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Budapest University of Technology and Economics
Faculty of Mechanical Engineering
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Modelling blood flow in the arteries and veins of the systemic circulation

Thesis booklet of the PhD dissertation

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1 Introduction

The circulatory system is one of the basic systems in the human body. Its main role is to supply the cells with oxygen and nutrients and to remove carbon dioxide and waste material. The cardiovascular system plays important roles in several other functions which are described books on physiology.

The analysis of the cardiovascular system from a biomechanical point of view is called hemodynamics. The circulatory system is divided into two parts: systemic and pulmonary circulation. The pulmonary circulation, however, delivers blood enriched with carbon dioxide to the lungs. From here oxygenated blood is returned to the heart. Systemic circulation is divided into arterial and venous parts. The arterial system is responsible to distribute blood betweent the cells of the body. The venous system collects blood and returns it to the heart.

Nowadays the diseases of the circulatory system are major causes of deaths in the developing and developed countries. According to the mortality database of the WHO in the year 2005 52 percent of deaths in Hungary were caused by diseases of the circulatory system. This indicates the important role of research on the analysis and treatment of these diseases. The reasearch is mainly the task of the physiciab but the mechanisms of blood flow has also to be investigated. In this field fluid mechanics can aid medicine.

At the Budapest University of Technology and Economics, Department of Hydrodynamic Systems hemodynamical research has been going since 2002. Within hemodynamics several, research fields have developed. One of these fields is the modelling the arterial network. Using the basic equations of fluid mechanics (momentum and continuity equations) and the constitutive equations of the vessel wall behaviourl a model of the arterial system can be set up. Computer simulations can be carried out on the arterial network model in order to investigate the various properties of the circulation. Pressure and velocity, wave propagation speed and deformation can be calculated in the arterial branches. One task of the current work is the selection of a proper material model for the vessel wall and its insertion into the arterial model.

Another important topic in medicine is the determination of the central aortic pressure curve. Central aortic pressure is measured using invasive techniques that are difficult and could lead to complications. To bypass this process, the central aortic pressure is usually estimated with blood pressure measurement on the upper arm. Several methods exist for this estimation process, however all of them have an empirical basis. Based on the arterial network model a new method has been developed relying on the basic equations of fluid mechanics and on the selected material model. The method is capable of estimating central aortic blood pressure within the arterial network model and is presented in the current dissertation.

In order to get more information of a vessel segment it has to be investigated separately from the circulatory system. In order to carry out Fluid-Structure Interaction (FSI) calculations, a proper FSI module has been developed. The most important part of this module are the used interpolation methods. In the current work the commonly used conservative interpolation method is investigated. Limitations of this method lead to the development of a new interpolation method. The FSI module is validated using measurement data taken from literature.

Venous circulation differs from the arterial circulation in several details. The average blood pressure in the venous system is less than that in the arterial system. The Young modulus of the vein walls is less, therefore these vessels are more compliant. Venous blood flow is induced by multiple mechanisms. Pressure difference generated by the heart plays a much less important role than in the arterial system. The pulsations of the surrounding muscles and arteries excite the vein walls. With the periodic collapse of the venous vessels, blood flow is induced. The so-called venous valves prevent the blood from backflow. To estimate the amount of blood that flows out of the vessel during contraction, the collapsed vessel shape has to be calculated. Two- and three dimensional models are created for the investigation of the shape. Both methods are presented and compared with each other in the dissertation. Furthermore a measurement rig was designed and built for the practical investigation of the collapse. The main part of the rig is a tank filled with water. The pressure in the tank can be held constant or varied periodically. Several measurements were carried out which are presented and discussed.

2 Viscoelastic material model

The behaviour of the vessel walls can be described properly by a model that is able to treat the time dependence of the deformation and the hysteresis. The so called Stuart model is selected for this purpose. The Stuart model had to be integrated into the Transient Simulator (our in-house code) that is capable of calculating one-dimensional unsteady flows in pipe networks.

Both lumped and distributed parameter models exist in the Transient Simulator. The viscoelastic material model was inserted into the distributed part of the model.

The resulting algorithm was validated using a previous measurement carried out by Till and Hegedűs [3]. A silicone tube was attached to an oil tank with a piston. With the excitation of this piston, pressure signals were generated. Pressure was measured using pressure transducers at the upstream and downstream ends of the tube. The upstream end pressure was set as a boundary condition in the code. Geometrical and material properties were set according to the silicone tube used for the measurement. A separate algorithm was implemented for tuning the parameters of the Stuart model. A genetic algorithm (GA) was used for this purpose. The agreement between the measured and calculated pressure curves is fairly good.

As for the arterial network model a network consisting of 45 viscoelastic branches was created (Figure 1). The calculation results of the network are in concordance with the experiences of physicians (Figure 2).

