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Transient CFD Modeling of Blood Flow Simulation through Left Coronary Artery

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Abstract: ***From the past twenty years, cardiac strokes have become one of the most common causes of death. Coronary artery disease means blockage of at least one course that supplies blood to the heart. These blockages are caused by the accumulation of plaque in the artery walls. That LDL cholesterol can restrict blood flow in the arteries and it directs to myocardial infarction; The CFD simulation is increasingly performed to study fluid phenomena inside the human vascular system. This simulation aims to develop a CFD model of coronary arteries and observe blood through the arteries, Estimate some of the hemodynamic parameters of the blood during the systolic and diastolic phase hemodynamic parameters were quantified and flow patterns are visualized visualized using CFD in the presence of plaques and forecast pressure rate, velocity in the arteries.***

Keywords: CFD modeling, coronary artery, blood flow analysis, plaque, transient flow.

I. INTRODUCTION

The cardiac muscle needs oxygen-rich blood to work like all the other cells in the body. Coronary arteries provide blood to the heart. The coronary is surrounding the exterior of the heart tiny branches reach into the muscle of the heart. Coronary flow in other areas of the arterial system is unique from flow as it is caused by heart contraction and relaxation. Early hemodynamic studies began with larger arteries, such as the aorta, and theories were developed that combine WSS and plaque accumulation. When plaque clogs our arteries, it is called atherosclerosis. These deposits consist of LDE cholesterols. As plaque accumulates, the thickness of the blood vessel wall shrinks the artery channel, restricting the flow of blood. In turn it reduces the amount of oxygen and other nutrients entering the body. Plaque may partially or completely block Blood flow in the heart, brain, pelvis, legs, arms, or kidneys through large or medium-sized arteries.

Zhou Y, kassab [1] mentioned coronary distribution in their work had a design constraint. The length of the sub tree from a subdivision determined the diameter of an arterial segment. It could be said that the variations in the diameter of the inner individual branch and the flow rate would be consistent with the variations of myocardial perfusion models. Asakura and Karino [2] treated the corpses of the CA so that they could be transparent and the flow of internal fluid could be observed. They observed that lesion accumulation occurred in two points: the outer and inner walls of bifurcation along the curvature. Vignon-Clemente [3] explains the correlation between pressures as a function of flow rate or coupling speeds it is possible to determine the interfaces, therefore it is possible to derive the conditions around the outflow for that downstream domain. In this analysis, various methods developed to accurately represent the boundary conditions of the Outflow, the state of impedance limit,a model based on the concept of electricity impedance.

B.M. Johnston, P.R. Johnson, S. Corney, and D.Kilpatrick,[5] the liquid to be incompressible and Non-Newtoninan and in the geometry of the of the arteries affects hemodynamic, including flow reactions,etc., but the inlet and outlet conditions affect large-scale flow characteristics such as the flow ratio at the branching site and the pressure distribution.

Biyue Liu, Jie Zheng, Richard Bach and Dalin Tang.M[7]. the flow patterns, pressure and velocity graphs are quite similar to the left and right sides of the arteries. In this analysis, various methods developed to accurately represent the boundary conditions.

II. GEOMETRY CREATION

For complex anatomical models, CT and MRI scans are preferable methods for geometry creation. The patient's image data is taken from the patient's CT scan reports (the one who suffering from CAD). In the beginning, the geometry is smoothened by using the blender. Then this geometry was imported to ANSYS Space Claim for convert to CAD model. Using skin surface tool in ANSYS Space Claim, several faces were created on the surface of the body, the solid model was generated by filling surfaces.

Figure1: The patient's image data is first converted into 3D Model and then the reverse engineering process extract a solid CAD model.

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Figure 2: Geometry

III. MESHING

Meshing was done by using Element size is 0.5 mm Inflation was done at inlet and aorta outlets,

Transition ratio=0.272
Maximum layers =5
Growth rate=1.2

The mesh of the geometry consists of 314207 elements, 83,274 nodes

After meshing, the geometries were imported to ANSYS FLUENT for numerical implementation and solution. This step involves assigning appropriate material properties, choosing governing equations, imposing relevant boundary conditions and finally running calculation. Steady flow analysis was performed and then time dependent boundary conditions were imposed for transient flow analysis.

IV. MATHEMATICL MODELING

Navier-Stokes:
EquationsIt is assumed that the Navier-Stokes Equations
regulate the flow of blood in the artery. The equation of continuity for an incompressible fluid

$$\nabla . \vec{V} = 0$$

Momentum-Equation:

$$\rho \frac{D\vec{V}}{Dt} = -\nabla p + \rho \vec{g} + \mu \nabla^2 \vec{V}$$

Where

$\frac{D\vec{V}}{Dt}$ - Material derivative of velocity, ∇p - Pressure force, $\rho \vec{g}$ - Body force, $\mu \nabla^2 \vec{V}$ - Viscous force

In this, writing the Navier-stokes equations allows the versati lity ofusing an arbitrary non-Newtonian blood model. B.M. Johnston, P.R. Johnson, S. Corney, and D.Kilpatrick[5] the liquid to be incompressible and Non-Newtoninan.

V. SETUP AND FLOW SPECIFICATIONS

The equations are solved by some assumptions such as blood was considered as incompressible and non-Newtonian fluid in order to solve governing equations. At all inlet and outlet limits, the equations are solved with no-slip boundary conditions and the gauge pressure is to be steady-state aorta outlet at zero

A. Power Law Model : $\eta=\eta 0\ \gamma\ n-1$

Where

- η0 Is the viscosity of zero-shear rate and n is a dimensionless parameter.
- Zero shear rate limit (η0) =0.035 Pa-s
- Relaxation time constant (γ) =3.313 s
- Power law index (n) =0.7
- Blood specific hear capacity =3617 J Kg-1 k -1
- Blood conductivity =0.52 W M-1 k -1
- Density= 1054 Kg/m3

B. Boundary Conditions: The working fluid in all cases is fluid.

Generally Coronary circulation is 5% of the total cardiac output.
The cardiac output is 5.5 l/min,
so coronary circulation is 275 ml/min.

