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Effects of the wall motion in artificial abdominal aortic aneurysms

Research Paper for Scientific Student Competition

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1. Introduction

Studying the human body has thousands of years of tradition. The technical progress results in continuously better methods and techniques to do research on ourselves. Even though, rapid development is observed in this field, there are several open questions.

An aneurysm is a localized blood-filled balloon on the wall of a blood vessel. According to their location, we differentiate between intracranial and abdominal aneurysms. Not all aneurysms are dangerous. According to the traditional view, when the size reaches a certain level, there is an increasing risk of rupture that can be fatal. There are some surgical intervention techniques [1] that may decrease the risk of the rupture. However, these interventions themselves carry some risk. Reason for the development of aneurysms is still unknown but researchers found that there is a strong biological, chemical, and physical interaction between the vessel wall and the blood flow [1]. On the one hand, flow-induced pressure periodically deforms the wall. On the other hand, the wall shear stress influences the structure of the vessel. The receptors on the inner wall surface send biological signals to the vessel wall according to the physical parameters of the flow. This way the cellular structure and the mechanical properties of the vessel are controlled [1].

CFD (Computational Fluid Dynamics) is nowadays reliable enough to be used as a tool to examine the flow field in various cases. FEM (Finite Element Method) is also a popular method to calculate the necessary variables in loaded and deformed solid bodies. Software developers are constantly working on the coupling of the two systems, so there are some available programs that can manage the coupling. These programs offer an excellent chance to investigate the blood flow in the vessels, and examine the interaction of the vessel wall and the blood flow.

The aim of this work is to assess the differences between two-way and one-way coupling of wall deformation and flow in abdominal aortic aneurysms. For this task two-way FSI (Fluid Structure Interaction) and rigid wall simulations must be compared. The chemical processes were neglected, and only the mechanical interaction between the elastic wall and the fluid was investigated. There are some articles, which negotiate the differences between one-, and two-way couplings [2] but no work was found that attempts to analyze the reasons of differences. It matters whether the differences are caused by the wall motion, or by the change of the geometry. The results may lead to a simplified treatment of highly resource-intensive two-way coupled FSI simulations.

2. Literature survey

2.1. Biological background

The systemic circulation is responsible for the metabolism of the cells. The engine of this system is the heart. It works like a volumetric pump and circulates the blood in our blood vessels [3].

The vessel wall has more layers: tunica intima, tunica media and tunica adventitia (Figure 1). The physical parameters of the wall are functions of the structure and the biological composition. For example, the elastin and collagen content influences the elasticity of the wall [4]. The blood flow plays also a great role in the properties of the wall [1] but these processes are not well known. All in all it is experienced that the structure of the vessel wall and the flow is changed near an aneurysm. This leads to the conclusion that the aneurysm growth may be initiated by changes in the flow field. It was found that wall shear stress (WSS) peak and the WSS gradient determines clearly the location of the emergence of intracranial aneurysms [5]. Moreover the slowed flow in aneurysms facilitates the formation of thrombus. Thrombus is the final product of the blood coagulation.

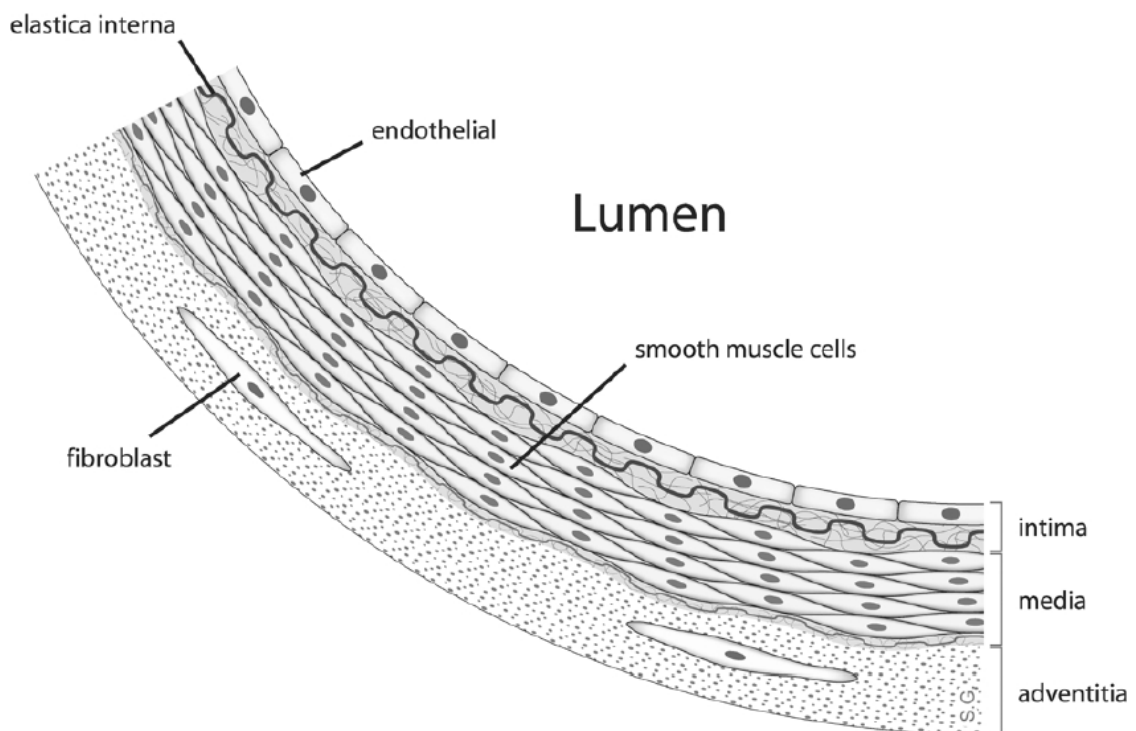


Figure 1: Anatomy of the arterial wall [6]

Cardiovascular diseases (including aneurysms) are the leading cause of death before the age of 65 in Europe [7]. The problem is that aneurysms have no symptoms and they are perceived often just after the rupture. The rupture leads to death in almost 90% of the cases

[8]. The medical doctors' decision about an operation is based on the size of the aneurysm. It is not a perfect method, because some bulges were found under the critical size, which ruptured and some were found over the critical size, which remained intact (giant aneurysms) [9]. A better indicator is necessary to size up the risk of the rupture. It is hoped that fluid mechanics research can give a solution.

2.2. Simulations of other authors

Making rigid wall simulations becomes a common practice in the field of vascular research. For example, Sheidaei et al. [10] modelled the growth of a realistic abdominal aortic aneurysm based on the pressure and wall shear stress distribution with a rigid wall assumption. Modelling of intracranial aneurysms can also be carried out using the rigid wall assumption [5, 9]. The effect of the wall motion is often negligible, because smaller arteries (further away from the heart) have a larger stiffness or the vessel wall is supported by harder tissues from every direction in some cases (typical for intracranial aneurysms). The simulation of the aortic flow requires taking the wall motion into consideration. The radial and axial inhomogeneity of the wall is usually neglected and a homogeneous material is assumed with constant thickness [11], although in some studies great emphasis is laid on the modelling of patient-specific wall [12]. Some authors modelled the intraluminal thrombus with its own Young modulus [13]. There are works which compare steady and transient results [14]. A frequent aim of these studies is to establish relation between physical parameters [5] (WSS, von Mises stress...) or the geometrical shapes [15] and the risk of rupture, to model the growth process [16], to investigate the effect of treatments [17], or to search for an easier way to model these complex problems [2, 14].

3. Theoretical background of the simulations

Biomedical research is a typical field of the application of FSI. FSI analysis involves performing a fluid analysis taking into account the interaction with the corresponding structural analysis. There are two types of FSI:

1; One-way FSI: the result of a CFD analysis is applied as a load to the mechanical analysis. The boundary displacement values resulted of the mechanical application is not passed back to the fluid model. It is a passable assumption if the deformation of the solid body does not influence the flow significantly. In this case the solver of the solid model is used after the CFD calculation.

2; Two-way FSI: the results from a CFD analysis are applied as a load to the mechanical analysis and there is a reverse information transfer on the deformed geometry too from the mechanical side to the CFD analysis. In this case the two solvers run parallel and the calculation will continue iteratively until overall equilibrium is reached between the mechanical and CFD solution. Two-way FSI is much more resource-intensive [18].

3.1. The solvers

The used software can solve the basic differential equations of the deformation of solid bodies and the motion of fluids numerically. CFX is a computational fluid dynamics software in the ANSYS package, and Mechanical is the responsible program for the solid body mechanics. CFX is based on the finite volume method and Mechanical is based on the finite element method.

The necessary equations for the fluid simulations are the continuity as the mass conservation equation and the Navier-Stokes equation as the equation of motion [19].

Equation (1) is the continuity equation for incompressible fluids where $\bar{v}$ is the velocity vector.

$$\text{div} \bar{v} = 0 \quad (1)$$

Equation (2) is the vectorial form of the Navier-Stokes equation if the density and the viscosity are constant. (Of course there is a form for non-Newtonian fluids too.) In this expression $\bar{v}$ is the velocity vector, t is the time, $\bar{g}$ is the field strength vector, ρ is the density, p is the pressure, μ is the dynamic viscosity,

$$\frac{d\bar{v}}{dt} = \frac{\partial \bar{v}}{\partial t} + \bar{v} \cdot \nabla \bar{v} = \bar{g} - \frac{1}{\rho} \nabla p + \frac{\mu}{\rho} \nabla^2 \bar{v} \quad (2)$$

The finite element solver solves the stiffness equation (3). It is a simplified matrix equation, where $\bar{\bar{K}}$ is the stiffness matrix of the body, $\bar{U}$ is the nodal displacement vector, and $\bar{\bar{F}}$ is the nodal load matrix [20].

$$\bar{\bar{K}} \bar{U} = \bar{\bar{F}} \quad (3)$$

3.2. The coupling

The interaction between the two analyses typically takes place at the fluid-solid interface. The results of one analysis are passed to the other analysis as loads across the fluid-structure interface. The surface mesh of the models can be different, so the data transfer is not simple in all cases. Each data transfer incorporates the following algorithmic components:

➤ Mapping: this is the first component. It contains the pairing of the source and the target location to generate weights. ANSYS uses two different mapping algorithms. The ‘Smart Bucket Algorithm’ divides the target mesh into a relatively coarse grid of imaginary buckets (each target nodes will be in a bucket). The nodes of the source mesh are associated with a bucket corresponding to the target mesh. Consequently a bucket can involve a number of source nodes (non-empty), or it remains empty. If the bucket is non-empty, the mapping algorithm tries to match the source nodes to one or more target elements in the bucket based on the distance between the source and target node. It works with a tolerance, so some source nodes could remain unmapped, if they are out of the tolerance. If the bucket is empty, then the algorithm works with the nearest non-empty bucket.

Another way is to generate a grid interface. This method divides the target and the source side into sub-faces (lines marked with X in *Figure 2*). These 3-dimensional sub-faces will be projected on to each other and they create the 2-dimensional control surface from the overlapping areas. The ratio of the areas determines the final mapping weights. *Figure 2* helps to imagine this procedure.

➤ Interpolation: this is the second component, and involves the use of the generated weights to project source data onto target location. Equation (4) presents the interpolation. φ is the value on the target side, φ_i is the i^{th} touched source value, and w_i is the associated weight.

$$\varphi = \sum_{i=1}^n w_i \varphi_i \quad (4)$$

➤ Interpolated data post-processing: this is the final component of the data transfer. It could involve explicit under-relaxation of the target data. The under-relaxation increases the simulation time, but makes the calculation more stable. Equation (5) shows the data transfer with under-relaxation between two coupling iterations, where ω is the under-relaxation factor.

$$\varphi_{\text{new}} = \varphi_{\text{old}} + \omega \cdot (\varphi_{\text{new}} - \varphi_{\text{old}}) \quad (5)$$

It is perceptible that the data transfer influences the accuracy of the simulation. The interpolations can ‘smear’ the interface load. It is also offered by ANSYS to generate the surface meshes identical at the interface.

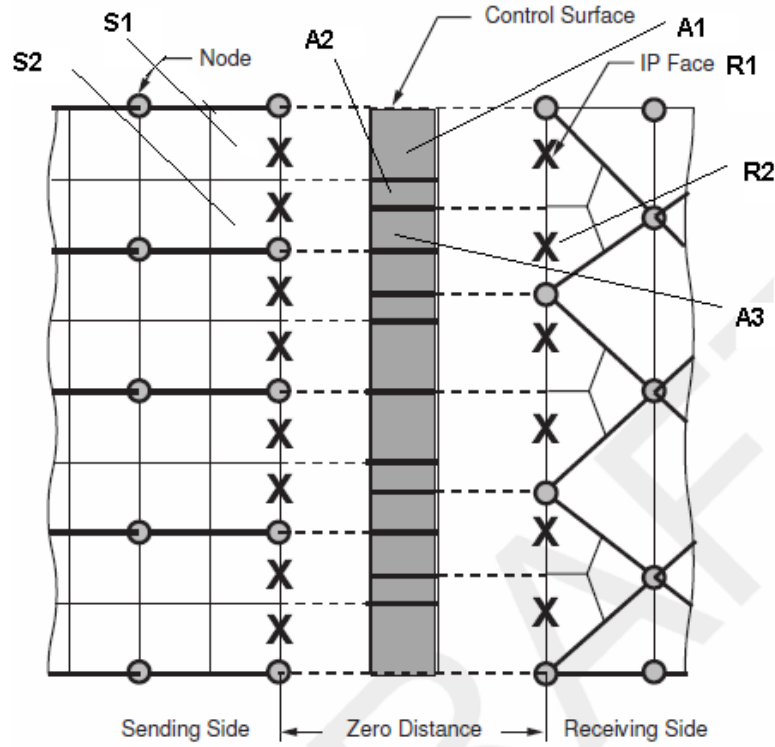


Figure 2: Mapping with a grid [18]

4. Pre-processing

4.1. Geometries

Four artificial aneurysm-like geometries were designed with a CAD program (CATIA v5). The geometries have in every case a ten diameter long initial part that is a rigid wall straight tube. Paál et al. [21] found that a tube like this is adequate for a Womersley profile to develop by the beginning of the investigated artery section. Without this there can be great differences between the real and the modelled flow [22].