The artiosclerosis of the iliaca was investigated as a next step. The results of the calculations (see Figure 3) show that reducing the diamater of the stenosis with $72,8\ \%$ the volumetric flow rate decreases by only $22,2\ \%$ which is in concordance with the experiences of physicians.

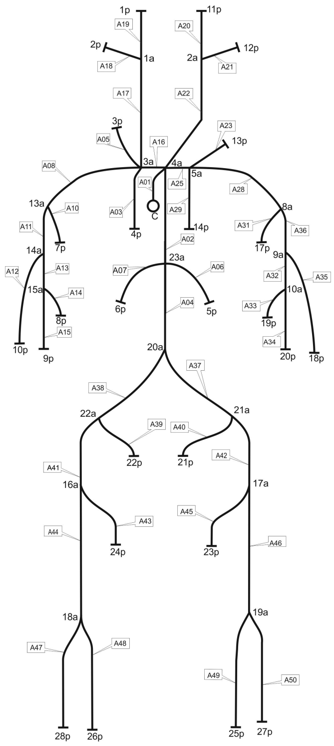
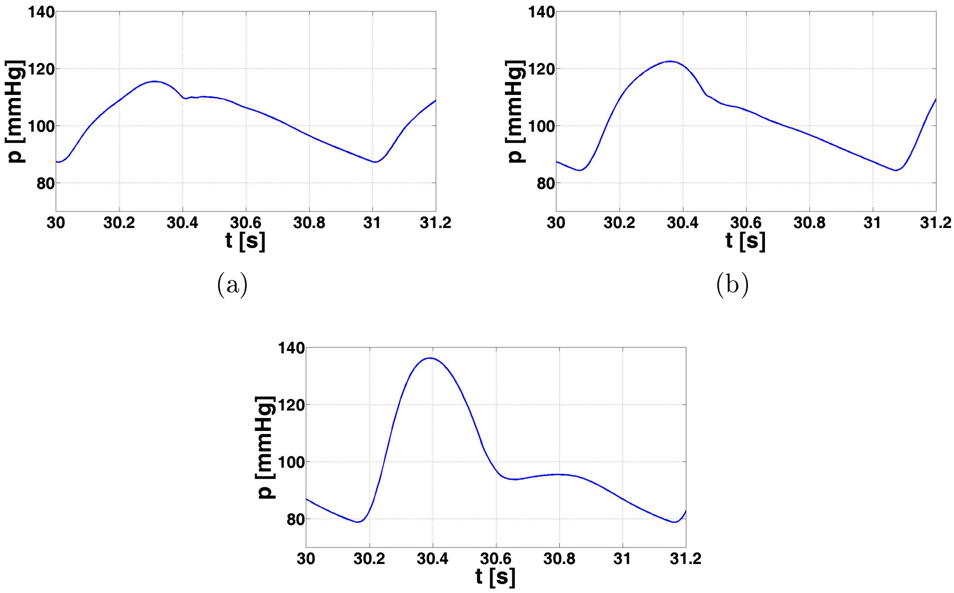


Figure 1: One-dimensional arterial model.

3 Calculation of the central aortic pressure ("backward" calculation)

A novel method for the estimation of the central aortic pressure curve was developed. The characteristic equations of the "forward" calculation cannot be used in this case since boundary conditions are only known at the downstream end of the selected vessel. The method of characteristics had to be generalized in order to carry out this kind of calculations. The method is capable to calculate pressure and velocity curves vessel by vessel in the arterial network model. To carry out a "backward" calculation pressure and velocity curves at the downstream end of the selected tube have to be known. However in practice usually only the former is measured (blood pressure measurement). Therefore another method had to be implemented to estimate the missing velocity curve.

"Backward" calculation was tested with the following method. A simple "forward" calculation was carried out in the arterial network as a first step. Therefore pressure and velocity curves were calculated for every arterial branch. Node $4a$ was selected. Only the pressure curve belonging to this node was stored, all other results were deleted. The central aortic pressure curve had to be estimated out of this single pressure curve. "Backward" calculation was carried out. The resulting estimation of the central aortic pressure curve shows good agreement with the central aortic pressure curve from the "forward" calculation (Figure 4).



(a) (b) (c)

Figure 2: Calculated blood pressure curves in the arterial network model in branches $A01$, $A37$ and $A50$. One can observe the increase of the systolic pressure and the slight decrease of diastolic pressure along the branches with direction to the peripheries.