Then this coronary circulation given at both LCA and RCA outlets.

Inlet (mass flow rate) = 5.5 l/min.

1) Outlets
a) At LCA (mass flow rate) =235 ml/min
b) RCA (mass flow rate) = 40 ml/min
c) Aortic (pressure) =100 mm-hg

Transitory state, time-dependent boundary conditions have been imposed in both theories. This time-dependent data are collected from the reference journal paper. The data was collected in a worksheet format for Microsoft Excel. The data was extracted in a Microsoft Excel worksheet format. So this was converted to comma-separated values (.csv). These files are imported into ANSYS. For the LCA, for the conditions surrounding the outflow, we take the average of the LCA and LCX flow rates.

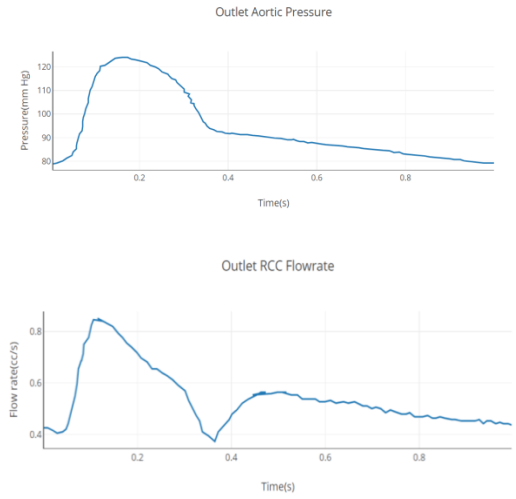

Figure 3: Transient boundary conditions Flow rates

C. Calculation:

For transient flow analysis, the following data was given to perform the calculation for 1 second.

- Time step size = 0.055 s
- Number of time steps = 510
- Number of iterations per time step = 10

VI. RESULT AND DISCUSSION

A) Steady state:

Volume description is taken with velocity as variable. And the pressure and velocity through artery reading are obtained by the steady state boundary conditions applied at outlets and the pressure distribution was observed; the maximum pressure obtained is 1.334e+04 Pa, maximum velocity obtained is 1.434 m/s

B) Transient state:

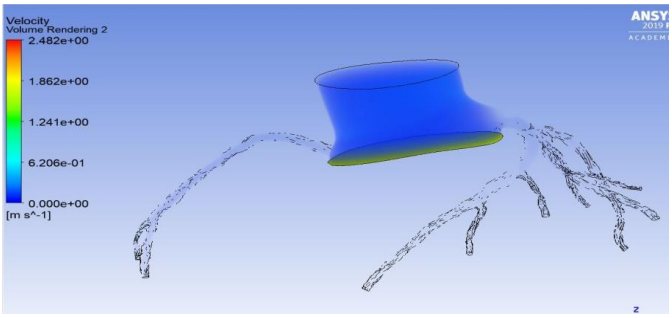

Figure 4: Volume rendering taking velocity as variable

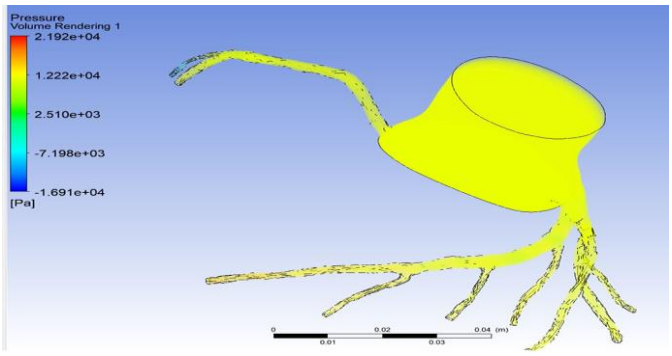

Figure 5: Volume rendering taking pressure as variable

At transient state boundary conditions, applied at all outlets and observed the pressure distribution, the maximum pressure obtained is 2.482e+04 Pa at LCA and at the pressure at RCA is 1.982e+04 pa. The blood flow rate in the LCA is more restricted than in the RCA.
Because the LCA
has more complex geometry and plaques occur in
regions near to bifurcations. The maximum velocity obtained at the LCA is 2.082e+00 m/s.

Transient state	Mass flow Flow rate outlet (ml/min)	Pressure outlet (mm of hg)	Velocity outlet (m/s)
Left coronary	235	186.165	2.082
Right coronary	40	148.66	1.766

VII. CONCLUSION

At steady boundary conditions, the blood pressure at the arte ries are to 98 mm of hg it is nearer to input value and at the time varying boundary conditions the blood pressure is more compare to the steady state boundary conditions and the maximum velocity obtained at LCA is 2.082e+00 m/s and the difference between LCA and RCA is 38mm of hg, by considering the systolic and diastolic pressures on arteries, the pressure difference is goes up to 41mm of hg.
The LCA blood pressure pressure is high compare to the RCA due to plaque accumulation at the walls of the arteries. The plaque deformation in the heart is restricts the blood flow to the heart muscles and rises the blood pressure in the arteries. In these type of cases, the heart doesn't get enough rich oxygenated blood; the patient may feel discomfort in the chest pain (angina) or it leads to myocardial infarction.

FUTURE SCOPE

The geometry which we have developed is used for further investigations for the flow pattern for the blockage of arteries. From these study doctor may come up with the individualized treatment plans.

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