The inlet and the outlet diameters are $D = 20$ mm (mean aorta diameter 21.3 mm Allison et al. [23]). The inlet and the outlet surfaces are parallel, and the distance between them is 370 mm. Fixed constraints were applied on them. (The abdominal aorta was originally modelled as a 170 mm long straight tube.) Two geometries (axisymmetric, asymmetric) are taken from Egelhoff et al. [8], and two (S-tube, S-tube with bulge) are idealized geometries based on the patient-specific image of the Vascular Surgery Clinic. *Appendix 1* includes a picture of the geometries, and shows the locations, where the physical parameters were compared. (The shape of an aorta changes sometimes if a person gets older. The blood vessels elongate. The S-tube intended to model this elongated state.)

4.2. The meshes

A good quality mesh is essential for numerical calculations. A hexahedral mesh was generated for every artificial fluid region with ICEM CFD 14.0. The mesh of the axisymmetric and asymmetric geometry is block-structured. For the S-tube, and the S-tube with aneurysm, a block-structured part was created, and the mesh was extruded from this. In every case there is an inflation layer near the wall with a growth ratio 1.2. The height of the first element was 0.05 mm.

The solid mesh forms a shell around the flow field. It was extruded from the surface elements of the interface (this way the program does not need to use any interpolation between the fluid and solid solver). A uniform (1.75 mm) vessel wall thickness was assumed following statistical data [24]. The wall was divided into two elements in the radial direction on the entire length. *Table 1* includes the number of elements; *Figure 3* shows the outlet surfaces, the fluid and the solid meshes, without the ten diameter long straight inlet tubes.

	number of elements in the circumferential direction	number of elements in the fluid zone	number of elements in the solid zone
axisymmetric	80	401280	28120
asymmetric	84	375192	26928
S-tube	80	469205	38000
S-tube with aneurysm	80	421135	33896

Table 1: Grid parameters

5. Model parameters

5.1. Blood

The blood was modelled as an incompressible, Newtonian fluid. Some non-Newtonian fluid models exist in the literature (for example Carreau-Yasuda model [17]) but the non-Newtonian behaviour of blood is negligible in large vessels like the aorta [4, 25, 26,]. It is a common assumption, however Khanafer et al. [27] compared Newtonian and non-Newtonian fluid flow in AAAs, and their conclusion was that WSS is significantly affected. All in all the non-Newtonian behaviour of blood may not have a great role from the comparison point of view, but it can be worth examining it. The dynamic viscosity of blood was $\mu = 0.003$ Pas, and the density was $\rho = 1050$ kg/m³ [5, 15, 21].

5.2. Vessel wall

The vessel wall was modelled as a homogeneous linear material. A better approximation would be the hyperelastic wall, for example the Mooney-Rivlin model [28], but it does not have such a great role from the comparison point of view. It is planned to implement a viscoelastic material model [29] that can take the hysteresis into consideration. The physical properties of the wall are presented in *Table 2*.

Density (ρ)	2000 kg/m ³
Young's modulus (E)	2.7 MPa
Poisson's ratio (ν)	0.45

Table 2: Properties of the vessel wall [12]

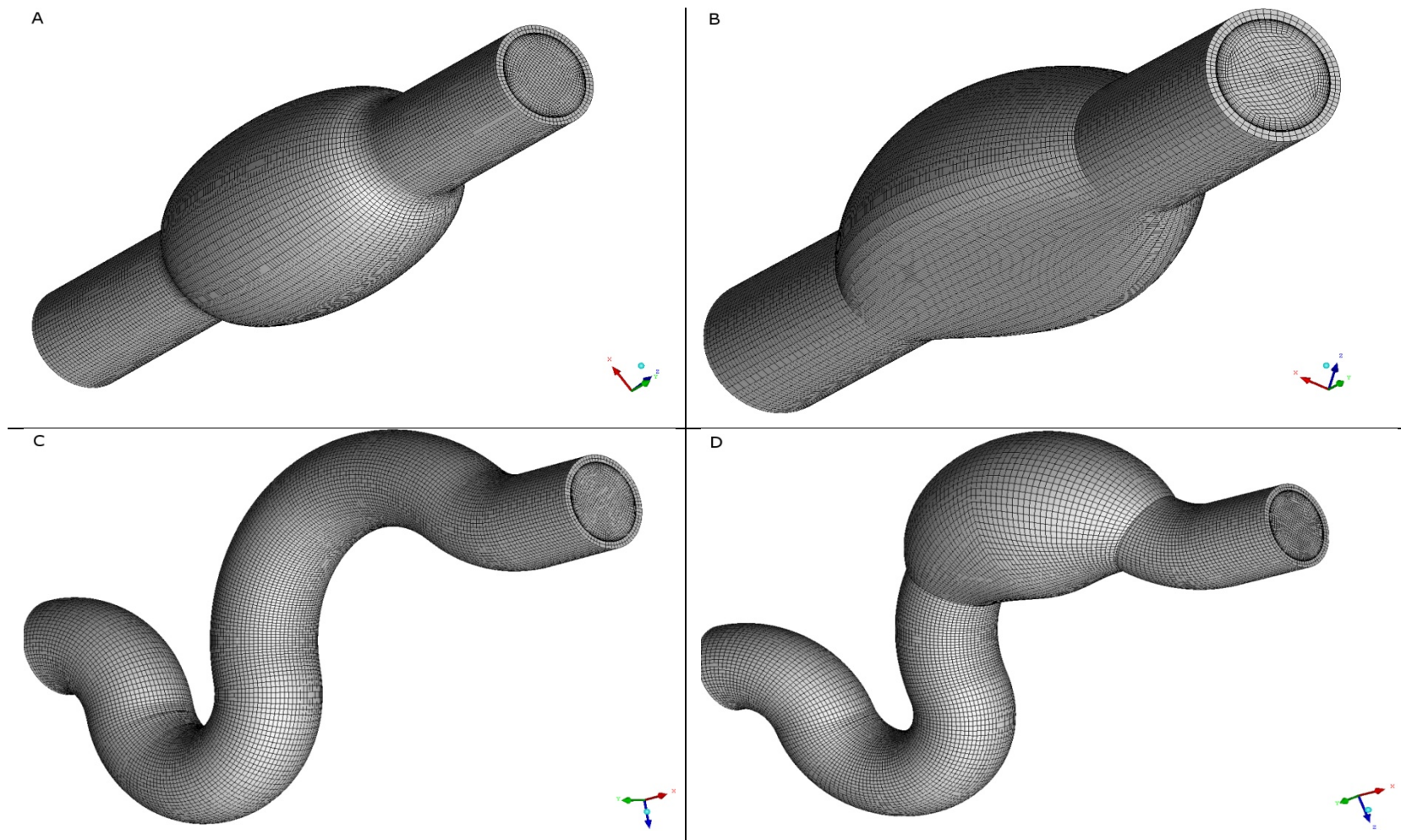


Figure 3: Meshes of the artificial geometries: A) axisymmetric, B) asymmetric, C) S-tube, D) S-tube with bulge

5.3. Strategy of the simulation

The simulations for the comparison consist of three heart cycles and the last cycles were compared. The period time of one heart cycle is in this case $T = 1$ s [2, 29]. It is shown below that the static pressure in the lumen is the sum of a constant pressure, and a pulsatile pressure (*Figure 5*). A ‘pre-run’ was made (two heart cycles) to obtain the deformation. The deformation also has a constant and a pulsatile part, and after the ‘pre-run’ the constant part was separated. The separated part is called the ‘minimum deformed’ geometry. Because the work aims to investigate the effect of the wall motion, the constant part was kept as starting geometry, and the minimum deformed geometry was loaded with the pulsatile pressure. Sometimes it was necessary to remesh the deformed solid and the fluid geometry too. (During the pre-simulation the solver wrote out the deformed mesh, but sometimes it got malformed. After a quality check it is easy to decide if a remeshing is necessary.) The deformed solid mesh was extruded again from surface elements of the interface with a thickness of 1.75 mm. Finally it became possible to do rigid wall simulations with the minimum and maximum deformed geometry, and to make a two-way FSI only for the pulsating wall motion without the constant part of the deformation.

The applied Workbench project includes several modules. Transient Structural module is the transient simulator of solid bodies in the Mechanical software. These modules are responsible for the vessel wall, and Fluid Flow (CFX) modules are responsible for the blood flow. Finite Element Modeler helps to import the mesh into Transient Structural. (Finite Element Modeler was created to make the importation of a mesh easier, and it makes possible to transform the mesh to geometry). The post-processing unit is CFD-Post. A draw about the Workbench project can be found in *Appendix 2*.

The Womersley number (Wo) can be calculated from the data that are sketched above. Womersley number is a dimensionless number in biofluid mechanics. It is an expression of the pulsatile flow frequency in relation to viscous effects [6]. In equation (6) the frequency is represented by the period time T :

$$Wo = \frac{D}{2} \sqrt{\frac{2 \cdot \pi \cdot \rho}{T \cdot \mu}} = \frac{0.02}{2} \sqrt{\frac{2 \cdot \pi \cdot 1050}{1 \cdot 0.003}} \approx 14.8 \quad (6)$$

Stalder et al. [30] used the same formula and the Womersley number was calculated from data measured on volunteers. 14.8 correlates well with the realistic numbers (12÷16.1).

5.4. Solver

The CFX-Solver Manager offers several options to transact a coupled simulation. This part of the program is responsible for the coupling too. The main options of the solver can be found below.

- Time step: 0.01 s (same in the solid and the fluid modules). This value was satisfactory, but in the literature the largest time step value was 0.005 s [13]. A time step convergence study is recommended for further research.
- Iterations: 1÷40 coefficient loops/ time step in rigid wall cases, 1÷2 coefficient loops and 5÷200 coupling iteration/ time step in FSI cases. (The number of coupling iterations seems high, but the solid part needs that to converge in the first steps.)
- Under-relaxation factor: 0.1
- Convergence target: 10^{-2} (default value for the solid part)
- Residual target: $2.5 \cdot 10^{-5}$ (for the fluid part)
- Advection scheme: High resolution
- Transient scheme: Second order backward Euler
- Turbulence model: laminar (Khanafer et al. [28] emphasizes the turbulence modelling in AAAs, and turbulence was recorded in humans by Bluth et al. [31], while other researchers [13, 21, 29] prefer laminar models, and laminar computations were validated using PIV (Particle Image Velocimetry), and LDA (Laser Doppler Anemometry) by Ugron et al. [32] in a realistic intracranial aneurysm model.)
- Reference pressure: 1 atm

In addition to the residuum of the momentum and mass, the imbalances, the pressure and velocity convergences were also monitored.

6. Boundary conditions

The definition of the boundary conditions is one of the most critical parts of the work in numerical simulations. There are some well-tried methods, when intracranial aneurysms are modelled. One of them is to define an inlet velocity profile and describe a constant pressure by the inlet after a porous zone [5, 15]. It can be realistic from the fluid mechanics point of view because in that case peripheral resistance (level of the capillaries) renders the pressure constant near the aneurysm. The porous zone represents this pressure drop. This method was tested in the aorta geometries, and it did not work. The pressure before the porous zone did not follow the temporal history of the aortic pressure. The level of capillaries may be too far from the aorta, so the porous zone approach is not satisfactory. One way is to define constant

pressure (without porous zone) as an outlet boundary condition, but it is not realistic in the aorta. Others use measured data as boundary conditions [27]. It is a precise solution, but it is very hard to measure pressure and volumetric flow rate or velocity in the blood vessels. Also, there are no measured data everywhere in the systemic circulation. Another, generally usable way must be found.

6.1. Fluid

The source of the boundary conditions on the fluid side was the one-dimensional in-house transient simulator [29]. The main parameters of the necessary Matlab simulation are expounded in this paragraph. The solver uses the method of characteristics to solve partial differential equations. The basic equations of the solver are the continuity, the equation of the motion, and the equation for the wave propagation velocity on the fluid side. The correlation between the stress and the deformation is available on the solid side. Laplace's law makes relationship between the solid and fluid parts. The results are physical quantities of the blood flow in the nodes of the systemic circulation model (*Appendix 3*). In *Appendix 3* 'C' symbolizes the heart and the branches (for example A04) are the blood vessels. The branch A04 models the abdominal aorta. The applied blood vessel wall in the transient simulator is a viscoelastic material showed in *Figure 4*. The parameters of the wall (E_1 , E_2 , η_2) remained the same as in the dissertation of Bárdossy [29]. The geometric dimensions of A04 were changed: the diameter was increased to 20 mm, and the wall thickness to 1.75 mm. These values are in the physiological realistic range, and they are better from the FSI point of view.

