Aortoiliac Aneurysm: Examination of Biomechanical Characteristics for an Individual Patient

Smiljana Djorovic*, **, Igor Koncar***, Lazar Davidovic***, Nenad Filipovic*, **
* Faculty of Engineering, University of Kragujevac, Serbia
** BioIRC Research and Development Center for Bioengineering, Kragujevac, Serbia
*** Clinic for Vascular and Endovascular Surgery, Serbian Clinical Center, Belgrade, Serbia
e-mails: smiljana@kg.ac.rs
dr.koncar@gmail.com
davidovic.lazar@gmail.com
fica@kg.ac.rs

***Abstract*—The aim of this work was to computationally examine the biomechanical characteristics of the patient-specific AortoIliac Aneurysm (AIA) employing the fluid-solid interaction (FSI) and finite element method (FEM). Beside clinical assessment, computational modeling and simulations have great importance for improving aneurysmal examination and patient monitoring. In that order, the patient-specific 3D model of AIA was created, based on the computed tomography (CT) data. The blood flow simulation and interaction with the aortic wall, considering existence of the intraluminal thrombus (ILT), gave a more specific and more complete analysis of aneurysmal biomechanical characteristics. Assuming the blood as incompressible, viscous and laminar fluid, with an average properties and parabolic flow, the computational simulation of cardiac cycle was performed. The Von Mises stress, as the main parameter for evaluation of aortic wall degradation and rupture risk, was analyzed at the peak systolic moment. Presence of high stress at the aortic bifurcation and aneurysmal neck indicated the need for operative treatment. Furthermore, the obtained blood flow characteristics showed presence of stagnant blood flow at the maximal aneurysmal diameter, which affects the ILT and aneurysmal growth. In order to include the computational simulations in a daily clinical practice, which may lead to better prevention of aneurysmal rupture and evaluation of operative treatment need, a larger number of patients should be investigated in further studies.**

I. Introduction

An aneurysm, a localized dilation of the artery's wall, most frequently appears in the abdominal aorta and in the brain vasculature. Abdominal aortic aneurysms (AAAs) are formed preferentially in infrarenal area, before aortic bifurcations. About 3% (1–7%) of the population aged over 50 are affected by an AAA [1]. The AAAs can be extended towards one or both iliac arteries, and in that case they are called AortoIliac Aneurysms (AIAs, Fig. 1). Approximately 20%-40% of patients with AAAs can have unilateral or bilateral iliac artery aneurysms [2]. Understanding AIA pathogenesis is extremely important to improve aneurysmal examination and patient monitoring.

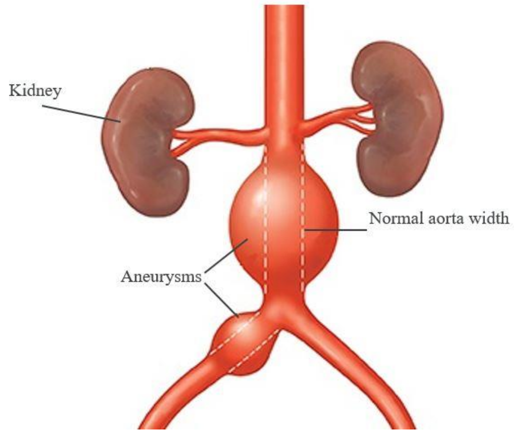

Figure 1. Graphical representation of Abdominal Aortic Aneurysm (AAA) with an iliac aneurysm that is called AortoIliac Aneurysm (AIA).

There are various endovascular treatments [3] of AIAs that prevent their rupture and fatal outcome. In order to prevent aneurysmal rupture and evaluate a need for operative treatment or determine the impact of the aortic treatment to blood flow, the novel approaches which combine clinical assessment and computational modeling and visualization are essential [4,5].

