MASTER’S PROJECT II REPORT

MDC 20112

FSI ANALYSIS OF ARTERY SUFFERING ATHEROSCLEROSIS

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ABSTRACT

From the statistics of international death rates, United States shows 169.4 per 100,000 of population death in year 2005 because of coronary heart disease (CHD). An atheroma is an accumulation and swelling an artery walls that made up of (mostly) macrophage cells, or debris, that contain lipids (cholesterol and fatty acids), calcium and a variable amount of fibrous connective tissue. It results the artery wall thickens and hardens. This is called as atherosclerosis or “hardening of the arteries“. Thus, will resulting in a heart attack and ensuing debility. In this project, a Fluid – Structure Interaction analysis is done on a blood flow in an artery which suffers atherosclerosis disease. The most often atherosclerotic plaque accumulate is at the abdominal aorta. This analysis is done on a simplified model of abdominal aorta. The conditions to be analyze is regarding to the size and the stiffness of the atheroma. The software tools that using in this research is the ANSYS Workbench, with the coupling of ANSYS and CFX analysis. The output result of structural analysis of the abdominal aorta shows that the abdominal with the size of 15% atheroma shows the highest value of the Maximum Total Mesh Displacement and Von Mises Stress. While the result of fluid analysis of the blood flow in abdominal aorta shows that the increases of size and stiffness of atherma will increases the absolute gradient for the pressure and velocity of blood flow at the point 3 in analysis. However, the increases of absolute gradient of velocity and pressure of blood flow for the increases of atheroma stiffness is lower than the increases of absolute gradient of velocity and pressure of blood flow for the increases of atheroma size. Thus, the visualization of the interaction between blood flow and artery wall in the vicinity of atheroma will help medical researchers and doctor understand atherosclerosis better.
ABSTRAK

CHAPTER I  INTRODUCTION  1

1.1 Background of Study  1

1.2 Definition of Terminology  4

1.2.1 Atheroma  4

1.2.2 Atherosclerosis  4

1.2.3 Finite Element Method  5

1.2.4  Ansys  5

1.2.5 Optical Coherence Tomography (OCT)  6

1.3 Significant of Study  6

1.4 Objective  7

1.5 Scopes of Study  7

1.6 Expected Result  8
1.7 Flow Chart of Project Planning

CHAPTER II LITERATURE REVIEW

2.1 Introduction
2.2 Previous Research

CHAPTER III METHODOLOGY

3.1 Introduction
3.2 Methodology Flow Chart
3.3 Geometrical Model of Artery Vessel
   3.3.1 Selection of Artery Vessel
   3.3.2 Geometry of Artery Vessel
   3.3.3 Properties of Artery Vessel and Atheroma
3.4 Parameter Assumptions and Blood Properties
3.5 Governing Equation and Boundary Conditions
3.6 Finite Element Modeling
   3.6.1 CAD approach to solid model construction
   3.6.2 Meshing
3.7 Fluid-Structure Coupling
3.8 Model Validation and Analysis
3.9 Sampling
   3.9.1 Artery Blood Vessel
CHAPTER IV RESULTS AND DISCUSSIONS

4.1 Introduction

4.2 Blood Velocity

4.3 Healthy Abdominal Aorta (AA)
   4.3.1 Blood Velocity Against Time Graph
   4.3.2 Blood Pressure Against Time Graph
   4.3.3 Blood Flow Pressure
   4.3.4 Blood Total Mesh Displacement
   4.3.5 Von Mises Stress of Abdominal Aorta

4.4 Abdominal Aorta (AA) With Different Size of Atheroma

4.5 Abdominal Aorta (AA) With Different Stiffness of Atheroma

4.6 Graph Analysis of AA Structural Analysis
   4.6.1 Analysis Of Total Mesh Displacement of AA for all Conditions
   4.6.2 Analysis of the Von Mises Stress For AA for All Conditions

4.7 Graph Analysis of Blood Flow Analysis
   4.7.1 Velocity Graph for The Healthy and With
Atheroma Size of AA 70

4.7.2 Velocity Graph for The Healthy and With

Atheroma Stiffness of AA 80

4.7.3 Pressure Graph for The Healthy and With

Atheroma Size of AA 90

4.7.4 Pressure Graph for The Healthy and With

Atheroma Stiffness of AA 100

CHAPTER V CONCLUSION AND RECOMMENDATION 110

5.1 Conclusion 110

5.2 Recommendation for Future Works 111

REFERENCES 112

APPENDIX A: FLOW CHART: PROJECT PLANNING FOR MP 1 116

APPENDIX B: FLOW CHART: PROJECT PLANNING FOR MP 2 118

APPENDIX C: GANTT CHART: PROJECT PLANNING FOR MP 1 120

APPENDIX D: GANTT CHART: PROJECT PLANNING FOR MP 2 122

APPENDIX E: RESULT OF FSI (FLUID STRUCTURE INTERACTION) ON ABDOMINAL AORTA AND BLOOD FLOW ANALYSIS 124-155
### LIST OF FIGURES

<table>
<thead>
<tr>
<th>FIGURE</th>
<th>TITLE</th>
<th>PAGE</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.1</td>
<td>The arterial layers of a human body [2]</td>
<td>2</td>
</tr>
<tr>
<td>1.2</td>
<td>Atheroma [31]</td>
<td>4</td>
</tr>
<tr>
<td>1.3</td>
<td>Flow Chart of Project Planning</td>
<td>9</td>
</tr>
<tr>
<td>2.1</td>
<td>distribution of magnitude of surface traction average over one cardiac cycle</td>
<td>11</td>
</tr>
<tr>
<td>2.2</td>
<td>Evolution of surface traction over one cardiac cycle</td>
<td>12</td>
</tr>
<tr>
<td>2.3</td>
<td>Illustration of the “Coupled Multidomain Method”, where information from the downstream (analytical) domain is incorporated in the upstream (computational) domain through a boundary map. This approach accommodates a variety of downstream models ranging from pure resistive to complex electrical analogs and impedance models.</td>
<td>13</td>
</tr>
<tr>
<td>2.4</td>
<td>Comparison of flow and pressure wave forms in a carotid artery with a periodic inlet flow and constant pressure, resistance and impedance outlet boundary conditions. For the pressure boundary condition, inlet pressure varies little from the prescribed constant outlet pressure and the peak pressure precedes the peak flow. The resistance boundary condition gives rise to an unrealistically large pressure amplitude. In addition, pressure and flow are in phase at the outlet. For the impedance boundary condition, the range of pressure is from approximately 85–115 mmHg</td>
<td></td>
</tr>
</tbody>
</table>
and the pressure lags flow along the length of the vessel. $1 \text{ mmHg} = 133.3 \text{ Pa}$, $1 \text{ cc/s} = 10^{-6} \text{ m}^3/\text{s}$.