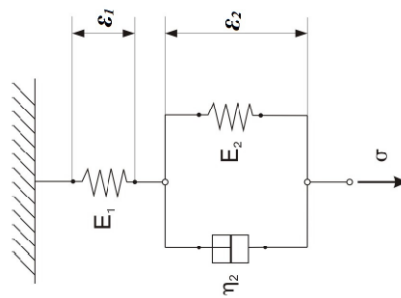


Figure 4: The viscoelastic material model used in 1D simulation [29]

It was easy to obtain the volumetric flow rate data (node 23a), and the pressure data (node 20a) from the solver. The mean velocity (v) was calculated from the volumetric flow rate. The maximum velocity in the middle of the circular inlet was twice the mean value

because of the parabolic inlet profile. After a high frequency filtration the data became suitable. *Figure 5* shows the filtered velocity and pressure signals.

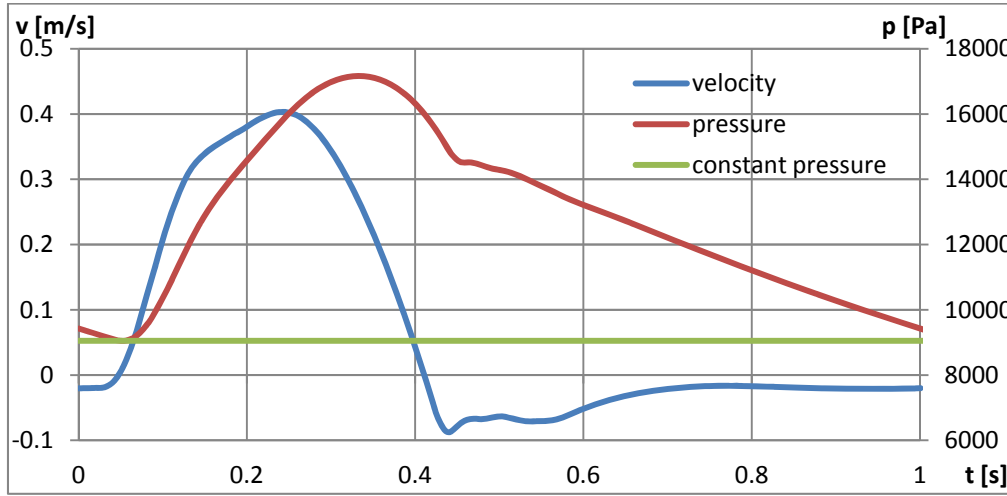


Figure 5: Pressure and mean velocity signals at the inlet and outlet boundaries

The prescribed boundary conditions:

- Pulsatile parabolic velocity profile as opening boundary condition at the inlet obtained from the 1D simulation (opening is necessary for the backflows)
- Time-dependent static pressure as opening boundary condition at the outlet obtained from the 1D simulation
- No slip wall
- rigid wall at the initial straight tube
- mesh motion from the mechanical module at the interface, with linearly elastic wall

The data were compared with measurement data [33]. The mean velocity at the inlet boundary condition is negative after ≈ 0.41 s in our case. Although the velocity is small in the second part of the cycle, it was found that it is partially positive by other authors [11, 17, 33]. As a result of to the pumping motion of the wall the volumetric flow rate and thus the mean velocity become partially positive at the outlet in our case. (The same phenomenon can be found in the 1D simulation too.) *Figure 6* shows the modification effect of the wall motion on the volumetric flow rate in the axisymmetric geometry. This important role of the wall motion is lost in the case of the rigid wall simulations. The pressure signal is similar to the boundary condition applied by other authors [11, 28, 33].

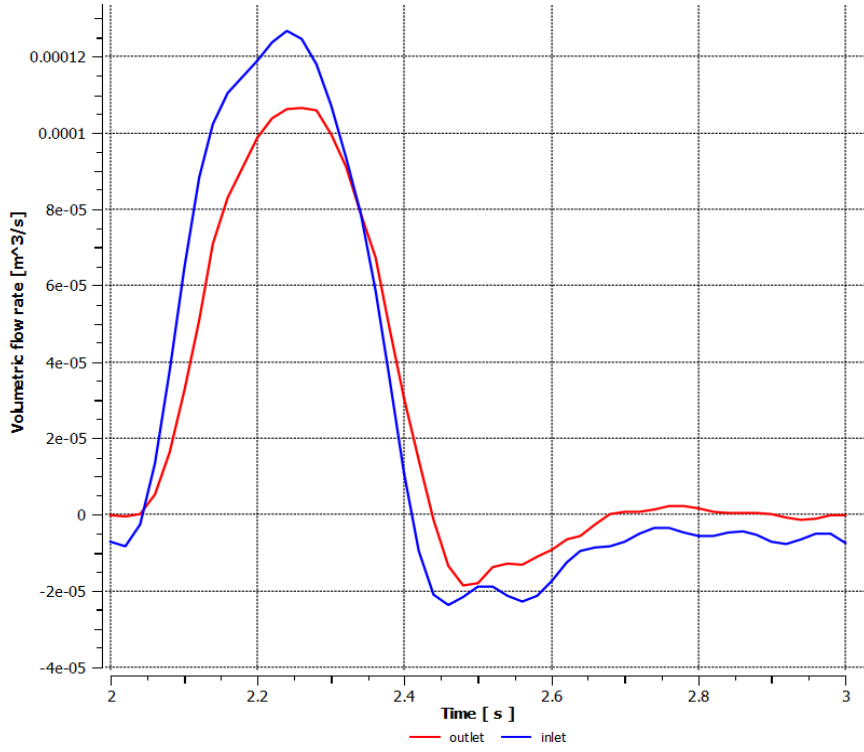


Figure 6: Volumetric flow rate at the inlet and outlet boundaries of the axisymmetric geometry in the third heart cycle

The inlet boundary condition can be checked with the mean and maximum Reynolds number (Re) (7). (v is the spatial mean value of the velocity at the inlet.) Reynolds number is a dimensionless number, the ratio of the inertial and viscous forces. $Re_{\text{mean}} \approx 525$ and $Re_{\text{max}} \approx 2823$. The order of magnitude of the mean and maximum Reynolds number is around the measurement data [33].

$$Re = \frac{Dvp}{\mu} \quad (7)$$

The outlet boundary condition can be checked with the performed measurement. (See below in chapter 9.) The systolic and diastolic pressure at the outlet was $p_{\text{dia BC}} = 68 \text{ mmHg}$, $p_{\text{sys BC}} = 129 \text{ mmHg}$. Equation (13) shows the expression of the mean pressure during a heart cycle. In this case $p_{\text{mean BC}} = 88 \text{ mmHg}$. The characteristic pressure values (mean and diastolic pressure) are really close to the measured data.

The Matlab model has not been validated yet quantitatively with patients. The validation is in progress, but it is a really a complicated task, because pressure and velocity data are indispensable in several points of the systemic circulation. The obtained data were compared with artificial systems, checked qualitatively by Bárdossy [29], and it correlates well with the data of the literature.

In our case, the inlet and the outlet cross section, and the initial straight tube was stationary, it did not change during the simulations. Mesh displacement was allowed only at the examined parts of the geometries. The origins of the mesh motion were in all case the changes of the solid-fluid interface.

6.2. Solid

Based on earlier works the solid geometry was fully constrained at both ends of the model [11, 28, 34]. The interface was loaded with the actual transmural pressure, and the wall shear stress in every time step. Environmental pressure was prescribed on the outer wall surface.

8. Validation and results

8.1. Validation, checking

Before we start to talk about the results of a simulation, we cannot forget to speak about the validity of the solutions. We can check the exactness of the simulations with some obvious methods.

	number of elements in the circumferential direction	number of elements	height of the first layer by the wall
mesh x 0.5	64	208680	0.1 mm
basic mesh	80	401280	0.05 mm
mesh x 2	100	935040	0.025 mm

Table 3: Parameters of the meshes for the mesh convergence study

The mesh convergence is one of the basic requirements. The convergence was checked separately on the fluid and solid side and stationary simulations were accomplished only with the axisymmetric geometry to save time. The load on the wall was the systolic pressure, the number of the radial elements was increased from two to five, and the results were compared. The inlet and the outlet boundary conditions were the pressure and velocity value at the highest velocity phase of the cycle on the fluid side, furthermore it was a rigid wall simulation. The element number of the fluid mesh was increased and decreased too. *Table 3* includes the parameters of the used meshes. The distribution of the higher priority parameters are listed in *Figure 7, 8, and 9*.

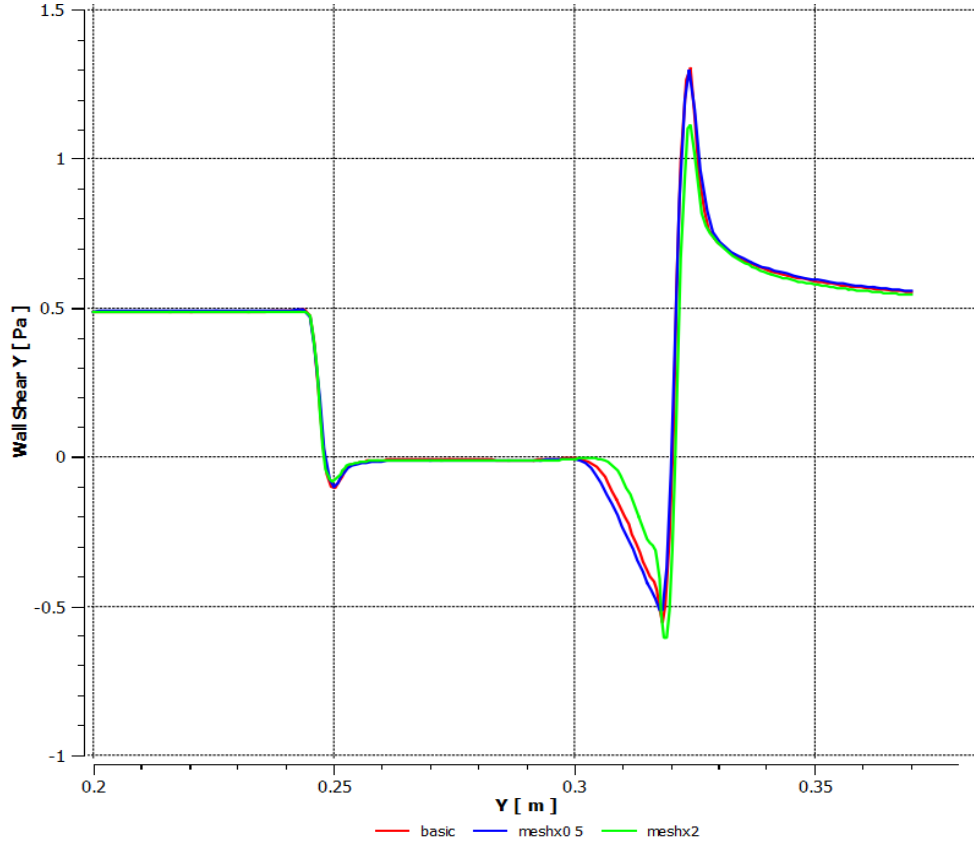


Figure 7: Y component of the wall shear stress as the result of the mesh convergence study along the curve indicated in Appendix 1

Pressure and velocity distribution was examined inside the bulge but the most important parameters in our case are the displacement, the wall shear stress, and the velocity. The pressure distributions show a similarly good agreement for the different meshes as the velocity, and they demonstrate that the number of elements in the investigated range inside the fluid domain has no significant effect on the field variables. In contrast, the thickness of the first layer influences the magnitude of the wall shear stress peaks relatively strongly. It looks as if it should be decreased in later works. The relative error by the positive peak (at the neck of the aneurysm) is $\approx -17\%$ between the basic and the refined mesh. There is a large error before the positive peak, where a vortex can be detected in the fluid domain. The relative error by the negative peak is $\approx 8.5\%$ between the basic and the refined mesh. *Figure 9* proves that two radial layers are sufficient on the solid side.

