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Computational Hemodynamic Analysis of a Patient Specific Abdominal Aortic Aneurysm

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ABSTRACT

Abdominal aortic aneurysm (AAA) is a cardiovascular disease caused by the enlargement of the aorta in the abdomen over time. Unless treated, the growth of AAA continues, resulting in 80% death in the case of rupture. Today, the width of the aneurysm diameter is taken into account in clinical practice to examine the status of AAA. Although there are aneurysms that do not rupture despite reaching a diameter of 9 cm, it is reported that aneurysms with a diameter of 3 cm are ruptured in several cases. Therefore, analyzing only the AAA diameter is not a reliable method, and a deeper investigation is necessary for the rupture risk assessment. In this study, a patient's situation is analyzed using computational fluid dynamics (CFD) simulations, which allows to elucidate the flow dependent parameters such as velocity, vorticity, pressure, and wall shear stress (WSS). First, the patient-specific geometry was obtained and boundary conditions were defined at the inlet and the outlet of the flow domain. The effects of intraluminal thrombus (ILT) formation and patient's effort conditions were also included in the analysis. According to the results, WSS and vorticity increase with the increasing blood flow velocity. In terms of the rupture risk, it has been found that the effect of patient's effort level is more critical than the amount of ILT in the AAA.

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1. Introduction

Abdominal Aortic Aneurysm (AAA) is a cardiovascular disease caused by the abnormal enlargement of the abdominal aorta over time [1]. Unless treated, growth continues and may result in rupture. In the case of rupture, a sudden death can occur at a rate of 80% [2]. The incidence of AAA is reported between 4% and 8% in men over the age of 50, while it is stated between 0.5% and 1% in women [3]. For the ages over 65 years, this probability is between 5% and 9%, but smoking increases this probability by 4 times [3]. Considering the incidence of AAA, the consequences of the rupture highlight the importance of diagnosis and treatment at an early stage.

The diameter of the healthy abdominal aorta ranges from 2 cm to 2.5 cm [4]. If the vessel diameter becomes 50% larger than the healthy value, this condition can be defined as a clinical case and diagnosed as aneurysm. AAA diameters may be larger than 5.5 cm depending on the patient's condition, and these patients are considered as critical in terms of rupture, and a surgical operation should be planned. Today, the width of the aneurysm diameter is taken into account in clinical practice to examine the status of AAA, but this examination criterion does not work well for all cases. Although there are aneurysms that do not rupture despite reaching a diameter of 9 cm, aneurysms with a width of 3 cm can rupture. In addition to the AAA vessel diameter, in order to determine the physical risk of the patient, it is necessary to consider the past history affecting the internal vascular structure, such as age, gender, smoking, enzymatic status of proteins in the microscale, and the patient's disease history. On the other hand, considering the engineering approaches, computational investigations can be conducted on the vascular structure, the state of the clot in the vessel, and the fluid properties [5, 6, 7].

In order to prevent the rupture, the rate of increase in vessel width may be important in addition to the aneurysm diameter in patients diagnosed with AAA. According to European Society for Vascular Surgery (ESVS) and American Heart Association (AHA), vessel diameters that grow more than 1 cm/year or exceed 5.5 cm may require medical intervention [8]. However, open surgery of aneurysm has high risks, so it is preferred only in cases that are considered to be the most risky. It should be noted that the open surgery is not the only treatment method, and in some cases, it can be planned to progress through the vein with endovascular methods and return the diameter of the flow in the vein to normal levels with an endovascular stent [5].

In many previous studies, using numerical methods, blood flow in the aneurysm and strains in the vessel wall were obtained and the condition of the AAA was examined [9, 10]. It has been observed that the enlarged vessel with an aneurysm causes the flow volume to expand [10, 11]. As a result, the shear stress on the inner surface of the vessel decreases due to the drop in the blood flow velocity [4]. Due to the reduction of the shear stress on the inner surface of the vessel, the vessel cells trigger the abnormal development of blood vessel in response. In the light of this information, it is of great importance to examine the patient's condition by taking into account the vascular deformations and internal forces [12]. Intravascular pressure is the main source of rupture for transient and static conditions in vessel rupture, and most academic research has focused on resolving these forces using wall shear stress and pressure [13]. It has been observed that if the wall stress is greater than 65 N/cm² (650 kPa), the AAA structure cannot withstand the mechanical rupture [5]. In addition to these, the formation of clots in the vessel also changes the physical properties of the artery, as well as chemically causes the inner part of the artery to be insufficiently nourished and weakened biologically [12]. When past autopsy reports were reviewed, clots were found in 80% of the ruptured aneurysms. In this study, the wall shear stress, blood pressure, velocity and vorticity in the blood vessel were determined using a patient specific aneurysm model considering the level of clot in the vessel, and activity status of the patient.