2.5 (a) Geometric model and (b) flow distribution between normal iliac artery and an iliac artery with a stenosis that reduces the cross-sectional area by 75% for constant pressure and impedance boundary conditions. For a constant outlet pressure, the flow split is dictated solely by the resistance to flow due to the geometry of the computational domain (approximately 70% to the normal side, 30% to the stenosed side). For an impedance boundary condition with a physiologic level of resistance, the flow split (50%/50%) is principally determined by the downstream demands (represented by equal outlet impedances). These results are consistent with clinical observations for resting flow conditions and iliac artery stenoses with less than an 85% area reduction.

2.6 Flow diagram of the algorithm for the calculation of Womersley developed profiles

2.7 Flow diagram of FIDAP-W0MER.f subroutine interface

2.8 Geometrical model of vessels

2.9 Finite element modeling process for a lipid rich coronary plaque cross-section. The OCT image (A) and histology image (B) are segmented (C,D) into lipid rich, fibrous plaque, and arterial wall regions. Each segmented image is used to create a finite element mesh (OCT mesh, E, with closeup in F). Application of an internal pressure load results in elevated stress at systolic pressure (G,H) and cyclic strain (I,J) distributions. Results for the OCT-based model are shown in the left column; results for the histology based model are shown in the right column.
2.10 Lipid positive probability map for the 30 to 34 age group
2.11 A schematic of the segment of femoral artery
2.12 Finite element model
2.13 Flow waves at four different locations in Figure above.
2.14 Pressure waves at four different locations in Figure above
2.15 Flow waves at the distal popliteal for 0, 50, 75 and 100 percent stenosis
2.16 Pressure wave at the distal popliteal for 0, 50, 75 and 100 percent stenosis
2.17 Pressure waves for no shear case and different blood models at the distal popliteal.
2.18 The simulated relationships between arterial pressure and arterial volume compliance for the artery with 100 kPa unstressed circumferential elastic modulus.
2.19 The simulated relationships between arterial pressure and pulse propagation time for the artery with 100 kPa unstressed circumferential elastic modulus.
2.20 Experimental results for the relationship between recorded arterial pressure changes and arm pulse propagation time changes (A) and arterial volume compliance changes (B), which were referred to horizontal values.
2.21 Analytical and predicted radial displacements at the mid-surface of a thin-walled cylinder under uniform pressure with fixed ends.
2.22 Analytical and predicted axial displacements at the mid-surface of a thin-walled cylinder under uniform pressure with fixed ends.
2.23 Analytical and predicted radial displacement at the mid-surface of a thin-walled vertical cylinder (with fixed bottom) filled with liquid.
2.24 Analytical and predicted axial velocities in an expanding pipe (Re=0.1).

2.25 Analytical and predicted radial velocities in an expanding pipe (Re=0.1).

2.26 Predicted radial displacement in the wall at the mid-way of the artery

3.1 Methodology Flow Chart

3.2 The abdominal aorta of a human [19]

3.3 Abdominal aorta model with branches identified [9]

3.4 Seven arterial segments studied include: AsA; PA and DA, descending proximal and distal thoracic aorta; AA, distal abdominal aorta; FA, left femoral artery; BT; right CA.