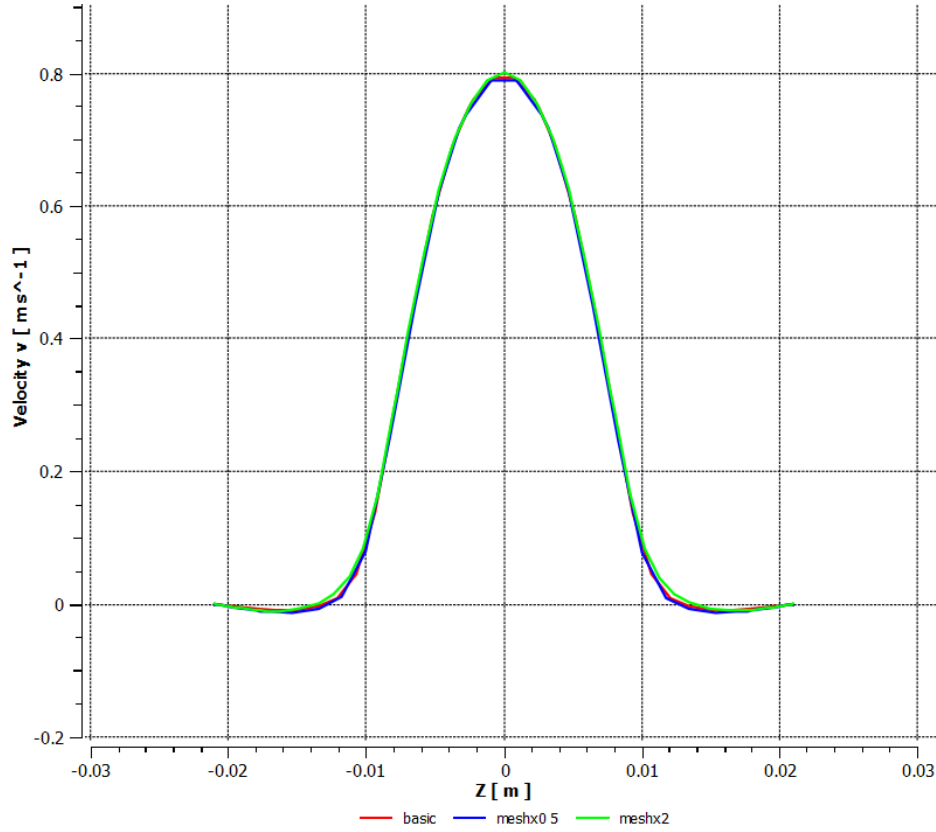


Figure 8: Y component of the wall velocity as the result of the mesh convergence study along a radial line in the middle of the axisymmetric geometry

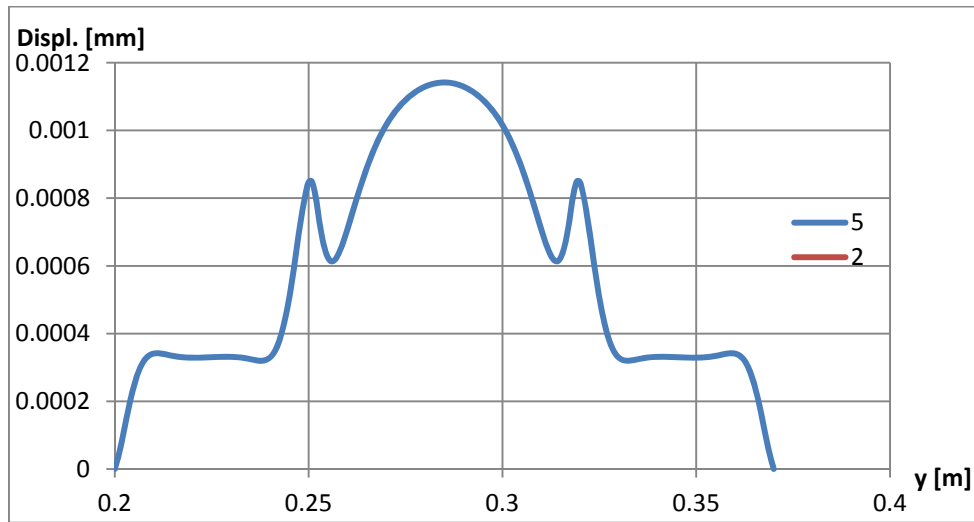


Figure 9: Total mesh displacement as the result of the mesh convergence study along the curve indicated in Appendix 1 (the lines are so close that they cannot be distinguished)

A qualitative validation of the fluid domain was possible because two of the used geometries (asymmetric, axisymmetric) were taken over from the literature [8]. Appendix 4 involves figures of our simulations, and of flow visualization from other publication written about measurement techniques on artificial geometries with the same parameters. The

comparison was carried out in the same typical points of the cycle. The vortex structures look similar.

Laplace's law was used to check the Mechanical solver. Equation (8) shows the stress σ caused by the pressure p in the wall of a simple cylindrical tube with an inner diameter D and thickness t .

$$\sigma = \frac{Dp}{2t} = \frac{0.02 \cdot 17168}{2 \cdot 0.00175} = 98.1 \text{ kPa} \quad (8)$$

The relative deformation (ε) was defined in equation (9), where D_d is the inner diameter of the deformed tube.

$$\varepsilon = \frac{D_d - D}{D} \quad (9)$$

From Hooke's law (10) it is possible to obtain the relative deformation thus the deformed diameter (11).

$$\sigma = E\varepsilon \quad (10)$$

$$D_d = \frac{\sigma}{E} D + D = \frac{98100}{2.7 \cdot 10^6} 0.02 + 0.02 = 0.02073 \text{ m} \quad (11)$$

The static simulation of the straight tube gets $\sigma = 92.2 \text{ kPa}$ in the wall ($\approx -6\%$ relative error), and $D_d = 0.02073 \text{ m}$. The mechanical solver can be considered from the deformation point of view highly accurately, and it gives a good approximation for the stresses too.

8.2. Velocity field

The velocity fields were investigated in several planes for comparison to analyze the vortex structures of the flow, and to compare the rigid wall simulations with the FSI simulations. The limitation of this paper does not allow the detailed analysis of all the geometries, thus the axisymmetric geometry will be presented extensively, and some interesting findings will be highlighted from the simulations of other geometries.

The used inlet velocity history can be divided into two characteristic parts in time: accelerate-decelerate part (0-0.44 s), and a decay part (0.44-1 s). The first part lasts approximately until the minimum of the inlet velocity. In the second part the velocity is smaller about one order of magnitude (*Figure 5*).

In the asymmetric geometry a vortex can be seen near the highest velocity phase at the immediate neck of the aneurysm in the x-y plane (*Figure 10*).

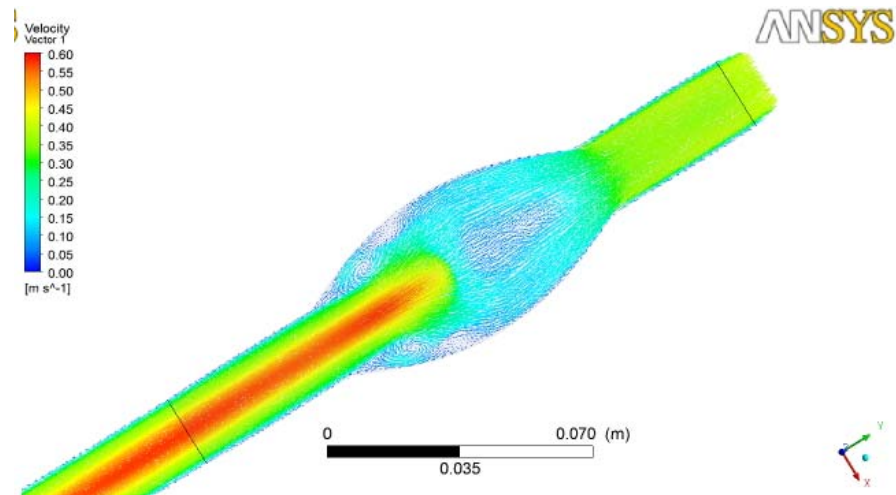


Figure 10: Velocity field in the x-y plane of the asymmetric geometry at 0.3 s

The growth of this vortex is inhibited (Figure 11) by another vortex that separates from the downstream flat wall surface, and moves backwards (Figure 12).

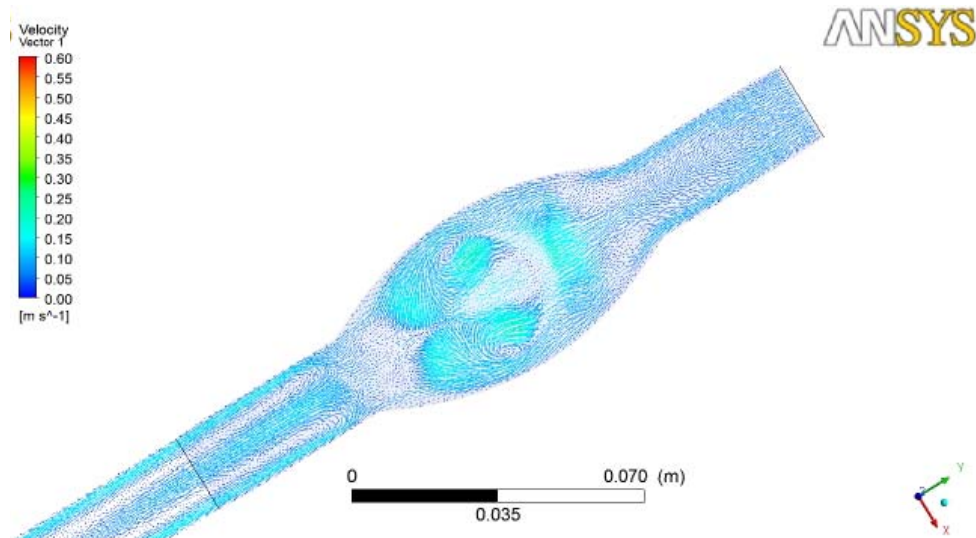


Figure 11: Velocity field in the x-y plane of the asymmetric geometry at 0.6 s

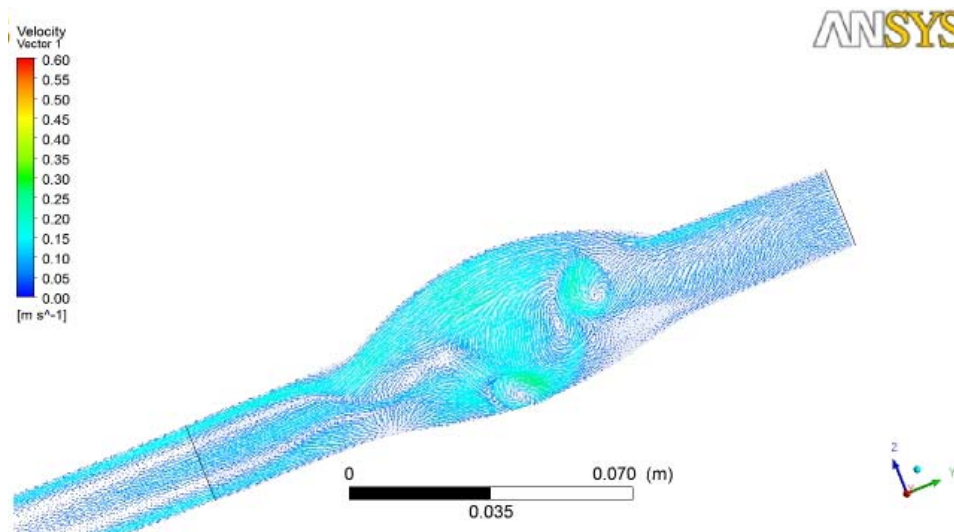


Figure 12: Velocity field in the symmetry y-z of the asymmetric geometry at 0.6 s

Later several smaller vortices are formed (Figure 13). No significant qualitative difference was found between the rigid wall and the FSI simulations from the velocity field point of view, but there are some smaller deviations in the second part of the cycle. In *Appendix 5* in figures like these about the S-tube with bulge geometry, some small differences can be noticed between the vortex structure of the rigid wall and FSI simulations.

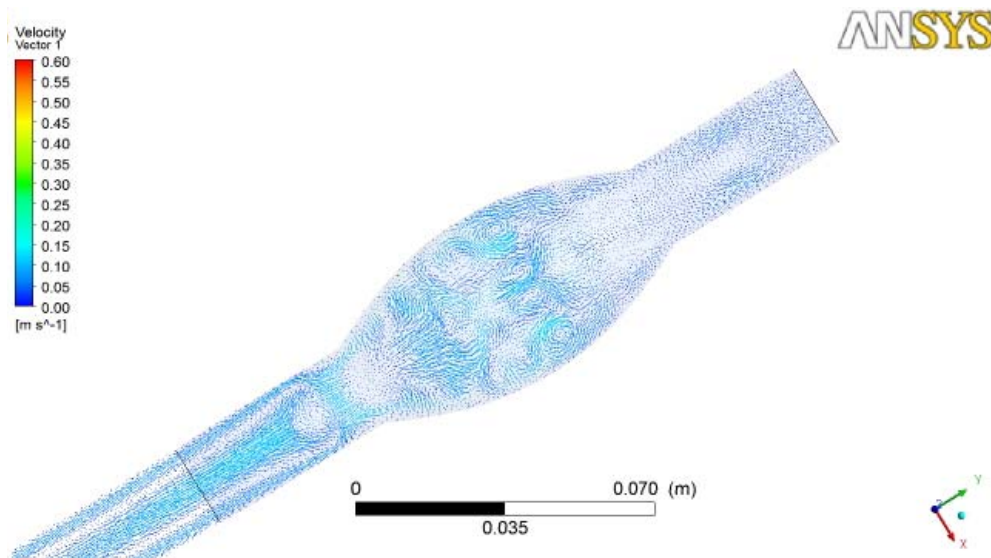


Figure 13: Velocity field in the symmetry plane of the axisymmetric geometry at 0.8 s

It is worth watching *Figure 12* very carefully. There is a small vortex on the flat surface that moves and rotates relatively fast. In fact this flat surface is supported by the spine, and the rupture is happened in this direction every time. Maybe this small fast vortex has a role in the risk of rupture.

We shall learn below that the WSS peak on the S-tube, and S-tube with bulge geometries is nearly twice as high as on the axisymmetric and asymmetric geometries. It is important to investigate the velocity field (*Figure 15*) of the S-tube geometry to understand this remarkable difference. The tortuous geometry causes some separation zones, and these zones narrow the cross section area of the flow. Thus the blood speeds up at these parts. We shall see later that WSS peaks are located on one of these parts of the wall.

Another important remark can be that secondary flow is noticed in the S-tube, and S-tube with bulge geometries (*Figure 14*). It increases WSS perpendicular to the flow direction. These two important differences (separation zones and secondary flow) are probably observed in every prolonged and wrapped old aorta.