2. Methodology

2.1 Geometry

The simulations for AAA analysis can be split into two groups. These are realistic models and idealized models. Idealized models have the ability to examine the effects of the parameters of interest independently while keeping the other parameters constant, but realistic models give results closer to reality [4]. For this reason, it is possible to obtain more realistic flow rates and pressure distributions by using realistic models instead of the idealized ones. Realistic models are obtained by rendering the vessel geometry of patients in 3D using the medical imaging methods.

In a realistic AAA model, there may be a clot deposit as it can be found in most of AAA patients [6, 7, 9]. Three different models were created for this situation. The vessel structures and geometries used in this study are provided in Figure 1 and Figure 2 [6]. When Figure 1 is examined, there is a fixed vessel wall on the outside, while there is a blood clot that restricts the blood flow. As the importance of blood clots, namely intraluminal thrombus (ILT), is stated in the introduction, it is quite possible to encounter a blood clot deposit in AAA patients, since blood clots are seen in many AAA autopsies [14]. However, different clot levels were also analyzed to examine the effect of ILT on flow variables. These analyses were conducted on the models with no clots (No ILT), light ILT, and dense ILT. The blood flow volumes for the vessels are given in Figure 2 in order to provide a comparison for better understanding the difference between the models [6]. From these figures, it can be clearly seen that the flow volume will decrease proportional to the increase in the amount of ILT.

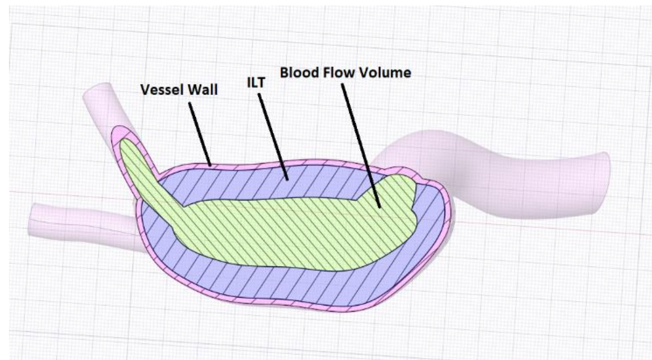

Figure 1. *Abdominal aortic aneurysm (AAA) layers.*

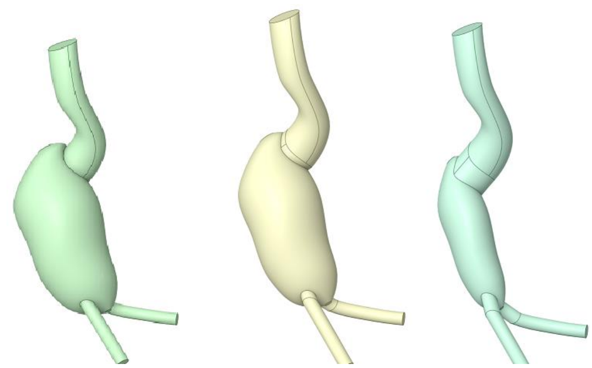

Figure 2. *Blood flow volumes for the no ILT (left), light ILT (middle), and dense ILT (right).*

2.2 Blood model

Blood is a non-Newtonian fluid; that is, its viscosity changes according to the shear stress on the fluid [4, 15]. However, in many studies in the literature, blood has been modeled as a Newtonian fluid due to its simplicity [4, 15]. In this study, the blood was modeled as a non-Newtonian fluid by employing the Carreau model [6, 9]. The coefficients of this model are obtained from previously published academic studies and tabulated in Table 1 [6]. In Figure 3, the viscosity changes are shown as a function of strain rate. Laminar flow analyses are performed using the Carreau model in ANSYS Workbench 2019 R2 platform by using the Fluent solver with laminar viscous flow model.