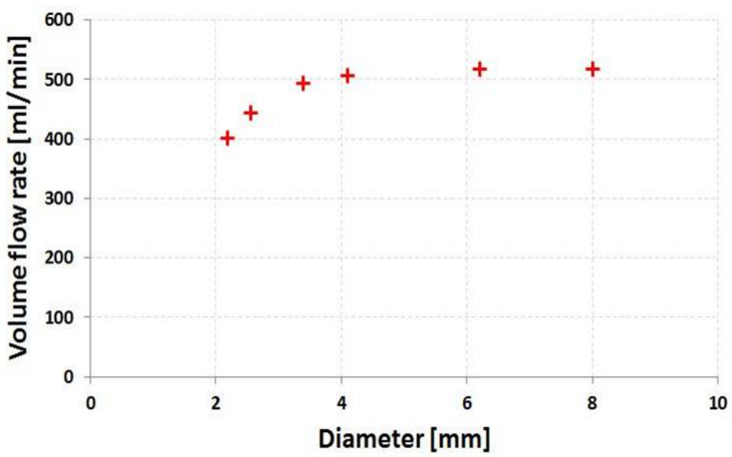


Figure 3: Volumetric flow rate as a function of the stenosis diamater in the iliaca.

4 Modelling collapsible vessels

With the Transient Simulator one-dimensional flow can be calculated properly in case of positive transmural pressures. The algorithm assumes that the expansion of the vessels is concentric. In case of positive transmural pressures this is a suitable assumption. However the vessels collapse if negative transmural pressures occur, in that case the previous

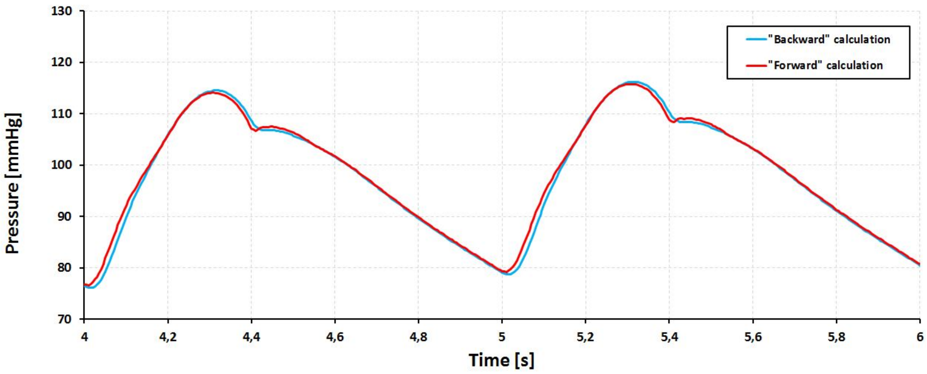
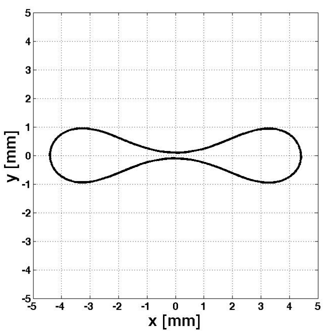


Figure 4: Pressure curves calculated with "forward" calculation and "backward" calculation from node $4a$ (brachial artery).

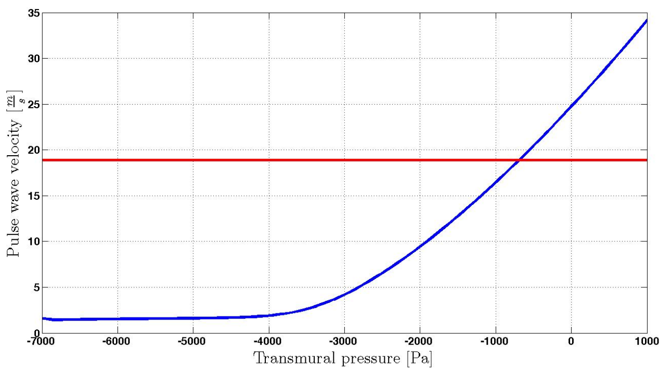
assumption is inaccurate.

The two-dimensional mathematical model of the collapsing vessels was created according to the work of Kresch and Noordergraaf [1]. Using the model of the authors the shape of the vessel cross-section can be calculated. Furthermore the pulse wave velocity can be estimated.

A vessel section was defined for testing calculations. Geometrical and material data are within range of the venous vessels: the Young modulus is set to $E = 5MPa$, the vessel diameter is 7 mm and the wall thickness $0,5$ mm. The initial shape was selected to be elliptical. The transmural pressure was reduced stepwise from 1000 Pa to -7000 Pa. The resulting shape at the pressure minimum is shown in Figure 5(a).



(a)



(b)

Figure 5: Figure 5(a): Collapsed cross-section (Transmural pressure set to -7000 Pa). Figure 5(b): Pulse wave propagation as a function of transmural pressure. Blue curve: according to the Young forumla. Red curve: constant pulse wave propagation speed obtained from the Moens-Korteweg formula.

Using the Young formula the pulse wave propagation speed can be estimated as a function of the transmural pressure. This graph is shown in Figure 5(b). The constant pulse wave propagation speed after Moens-Korteweg is also illustrated as a comparison.

Three-dimensional FSI calculations were carried out in order to verfiy the results of the two-dimensional model. The Ansys® Multiphysics code was used for this purpose. A vessel section similar to the one used for the two-dimensional calculations was created.