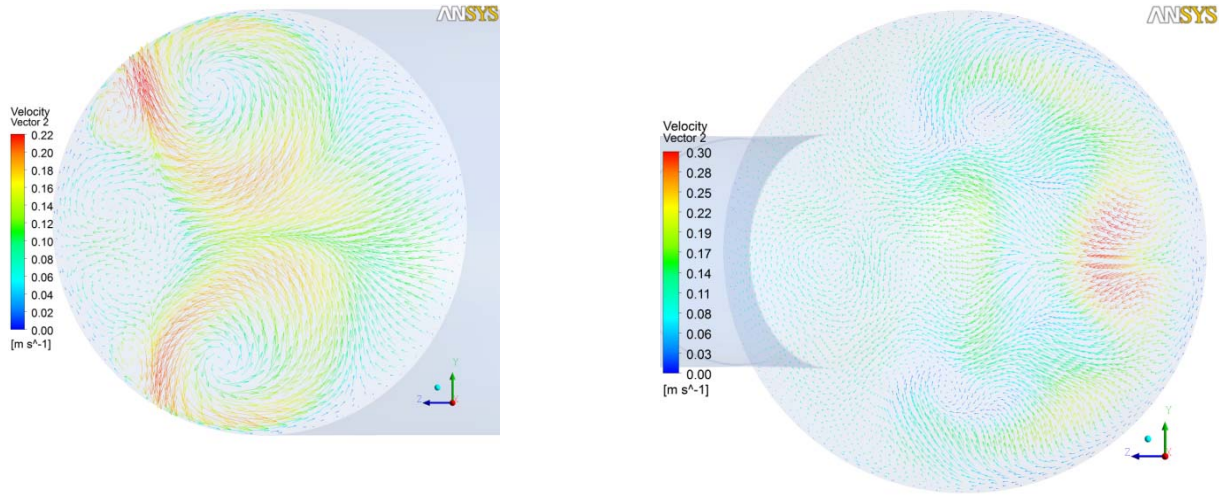


Figure 14: Secondary flows in the S-tube with bulge after the first bend (left) and in the middle of the bulge (right) at 0.6 s

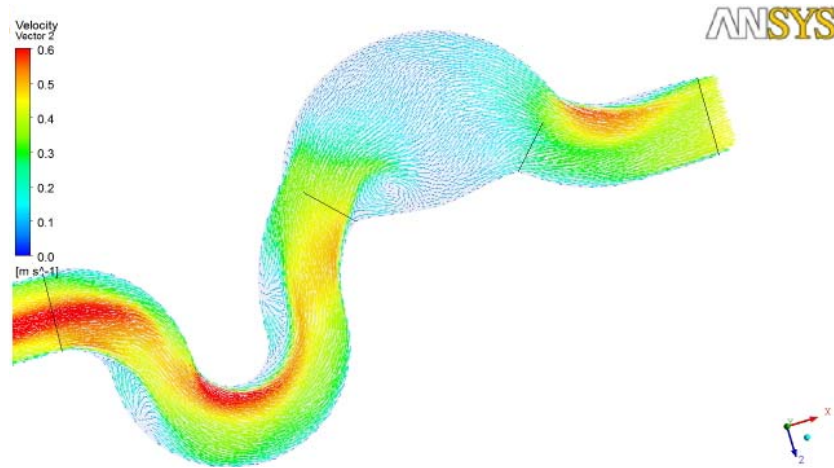


Figure 15: Velocity field in the S-tube with bulge geometry on the plane of the symmetry at 0.24 s

8.3. Wall shear stress on the wall

Equation (12) is the expression of the wall shear stress where n is the wall-normal coordinate and $v_{||}$ is the velocity component parallel to the wall. The component of WSS in the flow direction was examined generally.

$$\tau = \mu \frac{\partial v_{||}}{\partial n} \quad (12)$$

The positive WSS peak is in artificial geometries with bulge at the distal neck of the aneurysm at the higher velocity phase. In S-tube with bulge geometry the WSS peak is at the immediate bend, which has small curvature, and a smaller peak is at the distal neck (Figure 16). It was remarked a negative peak too, which is usually on the wall of the bulge.

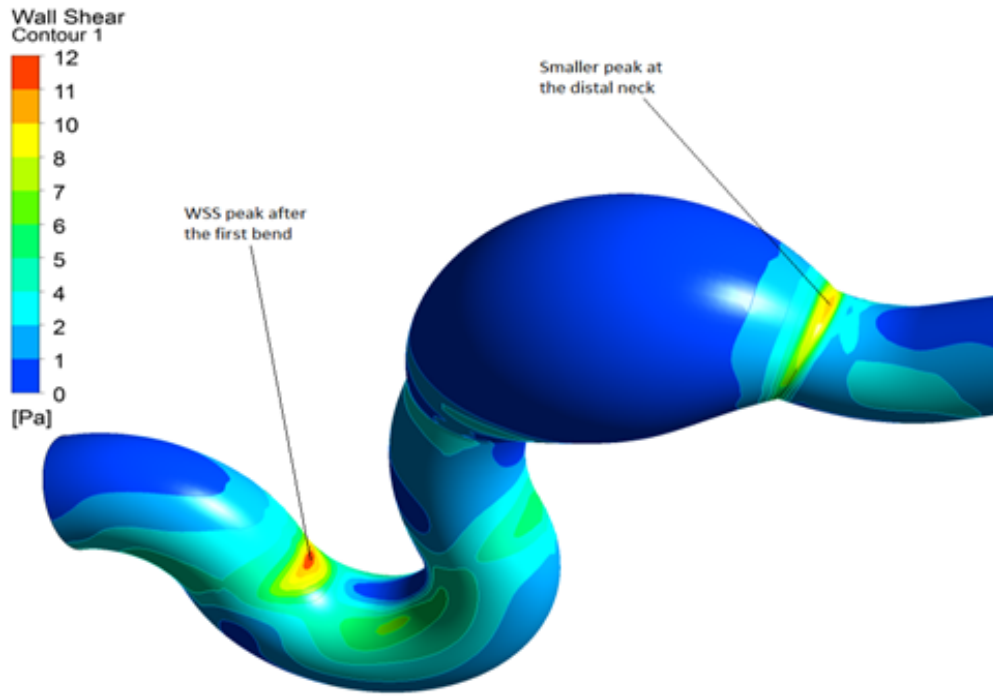


Figure 16: WSS distribution on the wall of the S-tube with bulge geometry at 0.24 s

On the wall of the asymmetric geometry the positive peak is not on the flat surface (Figure 17), but the negative peak is on the flat side (Figure 18). The negative WSS peak is caused by the vortex near the wall in Figure 12. The emergence of this vortex is presented at the immediate neck in Figure 17. This solution suggests that the sign of the WSS can play also role in the diseases of the vessel wall.

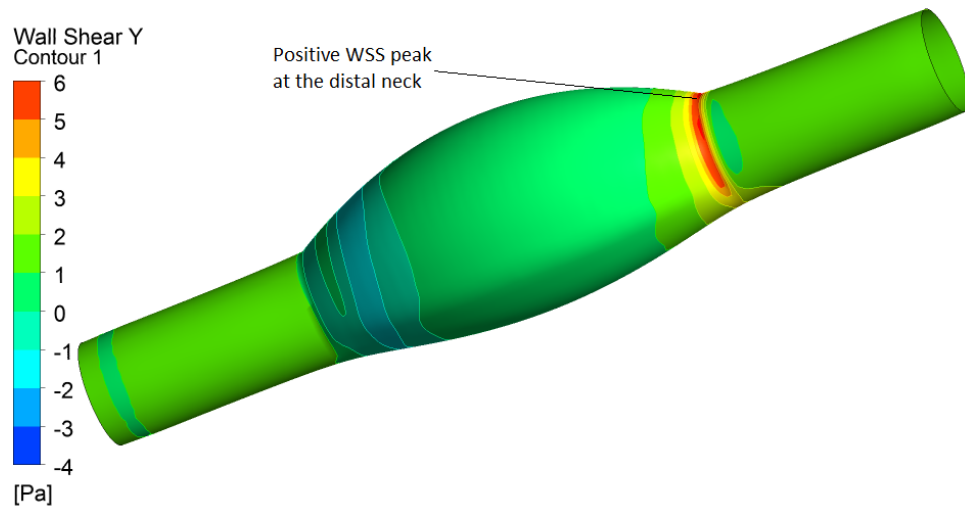


Figure 17: WSS distribution on the wall of the asymmetric geometry at 0.24 s

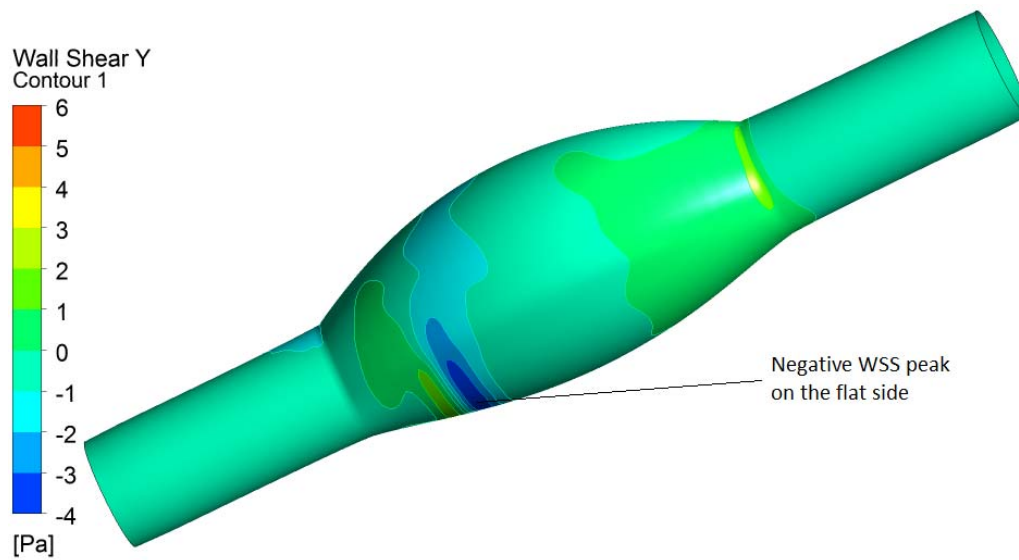


Figure 18: WSS distribution on the wall of the asymmetric geometry at 0.36 s

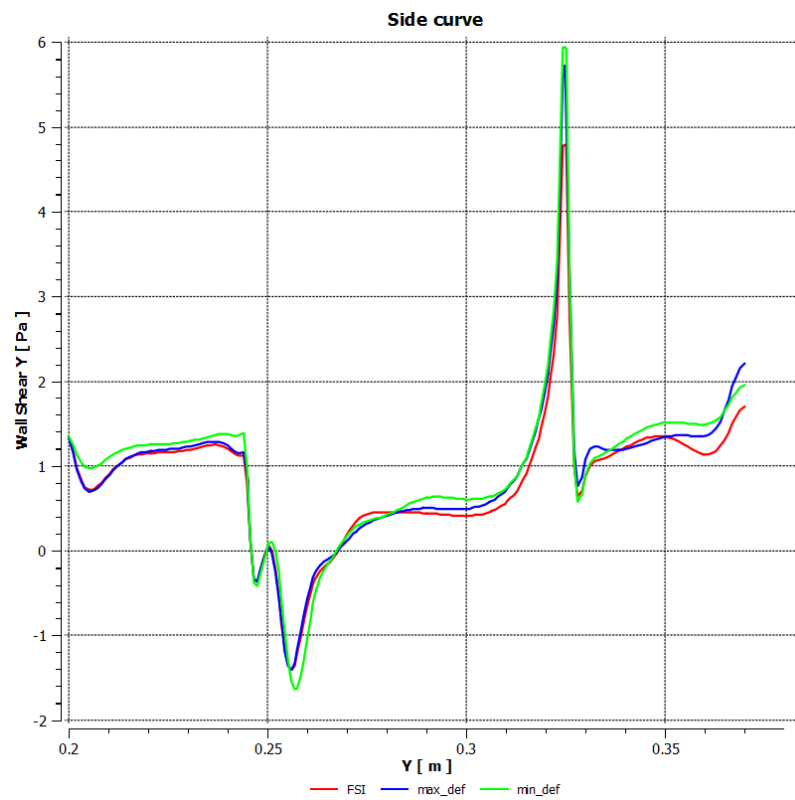


Figure 19: WSS distribution along the side curve on the asymmetric geometry at 0.24

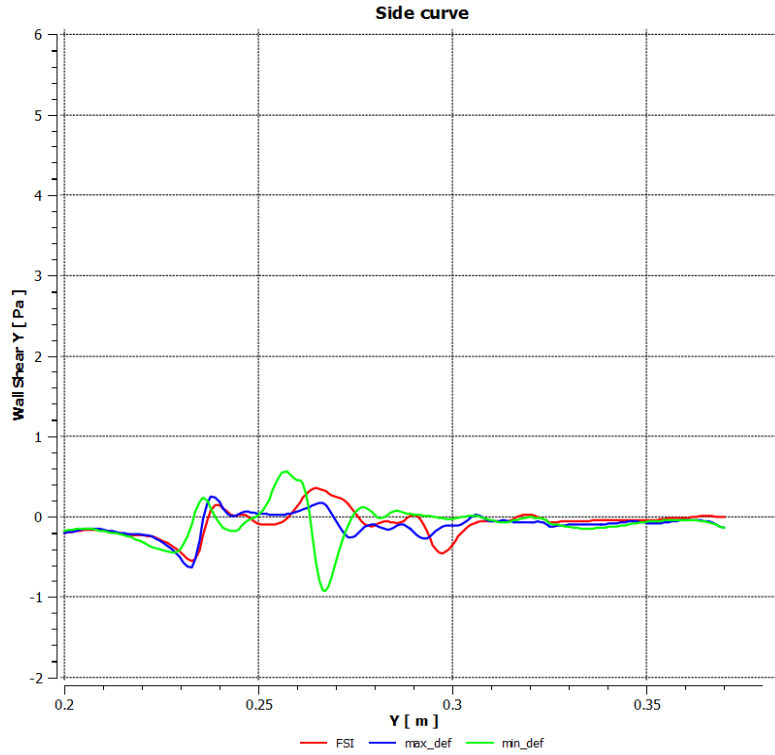


Figure 20: WSS distribution along the side curve on the asymmetric geometry at 0.8

It was found that there are no major qualitative differences between the rigid wall and FSI simulations in the first part of the cycle, but in the second part, the maximum deformed geometry follows the FSI WSS history both qualitatively (*Appendix 6*) and quantitatively better (*Figure 19-20*). The magnitude of both the positive and negative WSS peaks is generally overestimated by the rigid wall simulations. It is obvious in the case of the minimum deformed geometry because there are the smallest cross sectional areas (continuity equation), but it cannot be explained for the maximum deformed geometry. *Table 4* summarizes the solutions from the WSS point of view in the cases of asymmetric and S-tube with bulge geometries.