3.5 Abdominal aorta inflow in three types of modes [9]

3.6 Close-up of abdominal aorta finite element mesh

3.7 Block diagram for the Fluid-Structure Interaction analysis

3.8 Selecting the coupling software in Workbench

3.9 Set up columns

3.10 Example result of Fluid-Structure Interaction in Ansys analysis

3.11 Resting conditions waveforms

3.12 Exercise conditions waveforms

4.1 Healthy abdominal aorta

4.2 Location of points at abdominal aorta

4.3 Blood velocity graph for healthy abdominal aorta

4.4 Blood velocity of resting and exercise condition for AAA

4.5 Blood pressure graph for healthy abdominal aorta

4.6 Blood pressure for healthy abdominal aorta

4.7 Total Mesh Displacement of Blood for healthy abdominal aorta

4.8 Von Mises Stress for healthy abdominal aorta in cross sectional view
4.9  Von Mises Stress for healthy abdominal aorta  63
4.10  Abdominal aorta with different percentage atheroma size  64
4.11  Abdominal aorta with different stiffness of atheroma  65
4.12  Total Mesh Displacement graph for abdominal aorta in all conditions  66
4.13  Von Mises Stress graph for abdominal aorta (AA) in all conditions  68
4.14  Blood velocity of AA for healthy and with atheroma size at time, t = 4s  70
4.15  Blood velocity of AA for healthy and with atheroma size at time, t = 5s  72
4.16  Blood velocity of AA for healthy and with atheroma size at time, t = 6s  74
4.17  Blood velocity of AA for healthy and with atheroma size at time, t = 7s  76
4.18  Blood velocity of AA for healthy and with atheroma size at time, t = 8s  78
4.19  Blood velocity of AA for healthy and with atheroma stiffness at time, t = 4s  80
4.20  Blood velocity of AA for healthy and with atheroma stiffness at time, t = 5s  82
4.21  Blood velocity of AA for healthy and with atheroma stiffness at time, t = 6s  84
4.22  Blood velocity of AA for healthy and with atheroma stiffness at time, t = 7s  86
4.23  Blood velocity of AA for healthy and with atheroma stiffness at time, t = 8s  88
4.24  Blood pressure of AA for healthy and with atheroma size at time, t = 4s  90
4.25  Blood pressure of AA for healthy and with atheroma size at time, t = 5s  92
4.26 Blood pressure of AA for healthy and with atheroma size at time, $t = 6s$  
4.27 Blood pressure of AA for healthy and with atheroma size at time, $t = 7s$  
4.28 Blood pressure of AA for healthy and with atheroma size at time, $t = 8s$  
4.29 Blood pressure of AA for healthy and with atheroma stiffness at time, $t = 4s$  
4.30 Blood pressure of AA for healthy and with atheroma stiffness at time, $t = 5s$  
4.31 Blood pressure of AA for healthy and with atheroma stiffness at time, $t = 6s$  
4.32 Blood pressure of AA for healthy and with atheroma stiffness at time, $t = 7s$  
4.33 Blood pressure of AA for healthy and with atheroma stiffness at time, $t = 8s$  
A.1 Flow Chart of Project Planning for Master Project 1  
B.1 Flow Chart of Project Planning for Master Project 2  
C.1 Gantt Chart of Master Project 1  
D.1 Gantt Chart of Master Project 2  
E.1 Healthy abdominal aorta  
E.2 Blood velocity graph for healthy abdominal aorta  
E.3 Blood pressure graph for healthy abdominal aorta  
E.4 Blood flow pressure for healthy abdominal aorta  
E.5 Total Mesh Displacement for healthy abdominal aorta  
E.6 Von Mises Stress for healthy abdominal aorta in cross sectional view  
E.7 Von Mises Stress for healthy abdominal aorta  
E.8 Abdominal aorta with 10% atheroma size  
E.9 Blood velocity graph for abdominal aorta with 10% atheroma size
E.10  Blood pressure graph for abdominal aorta with 10% atheroma size  
E.11  Blood flow pressure for abdominal aorta with 10% atheroma size  
E.12  Blood Total Mesh Displacement for abdominal aorta with 10% atheroma size  
E.13  Von Mises Stress for abdominal aorta with 10% atheroma size in cross sectional view  
E.14  Von Mises Stress for healthy abdominal aorta with 10% atheroma size  
E.15  Abdominal aorta with 15% atheroma size  
E.16  Blood velocity graph for abdominal aorta with 15% atheroma size  
E.17  Blood pressure graph for abdominal aorta with 15% atheroma size  
E.18  Blood flow pressure for abdominal aorta with 15% atheroma size  
E.19  Blood Total Mesh Displacement for abdominal aorta with 15% atheroma size  
E.20  Von Mises Stress for abdominal aorta with 15% atheroma size in cross sectional view  
E.21  Von Mises Stress for healthy abdominal aorta with 15% atheroma size  
E.22  Abdominal aorta with 20% atheroma size  
E.23  Blood velocity graph for abdominal aorta with 20% atheroma size  
E.24  Blood pressure graph for abdominal aorta with 20% atheroma size  
E.25  Blood flow pressure for abdominal aorta with 20% atheroma size  
E.26  Blood Total Mesh Displacement for abdominal aorta
E.27 Von Mises Stress for abdominal aorta with 20% atheroma size in cross sectional view
E.28 Von Mises Stress for healthy abdominal aorta with 20% atheroma size
E.29 Abdominal aorta with 25% atheroma size
E.30 Blood velocity graph for abdominal aorta with 25% atheroma size
E.31 Blood pressure graph for abdominal aorta with 25% atheroma size
E.32 Blood flow pressure for abdominal aorta with 25% atheroma size
E.33 Blood Total Mesh Displacement for abdominal aorta with 25% atheroma size
E.34 Von Mises Stress for abdominal aorta with 25% atheroma size in cross sectional view
E.35 Von Mises Stress for healthy abdominal aorta with 25% atheroma size
E.36 Abdominal aorta with 30% atheroma size
E.37 Blood velocity graph for abdominal aorta with 30% atheroma size
E.38 Blood pressure graph for abdominal aorta with 30% atheroma size
E.39 Blood flow pressure for abdominal aorta with 30% atheroma size
E.40 Blood Total Mesh Displacement for abdominal aorta with 30% atheroma size
E.41 Von Mises Stress for abdominal aorta with 30% atheroma size in cross sectional view
E.42 Von Mises Stress for healthy abdominal aorta with 30% atheroma size
E.43 Abdominal aorta with 35% atheroma size 140
E.44 Blood velocity graph for abdominal aorta with 35% atheroma size 140
E.45 Blood pressure graph for abdominal aorta with 35% atheroma size 141
E.46 Blood flow pressure for abdominal aorta with 35% atheroma size 141
E.47 Blood Total Mesh Displacement for abdominal aorta with 35% atheroma size 141
E.48 Von Mises Stress for abdominal aorta with 35% atheroma size in cross sectional view 142
E.49 Von Mises Stress for healthy abdominal aorta with 35% atheroma size 142
E.50 Blood velocity graph for abdominal aorta with atheroma stiffness of 2 x10^4 Pa 143
E.51 Blood pressure graph for abdominal aorta with atheroma stiffness of 2 x10^4 Pa 143
E.52 Blood flow pressure for abdominal aorta with atheroma stiffness of 2 x10^4 Pa 144
E.53 Blood Total Mesh Displacement for abdominal aorta with stiffness of 2 x10^4 Pa 144
E.54 Von Mises Stress for abdominal aorta with stiffness of 2 x10^4 Pa in cross sectional view 144
E.55 Von Mises Stress for abdominal aorta with stiffness of 2 x10^4 Pa 145
E.56 Blood velocity graph for abdominal aorta with atheroma stiffness of 10 x10^4 Pa 145
E.57 Blood pressure graph for abdominal aorta with atheroma stiffness of 10 x10^4 Pa 145
E.58 Blood flow pressure for abdominal aorta with atheroma stiffness of 10 x10^4 Pa 146
E.59 Blood Total Mesh Displacement for abdominal aorta with stiffness of $10 \times 10^4$ Pa

E.60 Von Mises Stress for abdominal aorta with stiffness of $10 \times 10^4$ Pa in cross sectional view

E.61 Von Mises Stress for abdominal aorta with stiffness of $10 \times 10^4$ Pa

E.62 Blood velocity graph for abdominal aorta with atheroma stiffness of $50 \times 10^4$ Pa

E.63 Blood pressure graph for abdominal aorta with atheroma stiffness of $50 \times 10^4$ Pa

E.64 Blood flow pressure for abdominal aorta with atheroma stiffness of $50 \times 10^4$ Pa

E.65 Blood Total Mesh Displacement for abdominal aorta with stiffness of $50 \times 10^4$ Pa

E.66 Von Mises Stress for abdominal aorta with stiffness of $50 \times 10^4$ Pa in cross sectional view

E.67 Von Mises Stress for abdominal aorta with stiffness of $50 \times 10^4$ Pa

E.68 Blood velocity graph for abdominal aorta with atheroma stiffness of $100 \times 10^4$ Pa

E.69 Blood pressure graph for abdominal aorta with atheroma stiffness of $100 \times 10^4$ Pa

E.70 Blood flow pressure for abdominal aorta with atheroma stiffness of $100 \times 10^4$ Pa

E.71 Blood Total Mesh Displacement for abdominal aorta with stiffness of $100 \times 10^4$ Pa

E.72 Von Mises Stress for abdominal aorta with stiffness of $100 \times 10^4$ Pa in cross sectional view

E.73 Von Mises Stress for abdominal aorta with stiffness of $100 \times 10^4$ Pa

E.74 Blood velocity graph for abdominal aorta with atheroma
stiffness of $150 \times 10^4$ Pa

E.75 Blood pressure graph for abdominal aorta with atheroma
stiffness of $150 \times 10^4$ Pa

E.76 Blood flow pressure for abdominal aorta with atheroma
stiffness of $150 \times 10^4$ Pa

E.77 Blood Total Mesh Displacement for abdominal aorta with
stiffness of $150 \times 10^4$ Pa

