flow in a scaled model of an aorto-coronary bypass graft derived from *ex vivo* MR imaging of a cast demonstrates the applicability of this procedure to well resolved *in vivo* data. The ability to reproduce *in vivo* geometries in this manner allows parallel computational and experimental investigations of physiologically relevant flows. This approach provides a valuable tool for detailed flow investigations in difficult to image vessels, due to their small size or pattern of motion or both (coronary arteries). Preliminary results from a flow investigation based on this approach for an aorto-coronary bypass graft indicated the strong influence of the flow waveform applied at the graft inlet on the perianastomotic flow field. It was also shown that a reduction of the graft calibre near the suture line produced a significant increase in the local wall shear stress that was 5 times higher than the fully developed wall shear stress calculated at the inlet. Further work will extend current flow investigations in models of *in vivo* extracted geometries with inlet flow waveforms also measured *in vivo*.

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References