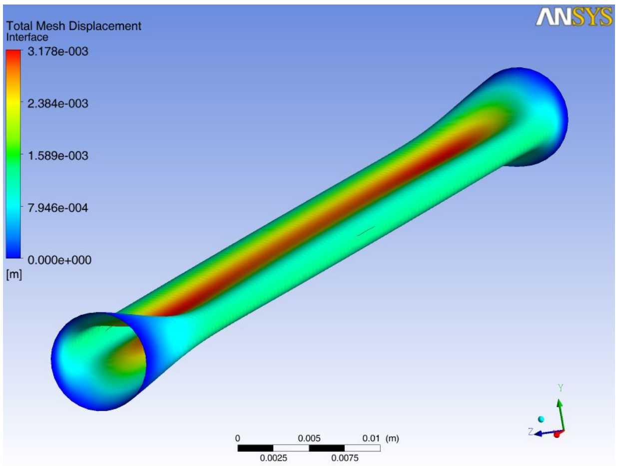


Figure 6: Results of the FSI simulation - deformation of the vessel section (Transmural pressure set to $-5200\ Pa$).

The area of the vessel cross-section was calculated at the midpoint of the axis. This area was compared to the results of the two-dimensional calculations (Figure 7).

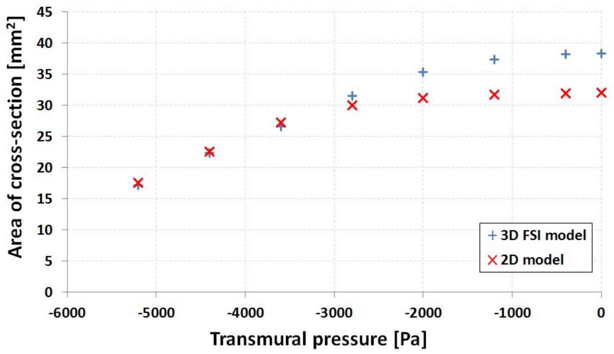
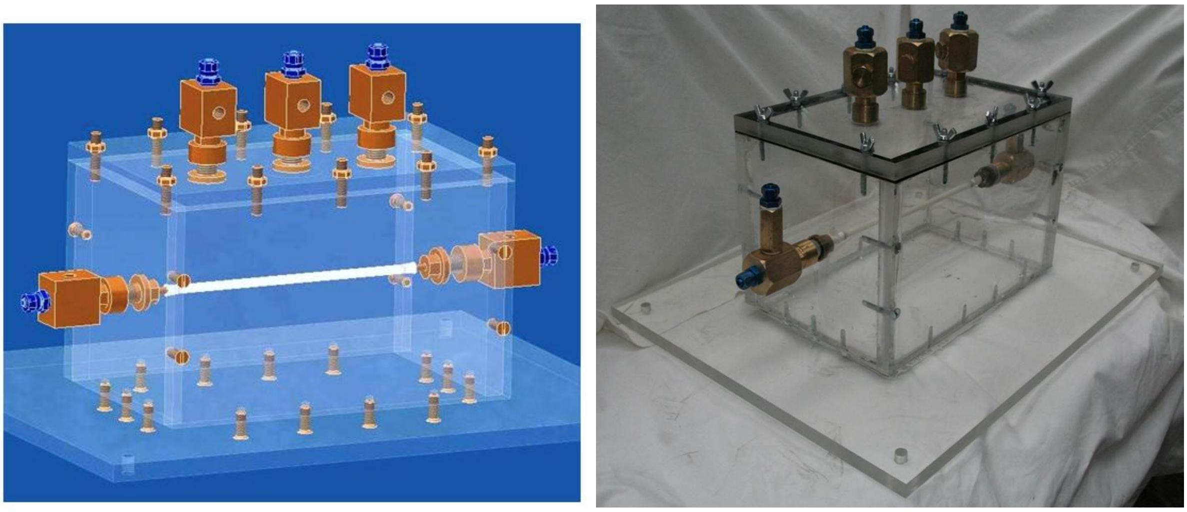


Figure 7: Area of the collapsed cross-section as a function of the transmural pressure. Comparison of the two-dimensional mathematical model (red marker) and the FSI simulation (blue marker).

A measurement rig was designed for the experimental investigation of collapsible tubes. The main component of the rig is a water tank. The pressure inside the tank can either



(a) (b)

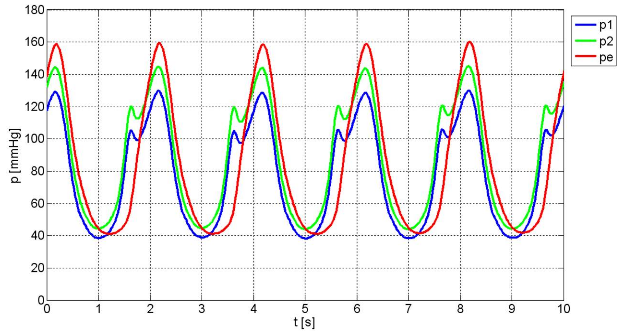
Figure 8: The 3D model (8(a)) and the photo (8(b)) of the measurement rig.

be held constant or varied periodically. Therefore the calf muscle pump can be modeled with the measurement rig.