E.78 Von Mises Stress for abdominal aorta with stiffness
of $150 \times 10^4$ Pa in cross sectional view

E.79 Von Mises Stress for abdominal aorta with stiffness
of $150 \times 10^4$ Pa

E.80 Blood velocity graph for abdominal aorta with atheroma
stiffness of $200 \times 10^4$ Pa

E.81 Blood pressure graph for abdominal aorta with atheroma
stiffness of $200 \times 10^4$ Pa

E.82 Blood flow pressure for abdominal aorta with atheroma
stiffness of $200 \times 10^4$ Pa

E.83 Blood Total Mesh Displacement for abdominal aorta with
stiffness of $200 \times 10^4$ Pa

E.84 Von Mises Stress for abdominal aorta with stiffness
of $200 \times 10^4$ Pa in cross sectional view

E.85 Von Mises Stress for abdominal aorta with stiffness
of $200 \times 10^4$ Pa
## LIST OF TABLES

<table>
<thead>
<tr>
<th>TABLE</th>
<th>TITLE</th>
<th>PAGE</th>
</tr>
</thead>
<tbody>
<tr>
<td>2.1</td>
<td>Material properties for the element in blood vessel</td>
<td>24</td>
</tr>
<tr>
<td>2.2</td>
<td>Content of element maximum stress of artery according to age</td>
<td>25</td>
</tr>
<tr>
<td>3.1</td>
<td>Maximum and Minimim Pressure $p_{\text{max}}$, $p_{\text{min}}$ (in millimeter of mercury) and Radii $r_{\text{max}}$, $r_{\text{min}}$ (in centimeters), in Vitro (postexcised) Length $l_e$ (in centimeters), Axial Stretch Coefficient $s_c$, and Wall Thickness $h$ (in centimeters) Average for each Location.</td>
<td>40</td>
</tr>
<tr>
<td>3.2</td>
<td>Material properties</td>
<td>41</td>
</tr>
<tr>
<td>3.3</td>
<td>Material properties for the element in blood vessel</td>
<td>41</td>
</tr>
<tr>
<td>3.4</td>
<td>Flow conditions</td>
<td>45</td>
</tr>
<tr>
<td>3.5</td>
<td>Analysis for different size of atheroma in abdominal aorta</td>
<td>53</td>
</tr>
<tr>
<td>3.6</td>
<td>Analysis for different stiffness of atheroma in abdominal aorta</td>
<td>54</td>
</tr>
<tr>
<td>4.1</td>
<td>Maximum Total Mesh Displacement of abdominal aorta for all conditions</td>
<td>66</td>
</tr>
<tr>
<td>4.2</td>
<td>Maximum Von Mises Stress of abdominal aorta (AA) for all conditions</td>
<td>68</td>
</tr>
<tr>
<td>4.3</td>
<td>Blood velocity of AA for healthy and with atheroma size at time, $t = 4s$</td>
<td>70</td>
</tr>
<tr>
<td>4.4</td>
<td>Blood velocity of AA for healthy and with atheroma size at time, $t = 5s$</td>
<td>72</td>
</tr>
<tr>
<td>4.5</td>
<td>Blood velocity of AA for healthy and with atheroma size at time, $t = 6s$</td>
<td>74</td>
</tr>
</tbody>
</table>
4.6 Blood velocity of AA for healthy and with atheroma size at time, $t = 7s$  
4.7 Blood velocity of AA for healthy and with atheroma size at time, $t = 8s$  
4.8 Blood velocity of AA for healthy and with atheroma stiffness at time, $t = 4s$  
4.9 Blood velocity of AA for healthy and with atheroma stiffness at time, $t = 5s$  
4.10 Blood velocity of AA for healthy and with atheroma stiffness at time, $t = 6s$  
4.11 Blood velocity of AA for healthy and with atheroma stiffness at time, $t = 7s$  
4.12 Blood velocity of AA for healthy and with atheroma stiffness at time, $t = 8s$  
4.13 Blood pressure of AA for healthy and with atheroma size at time, $t = 4s$  
4.14 Blood pressure of AA for healthy and with atheroma size at time, $t = 5s$  
4.15 Blood pressure of AA for healthy and with atheroma size at time, $t = 6s$  
4.16 Blood pressure of AA for healthy and with atheroma size at time, $t = 7s$  
4.17 Blood pressure of AA for healthy and with atheroma size at time, $t = 8s$  
4.18 Blood pressure of AA for healthy and with atheroma stiffness at time, $t = 4s$  
4.19 Blood pressure of AA for healthy and with atheroma stiffness at time, $t = 5s$  
4.20 Blood pressure of AA for healthy and with atheroma stiffness at time, $t = 6s$
4.21 Blood pressure of AA for healthy and with atheroma stiffness at time, $t = 7s$

4.22 Blood pressure of AA for healthy and with atheroma stiffness at time, $t = 8s$
# LIST OF SYMBOLS AND ABBREVIATIONS

<table>
<thead>
<tr>
<th>SYMBOL</th>
<th>DESCRIPTION</th>
</tr>
</thead>
<tbody>
<tr>
<td>AA</td>
<td>Abdominal Aorta</td>
</tr>
<tr>
<td>ρ</td>
<td>Density</td>
</tr>
<tr>
<td>A</td>
<td>Cross sectional area</td>
</tr>
<tr>
<td>E</td>
<td>Young’s Modulus of Elasticity</td>
</tr>
<tr>
<td>r</td>
<td>Radius</td>
</tr>
<tr>
<td>V</td>
<td>Volume</td>
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<td>V</td>
<td>Velocity</td>
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<td>t</td>
<td>Time</td>
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<td>m</td>
<td>Meter</td>
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<tr>
<td>kg</td>
<td>Kilogram</td>
</tr>
<tr>
<td>s</td>
<td>Second (time)</td>
</tr>
<tr>
<td>π</td>
<td>Pai, the constant</td>
</tr>
<tr>
<td>a</td>
<td>Acceleration</td>
</tr>
<tr>
<td>Pa</td>
<td>Pascal (pressure)</td>
</tr>
<tr>
<td>mmHg</td>
<td>Milimeter height of Mercury (pressure)</td>
</tr>
<tr>
<td>Cp</td>
<td>Specific heat capacity</td>
</tr>
<tr>
<td>μ</td>
<td>Viscosity</td>
</tr>
<tr>
<td>T</td>
<td>Temperature</td>
</tr>
<tr>
<td>K</td>
<td>Thermal conductivity</td>
</tr>
<tr>
<td>Q</td>
<td>Flow rate</td>
</tr>
</tbody>
</table>
CHAPTER I

INTRODUCTION

This chapter will describe about the introduction of the project. In the section of the Background of Study, the problem in this study is related to the nature and real problem of the human life, which suffering from the atherosclerosis disease. Besides, the more specific problem and method of this study is also been mention for better understanding of this area. There are several bio-medical terms is explained in the section of the Definition of Terminology. Then Significant of Study, Objectives, Scopes of Study and Expected result will be show for the detail of this study.