Numerous prior studies have identified many factors that potentially contribute to the development, enlargement and rupture of AAAs [6]. However, the links between these factors and the underlying mechanisms responsible for the formation, growth and stabilization or rupture of AAAs/AIAs are still insufficiently investigated. It is generally accepted that the evolution of aneurysms is driven by flow-induced progressive degradation of the vessel wall [7, 8]. Since the internal mechanical forces in the aorta are maintained by dynamic action of blood flow, the quantification of the AAAs/AIAs hemodynamics is essential for characterization of their biomechanical behavior. Furthermore, it is important to include both the dynamics of blood flow as well as the wall motion response associated with the pulsatile nature of the flow to accurately model the AIA. With that goal, the Fluid Solid Interaction (FSI) approach based on Finite Element Method (FEM) was performed in computational simulations employing simplified (parametric) model and blood-aortic wall domains [9], or patient-specific geometrical model and blood-intraluminal thrombus (ILT)-aortic wall domains [10].

In presented work we created the patient-specific model of extended abdominal (aortoiliac) aneurysm and performed computational simulation which made the connection between hemodynamic variables and wall mechanical properties. Thus, this work included blood flow simulation and interaction with the aortic wall, considering existence of intraluminal thrombus (ILT), which gives a more specific and more complete analysis of aneurysmal biomechanical characteristics and rupture evaluation.

The rest of the paper is organized as follows. The creation of a 3D patient-specific model and explanation of the FSI numerical simulation based on FEM, with employed appropriate material characteristics, are included in the part II. Part III covers discussion of obtained results for employed model. In the part IV, main conclusions are given with plans for further improvements of presented FSI analysis.

II. Material and Methods

A. Geometrical model

In order to perform the FSI analysis, the patient-specific model of AIA was created. Process of obtaining an appropriate geometrical model included several phases that will be described. First, the initial model was created in the segmentation software using computed tomography (CT) scan images. The set of DICOM images, with resolution of 512 × 512 *mm* and the slice increment of 0.625 *mm* were imported in the Mimics software [11]. A user-friendly interface enabled to extract the region of interest such as part from the infrarenal aorta to the end of the common iliac arteries. This specific patient had the AIA and, in this case, it was extended AAA towards the left common iliac artery. Considering this, the different solid and fluid domains (aortic wall, ILT formation and blood) were created. The stereolithography (STL) representation set the boundary surface for fluid and solid domains.

In the second phase, the obtained 3D description of the AIA was exported from Mimics as a surface triangulation (STL format), which mainly consists of a triangular mesh not suitable for computational analysis due to overlapping and distorted elements. For this reason, the STL had to be processed to further optimization of the surface meshes that was performed in Geomagic [12] software. The final geometry of AIA model consisted of finite element mesh is presented on Fig. 2. It was difficult to accurately capture the aortic wall thickness in patient-specific CT images due to calcification, thrombus and unclear boundary between inner and outer wall surfaces. Therefore, a uniform and average aortic wall thickness of 2 mm was prescribed in modeling of individual AIA [10].

After the AIA model creation, the surface STL model was exported from Geomagic to Femap software [13] in order to create volumetric model. Further, Femap was used in combination with our *in-house* software for the conversion of tetrahedral elements to eight-node brick elements. The dataset was imported into the next *in-house* software PAK for further computational FSI simulation [14].

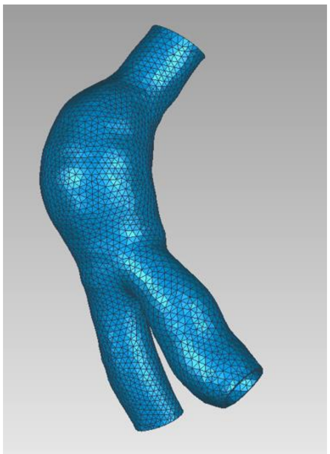

Figure 2. Geometrical model of AIA consisted of triangular finite element mesh (anterior view).

B. Material Characteristisc

After creation of geometrical model, appropriate material characteristics were prescribed in order to prepare the AIA model for computational simulation. The material characteristics (wall thickness, dynamic viscosity, Young's modulus, Poisson coefficient, density) of fluid and solid domains are given in Table 1.