	FSI [Pa]	Min. def. geom. [Pa]	Max. def. geom. [Pa]	Time [s]	Notes	
positive peak	4.9	6.4	5.8	0.24	at the distal neck	Asym. geometry
negative peak	-3.1	-3.3	-3.9	0.36	on the flat surface	
positive peak	9.7	12.0	11.0	0.24	at the first bend	S-tube with bulge geometry
smaller positive peak	6.1	10.8	9.8	0.24	at the distal neck	
negative peak	-4.8	-4.2	-5.5	0.42	at the immediate neck	

Table 4: The locations and the values of the minimum and maximum WSS

Based on *Table 4*, the rigid wall simulations estimate the value of the WSS peaks relatively wrong (6.5-60 % differences), but *Figure 19-20* shows that they save characteristic of the distribution well. (It is important that an intermediate deformed geometry was also checked in the asymmetric case, and it was not a better approximation.) Based on these solutions it seems that mainly the change of the geometry cause the differences between rigid wall and FSI simulations.

9. Measurement

During my work I had the pleasure to get to know some of the modern non-invasive medical measurement tools, and to carry out an interesting measurement with them. These devices are useful not only for the medical diagnosis, but they can be applied for the validation of the one-, and three-dimensional simulations. The aim of the measurement was to collect data for the validation process, and to examine the behaviour of the vessel wall in the human body.

9.1. Measurement equipments

The measurement was performed with the help of the Institute of Human Physiology and Clinical Experimental Research, and the instruments belong to them. *Figure 21* shows the schematic draw of the setup.

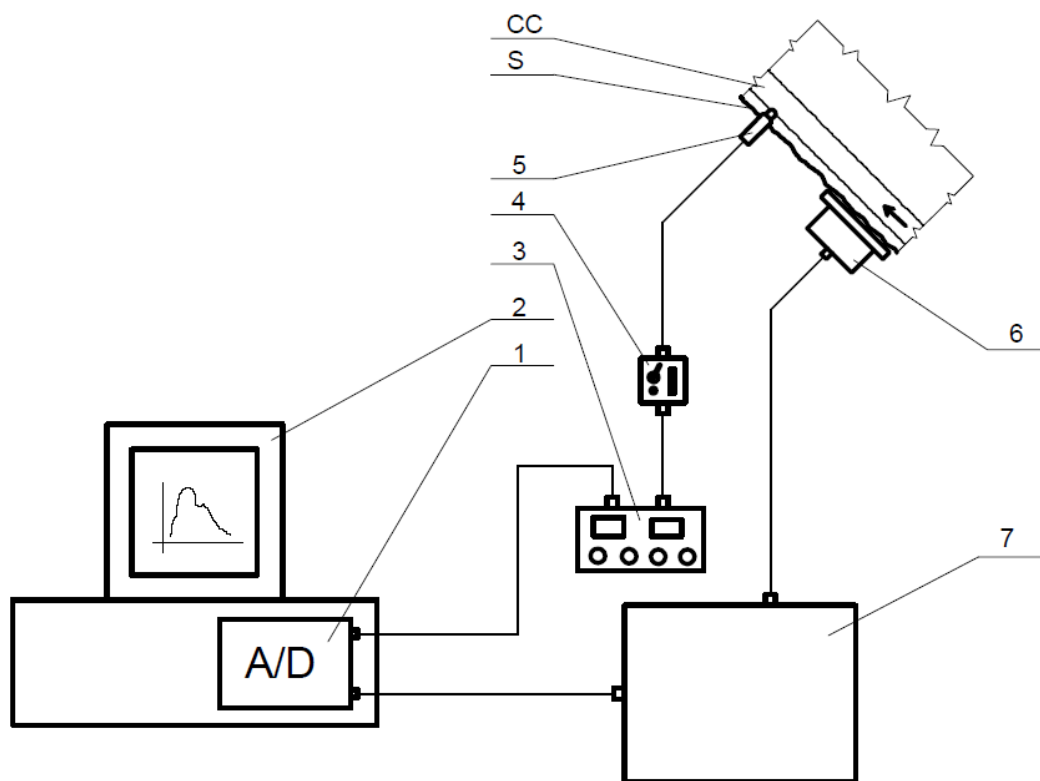


Figure 21: Arrangement of the measurement instruments

In *Figure 21* the arrow indicates the direction of the blood flow. The numbers are:

- 1; Analogue-digital converter (as a card in the computer)
- 2; Computer
- 3; Signal amplifier
- 4; Transducer for the calibration
- 5; Tonometer
- 6; Ultrasound probe
- 7; Ultrasound System (has an own monitor for the ultrasound picture)
- CC Arterie carotis communis of the human (by the neck)
- S Skin of the human proband.

A digital oscillometric blood pressure measurement instrument was also necessary to get the blood pressure in the arteria brachialis (by the arm). *Table 5* includes the list of the used instruments. The instruments are in regular use, thus they are calibrated.

No.	Name	Type
3.	Signal amplifier	product of GOULD Electronics GmbH
4.	Transducer	Millar TC 510
5.	Tonometer	Millar SPT-301
6.	Ultrasound probe	Esaote L10-5
7	Ultrasound system	Esaote Picus Pro
8	Digital oscillometric instrument (with arm cuff)	Collin CBN 7000

Table 5: List of the measurement instruments

The sampling frequency was 30 Hz. The tonometer measures approximately in one cross section of the artery. The ultrasound probe involves several senders and receivers. The computer saves an average diameter along a 40 mm long part of the vessel (40 mm is the length of the probe).

9.2. The theoretical background of the used instruments

Tonometry is applied basically to determine the eye pressure, but it becomes an accepted method to measure the blood pressure. The used tonometer looks like a pen. The medical researcher presses it on the patient's vessel. (The vessel must be close to the skin surface.) The position of the pen is assumed stationary during the measurement. The force acting on the pen can be measured, for example, with a strain gauge. The cross sectional area

of the pen is known, so the pressure can be calculated as the ratio of the force and the area. Strain gauges transform the input displacement signal to voltage for example with a piezoelectric sensor, thus the output voltage is proportional with the blood pressure [3].

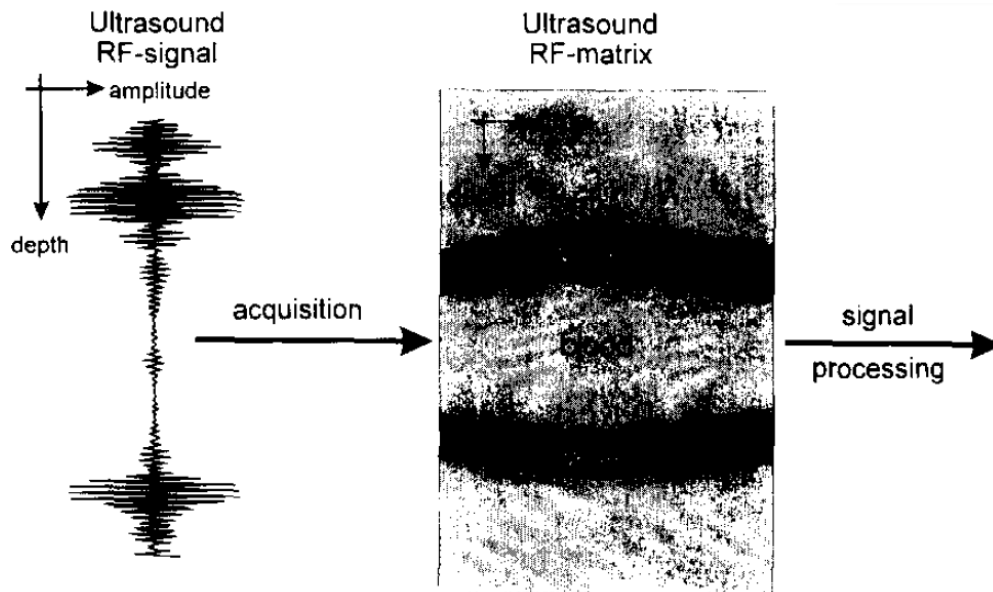


Figure 22: The results based on the ultrasound measurement [35]

The operation of ultrasound diameter and wall thickness measurement like this is described by Brands et al. [35]. The instrument sends ultrasound waves, and the receivers detect the reflected waves. The amplitude of the echo sounds is recorded as the function of the time. The wave propagation velocity of the sound is known in the human body, so the distance can be calculated between the reflective surfaces, as the wave propagation velocity times the elapsed time. With this method the internal diameter and the thickness of the tunica intima and media can be determined. The adventitia does not have a sharp limit like the surface between media and adventitia. A typical result of the ultrasound measurement is shown in Figure 22. The medical researcher must designate the lumen on the monitor at the start of the measurement. The measurement of the blood or tissue velocity is also possible with the used instrument based on Doppler method.

9.3. The measurement process

The measurement is carried out on a lying human. After a short preparation the measurement has the next steps:

1. Blood pressure measurement by the arteria brachialis minimum three times.
2. The neck must be daubed with ultrasound gel.

3. A medical doctor places the ultrasound probe (6) on the neck (S) and finds the arteria carotis communis (CC) with the help of the monitor of the ultrasound system (7). The lumen can also be defined on the monitor.
4. Another medical doctor feels the arteria carotis communis, and pushes the head of the tonometer (5) on it. The tonometer is placed next to the ultrasound probe. This researcher must try to keep the head of the tonometer with constant force.
5. The sampling can be started. It lasts at least for a minute. It is important to pay attention to the simultaneous measurement.
6. It is offered to repeat the steps 3-5 more times.
7. The data can be checked with ART-lab software, and it can be saved in a readable format for other programs.

9.4. Measurement results

The results of the tonometry are influenced by the medical researcher and this effect must be eliminated somehow. It is known that the pressure value of the diastole (p_{dia}) and the mean pressure (p_{mean}) during a heart cycle does not change in the larger arteries from artery to artery (only the systolic peak (p_{sys}) changes). The mean pressure can be expressed with p_{dia} and Δp (13), where $\Delta p = p_{sys} - p_{dia}$. This way the voltage caused by the medical researcher can be separated, because there is a pressure from the mean value of the measured brachialis blood pressure too. In our case from the three pressure measurement by the arm $p_{dia} = 65$ mmHg, $p_{sys} = 120$ mmHg.

$$p_{mean} = p_{dia} + \frac{\Delta p}{3} = 83 \text{ mmHg} \quad (13)$$

These pressure values above help to transform the measured voltage signal to pressure. The output data of the ultrasound instrument is basically the diameter. The results of the measurement are depicted in *Figure 23*. Three complete heart cycles are visible. The appreciable fluctuation of the systolic peak is probably the effect of the breathing. On the one hand the diagram shows that there is no significant time shift between the peaks of the pressure and the diameter signals. On the other hand the effect of the hysteresis is not negligible, and clearly visible. This follows from the fact that from the diastolic to the systolic peak the pressure is perfectly followed by the deformation. It can be observed that the signals separate from each other after the systole, and the diameter signal is delayed relative to the pressure signal.