1.1 Background of Study

In human body, blood flows from heart to arteries, which branch and narrow into the arterioles and capillaries. After the tissue has been perfused, capillaries join and widen to become venules and then widen more to become veins, which return blood to the heart.

Arteries are blood vessels that carry blood away from heart to entire of the human body. This blood is normally oxygenated, exceptions made for the pulmonary and
umbilical arteries. The main function of the arteries blood vessels are deliver the oxygen and nutrients to all cells, as well as transportation for the removal of carbon dioxides and waste products, maintenance of optimum pH, and mobility of the elements, proteins and cells of the immune system. The arterial system is the higher-pressure portion of the circulatory system. Thus, we can see that arteries are more muscular than veins. Arterial vessels have the peak pressure during heart contraction, which called the systolic pressure, and the minimum, or diastolic pressure between contractions, when the heart expands and refills. This pressure variation within the artery produces the pulse which is observable in any artery, and reflects heart activity. Arteries also aid the heart in pumping blood [1].

All arteries are comprised of three distinct layers, intima, media and adventitia, but the proportion and structure of each varies with the size and function of the particular artery. The figure below shows the figure of arterial layers of a human body:

![Arterial layers of a human body](image)

**Figure 1.1** The arterial layers of a human body [2]

An atheroma is an accumulation and swelling an artery walls that made up of (mostly) macrophage cells, or debris, that contain lipids (cholesterol and fatty acids), calcium and a variable amount of fibrous connective tissue. Then, it results the artery wall thickens and hardens. This is called as atherosclerosis or “hardening of the arteries“. Thus, will resulting in a heart attack and ensuing debility [3]. Besides,
Atherosclerosis is also an underlying cause of stroke, which afflicts about 795,000 Americans every year [4]. From the statistics of international death rates, United States shows 169.4 per 100,000 of population death in year 2005 because of coronary heart disease (CHD) [5]. The atherosclerosis disease statistics increases as the group of human age increases [6].

In clinically, doctors are unable to start any medical surgery of the artery which affected by atherosclerosis disease in human body without any visualisation of the artery problems. Thus in this project, the research of blood flow condition in the normal artery and effect of atheroma in the artery system is done by using the commercial finite element analysis software, ANSYS. The ANSYS analysis of artery problem may help doctor to understand the condition of the artery vessel in the normal or with arthrosclerosis disease. This research main objective is study blood flow interaction with the artery on the variation of the atheroma size and stiffness. The interaction is expected to alter the variation of blood pressure, artery wall stress and displacement of artery wall compared to healthy artery. The reason for study of atheroma size and stiffness is because the bigger size and higher stiffness of atheroma may due to the high blood velocity and pressure in the artery vessel which may increases the wall stresses. Besides, the high stiffness of atheroma may due to the rupture of the artery vessel easily and causing the death. This analysis can be done by using ANSYS multiphysics, fluid-structure interaction environment. The geometry and material properties of the blood and artery wall will be set as close as possible to the real artery which can be obtained from the literature studies. After the simulation study of the artery problem, we may understand the artery wall condition, blood pressure, upstream and down stream condition due to the parameter of atheroma which significant for this study. The visualization of the interaction between blood flow and artery wall in the vicinity of atheroma will help medical researchers and doctor understand atherosclerosis better.

1.2 Definition of Terminology
1.2.1 Atheroma

In pathology, an atheroma is an accumulation and swelling in artery walls that is made up of (mostly) macrophage cells, or debris, that contain lipids (cholesterol and fatty acids), calcium and a variable amount of fibrous connective tissue. Atheroma occurs in atherosclerosis, which is one of the three subtypes of arteriosclerosis; atherosclerosis, Monckeberg's arteriosclerosis and arteriolosclerosis. In the context of heart or artery matters, atheromata are commonly referred to as atheromatous plaques. It is an unhealthy condition, but is found in most humans [30].

![Atheroma Diagram](image)

Figure 1.2: Atheroma [31]

1.2.2 Atherosclerosis

Atherosclerosis (also known as arteriosclerotic vascular disease or ASVD) is a condition in which an artery wall thickens as the result of a build-up of fatty materials such as cholesterol. It is a syndrome affecting arterial blood vessels, a chronic inflammatory response in the walls of arteries, in large part due to the accumulation of macrophage white blood cells and promoted by low-density lipoproteins (plasma proteins that carry cholesterol and triglycerides) without adequate removal of fats and cholesterol from the macrophages by functional high density lipoproteins (HDL), (see
apoA-1 Milano). It is commonly referred to as a hardening or furring of the arteries. It is caused by the formation of multiple plaques within the arteries [32].

### 1.2.3 Finite Element Method

The finite element method (FEM) (its practical application often known as finite element analysis) is a numerical technique for finding approximate solutions of partial differential equations (PDE) as well as of integral equations. The solution approach is based either on eliminating the differential equation completely (steady state problems), or rendering the PDE into an approximating system of ordinary differential equations, which are then numerically integrated using standard techniques such as Euler's method, Runge-Kutta, and others [33].

### 1.2.4 ANSYS

ANSYS, Inc. is an engineering simulation software (computer-aided engineering, or CAE) developer that is headquartered in Canonsburg, Pennsylvania, United States. The company was founded in 1970 by Dr. John A. Swanson and originally named Swanson Analysis Systems, Inc [34].

The core product of Ansys Inc is its ANSYS Multiphysics/Structure mechanics module. This code is based on the Finite element method and is capable of performing static (stress) analysis, thermal analysis, modal analysis, frequency response analysis, transient simulation and also coupled field analysis. The Ansys multiphysics can couple various physical domains such as structural, thermal and electromagnetics. Many researchers and engineers prefer this module because of its parametric language known as Ansys Parametric Design Language (APDL). The APDL allows users to execute all the commands required to pre process, solve and post process the problem,
from a separate text file known as macro.

### 1.2.5 Optical Coherence Tomography (OCT)

Optical coherence tomography (OCT) is an optical signal acquisition and processing method. It captures micrometer-resolution, three-dimensional images from within optical scattering media (e.g., biological tissue). Optical coherence tomography is an interferometric technique, typically employing near-infrared light. The use of relatively long wavelength light allows it to penetrate into the scattering medium. Depending on the properties of the light source (superluminescent diodes and ultrashort pulsed lasers have been employed), Optical coherence tomography has achieved sub-micrometer resolution (with very wide-spectrum sources emitting over a ~100 nm wavelength range) [35].

