Cardiovascular Flow Simulation by Correlation based Optical Flow

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Introduction

World Health Organisation statistics showed 17.3 million death was imputed to cardiovascular diseases in 2008 [1]. For diagnosis of them, Computed Tomography (CT), Magnetic Resonance Imaging (MRI) and Ultra Sound (UL) techniques are used. The former two have advantage in solution though 2D UL is the most widely spread because of its cost effectiveness. In this research, voxel geometry, which is modeled from 2D UL images, driven unsteady computational fluid dynamic (CFD) was presented. The accuracy of the proposed technique is considered to be inferior to analysis using native 3D data. However, for an initial heart function assessment, the suggested method can be candidate to judge necessity of detailed examination. To initiate computational fluid calculation, region based optical flow calculated velocities were applied on fluid / solid interaction boundary. Using them, Navier-Stokes equation was solved over the fluid segment.

Method

Since image is created from square cells called pixels, space inertial based Eulerian mesh is considered to be suited. In this method, a scalar parameter to judge fluid / solid was assigned to the cells. 24 sequentially photographed source images were chronologically interpolated to 96 images by 3rd order polynomial interpolation to ensure CFD stability. The segmented images were created manually using VSG Avizo. Figure 1 describes original image and schematic illustration of fluid / solid Eulerian mesh segmentation.

To construct 3D geometry from 2D UL images, thrree hypothesises were used. First, UL images were assumed to be on inertial 2D plane in 3D solution domain. Second symmetry in depth direction was assumed. Finally, depth is determined by s scalar depth determination parameter which is the solution of Laplace equation in the fluid region. Figure 2 illustrates, reconstructed 3D geometry by this procedure.

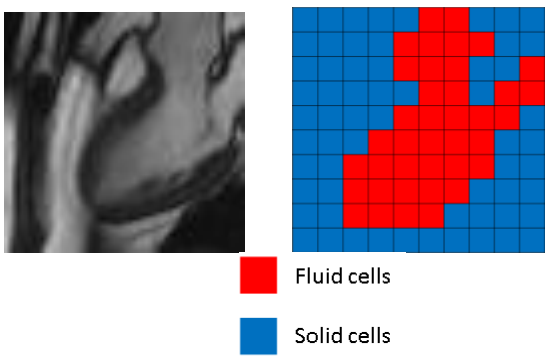

Figure 1: Source Ultra Sound Image (left) and 2D Eulerian Mesh Representation (right)

For the correlation calculation of block matching algorithm Pearson correlation was used to identify local similarities. The block match template size and search region was set to 5 and 7. Similar search configuration was used by Duan et al. [3]. For CFD scheme, Navier-Stokes equation was decoupled by highly simplified marker and cell method (HSMAC) [5]. Then advection term and diffusion term were spatially discritised by scheme second order upwind method and second order central difference scheme respectively. The overall calculation flow is described to be Figure 3. The communication between optical flow and CFD was process at boundary condition treatment. For numerical test, density of 1050 kg per cubic meter and 0.0035 Pa.s of viscosity were assigned to simulate blood.

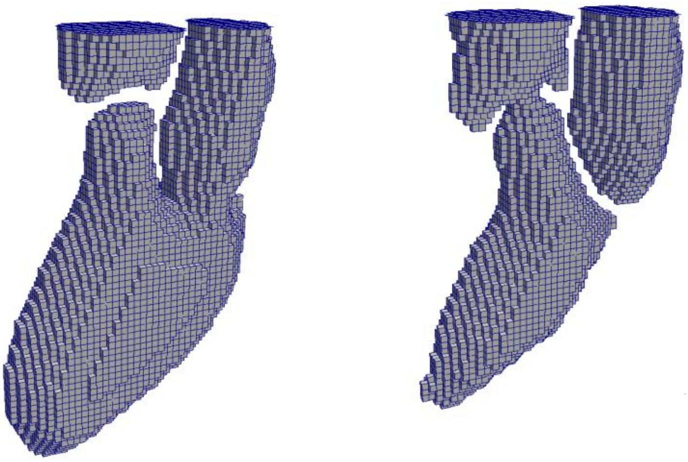

Figure 2: Constructed 3D Left Ventricle Geometry,

Left End of Diastole Phase, Right End of Systole Phase

For boundary condition, aorta and pulmonary vein were set free inlet/outlet boundary condition (yellow lines in Figure 3). For the wall boundary condition (black lines in Figure 3), a normal component of optical flow calculated heart wall velocity was substituted. For solid / fluid interaction boundary . Since regular grid and staggered grid were used for optical flow and CFD scheme respectively, to substitute optical flow velocity component interpolation scheme was required..