Solid domains (aortic wall and ILT) were modeled as linearly-elastic isotropic material [8], under the assumption that they are affected by average blood pressure (80-120 *mm Hg*) [10]. Although it is known that Young's modulus of 1.2 MPa is present in healthy arteries, experimental results show that Young's modulus has significantly higher values in case of abdominal aneurysms (between 3 and 5 MPa) [15]. In this work, Young's modulus of 5 MPa and Poisson coefficient of 0.45 were assumed for the aortic wall [10]. Furthermore, Young's modulus of 0.1 MPa and Poisson coefficient of 0.49 were adopted for ILT [10]. The densities of $1.12 \cdot 10^{-3}$ g/mm^3 and $1.121 \cdot 10^{-3}$ g/mm^3 were adopted for aortic wall and ILT, respectively.

In order to perform analysis which reflects the realistic human blood flow we used average blood properties: a dynamic viscosity of $3.5 \cdot 10^{-3}$ $Pa \cdot s$ and a density of $1.05 \cdot 10^{-3}$ g/mm^3 [4, 5, 16]. Flow is considered to be laminar, homogeneous (Newtonian) and viscous incompressible.

TABLE I. MECHANICAL PROPERTIES FOR DIFFERENT DOMAINS

Parameters	**Blood domain**	**Aortic wall**	**ILT**
Wall thickness [*mm*]	/	2	/
Dynamic viscosity [$Pa \cdot s$]	$3.5 \cdot 10^{-3}$	/	/
Young's modulus [*MPa*]	/	5	0.1
Poisson coefficient	/	0.45	0.49
Density [g/mm^3]	$1.05 \cdot 10^{-3}$	$1.12 \cdot 10^{-3}$	$1.121 \cdot 10^{-3}$

C. Numerical Simulation

The employed boundary conditions for fluid domain consisted of:

- prescribed velocity at the inlet of infrarenal aorta,
- zero-velocity at the ILT and aortic wall,
- physiological resistance pressure at the aortic outlets.

At the inlet of infrarenal aorta, a parabolic velocity profile was used together with a pulsatile waveform based on the measured data (Fig. 3) [17].

The employed boundary conditions for solid domains were next:

- top surface of infrarenal aorta and bottom surfaces of common iliac arteries were fixed.

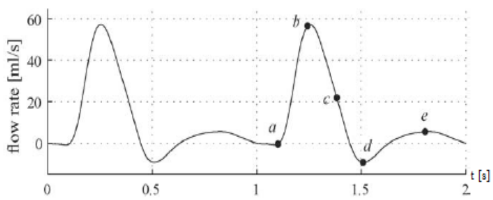

Figure 3. Prescribed blood flow rate [17]: (a) end-diastole (b) peak-systole; (c) late-systole; (d) end-systole; (e) mid-diastole.

D. Finite Element Analysis

After applied appropriate boundary conditions and material characteristics for the both models, the *in-house* software for FSI analysis [14] was used for the computational simulation, which calculated the velocity field of blood and Von Mises stress distribution. This software solves the mass and momentum conservation (Navier-Stokes) equations for the fluid domain and the equilibrium equations of elasticity for the solid domain using a finite element discretization method. Also, it implements an arbitrary Lagrangian Eulerian (ALE) approach. The fluid and solid domain meshes were independently converted, considering appropriate numeration of nodes and elements, and then imported into PAK for further analysis [13].

III. RESULTS AND DISCUSSION

The 3D finite element model of the AIA was created using the patient-specific CT scan images in combination with several different software packages. The fluid (blood flow) and solid domains (ILT and aortic wall) were created. Also, appropriate boundary conditions and equivalent material characteristics were prescribed for these domains. The employed FSI analysis provided a comprehensive insight into the examination of aneurysm-related complications, and, in chis specific case, enabled computation of stress distributions in the aortic wall and velocity field of blood. The analysis was performed using numerical methods and algorithms for FSI on computational finite element meshes. It was built on experiences of other groups having performed FSI simulations of AAA/AIA with obtained results which are in good agreement with references 8, 10, 18.

As result, the velocity field of blood within the AIA is presented on Fig. 4, while the Von Mises stress distribution in the ILT and aortic wall is presented on Fig. 5. Blood circulation, which was considered in the simulation, and distribution of mechanical stress have significant influence on aneurysmal growth and aortic wall behavior. Also, it is known that aortic wall suffers maximal loads at the peak systolic moment. Therefore, in cases of analyzed mechanical parameters, the model is presented at peak systolic moment, as the worst-case scenario, which is reached after about 0.3 s of the simulation.