### 1.3 Significant of Study

Due to this project, the blood flow interaction with the variation of the atheroma size and stiffness is studied. It helps students to understand the condition of atherosclerosis blood flow, and also the structure condition of the artery blood vessel. The project’s goal is to enhance the knowledge of mechanical field in the human blood artery system. The visualization of the interaction between blood flow and artery wall in the vicinity of atheroma will help medical researchers and doctor understand atherosclerosis better.

### 1.4 Objectives
1. To study the blood flow and its interaction with healthy artery.
2. To study the blood flow and artery wall interaction with the effect of 
   atherosclerosis disease, which affecting by the:
   a) size of the atheroma.
   b) stiffness of the atheroma.
3. To study the stress variation due to the atherosclerotic artery.
4. To provide visualization of the blood flow and its interaction with the artery 
   wall in the vicinity of atheroma.

1.5 Scopes of Study

The scopes of the study are :
1. Analyzing the blood pressure variation in the artery around atheroma.
2. The analysis is considered be linear analysis.
3. The normal blood viscosity is set to be normal human blood viscosity for the 
   analysis.
4. The blood is assumed as incompressible in the analysis.
5. Downstream boundary condition of the artery blood vessel assumed to be 
   neglected.

1.6 Expected Results
At the end of this study, the results may shows that:

1. The blood flow variation and interaction in the healthy artery is normal.

2. For the study of blood flow and artery wall interaction with the effect of *atherosclerosis* disease, the blood flow rate decreases due to the larger size and more stiffer of the *atheroma*.

3. The artery will get rupture easier due to the bigger size and higher stiffness of atheroma.

4. The stress of artery wall becomes higher due to the existence of the atheroma.

5. Ansys software able to simulate and visualize the blood flow and its interaction with the artery wall in the vicinity of *atheroma*.

1.7 Flow Chart of Project Planning
Literature review

Artery and Blood Parameters

Modelling and Simulation

Healthy blood flow in artery
Artery blood flow with effect of atherosclerosis disease

Size of atheroma
stiffness of atheroma

Analysis of data

Thesis preparation

**Figure 1.3:** Flow Chart of Project Planning
CHAPTER II

LITERATURE REVIEW

2.1 Introduction

In this chapter of literature review, most of the recent related researches had been reviewed as the reference of this study. The references are reviewed based on the artery type of blood vessel, atherosclerosis disease problem and the finite element analysis of the blood flow using numerical software. The parameters and boundary condition of blood flows is also important for the review in this study, because these information can be obtain for the analysis uses. Finally, the ideas and method of the previous researches are referred for the better understanding of this study.

2.2 Previous Research

From the research of references [7], it is to develop a 3D numerical simulation system for the clinical study for the cerebral aneurysm. The numerical system is described through the case of flow in the internal carotid artery, which called the carotid siphon. The paper also presented the mathematical model and numerical discretization using the finite element method. Since the blood flow in the artery is
considered to be incompressible, the mathematical model governing equation to the blood flow is the continuity and Navier-Stokes equations. Following is the governing equations:

\[
\frac{\partial u_i}{\partial x_i} = 0, \\
\rho \left( \frac{\partial u_i}{\partial t} + u_j \frac{\partial u_i}{\partial x_j} \right) = -\frac{\partial P}{\partial x_j} \delta_{ij} + \mu \frac{\partial^2 u_i}{\partial x_j \partial x_j} + f_i
\]  \hspace{1cm} (2.1)

The diameter of the internal carotid artery is about 0.5 cm on average, thus the blood flow is considered to be Newtonian. These assumptions and governing equation in the paper thus may relate to this study. The CT (computer tomography) angiography is used to construct finite element mesh for the 3D solid model.

From the analysis obtain in this paper, the physical properties of blood is set to be \(\rho = 1.00 \text{ g/cm}^3\) and \(\mu = 0.02 \text{ poise}\). The resulting Reynolds varies from \(Re = 300\) to 700. The blood flow in this paper is assumed as laminar since the maximum Reynolds number is 700. These physical properties and assumptions can be use as the support for the information for this study in the next chapter (Methodology). As from the results, the magnitude of surface traction becomes larger at the upstream region where the carotid siphon starts to bend as shows in the figure below. The aneurysm tends to be created at the location of bifurcation of the posterior communicating artery. The aneurysm is reported can be created at the location (1).

\[\text{Figure 2.1: distribution of magnitude of surface traction average over one cardiac cycle}\]
As the velocity of inlet becomes smaller, the overall magnitude of the wall shear stress starts to decrease as shown in figure below:

![Figure 2.2: Evolution of surface traction over one cardiac cycle](image)

Besides, the results from this paper also show that the evolution of the wall shear stress corresponds to the change in Reynolds number in the inlet. It is also found that the areas of the high wall shear stress show large variation of the wall shear stress with respects to time. The overall magnitudes of secondary velocity and axial velocity decrease associate with a decrease of velocity in the inlet. Thus, it was found that secondary flow becomes quite large downstream of the carotid siphon due to the curvature. The results shows the major agreement in the expected results in this study that mentioned in Chapter 1, which the wall shear stress will increases as the increase of blood velocity that varies by the size and stiffness of atheroma.

From the study of E.V. Irene et al [8], the paper is describe about the development and implement a method to prescribe outflow boundary conditions intended for 3D finite element simulations of blood flow based on the Dirichlet-to-Neumann and variational multiscale methods in the major arteries. Given the computational expense of three-dimensional numerical methods and the resolution limits imposed by current imaging technologies, the three-dimensional domain is constrained to the major
arteries, and models the downstream domains with simpler models as shown in figure below. The outlet boundary conditions are implemented implicitly resulting in good stability and convergence properties at physiologic pressures. The paper had also described the coupled multidomain method in three-dimensions and its specialization to resistance and impedance boundary conditions that might be related to this study. Then demonstrate this new method on a straight, cylindrical blood vessel, a bifurcation model with a stenosis on one side, and a subject-specific model of the human abdominal aorta.

**Figure 2.3:** Illustration of the “Coupled Multidomain Method”, where information from the downstream (analytical) domain is incorporated in the upstream (computational) domain through a boundary map. This approach accommodates a variety of downstream models ranging from pure resistive to complex electrical analogs and impedance models.

From the analysis, the nominal radius of 0.3 cm and the vessel length of 12.6 cm (length to diameter ratio of 20) were chosen to correspond approximately to that of the human common carotid artery. The solution was computed on a 45,849 element and 9878 node mesh with a time step of 0.002 s, for a total of five cardiac cycles. The pressure pulse in the carotid artery has been reported to be approximately 70–80/110–120 mmHg for a healthy person. The figure below depicts the results
obtained with constant pressure, resistance and impedance outflow boundary conditions. As this results obtained may be use as for the validation of the present study.