Table 1. *Carreau model parameters for the blood.*

Mass density [kg/m^3]	1050
Time constant, lambda [s]	10.976
Power-law index, n	-0.3216
Zero shear viscosity [kg/(ms)]	0.056
Infinite shear viscosity [kg/(ms)]	0.0033

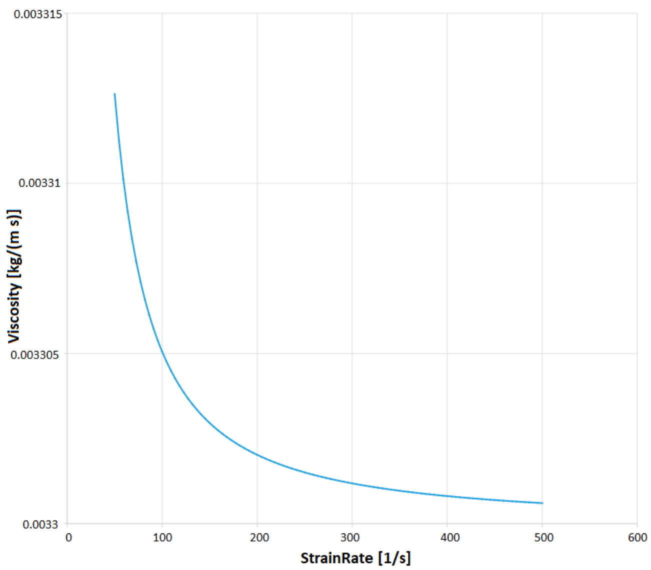

Figure 3. *The change in blood viscosity as a function of strain rate.*

2.3 Boundary conditions

In order to have realistic results, the boundary conditions should be chosen close to the physiologic flow conditions in AAA [4, 6]. For this purpose, an inlet flow profile and an outlet pressure profile are used as provided in Figure 4 and Figure 5, respectively [6, 9]. Figure 4 and Figure 5 show the time-dependent velocities and pressures used in the analysis [6, 9]. In addition, the inlet flow velocities during the effort, normal state, and resting conditions are provided in Figure 4.

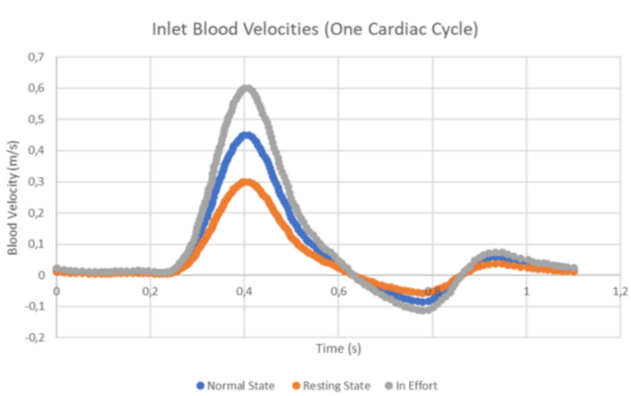

Figure 4. *Inlet blood velocity conditions during one cardiac cycle.*

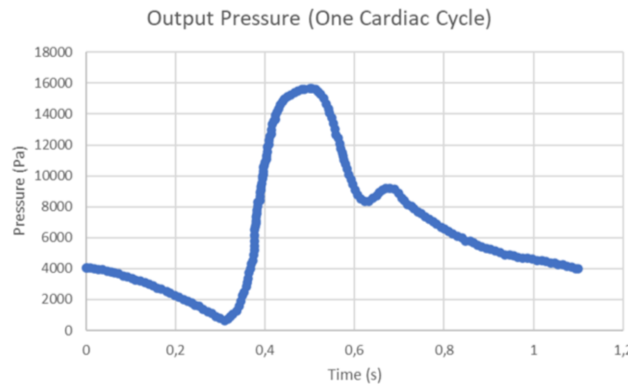

Figure 5. *Outlet pressure conditions during one cardiac cycle.*

3. Results

In this section, the results obtained are presented. A mesh independence study is performed and it is seen that a mesh composed of 450000 tetrahedral elements provide satisfactorily accurate results. Three cardiac cycles are simulated using a total of 600 time steps with 0.0055 s increments. The effects of ILT level and effort conditions are analyzed.