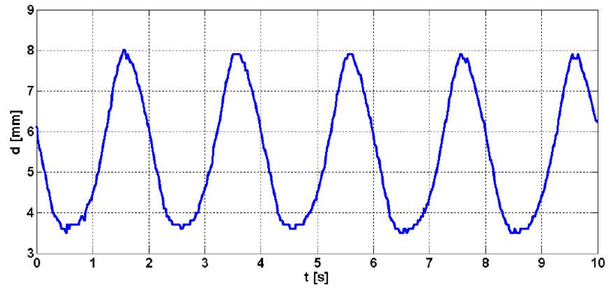
As a first step the funcionality of the measurement rig was tested. Also a validation was carried out while the pressure in the tank was held constant. As a next step the tank pressure was varied periodically using a so called pressure signal generator. The pressure was measured in several points of the system using pressure transducers. The periodical collapse of the tube was recorded with a high frequency camera. Unsteady volumetric flow was also measured using an experimental device.

In Figure 9(a) the measured tank pressure (p_e) and the upstream (p_1) and downstream end (p_2) pressures are presented. In Figure 9(b) the main dimension of the collapsing tube section is shown.

The measured volumetric flow rate curve is similar to the volumetric flow rate curve of a volumetric pump.



(a)



(b)

Figure 9: Tank pressure (pe); upstream ($p1$) and downstream ($p2$) end pressures. Main dimension of the collapsing tube (9(b)).

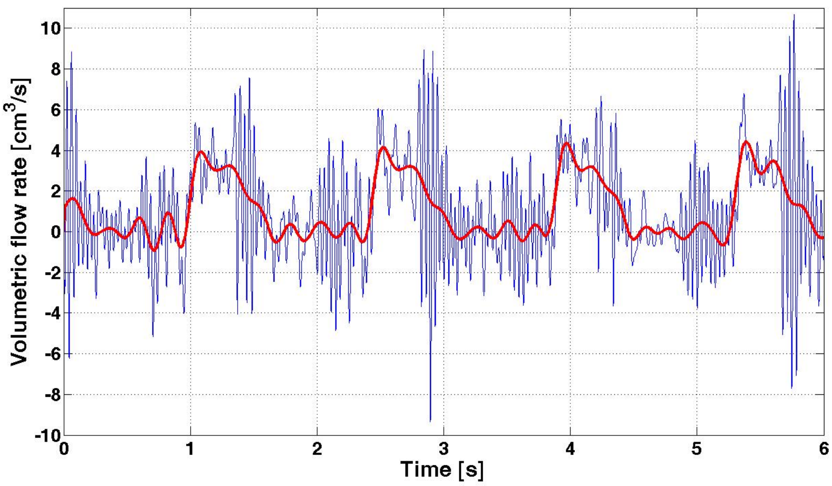


Figure 10: Volumetric flow rate as a function of time. The blue curve represents the measured signal, the red curve is the result after the application of a low-pass filter.

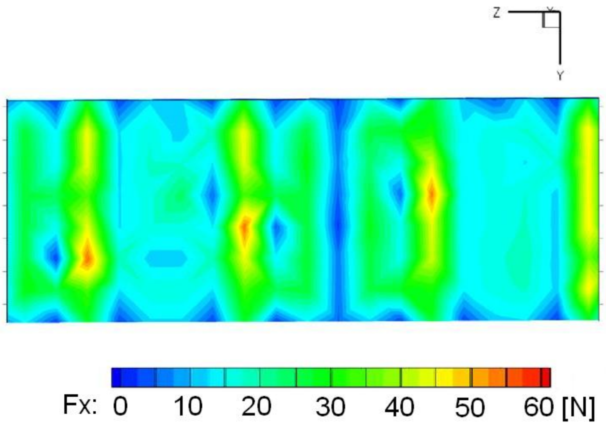


Figure 11: Uneven force distribution using conservative interpolation.

5 FSI calculations

The main task was to attach an existing in-house CFD solver to a FEM code. The work was carried out at the TU Karlsruhe (KIT), Fachgebiet Strömungsmaschinen.

The so called conservative interpolation method was implemented for transferring the fluid forces to the structure. The main advantage of this method is - as the name shows - that it ensures conservitivity i.e. the sum of the forces on the fluid side equals the sum of the transferred forces on the structural side. Good results can be achieved if the CFD mesh is finer than the structural mesh. In the opposite case the interpolation method showed problems in our test cases. A uniform force distribution was created on the fulid side of the contact interface. However in the latter case (fine FEM mesh, coarse CFD mesh) the resulting force distribution on the structure is uneven (Figure 11).