Figure 2.4: Comparison of flow and pressure wave forms in a carotid artery with a periodic inlet flow and constant pressure, resistance and impedance outlet boundary conditions. For the pressure boundary condition, inlet pressure varies little from the prescribed constant outlet pressure and the peak pressure precedes the peak flow. The resistance boundary condition gives rise to an unrealistically large pressure amplitude. In addition, pressure and flow are in phase at the outlet. For the impedance boundary condition, the range of pressure is from approximately 85–115 mmHg and the pressure lags flow along the length of the vessel. 1 mmHg = 133.3 Pa, 1 cc/s = 10^-6 m³/s.

To further illustrate the critical influence of boundary conditions, the paper performed numerical simulations in an idealized model of an abdominal aorta bifurcation to the iliac arteries with a 75% area reduction stenosis on one side, using realistic anatomic dimensions and inflow wave form for an abdominal aortic bifurcation. Figure below (a) shows the geometric of the idealized model, (b) depicts the dramatic differences in flow split depending on outlet boundary condition. Besides, the paper provides many ideas of methodologies that related to this study such as the related governing equations, coupling method, boundary conditions, geometrical modeling and finite element method,
**Figure 2.5**: (a) Geometric model and (b) flow distribution between normal iliac artery and an iliac artery with a stenosis that reduces the cross-sectional area by 75% for constant pressure and impedance boundary conditions. For a constant outlet pressure, the flow split is dictated solely by the resistance to flow due to the geometry of the computational domain (approximately 70% to the normal side, 30% to the stenosed side). For an impedance boundary condition with a physiologic level of resistance, the flow split (50%/50%) is principally determined by the downstream demands (represented by equal outlet impedances). These results are consistent with clinical observations for resting flow conditions and iliac artery stenoses with less than an 85% area reduction.

In the paper of C.A. Taylor et al [9], the major topic is the development of a software system to integrate the model construction, mesh generation, problem initialization, 3D pulsatile flow solution, scientific visualization, and quantitative data extraction aspects of computational vascular modeling. The second major topic is the application of the developed software system to the analysis of blood flow in complex system. The computational method employed was validated with an analytic solution for pulsatile flow in a straight artery, and with experimental data for steady and pulsatile flow in an end-to-side vascular graft anastomosis and pulsatile flow in a carotid artery. Next, the computational method developed was used to quantitatively characterize the hemodynamic conditions under simulated rest and exercise pulsatile flow conditions in an idealized model of an abdominal aorta. Finally, the computational method developed was applied to compute the flow in patient-specific models created from pre-operative and post-operative medical imaging data of an aneurysmal abdominal aorta. The example illustrates the potential application of
computational blood flow simulations to surgical planning.

Hemodynamic conditions, including velocity, shear, and pressure, play an important role in the modulation of vascular adaptation and the localization of vascular disease. As the models of vascular anatomy and physiology increase in complexity, the importance of having a comprehensive framework for computational vascular research will be of even greater import. Tasks that required weeks, such as the construction and modification of the model of the abdominal aorta, can now be performed in minutes. The geometric representation employed enables the use of an automatic mesh generator resulting in a high quality discretization with literally only seconds of user interaction. The finite element solver employed results in time accurate solutions to hundreds of thousands of equations and permits the resolution of flow features heretofore unobserved and unmeasurable. The scientific visualization techniques utilized allow the examination of vast amounts of data from multiple perspectives. The validation studies show the good agreement between the computational solution and the analytical Womersley solution was obtained for pulsatile flow in a long, straight artery. Very good agreement between the computational solution and experimental LDA data was obtained for steady and pulsatile flow in an end to side graft-artery anastomosis. A computational solution of this problem had not been previously reported. There have been no reported computational solutions and little quantitative experimental data for this important vascular blood flow in a model of an abdominal aorta. The information obtained regarding the effects of graded exercise on wall shear stress and particle residence times yields new insights into the effects of exercise and supports the growing body of evidence that moderate levels of exercise are beneficial in reducing vascular disease. These studies provide impetus to examine exercise hemodynamics in vivo using magnetic resonance imaging techniques, and to assess the validity of the assumptions made regarding the distribution of blood flow.
The studies about the pattern analysis of temporal changes in the carotid artery diameter under normal and pathological condition had been made by A. Blizhevsky et al [10]. The objective of the study was to analyse the age-related and cyclic temporal changes in the CCA (common carotid artery) diameters in order to characterize their pattern under normal and pathological conditions and investigate whether these patterns can indicate the existence of stenosis in the internal carotid artery.

Two groups of subjects were studied: (i) 11 healthy normotensive subjects and (ii) eight hypertensive subjects, with various degrees of stenosis in the left and/or right internal carotid artery. All subjects were studied in the supine position, and the arterial systolic and diastolic pressures were measured in the brachial artery using a blood pressure cuff. Cross-sectional images of the left and right CCA were acquired via an ultrasonic system fitted with a 7-MHZ linear array transducer. The system provided 25 frames per second. The images were recorded on a videotape, and then transferred to a PC using a special software package developed in our department” and a frame grabber (‘PCVISIONplus’).

The results of analysis show that the diastolic diameter increases linearly with age. There was no significant dependency of the ED (end diastolic) diameter on age. For the normal subjects, there is a linear dependency of the relative diameter decreases with age. There is no significant difference was found between the dependency of the RDC on age for the groups of hypertensive and normal subjects. For group of healthy subjects, the elastic index of the CCA increases with age. However, the mean value of the elastic index for the abnormal subjects is more than twice that of healthy subjects. For the normal subject, the diameter of the right CCA increases faster than the diameter of the left CCA during the systole, and the systolic duration of the right CCA is shorter than left CCA. The group healthy subjects have the positive delay (LRPDD) between the peak diameters. In 5 pathological subjects (out of 8) this delay was negative and other 3 had slight positive delay. In the DDloop analysis, 10 out of 11 healthy subjects had a positive value of are and only one of them had a small negative
area of DDloop. 3 subjects with a stenosis less than 50% from the abnormal group had a positive area, but these values very close to zero. However, there is disagreement views on the paper that studies by A. Blizhevsky et al. The results still shows some ambiguities of the analysis. 5 out of 8 pathological subjects doesn’t meant for the absolute result.

In the paper of A. Redaelli et al [11], a new method for the assignment of pulsatile velocity profile as input boundary conditions in finite element models of arteries is presented. The analytical solution of the Navier-Stokes equation with pulsatile boundary condition is implemented. Then, the comparison is done between the analytical method and the traditional numerical approach which requires the finite element simulation of the upstream region. The two main objective of this study is to easily calculate the correct inlet boundary condition and to reduce the computational time involved.