The Von Mises stress, as the main parameter for evaluation of aortic wall degradation and rupture risk, was analyzed with accent on the areas of high stress in the AIA's wall. The obtained stress distribution is affected by shape of the aneurysm formation, the material characteristics of the aortic wall and ILT, as well as by interaction between the fluid and solid domains. High stress at the aortic bifurcation and aneurysmal neck indicated the need for operative treatment. Furthermore, the obtained blood flow characteristics showed presence of stagnant blood flow at the maximal aneurysmal diameter, which affects the ILT and aneurysmal growth. As a confirmation of these findings, the patient underwent an operative treatment according to clinical assessment.

It should be underlined that the analyzed model used for this study involved some strong limitations. First, the linear material model did not fully describe biological aortic tissue, considering that aortic wall has nonlinear and anisotropic properties. Third, average material properties and wall thickness were adopted from the literature which did not replicate the patient-specific characteristics. Also, we assumed that the top surface of infrarenal aorta, as well as the bottom surfaces of common iliac arteries were fixed from motion. Due that fact, the presented model did not replicate the realistic movement. Furthermore, we prescribed average blood flow rate at the inlet of infrarenal aorta. Hence, the presented results can be applied only for this study.

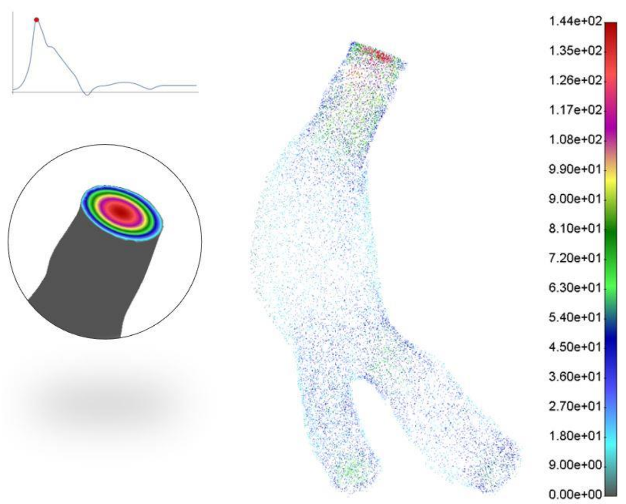

Figure 4. Velocity field at the peak systolic moment (Units: mm/s); anterior view.

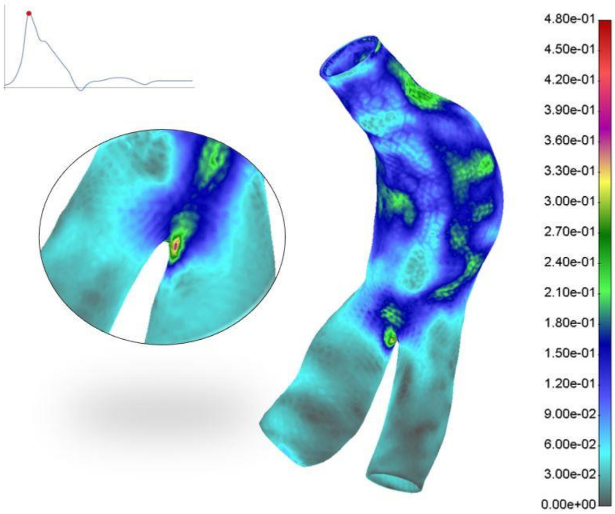

Figure 5. Von Mises stress distribution at the peak systolic moment (Units: MPa); aortic wall, posterior view.

IV. Conclusion

In the present work, the computational FSI analysis was selected to provide a better insight into the aneurismal condition due to difficulties in obtaining the presented mechanical parameters experimentally. Results of computational simulation indicated the need for operative treatment. In summary, this study proved that biomechanical parameters' determination can be useful for appropriate prevention of aneurysmal rupture and evaluation of operative treatment need, which have the great importance in the patient's health condition. The impact of hemodynamics on wall structure and mechanical properties should be further investigated in larger studies, in order to include the computational simulations in a daily clinical practice.