Therefore a new interpolation method had to be developed. This method takes the finer mesh as a basis for force transfer - in this case the FEM mesh is the basis. Figure 12 shows the force distribution on the structure in case of this new interpolation method.

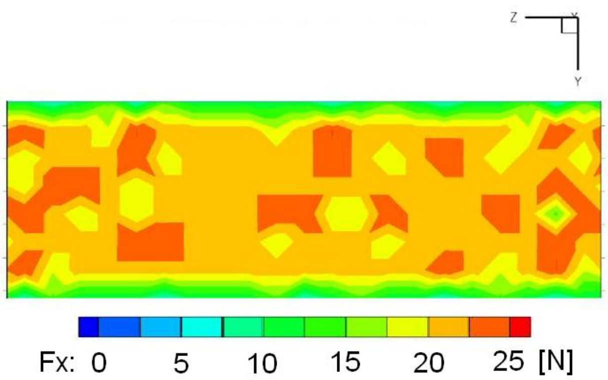


Figure 12: Uniform force distribution on the structure using the new interpolation method.

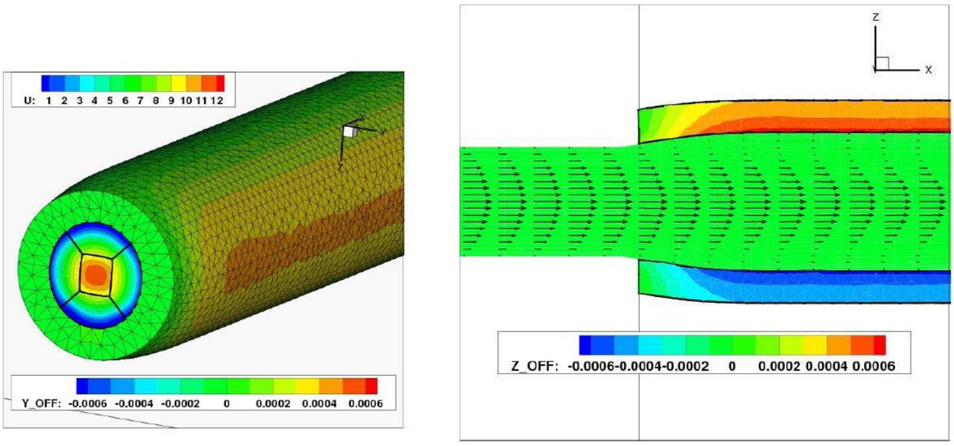


Figure 13: Deformation of the tube section.

The FSI module was validated with measurements carried out by Kylstra et al. [2]. The authors measured the deformation of a latex tube by raising the transmural pressure stepwise. The geometry (fluid and solid) of the tube was created and the material properties were set. Incompressible fluid was selected in the fluid domain. Steady FSI simulations were carried out for each transmural pressure value. The calculated average tube diameters were compared to the ones of the measurement. The results are shown in Figure 14.

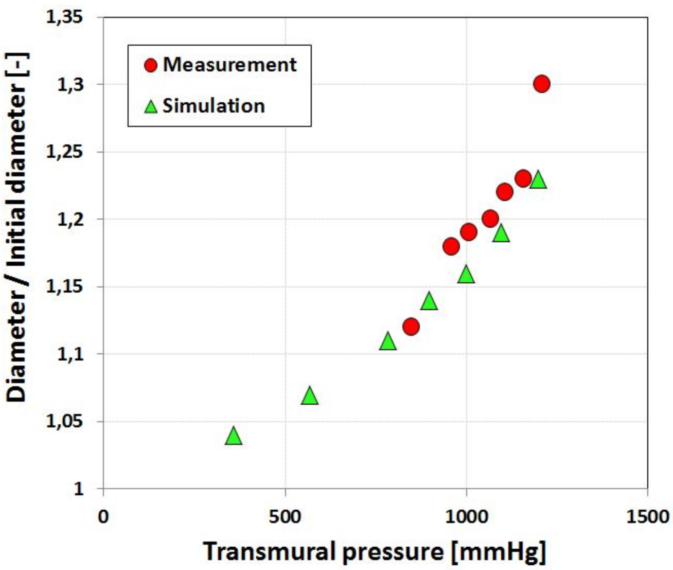


Figure 14: Specific diameter as a function of transmural pressure.

The calculation results show good agreement with the measurement.