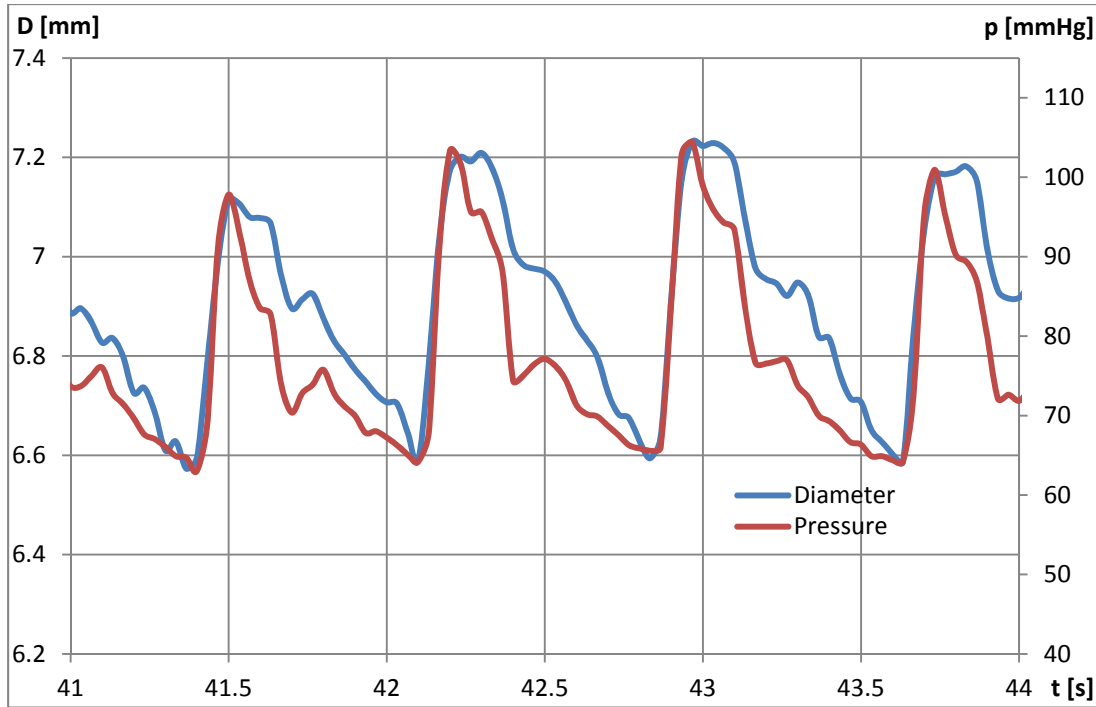


Figure 23: Results of the measurement

10. Conclusions

This work focuses on abdominal aortic aneurysms. The chemical processes were neglected, and only the mechanical interaction between the elastic wall and the fluid was investigated. After the discretization several numerical simulations were run in five typical artificial models with realistic geometrical and physical parameters. It was assumed that the flow is laminar, the blood is Newtonian, and the wall is linearly elastic.

The flow in abdominal aorta cannot be modelled with a porous zone but the one-dimensional simulations can be the new source of the boundary conditions. It is an easy, flexible method to define the physical parameters at the boundaries, but it must be validated with patient specific invasive and non-invasive (like above) measurements before practical implementation.

It was found that mainly the changes of geometries cause the differences between the rigid wall and FSI simulations. Thus, there is a chance to use an easier way to estimate the flow in abdominal aortic aneurysms. The simplification is possible only if the simulations of patient-specific geometry have similar behaviour.

During the post-processing the velocity fields, and the WSS distribution were analyzed on the wall, and the field variables of the minimal and maximal deformed geometries were compared with FSI-simulations. The differences were examined in space and time

qualitatively and quantitatively. It was found that well established rigid wall simulations have tolerable error by comparison the FSI.

The investigation of the velocity field proved very useful. It provides an explanation, how the aging caused strain and shape change of the vessel wall influences the wall shear stress, and maybe the risk of rupture too. In addition it yields ideas, on what types of vortices can be dangerous from the wall shear stress point of view. The results drew attention to the sign of the the wall shear stress in the flow direction.

The vessel wall has a default deformation, consequently a load between 80 and 120 mmHg is not realistic. The separation of the pulsatile pressure is a logical solution, and can be a well applied method in the later works.

The results of the modern non-invasive measurement of the carotis communis show clearly that the vessel wall can be modelled as a viscoelastic material. No time shift was found between the deformation and the pressure signal. Measurements like this can be useful in the future for a validation process of the coupled 3D and 1D simulations.

The development of the solid boundary conditions is necessary in later work to take into consideration the support caused by the spine and the tissues. The application of a viscoelastic material model would be also unique in the fields of the 3D FSI simulations.

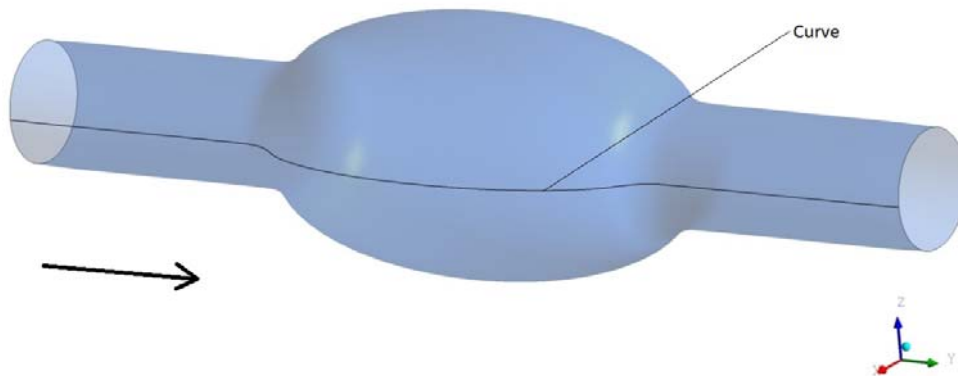
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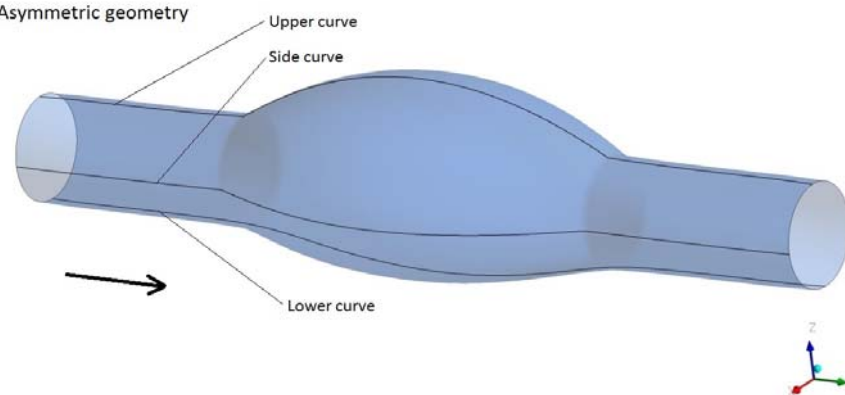
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Appendix 1