3.1 The effect of the blood velocity

It is important to examine AAA in different conditions since the heart does not constantly pump blood at a constant flow rate, and analysis results for slow (at rest), normal, and fast (in effort) conditions are provided in this section. In Figure 6, 7, and 8, the volume averaged vorticities, maximum wall shear stresses (WSS) on the AAA wall, and maximum blood pressure in the flow domain are investigated by considering different effort conditions, respectively.

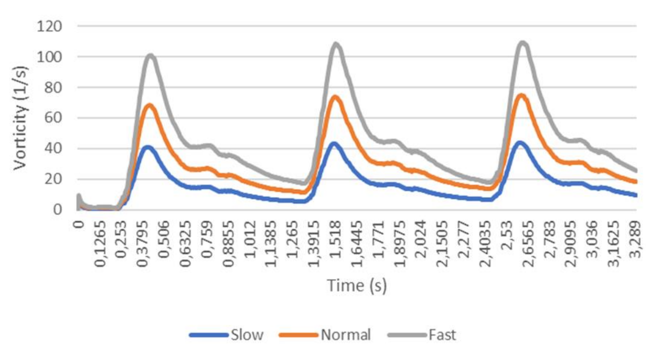

Figure 6. *Volume averaged vorticities for 3 consecutive cardiac cycles considering different effort conditions.*

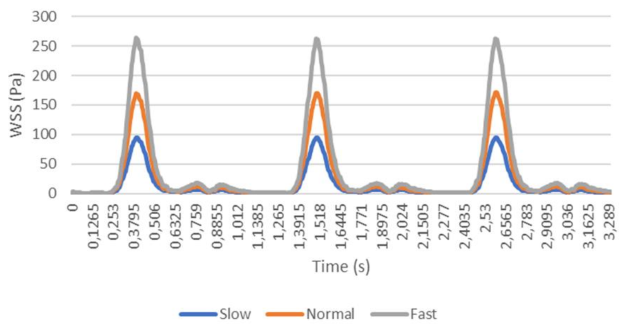

Figure 7. *Maximum WSS for 3 consecutive cardiac cycles considering different effort conditions.*

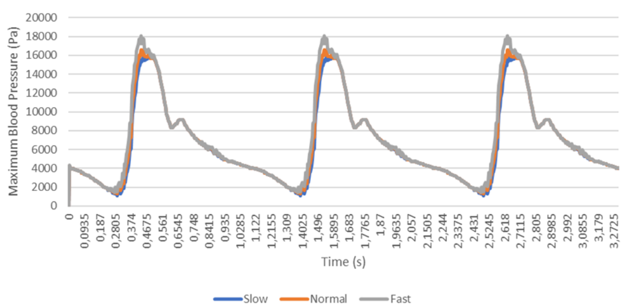

Figure 8. *Maximum blood pressures for 3 consecutive cardiac cycles considering different effort conditions.*

While the resting state is optimal in terms of loads on the vessel and flow parameters, a large increase has occurred in these values even in the normal state. The increase due to the effort conditions is high for the vorticity and WSS, on the other hand, blood pressure only varied by 3% depending on the effort conditions of the patient.

Vorticity is a metric that is defining the rotational behavior in the flow domain. With the increasing flow velocities depending on the patient activity, higher vorticity levels are observed in the aneurysm. This is due to the enlarged AAA section in the flow domain which is generating higher vorticity levels with increased flow velocities.

3.2 The effect of the ILT

The amount of ILT in the AAA limits the flow volume and changes the vascular structure. The effect of ILT on the WSS and vorticity is investigated in this section. In Figure 9, 10, and 11, the volume averaged vorticity, maximum WSS, and maximum blood pressure are investigated by considering different amount of ILT in the AAA, respectively.

