The aim of this research was to show superiority of using real geometries in simulations of blood flow through cardiovascular system.

Our model compared blood flow through an abdominal part of aorta reconstructed with a use of data from an AngioTK research with the 3DDoctor software to geometries with the same diameters at inlet and outlet as mention before but created only with the Gambit 2.2.30 software without data from AngioTK. Blood flow simulations were prepared with the Fluent 6.2.16 software. Calculations of flow through a real geometry allows to obtain realistic results of values connected with process of blood flow.

Results showed that calculations blood flow through a virtual geometry lasted two times longer than for a real geometry. Mesh for a real geometry consist about 600,000 elements and for a virtual geometry about 900,000 elements. Wall shear stress and blood velocity was higher for a real geometry and closer to that in human organism.

It was shown that calculating a virtual geometry vessel was to big simplification when investigating blood flow through a vessel. Application of mathematical models based on real geometries gives more realistic results than artificial geometries. Virtual models have lots of simplifications which results are far away from expectations. Simplifications depend on the model that is used.

Keywords and phrases: CFD, abdominal aorta, computer simulations, computer tomography, geometry.

Introduction

Mathematical methods have become popular among scientists who use them for describing experiments that illustrate dangerous conditions which may affect human body. Medicine is a branch of knowledge in which computer simulation are wildly use to predict side effects of different medical procedures. Sometimes it is difficult, dangerous or even impossible to perform in vivo experiments and measurements in biological fluids while computer simulations can provide insights into physiological processes without any harm for living organisms. Moreover, specialized software are used to project and simulate parts of skeletal system [1], or even to analyze blood coagulation process and factors that may influence it [2]. High performance computers provide the computational rates necessary for advanced biomedical computing. Nowadays, a common move is to analyze endovascular prosthesis placement with use of a specialist software, before inserting it into human [3]. Such process enable to delete elements which may collide after implantation at the project level. Blood flow process is a good example of application mathematical models, which are utilized to solve problems connected with liquid phase flow [4]. The CFD (computational fluid dynamic) is a perfect tool to simulate all kinds of processes connected with a domain of flow [5]. Results depend on geometry and boundary conditions. However, computer simulations are often burden with mistakes related to necessary simplifications that are made during calculation process. Usually the physical domain is divided into many subdomains by imposing a grid. Solving the mathematical model over a discretized domain involves obtaining the values of a certain physical quantity at every grid point for each time interval. A grid point is influenced only by the surrounding grid points. Each calculation step gives new values of the physical quantity for the next interval of the real time.
Aim

The aim of this article was to compare 3D real geometries and 3D virtual geometries used for simulations of blood flow through an Abdominal Aorta Aneurysm.

Material and methods

Computer simulation was used to examine flow through an abdominal part of aorta with the use of real and virtual geometries. Two types of a 3D geometry, real and virtual, were prepared.

Geometry

To get a real 3D model of Abdominal Aorta Aneurysm (AAA), AngioTK data of abdominal part of human body was used. Virtual 3D geometry was created with use of special Gambit 2.2.30 software.

CT scans with use of 3Ddococtor were segmented into a 3D geometry. On each scan area representing investigated object was marked (Fig. 1). At the beginning geometry of AAA represents only close surface without inlet and outlet boundary. Next step was to change investigated object from closed area to volume and create surfaces for inlet and outlet of blood. It was achieved with use of Gambit 2.2.30 software.

Final step, before exporting into calculation, was discretization. Volumetric schema of meshing was used with tetrahedral elements. Size of elements was in range between 1 to 2.5 mm. To obtain the smallest elements (1 mm) near the aorta’s wall and the biggest (2.5 mm) inside sizing function for discretization was used. Number of elements was around 600,000 (Fig. 2).

A virtual geometry was created with the use of Gambit 2.2.30 software (Fig. 2). It was assumed that inlet and outlet surfaces are circles. Their diameters were calculated with usage of inlet and outlet surfaces from a real geometry. Diameter of aneurysm for a virtual geometry was equal to maximum diameter of aneurysm in a real geometry. This simplification allows to create artificial aneurysm for a virtual 3D geometry. Table 1 summarizes obtained results.

A radius of inlet and outlet tubes did not vary. As a result of previously described simplification aneurysm of a virtual geometry had different size compared with a real geometry. Only high and maximum diameter were the same. Simplification for virtual model connected to obtained regular shape of aneurysm. A real geometry had irregular shape with changing of diameter for longitudinal section. Mesh consist of about 900,000 elements. Size of elements was the same as for a real geometry. A sizing

![Fig. 1. Preparation and segmentation of 3D geometry from CT scans.](image-url)
Andrzej Polanczyk, Aleksandra Piechota

function to deposit the smallest elements near the aorta’s wall and the biggest in side was also used.

Simulation

We analyzed the laminar blood flow [9] through a real and virtual 3d geometry with the usage of Fluent 6.2.16 software. At the inlet of both investigated 3d models of AAA we assumed pulsating character of flow as a boundary condition. The flow through these models was predicted using three-dimensional Navier-Stokes equations and the equations for conservation of mass. In classical form, these equations may be written as follows.

$$\nabla \cdot U = 0$$  \hspace{1cm} (1)

$$\nabla \cdot (U \cdot U) = - \nabla \cdot \frac{\nabla p}{\rho} + \nu \nabla^2 U$$  \hspace{1cm} (2)

where: $\rho$ — fluid density,  
\nu — kinematic viscosity,  
$U=(u,v,w)$ — fluid velocity,  
p — pressure.

