COMPARISON OF BLOOD FLOW PATTERNS FROM CFD AND MRI IN A BYPASS GRAFT MODEL

Joy P. Ku (1), Christopher J. Elkins (2), Charles A. Taylor (2,3)

(1) Department of Electrical Engineering
Stanford University
Stanford, CA

(2) Department of Mechanical Engineering
Stanford University
Stanford, CA

(3) Department of Surgery
Stanford University
Stanford, CA

INTRODUCTION

Taylor, *et al*. have described the use of computational fluid dynamics (CFD) in a simulation-based medical planning system that enables the preoperative assessment of alternate treatment plans for vascular disease [1]. Validation of these simulation methods against experimental studies demonstrated favorable agreement in blood flow rates and flow rate waveforms in bypass grafts [2]. However, flow patterns were not compared. Other validation studies have examined the accuracy of the blood flow patterns predicted by numerical simulation methods. The most complex of the geometries used in these studies have been of bifurcations [3, 4] and anastomotic junctions [5, 6].

To date, no studies have validated the flow patterns predicted by numerical simulation methods for the more complex geometry of a stenotic vessel with a bypass graft under conditions that are suitable for surgical planning purposes. Unlike many previous studies, the simulations in the following *in vitro* validation study assumed no *a priori* knowledge of the flow distribution between vessels. Furthermore, the model used in this study included both the proximal and distal anastomoses, thus incorporating both diverging and converging flow situations. Comparisons were made between the flow patterns predicted by the numerical simulations and those measured using magnetic resonance imaging (MRI) for this geometry.

MATERIALS AND METHODS

A phantom model was constructed out of a photoreactive resin using stereolithography [7]. The model consisted of a host vessel with a 75% stenosis, or narrowing in the vessel's cross-sectional area, and a bypass graft attached both proximally and distally to the stenosis [8], as shown in Figure 1. The fluid used in this experiment was a mixture of 39.8% glycerol, 59.7% distilled water, and 0.5% gadolinium by volume. This mixture's viscosity, as measured using a Cannon-Fenske viscometer (International Research Glassware, Kenilworth, NJ), was 0.039 dyn s/cm^2 at 22^{o}C, and the density was 1.1 g/mL. A blood flow pump (Harvard Apparatus, Holliston, MA) was used to generate a pulsatile flow waveform at the inlet of the model, and an electrocardiogram (ECG) simulator (Shelley Medical Imaging Technologies, London, Ontario, Canada) converted the pump's trigger signal to an ECG signal used by the MRI system.

Magnetic resonance angiography (MRA) provided the volumetric data for the numerical simulations, while two-dimensional (2D) cine phase-contrast magnetic resonance imaging (PC-MRI) was used to measure the three components of velocity at the planes indicated in Figure 1. The grad-warp corrected MRA data [9] was used to construct a solid model and finite element mesh, as described in [2], and the PC-MRI measured velocities were mapped onto the mesh inlet [10].

A no-slip boundary condition was applied to the walls and the outlet pressure was set to zero. The fluid was modeled as incompressible and Newtonian with a viscosity and density as measured above, and the walls were assumed to be rigid. Under these boundary conditions and assumptions, pulsatile flow was computed for 30 cycles using a previously validated finite element method [11]. The numerically computed velocities were then averaged over 8 cycles to produce relatively periodic results.

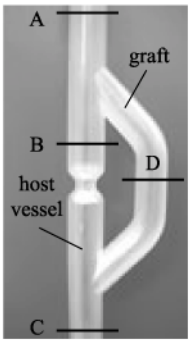

Figure 1. Phantom model used in the experiment. Inlet velocities are prescribed upstream (A) and comparisons of flow patterns are made at planes B (aorta), C (outlet), and D (graft).

RESULTS

Velocities averaged over cycles 14 to 21 from the simulations were compared against the PC-MRI-acquired velocities at three locations, as indicated in Figure 1: in the aorta (B), outlet (C), and graft (D). Figure 2 shows isocontours of the through-plane velocities at these locations at time points corresponding to the maximum and minimum flows. The Womersley number was 5.9, and based on the average inlet velocity, the Reynolds' number (Re) was 264.

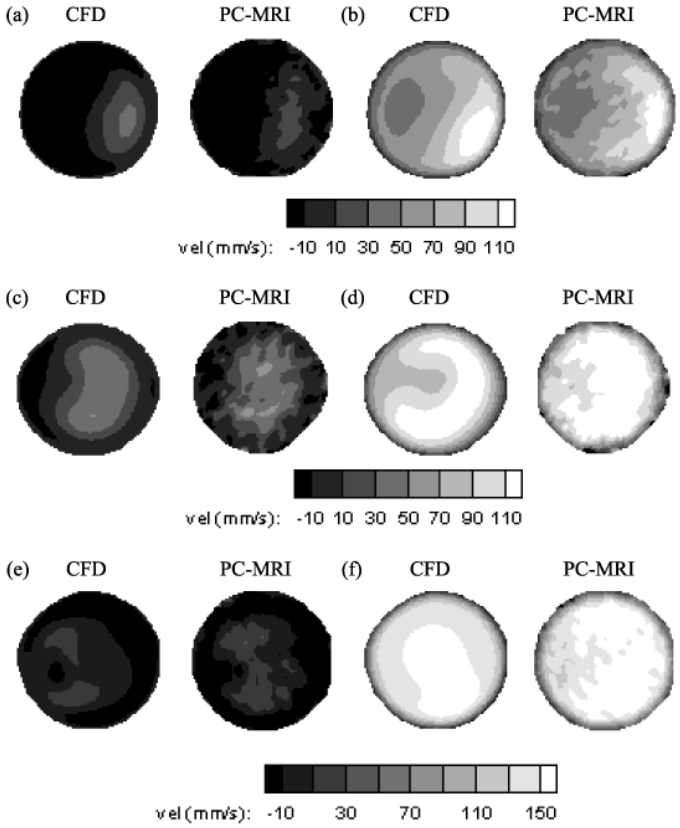

Figure 2. Comparison of isocontours of through-plane velocities in the aorta (plane B) at (a) minimum flow and at (b) maximum flow. (c) and (d) compare results in the bypass (plane D) at the minimum and maximum flow time points respectively, while (e) and (f) are comparisons at the outlet (plane C) at the minimum and maximum flow time points respectively.

DISCUSSION

There is reasonable agreement between the blood flow patterns measured using PC-MRI and those generated by the CFD methods, particularly at the bypass and aorta planes. The magnitudes and shapes of the isocontours at these two locations are similar. Although the velocity magnitudes at the outlet plane are also similar, there are more striking differences in the isocontour shapes. This would be expected since two flow streams converge at the outlet location, resulting in more complex flow patterns than at the aorta or bypass planes. The increase in complexity may be more difficult both to image with MRI as well as to model with CFD methods.

Numerous factors affect both the numerical simulations and the PC-MRI data. The angle of the PC-MRI plane and the temporal resolution influence the PC-MRI accuracy, while the numerical simulations are sensitive to changes in the geometry, particularly the stenosis size and the graft attachment angle. Investigation of parameters such as these could lead to even better agreement between the simulation results and the PC-MRI measurements.

The results presented in this investigation are promising, and future work would extend these comparisons to flows at higher Re. Current *in vitro* investigations have been limited to Re below 1000, while physiologically, the average Re in the human aorta is estimated to be in the range of 1325 to 2000 [12, 13]. The eventual goal is to achieve comparable agreement in *in vivo* experiments.

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