Acknowledgment

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References

[1] M. L. LeFevre, "Screening for abdominal aortic aneurysm: U.S. Preventive Services Task Force recommendation statement," *Ann Intern Med*, vol. 161, pp. 281–290, August 2014.

[2] R. Hobo, J. E. Sybrandy, P. L. Harris and J. Buth, "Endovascular repair of abdominal aortic aneurysms with concomitant common iliac artery aneurysm: outcome analysis of the EUROSTAR Experience," *J Endovasc Ther*, vol. 15, pp. 12–22, 2008.

[3] S. Duvnjak, "Endovascular treatment of aortoiliac aneurysms: From intentional occlusion of the internal iliac artery to branch iliac stent graft," *World J Radiol*, vol. 8, pp. 275–280, March 2016.

[4] N. Filipovic, D. Milasinovic, N. Zdravkovic, D. Böckler, H. von Tengg-Kobligk, "Impact of aortic repair based on flow field computer simulation within the thoracic aorta," *Comput. Methods Programs Biomed*. vol. 101, pp. 243-252, ISSN 0169-2607, February 2011.

[5] N. Filipovic and H. Schima, „Numerical simulation of the flow field within the aortic arch during cardiac assist", *Artifical Organs*, vol. 35, pp. 73-83, ISSN 1525-1594, April 2011.

[6] Z. Domagala, H. Stepak, P. Drapikowski, et al., "Biomechanical factors in Finite Element Analysis of abdominal aortic aneurysms," *Acta Angiologica*, vol. 22, pp. 164–171, 2016.

[7] M. F. Fillinger, M. L. Raghavan, S. P. Marra, et al., "In vivo analysis of mechanical wall stress and abdominal aortic aneurysm rupture risk," *J Vasc Surg*, vol 36, pp. 589–597, 2002.

[8] C. M. Scotti, E. A. Finol, "Compliant biomechanics of abdominal aortic aneurysms: A fluid-structure interaction study," *Comput Struct*, vol. 85, pp. 1097–1113, 2007.

[9] L. Shengmao, et al., "Fluid-Structure Interaction in Abdominal Aortic Aneurysm: Effect of Modeling Techniques," *Biomed Res Int*, vol. 2017, February 2017.

[10] E. S. Di Martino, G. Guadagni, A. Fumero, G. Ballerini, R. Spirito, P. Biglioli, A. Redaelli, "Fluid-structure interaction within realistic three-dimensional models of the aneurysmatic aorta as a guidance to assess the risk of rupture of the aneurysm," *Medical Engineering and Physics*, vol. 23, pp. 647–655, November 2001.

[11] Mimics 10.01, Materialise, Leuven, Belgium.

[12] Geomagic Studio 2013, Raindrop Geomagic.

[13] Femap v.11, Siemens PLM Software, USA.

[14] M. Kojic, N. Filipovic, B. Stojanovic, N. Kojic, "Computer Modeling in Bioengineering - Theoretical Background, Examples and Software," *John Wiley and Sons*, Chichester, England, 2008.

[15] M. J. Thubrikar, J. Al-Soudi, F. Robicsek, "Wall Stress Studies of Abdominal Aortic Aneurysm in a Clinical Model," *Ann. Vasc. Surg*., vol. 3, pp. 355-366, May 2001.

[16] F. Gao, M. Watanabe, T. Matsuzawa, "Stress Analysis in a Layered Aortic Arch Model Under Pulsatile Blood Flow," *BioMed. Eng. Online*. vol. 5, 2006.

[17] M. S. Olufsen, C. S. Peskin, W. Y. Kim, E. M. Pedersen, A. Nadim, J. Larsen, "Numerical Simulation and Experimental Validation of Blood Flow in Arteries with Structured-Tree Outflow Conditions," *Ann. Biomed. Eng*. vol. 28, pp. 1281-1299, 2000.

[18] S. Chandra, S. S Raut, A. Jana, R. W. Biederman, M. Doyle, S. C. Muluk, E. A. Finol, "Fluid-structure interaction modeling of abdominal aortic aneurysms: the impact of patient-specific inflow conditions and fluid/solid coupling," *J Biomech Eng*, vol. 135, August 2013.