Figure 2.6: Flow diagram of the algorithm for the calculation of Womersley developed profiles.
For the procedure to simulate pulsatile, developed velocity profiles, add a straight inlet tube to a model which simulates pulsatile flow in a vascular district, the flow develops and tends asymptotically to the analytical solution after an appropriate number of diameters. The velocity boundary conditions at the inlet section were assumed to be either flat or parabolic, typical of undeveloped steady flows, respectively. Further assumptions for the models were: homogeneous and Newtonian
fluid \((p = 1060 \text{ kg/m}^3, \mu = 3 \times 10^{-3} \text{ kglm})\), impermeable walls, no slip boundary conditions at the wall, uniform pressure at the outlet section, neglected gravitational effects.

The results presented show a good agreement between W0MER.f and FIDAP velocity calculations where an appropriate length for the vessel in FIDAP simulations is adopted. It should, however, be noted that after 40 diameters, the FIDAP simulations give slightly different values than those computed by the W0MER.f. The first reason for this may be that 40 diameters are not sufficient to reach stable solutions. Second reason, is that the instantaneous flow rates for both software are not the same. The differences in flow rate may be due to finite element codes and Fast Fourier Transform.

Besides, the paper also gave much information of the model properties related to this study, such as the density and the viscosity for a Newtonian fluid. The paper also gave an idea of the pulsatile flow that related on this study.

The numerical study of nonlinear pulsatile flow in S-shaped curved arteries is done in the paper [12] by A.K. Qiao et al. Two models of S-shaped curved arteries with different vessel diameters are considered in this paper. The temporal and spatial distributions of hemodynamic variables during the cardiac cycle such as velocity field, secondary flow, pressure, and wall shear stresses in the arteries were analyzed. The assumptions that made are:

a) the arteries are rigid, circular and equi-sectional Sshaped vessels;
b) the blood is incompressible Newtonian fluid;
c) the motion of blood in vessels is three-dimensional and unsteady laminar flow.
Geometrical model of S-shaped vessels is shown in figure below.

![Geometrical model of vessels](image)

**Figure 2.8:** Geometrical model of vessels.

The results of the numerical simulations showed that the secondary flow in the larger S-shaped curved artery is more complex than in the smaller arteries. Besides, the secondary flow is stronger in the downstream of the curved vessel than in the upstream. In the systolic phase, the pressure at the outer bend is larger than in the inner bend by the cause of the centrifugal force. While the pressure at the outer bend is very similar to inner bend during the diastolic phase. The pressure between outer and inner bend of LVM is larger than SVM. However, the wall shear stress (WSS) increases and decreases with the magnitude of the entrance velocity. During the systolic phase, the WSS levels are very high because of the large shear rates near the vessel wall. During the late part of the more quiescent diastolic phase, the WSS levels are low and exhibit little variation. The results shows again the agreement that the hypothesis of the expected result had been done in the Chapter 1.
In the paper [13], the first use of new imaging technique, Optical Coherence Tomography (OCT) is demonstrated as the basis for finite element analysis. The geometry of excised human coronary vessel is determined by using OCT imaging and the conventional histology method. Then, the finite element models are constructed and compared stress and strain distributions. Then the OCT’s limited depth penetration and subsequent outer contour ambiguity is investigated.

**FIGURE 2.9:** Finite element modeling process for a lipid rich coronary plaque cross-section. The OCT image (A) and histology image (B) are segmented (C,D) into lipid rich, fibrous plaque, and arterial wall regions. Each segmented image is used to create a finite element mesh (OCT mesh, E, with closeup in F). Application of an internal pressure load results in elevated stress at systolic pressure (G,H) and cyclic strain (I,J) distributions. Results for the OCT-based model are shown in the left column; results for the histology based model are shown in the right column.

There are two primary benefits of OCT relative to histology: 1) imaging is
performed without excessive tissue handling, providing a more realistic geometry than histology and avoiding structural artifacts common to histologic processing, and 2) OCT imaging can be performed in vivo, making it possible to study disease progression and the effect of therapeutic treatments in animal models and living patients.

The results for the paper noted that even though the OCT and histology images exhibit a close correspondence, they do not have identical geometry. In the OCT image, the lumen boundary is smooth with no jagged edge, but in the histology geometry there are many sharp edges in the lumen. Both OCT based modeling and accepted histology based modeling provide the about the same stress and strain distribution, but can yield disparate stress and strain magnitudes. The results of a segmentation sensitivity analysis show that the stress and strain predictions are not significantly affected by segmentation ambiguities associated with OCT signal attenuation. Since OCT can be performed in vivo and at multiple time points, the results suggest that OCT-based finite element analysis may be a powerful tool for investigating coronary atherosclerosis, detecting vulnerable plaque, and monitoring response to therapy in living subjects. However, in this study the geometric of the artery vessel are simplified geometry. The OCT and histology images might not be use in this research. By the way, the imaging results from the paper can be use as information in this study. The figure of illustration of the atherosclerosis pattern can be use in this study.

The paper [14] is study about the finite element modeling of atherosclerotic plaque. The 2D mathematical models of the coronary walls have been developed based on actual plaque morphology from a multi-center study of coronary disease (Pathological Determinants of Atherosclerosis in Youths, PDAY). The lesion geometry is represented as probability maps that were created from histological analysis of 625 coronary plaques (left anterior descending coronary arteries) excised
from trauma victims, which shows in the figure below.

![Lipid positive probability map for the 30 to 34 age group](image)

**Figure 2.10:** Lipid positive probability map for the 30 to 34 age group

The models were created and analyzed using the Cosmos (Structural Research, Inc.) finite element program. The models were isotropic, linear and incompressible. Each model consists of approximately 900, plane stress, 8-node, elements. Separate material properties defined the three areas of the artery and lesion and are shown in Table below which can be use as the properties information of the artery and atheroma in the Chapter 3. The models were internally pressure loaded ($P_{iv} = 120$mmHg).

**Table 2.1:** Material properties for the element in blood vessel

<table>
<thead>
<tr>
<th>Material</th>
<th>Young's Moduli $\mu$ (Pa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>artery</td>
<td>$17.4 \times 10^5$</td>
</tr>
<tr>
<td>lipid</td>
<td>$17.4 \times 10^5$</td>
</tr>
<tr>
<td>cellular cap</td>
<td>$5.1 \times 10^5$</td>
</tr>
<tr>
<td>fibrous cap</td>
<td>$9.0 \times 10^5$</td>
</tr>
<tr>
<td>calcified cap</td>
<td>$2.2 \times 10^6$</td>
</tr>
</tbody>
</table>

Under physiological loading ($P_{iv} = 120$mmHg) the resulting stress distributions give an indication of the physical changes that occur as the plaques develop. The results are shown in the table below:

**Table 2.2:** Content of element maximum stress of artery according to age
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