6 New scientific results

The new scientific results of the dissertation are summarized in the following theses:

1. A new calculation method was developed that is capable of determining time dependent pressure, velocity and deformation in the arterial network model (from the aorta to the arterioles). The viscoelastic effects of the vessel wall were taken into account using the Stuart model. The calculation method solves a system of partial and ordinary differential equations system numerically with the method of characteristics. A genetic optimization algorithm to tune the three parameters (E_1, E_2, η_2) of the Stuart model was developed. Publications related to this thesis: [20], [17], [12], [4], [13]

   1.1 It was shown that in the case of a proper boundary condition for the heart (volumetric flow rate as a function of time) the resulting pressure and velocity curves along the arterial network model deliver suitable results. The pressure curves agree well qualitatively with measured pressure curves in a human body. The value of the systolic and diastolic pressure, the shape of the pressure curves can be varied by modification of the Stuart model parameters. The effect of arteriosclerosis in the iliaca was investigated. It was proved that reducing the diameter of the stenosis to 50% of the vessel diameter does not lead to significant change in the volumetric flow rate of the iliaca. This is in concordance with the experiences of physicians.

   1.2 The viscoelastic material model was generalized such that the Young modulus of the linear elastic element (E_1) depends on the deformation. It was proved that the pulse wave velocity depends on the deformation and the transmural pressure. This is in agreement with measurement results of physicians.

2. A new method was developed based on the viscoelastic material model capable of estimating the central aortic pressure curve from a pressure curve given in an arbitrary point (mainly near the peripheries) of the arterial network model. Publications related to this thesis: [20], [19]

   2.1 With the generalization of the method of characteristics a new calculation method was developed calculating the upstream pressure and velocity curves of a vessel from the downstream end pressures and velocities ("Backward" calculation). The method was applied to the arterial network model. A part of this method is a procedure that estimates the velocity curve at the downstream end of the investigated vessel section.

   2.2 The functionality of the method was verified by testing calculations. The central aortic pressure curve is calculated from the "forward" calculation results with an accuracy of ± 1 $mmHg$. The systolic peak of the estimated central aortic pressure curve agrees well with the results of well tested estimation methods presented in physiological publications.

3. Venous blood flow is induced partly by the contraction of the surrounding muscles that deform the vein walls (calf muscle pump). The collapse of the vessel walls is investigated using two- and three dimensional models and a measurement rig designed for this purpose. Publications related to this thesis: [14], [16], [4], [15]

   3.1 A two-dimensional mathematical model and a three dimensional FSI simulation were set up for the investigation. It was demonstrated that the calculated shape of the collapsed cross section is similar for both models. Below a transmural pressure of $-2000\ Pa$ the difference between the cross section areas is below five percent. Using the more complex FSI simulations, the faster and more simple two dimensional mathematical model was validated.

   3.2 A new algorithm was developed for the solution of the two dimensional mathematical model. Using this algorithm the cross sectional shape of the collapsed vessel can be calculated with a short running time. From the cross-section shape area the wave propagation velocity can be well estimated.

   3.3 A measurement rig was designed to physically investigate the collapse phenomenon. According to the measured pressure and velocity curves it was demonstrated that the calf muscle pump acts similarly to a volumetric pump. The measurement system is capable of controlling and validating of the vessel wall collapse models.

4. FSI calculations were carried out to investigate the deformation of a vessel section in case of positive transmural pressures. Publications related to this thesis: [11], [9], [6]

   4.1 It was shown that in case of the finite element mesh of the structure is finer than the CFD mesh the usage of the conservative interpolation method can lead to problems. A uniform force distribution on the fluid side of the fluid-solid interface is distorted through the interpolation process. As a result, the force distribution on the structural side is distorted and uneven.

   4.2 A new interpolation method was developed for transferring the calculated forces from the fluid domain to the structure. It was shown that the previously discussed uniform force distribution is transferred properly to the structural side of the fluid solid interface. The conservativity of the force field is guaranteed by the method.

   4.3 The FSI module was validated using measurement data taken from literature. It was shown that a fair agreement exists between the measured and the calculated diameter - transmural pressure values.

Own publications

[1] Hős C , Pandula Z , Halász G , Kullmann L , Bárdossy, G: *Water demand forecast and on/off type pump schedule optimisation for water distribution systems*, Proceedings of the 23nd IAHR Symposium on Hydraulic Machinery and Systems. Tokyo, Japán, 2006

[2] Pandula Z, Szekeres G, Debreceni B, Bárdossy G: *Capacity Measurements of Drinking Water Wells*, Proceedings of the Fifth Conference on Mechanical Engineering. Budapest, Magyarország, 2006.05.25-2006.05.26.

[3] Bárdossy G , Halász G: *Prognosis of Urban Water Consumption*, Proceedings of the Fifth Conference on Mechanical Engineering. Budapest, Magyarország, 2006.05.25-2006.05.26.

[4] Licskó G , Bárdossy G: *Modelling blood flow in veins due to external pressure excitation*, The 1st IMACS International Conference on Computational Biomechanics and Biology: Short communications. Plzen, Csehország, 2007.09.10-2007.09.13.

