NUMERICAL MODELING OF FLUID SOLID GROWTH IN ABDOMINAL AORTIC ANEURYSM

Nino Horvat¹, Byron Zambrano², Seungik Baek², Igor Karšaj¹

¹ Faculty of Mechanical Engineering and Naval Architecture, University of Zagreb, Ivana Lučića 5, 10000 Zagreb, Croatia, {nino.horvat, igor.karsaj}@fsb.hr
² Department of Mechanical Engineering, Michigan State University, 2555 Engineering Building, East Lansing, MI 48824, USA, zambran2@msu.edu, sbaek@egr.msu.edu

SUMMARY

A framework for coupling finite element growth and remodeling (G&R) analysis of abdominal aortic aneurysm (AAA) with hemodynamics analysis is presented. Majority of AAAs harbor intraluminal thrombi (ILTs) which have biochemomechanical influence on aortic wall, and therefore active role in G&R of AAA. Based on time averaged wall shear stress from hemodynamics analysis, layers of ILT are added on luminal surface of AAA model. Analyses are loosely coupled and run in a time loop where G&R analysis has longer time scale (i.e. months) while hemodynamics analysis has shorter time scale (i.e. seconds).

Key words: abdominal aortic aneurysm, finite element method, hemodynamics, growth

1 INTRODUCTION

Abdominal aortic aneurysm (AAA) is a permanent dilatation of abdominal aorta that results mainly due to altered deposition and degradation of extracellular matrix. The majority of AAAs contain intraluminal thrombi (ILTs) which are 3D fibrin structure comprised of blood cells, blood proteins, platelets and cellular debris adhered to the expended aortic wall. Increasing evidence shows that ILT has biochemomechanically active role in growth and remodeling (G&R) of AAA [1]. In Virag et al. [2], a computational model of multi-layered ILT showed its diverse biochemomechanical effects on the AAA enlargement and emphasized the importance of modeling ILT in G&R simulations. However, an idealized, axisymmetric, cylindrical geometry was used, and every time step when the aneurysm enlarged, an additional ILT was deposited under assumption of constant luminal radius. Generally, ILT is not accumulated immediately as the AAA expands. One of the parameters that can be used to predict ILT accumulation is time averaged wall shear stress (TAWSS) (e.g., [3], [4]). Using finite elements we can model and study the ILT accumulation, and thus its influence on a 3D axisymmetric and asymmetric aneurysm geometry. Our aim is to develop a framework which will combine finite element G&R analysis of evolving AAA with included ILT and hemodynamics analysis.

2 METHODOLOGY

Growth and remodeling analysis of evolving AAA and corresponding hemodynamics analysis are run in an iterative time loop. In the first time step of G&R analysis healthy aorta is approximated by a straight cylinder. Aneurysm is assumed to develop due to local elastin loss. Therefore, elastin is degraded by a Gaussian spatio-temporal function (similar to [5]). For growth of axisymmetric (fusiform) aneurysm elastin loss depends on G&R time and axial location, while for asymmetric it also changes in circumferential direction. After a certain change in aortic geometry (i.e., sufficient expansion) STL geometry of aortic luminal surface is exported in order to perform a hemodynamics analysis. A computational fluid dynamics (CFD) is used to determine whether an ILT will be deposited, as well as to predict location and amount of its accumulation. Thrombus accumulation is predicted based on TAWSS on luminal surface. Since our primary focus, at this
stage, is development of a working algorithm for coupling hemodynamics and G&R simulation, we use a simple assumption that ILT is accumulated when TAWSS is less than 0.4 Pa [6]. Based on accumulation prediction, a new layer of finite elements representing ILT is locally added. The elements are deposited on deformed geometry and, at time of deposition, they are not pre-stretched and are in stress-free state. Symmetry boundary conditions are enforced on them when necessary, whereas the pressure is transferred from the old luminal surface to the new elements. After inclusion of the new ILT elements G&R analysis continues until next CFD simulation is needed (e.g., due to sufficient growth of AAA).

![Flowchart](attachment:flowchart.png)

**Fig. 1.** Time loop and information transfer between the hemodynamics and G&R simulations.

### 2.1 G&R Simulation

For G&R modeling of evolving AAA, model described in Karšaj et al. [7] is used. We implemented model in finite element analysis program FEAP through user defined subroutines. To enforce incompressibility we used Augmented Lagrange method. ILT development and its effects on the aortic wall are described using model from Virag et al. [2].

### 2.2 Hemodynamics Simulation

In order to avoid numerical instabilities and to minimize influence of boundary conditions in CFD analysis, an extension on the inlet side and a bifurcation with iliac branches on the outlet side are added to the STL geometry exported from G&R simulation (Fig. 2). Idealistic time-dependent volume flow rate and pressure waves, based on data presented in Olufsen et al. [8], are prescribed at the inlet and the outlets, respectively. After defining computational domain, model is meshed and hemodynamics simulations is performed with an assumption of laminar flow and rigid walls. Blood is modeled as incompressible ($\rho = 1060 \text{ kg/m}^3$) non-Newtonian fluid described by Carreau–Yasuda model [9]. In each simulation 5 cardiac cycles with 1000 time steps per cycle are analyzed. To minimize the influence of the initial conditions, first four cardiac cycles are discarded and results from the last cycle are used to obtain TAWSS. For axisymmetric FEM model of fusiform aneurysm, TAWSS is further circumferentially averaged.
Fig. 2. STL geometry of AAA luminal surface obtained from G&R analysis (left) and STL geometry used for hemodynamics analysis which is extended on the inlet side and with a bifurcation and iliac branches on the outlet side.

3 RESULTS AND CONCLUSIONS

Using implemented G&R model in finite element code and by Gaussian spatio-temporal elastin degradation in both axial and circumferential directions we are able to simulate development and expansion of axisymmetric (fusiform) and asymmetric aneurysms. An example of two aneurysm types is shown on Fig. 3.

Fig. 3. Geometry of developed axisymmetric fusiform (left) and asymmetric (right) aneurysm that resulted from finite element G&R simulation. Initial geometry was straight gray cylinder.

Although geometry of simulated axisymmetric aneurysm (Fig. 2 left) is symmetric with respect to plane perpendicular to centerline, distribution of circumferentially averaged TAWSS is not symmetric (Fig. 4). This can be partially explained by effect of iliac bifurcation on blood flow. Region with the lowest TAWSS corresponds to a region of maximum AAA diameter, and, in shown case, it fulfills previously mentioned criterion for thrombus accumulation. Distribution of time averaged blood pressure can also be extracted from CFD analysis of AAA, however at this stage we are assuming constant blood pressure.

We have not fully implemented ILT model in finite element code yet. Also, we are working on a method for depositing thrombus elements in finite element model of AAA. When completed, described fluid solid growth framework will enable us to analyze development and growth of AAA as well as ILT accumulation and its biochemomechanical influence on aortic wall in a 3D environment.
Fig. 4. Example of circumferentially and time averaged wall shear stress (TAWSS) obtained from CFD hemodynamics analysis of axisymmetric aneurysm shown in Fig. 2. The stress is plotted only for aortic segment used in FEM analysis. Bottom side of y axis (axial length) corresponds to aorta part close to the iliac bifurcation.

ACKNOWLEDGMENT

This work was supported by grant from the Croatian Science Foundation (project IP-2014-09-7382 I. Karšaj).

REFERENCES