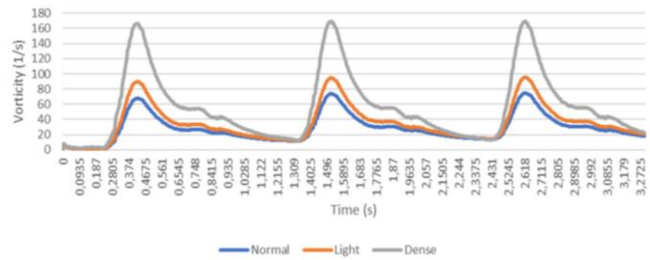

Figure 9. *Volume averaged vorticities for 3 consecutive cardiac cycles considering different ILT amounts.*

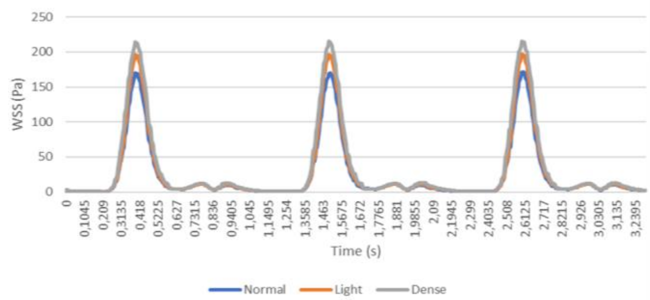

Figure 10. *Maximum wall shear stresses (WSS) for 3 consecutive cardiac cycles considering different ILT amounts.*

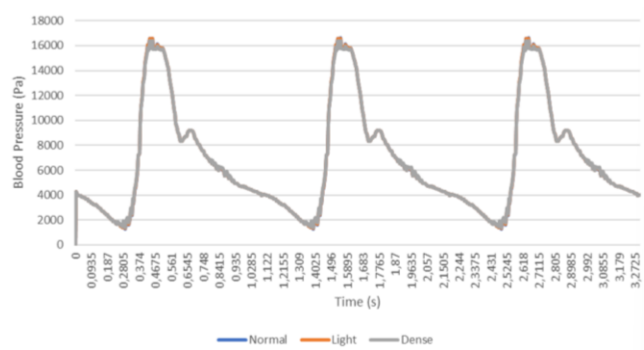

Figure 11. *Maximum blood pressures for 3 consecutive cardiac cycles considering different ILT amounts.*

4. Conclusion

In this study, the effects of effort conditions and ILT amounts are examined using computational fluid dynamics (CFD) analysis. As a general behavior, the increase in flow velocity caused an increase in vorticity and WSS levels.

There was no significant change in pressure in the investigated cases. This condition may be due to the applied outlet pressure boundary conditions in the analysis. Due to the lack of data in the literature, a common pressure boundary condition was used for three different activity conditions. In further studies, it is aimed to improve the models by implementing patient-specific boundary conditions in the CFD simulations.

In the numerical analysis, the most important parameter was considered to be the wall shear stress (WSS). It was observed that WSS levels were heavily affected by the flow rate. While the increase in the amount of ILT increased the amount of WSS by approximately 10%, there was a WSS increase of 47% and 28% due to the 50% and 33% increase in flow rate, respectively. This shows that the effort conditions lead to different flow rates which significantly affect the WSS environment in the AAA.

WSS can be determined by dividing the fluid-driven friction force to the local wall area and do not have a direct effect on the rupture. However, the abnormal levels of WSS reduces the functional behavior of the endothelial cells on the AAA wall, and this condition leads to degenerated AAA wall with lowered mechanical strength. Therefore, the abnormal WSS levels play a role in the rupture of AAA in the long-term exposure. For the investigated cases, it is seen that the WSS values did not exceed 300 Pa in the modeled cases. It should be noted that even if the rupture does not occur at these levels, some risk factors may overlap and trigger the rupture mechanism.

In the light of the results obtained, it can be concluded that the effect of effort conditions is found to be more prominent than the amount of ILT burden in the AAA. If the AAA is exposed to high effort conditions with the high ILT burden at the same time, the rupture risk significantly increases as can be seen from the excessively increased vorticity and WSS levels around the aneurysmal enlargement.

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