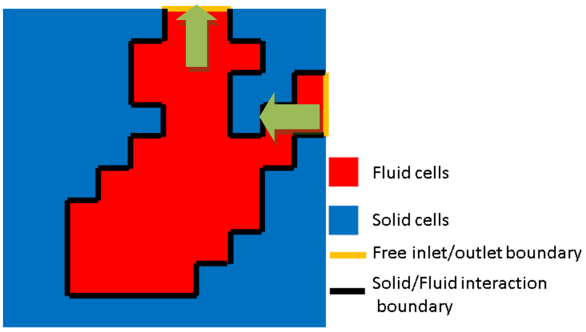

Figure 3: Calculation Domain and Boundary Condition

Result

The calculated wall velocity is described to be Figure 4. Figure 4 (a) illustrates velocity vector in systole phase and Figure 4(b) was taken from diastole phase. We can see inward velocity vectors in Figure 4(a) and opposite direction vector in Figure 4 (b). Figure 5 illustrates calculated heart wall movement velocities. Figure 5 (a) shows systole flow and Figure 5 (b) describes diastole flow. Figure 5 demonstrates, the contraction and expansion were tracked by the optical flow scheme. Figure 6 suggest, the characteristic of systole and diastole flow was recreated by this approach. The overall, outflow and inflow was agreed about 7% of difference assuming averaged cross sectional area over the cycle. The measurement lines were at the free flow boundary of aorta and pulmonary vein due to small geometrical change.

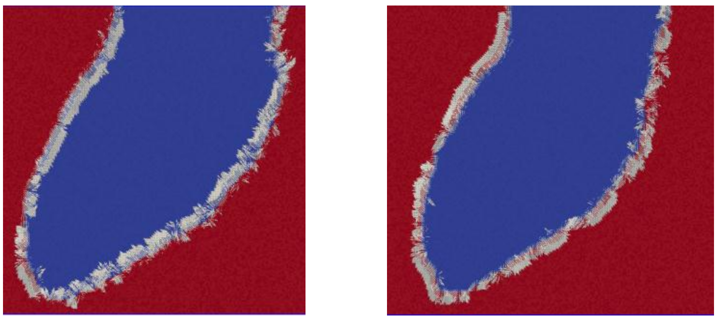

Figure 4: (a) Wall Velocity in Systole Phase   (b) Wall Velocity in Diastole Phase

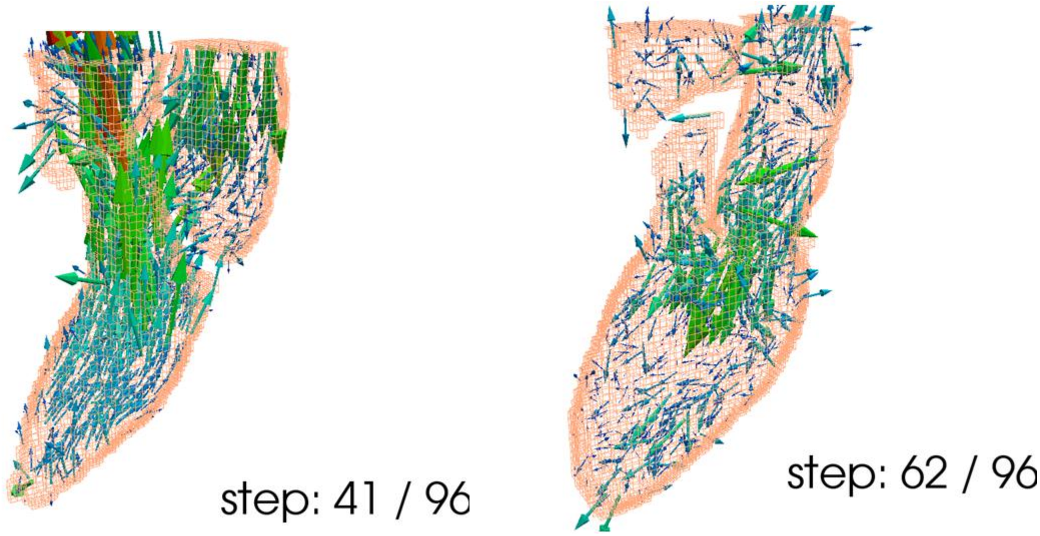

Figure 5: (a) Velocity Vector in Systole Phase (b) Velocity Vector in Diastole Phase

Conclusion

Cardiovascular simulation using the combination of computational fluid dynamics and block matching scheme was proposed in this work. Since Correlation method does not depends on pixel luminosity, more stable optical flow velocity than gradient based scheme was calculated by extracted images. The characteristic of cardiac flow was captured. The cause of disagreement of in/out flow is estimated geometrical change. Hence, exact cross sectional area near the valves need to be traced over the cycle.

References

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