Data from USG-Doppler research from one patient was used to define pulsating character of blood flow utilizing Fourier transformation (eq. 3).

$$F(t) = \frac{1}{2} a_0 + \sum a_n \cos(n\omega t) + b_n \sin(n\omega t)$$  \hspace{1cm} (3)

where: $a_0$ $a_n$ $b_n$ — coefficients of Fourier

The pulsating character of blood flow in human cardiovascular system was implemented in simulations. One heart’s cycle for healthy human is about 0.8 s. For calculations was taken 1 s period of time. This assumption allows shorten calculations but it had not influenced obtained result. At outlet was assumed 0 pressure boundary at the beginning of calculation. Our model presents one important simplification, a no-slip condition was applied at the interface between the blood and the endovascular prosthesis wall. Because of smooth structure of inner side of vessel’s wall roughness was neglected. Investigated parameters were wall shear stress WSS and blood velocity. Density of blood was assume 1050 kg/m$^3$. Blood was assumed to be non-Newtonian liquid [1]. Blood viscosity changes its property because of shear velocity. Including this assumption Casson model of viscosity was used (eq. 4) [2].

$$\mu = \left[ \left(0.0012(1 - Htc) \right)^2 \left(\frac{\gamma}{2} \right) \right]^{1/4}$$  \hspace{1cm} (4)

$$+ 2^{1/2} 0.1 \cdot 0.625 Htc^{3/2} \left(\frac{\gamma}{2} \right)^{3/2}$$

where: $Htc$ — hematocrit,  
\gamma — shear velocity.

Figure 3 shows changing value of viscosity depends on shear velocity for $Htc = 40\%$. With increasing value of shear velocity blood viscosity decreased. It happened because of increasing shear stress together with shear velocity. Shape of flow curve was typical for liquids diluted with shear.

![Fig. 3. Casson model of blood viscosity.](image-url)
Results
Calculation of blood flow through investigated abdominal part of cardiovascular system gave the distribution of blood velocity and wall shear stress. The most important features of results obtained are summarized in Table 2.

Table 2. Summary of blood flow calculations.

<table>
<thead>
<tr>
<th>Variable</th>
<th>Results of real 3d geometry</th>
<th>Results of virtual 3d geometry</th>
</tr>
</thead>
<tbody>
<tr>
<td>max. blood velocity m/s</td>
<td>1.55</td>
<td>1.31</td>
</tr>
<tr>
<td>min. blood velocity m/s</td>
<td>0.51</td>
<td>0.59</td>
</tr>
<tr>
<td>average blood velocity m/s</td>
<td>0.91</td>
<td>0.84</td>
</tr>
<tr>
<td>max wall shear stress Pa</td>
<td>33.82</td>
<td>26.68</td>
</tr>
<tr>
<td>min wall shear stress Pa</td>
<td>0.09</td>
<td>0.03</td>
</tr>
<tr>
<td>average wall shear stress Pa</td>
<td>3.16</td>
<td>3.13</td>
</tr>
</tbody>
</table>

Blood velocity
Figures 4a and 4b show blood velocity for a real and virtual 3d geometry. A real geometry obtain higher velocity of blood flow compare with a virtual geometry. Both models had pulsating character of blood flow. USG-Doppler research was similarly to real geometry results.

Shape of a virtual geometry was regular compare with a real geometry. No deformation on walls decreases blood velocity. It involved decreasing drag forces inside analyzing part of aorta. A real geometry had irregular shape. Increasing and decreasing of diameter, which induced increasing of blood velocity was visible (Fig. 5).

A real and virtual 3d model of Abdominal Aortic Aneurysm was partially validated with velocity data.
obtained from Doppler ultrasound measurements. The measured maximal velocity and the shape of the measured velocity were similar with theoretical calculations for a real geometry.

Wall shear stress
Spatial distribution of maximum, minimum and average WSS was shown in Figs. 6a and 6b. Maximum shear stress for a real geometry was approximately 35 Pa and for a virtual geometry about 27 Pa. Irregular shape of real geometry walls involved higher value of WSS. Cylindrical shape of virtual geometry was obtained with no drag forces on walls (Fig. 7).

There were no changes of the aorta diameter which was obvious in a real model (Fig. 7). Elimination of natural deformation of blood’s wall, because of flow, resulted in decreased WSS. Mesh for virtual 3d geometry consisted of more elements because of its regular shape and bigger size. Diameter of aneurysm was constant for a virtual geometry but it varied with longitudinal section in a real geometry aneurysm. That was a reason why the total volume of a real geometry was lower comparing to a virtual geometry.

Conclusion
In this article we presented CFD model of blood flow through an abdominal part of aorta with aneurysm. Calculated model includes pulsating character of blood flow. Simulation was performed for period of time 1s.

It was shown that calculating virtual geometry vessel was too big simplification when investigating blood flow.

Fig. 6. Wall shear stress for: a) real 3d geometry and b) virtual 3d geometry.

Fig. 7. Wall shear stress contours for real and virtual 3d geometry.
through a vessel. Also, it increased simulations time to about 25h and decreased calculations precision. Compare both, virtual and real geometries, it can be seen that only with real geometries results similar to human organism can be obtained. Virtual model has got smaller velocity values comparing to real geometries and USG-Doppler research. Elimination natural deformation vessels wall obtain decreased of WSS.

References