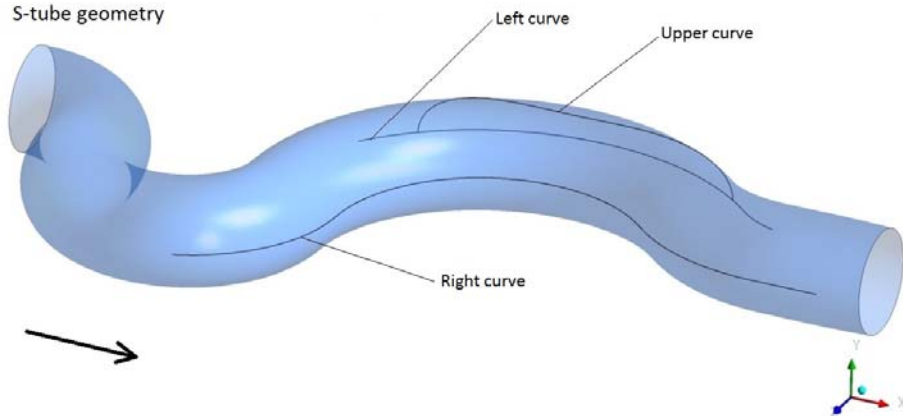
Symmetric geometry



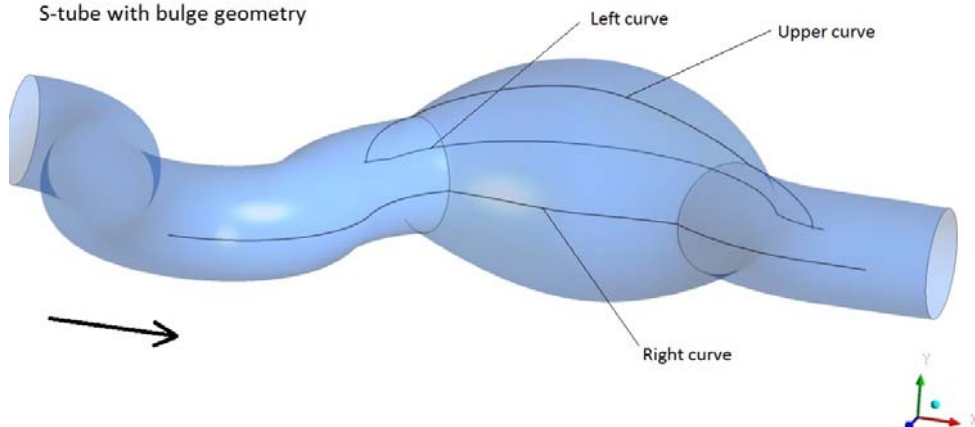
Asymmetric geometry



S-tube geometry

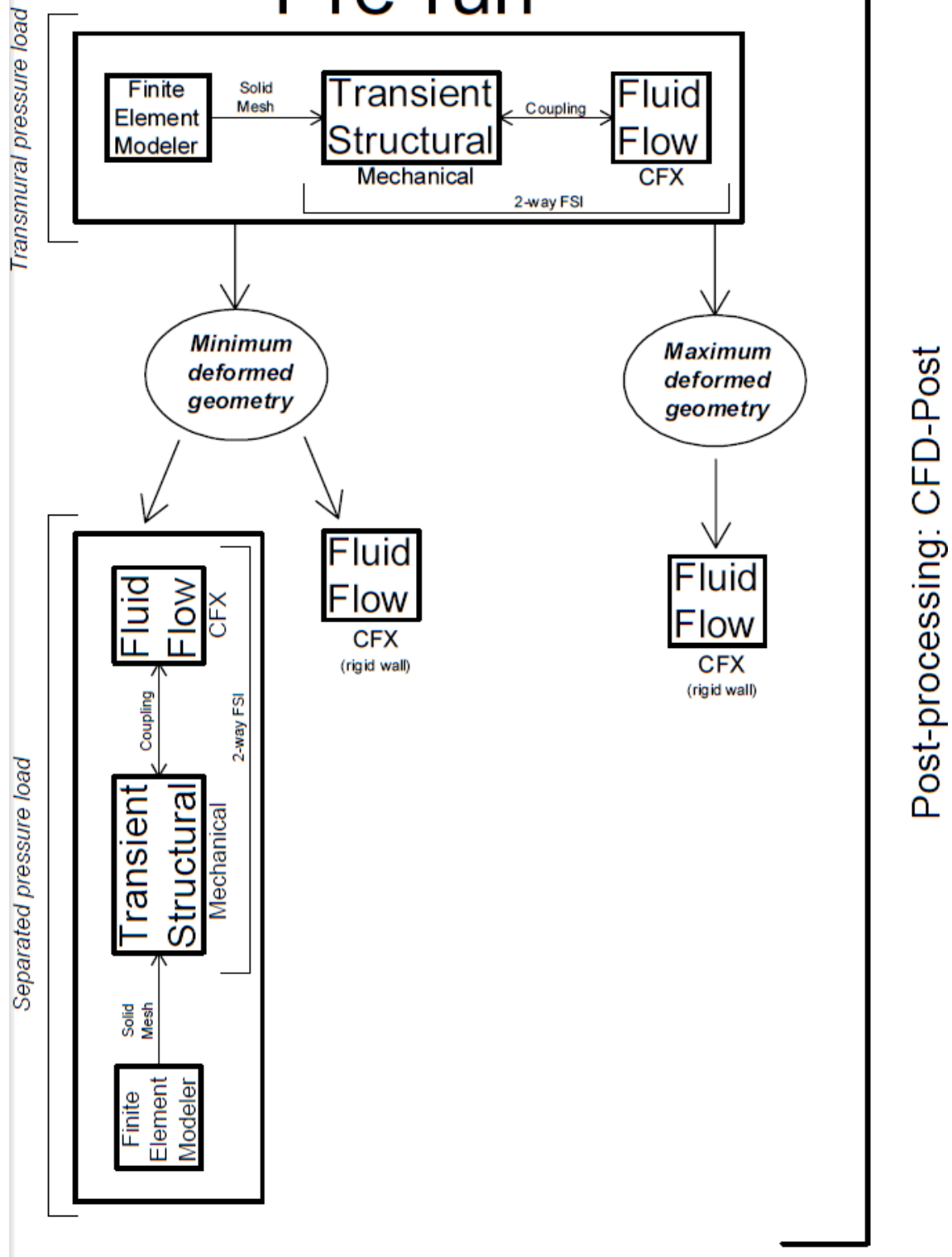


S-tube with bulge geometry

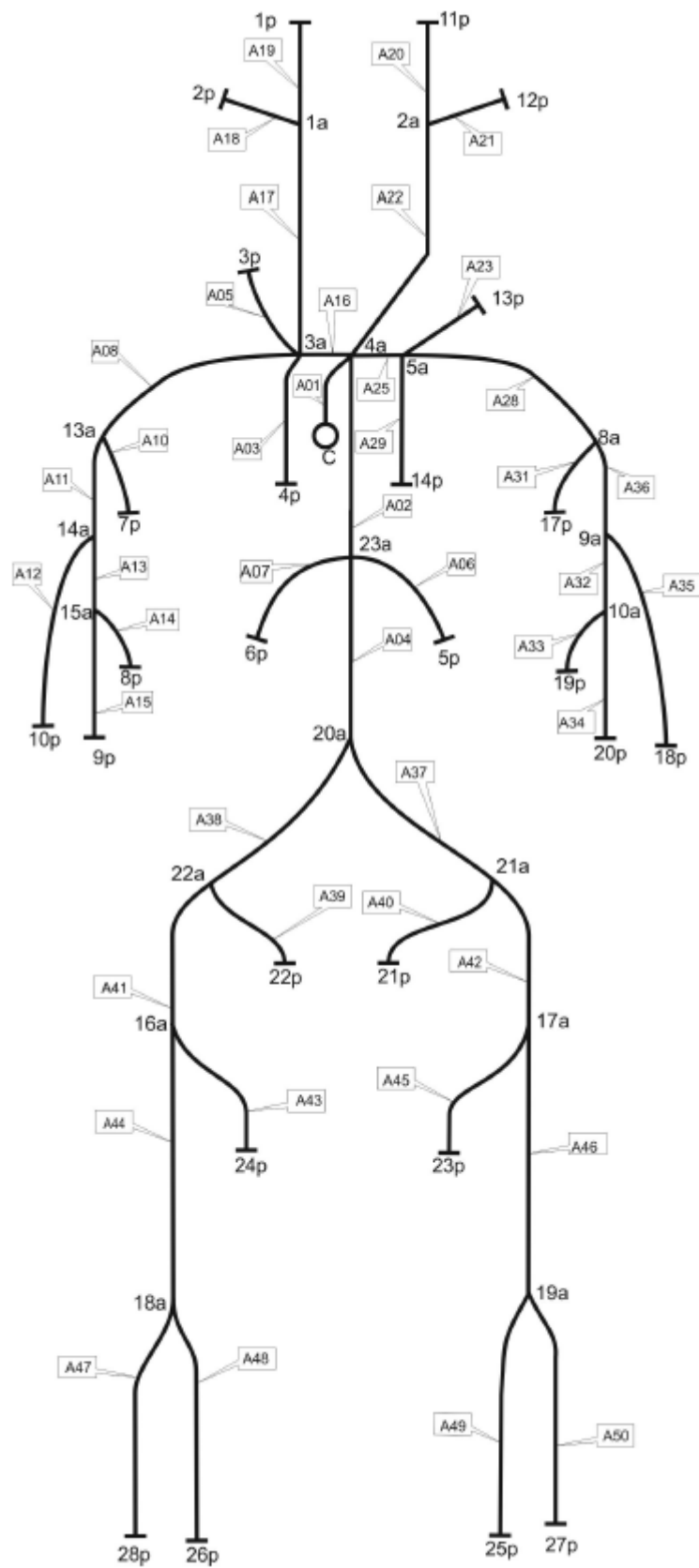


Appendix 2

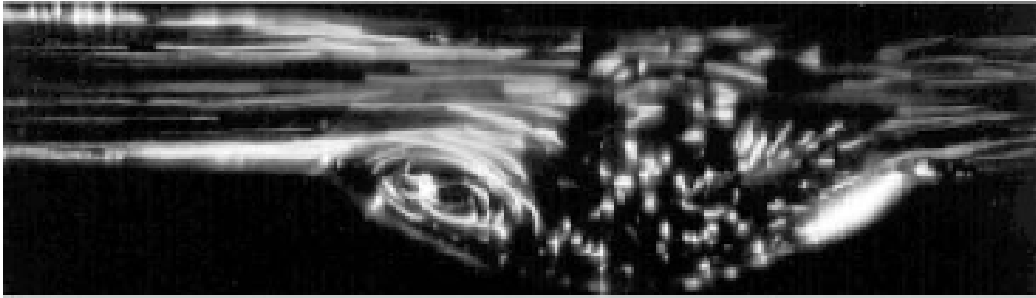
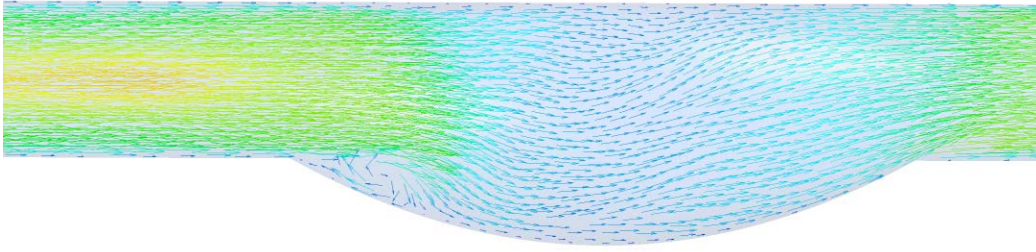
Pre-run



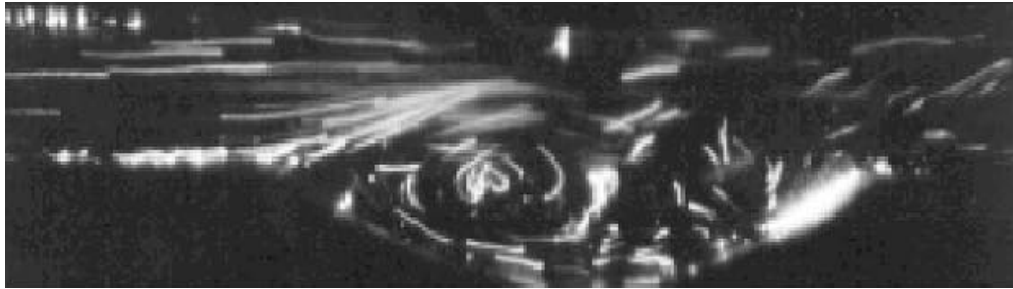
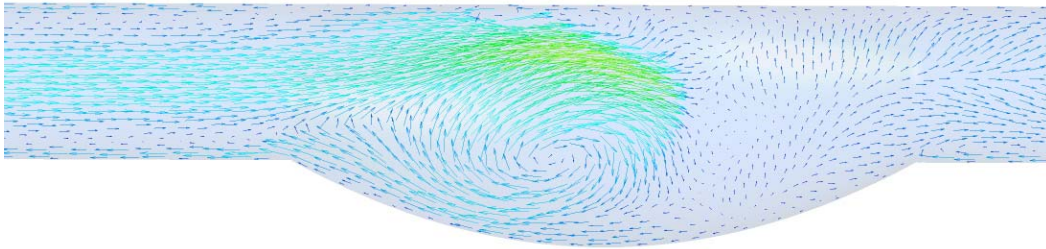
Appendix 3 [29]



Appendix 4

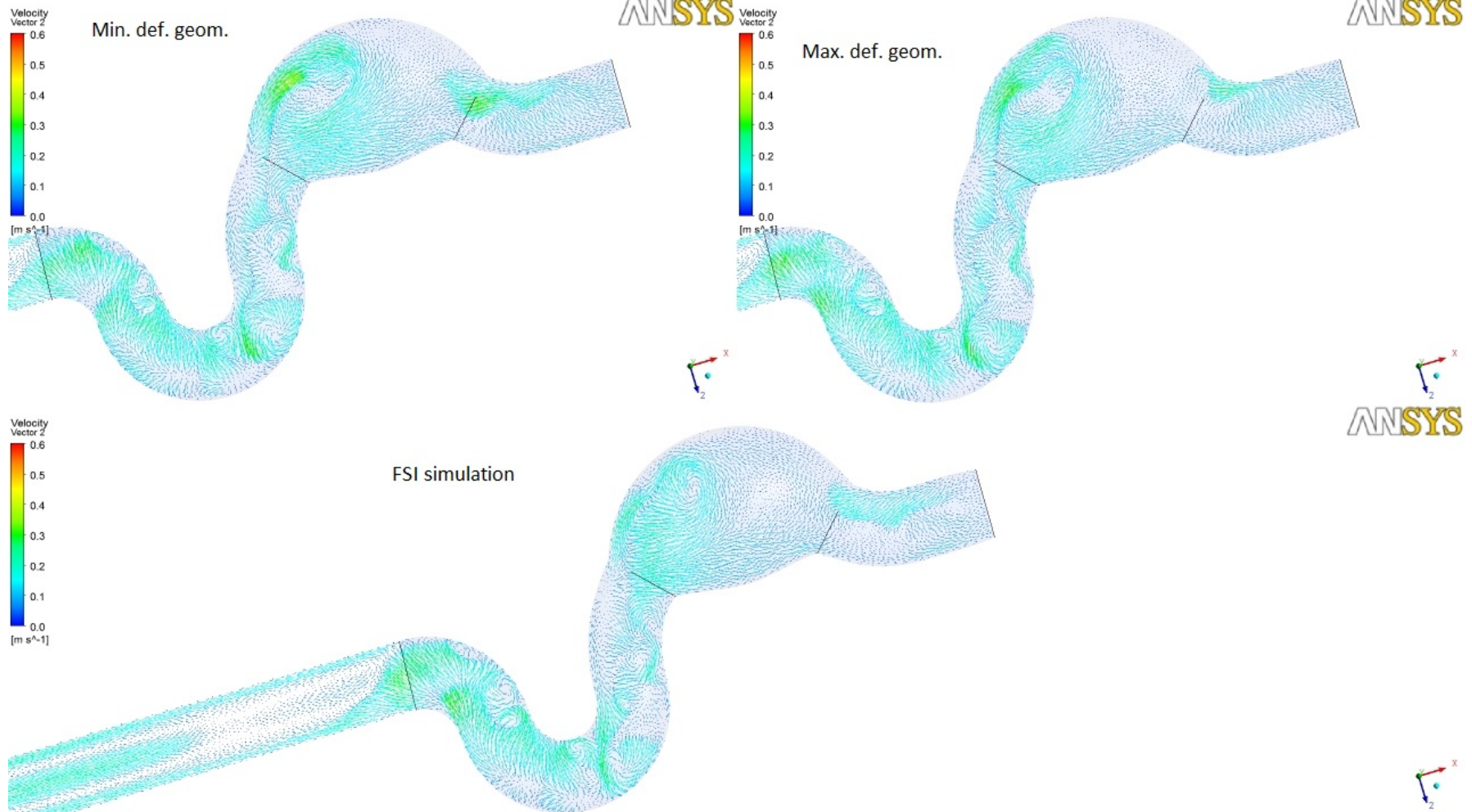


Velocity field in the phase of the highest velocity



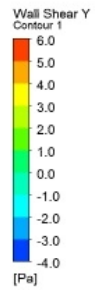
Velocity field after the highest velocity

Appendix 5

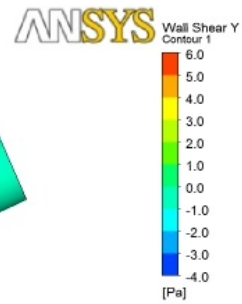
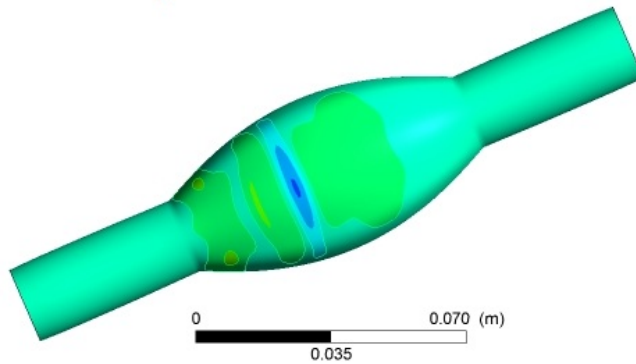


Vortex structures move backward in S-tube with bulge geometry at 0.52 s

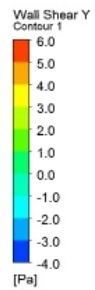
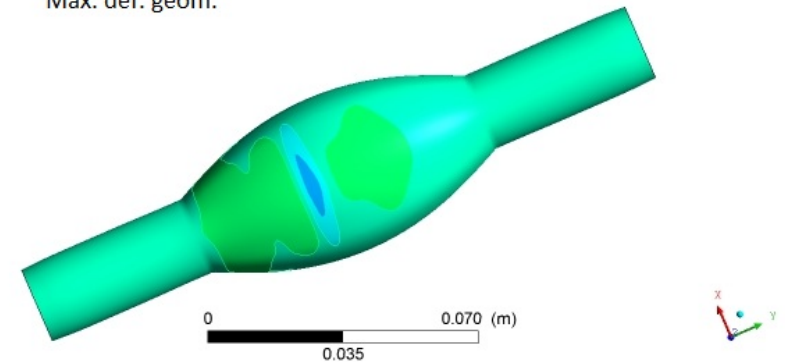
Appendix 6



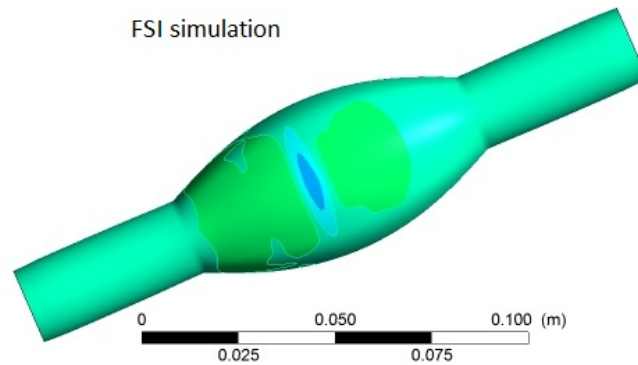
Min. def. geom.



Max. def. geom.



FSI simulation



Wall shear stress distribution on the flat surface of the asymmetric geometry at 0.55 s