[5] Bárdossy G, Halász G: *Városi vízfogyasztás előrejelzése hibrid fuzzy algoritmusok segítségével*, XI. Országos Víziközmű Konferencia. Sopron, Magyarország, 2007.06.12-2007.06.14.

[6] Pantle I, Bárdossy G, Gabi M: *Numerical Approaches for Flow Induced Structure Vibrations Aiming for Medical Applications*, Proceedings of the 2007 Congress and Exposition on Noise Control Engineering (INTERNOISE), Istanbul, Turkey, 2007

[7] Bárdossy G, Pandula Z, Halász G, Winter J, Márialigeti B: *Forecasting Water Consumption with Hibrid Fuzzy Algorithms: Theory and Operating Experiences*, Proceedings of Sixth Conference on Mechanical Engineering. Budapest, Magyarország, 2008.05.29-2008.05.30.

[8] Tompa F, Halász G, Hegedűs F, Bárdossy G, Pandula Z: *Computation of a District-Heating Network*, Proceedings of Sixth Conference on Mechanical Engineering. Budapest, Magyarország, 2008.05.29-2008.05.30.

[9] Pantle I, Bárdossy G, Gabi M: *Numerischer Ansatz zur Simulation von Strukturschwingungen induziert durch Fluidschwingungen*, Deutsche Jahrestagung für Akustik (DAGA08), Dresden, Germany, 2008

[10] Bárdossy G, Halász G: *Modelling Venous Blood Flow*, Third Hungarian Conference on Biomechanics. Budapest, Magyarország, 2008.07.04-2008.07.05.

[11] Bárdossy G, Halász G, Pantle I, Gabi M: *Fluid-structure-interaction: a hybrid CFD-FEM approach for multiple engineering and biomedical applications with special focus on venous blood flows*, Conference on Modelling Fluid Flow CMFF'09, Budapest, Magyarország, 2009.09.09-2009.09.12., pp. 144-151. Paper BF001. (ISBN: 978-963-420-987-4)

[12] Bárdossy G, Halász G, Winter J: *Prognosis of urban water consumption using hybrid fuzzy algorithms*, Journal of Water Supply Research and Technology-AQUA, 58(3) pp. 203-211. Paper 448EQ. (2009), IF: 0.849

[13] Bárdossy G, Halász G: *Viszkoelasztikus csövekből felépített vénás modellhálózat időfüggő áramlásának számítása*, OGÉT 2009 - XVII. Nemzetközi Gépészeti Találkozó. Gyergyószentmiklós, Románia, 2009.04.23-2009.04.26., pp. 48-51.

[14] Boros Á, Bárdossy G, Halász G: *Measuring the Deformation of a Viscoelastic Tube Using an Optical Method*, Proceedings of the Seventh International Conference on Mechanical Engineering. Budapest, Magyarország, 2010.05.25-2010.05.26.

[15] Bárdossy G, Lantos C, Halász G, Monos E, Nádasy LG: *Determination of venous blood flow velocity using digital videomicroscopy (A short methodical communication)*, Acta Physiologica Hungarica, 97(4) pp. 417-421. Paper 10.1556/APhysiol.97.2010.4.6. (2010), IF: 1.226

[16] Bárdossy G, Boros Á, Halász G: *Experimental modelling of the venous muscle pump*, Biomechanica Hungarica, III.(2) pp. 23-30. (2010)

[17] Bárdossy G, Halász G, Gondos T: *The diagnosis of hypovolemia using advanced statistical methods*, Computers in Biology and Medicine, 41(11) pp. 1022-1032. Paper 10.1016/j.compbiomed.2011.0.002 (2011), IF: 1.112

[18] Matolcsi K, Bárdossy G, Halász G: *Simulation of Saline Solution Injection into a Venous Junction*, IFMBE Proceedings (Vol. 37): 5th European Conference of the International Federation for Medical and Biological Engineering (EMBEC). Budapest, Magyarország, 2011.09.14-2011.09.18.

[19] Bárdossy G, Halász G: *Modelling blood flow in the arterial system*, Periodica Polytechnica - Mechanical Engineering Series, közlésre elfogadva (2011).

[20] Bárdossy G, Halász G: *A "backward" calculation method for the estimation of central aortic pressure wave in a 1D arterial model network*, Annals of Biomedical Engineering, közlésre benyújtva (2011).

References

[1] E Kresch és A Noordergraaf: Cross-Sectional Shape of Collapsible Tubes. *Biophysical Journal*, 12. évf. (1972) 3. sz. 274-294.

[2] JA Kylstra, T Wierbicki, ML Wolbarsht, MB Landers, és E Stefansson: The Relationship between Retinal Vessel Tortuosity, Diameter and Transmural Pressure. *Graefe's Archive for Clinical and Experimental Ophthalmology*, 224. évf. (1986) 5. sz. 477-480.

[3] S Till és F Hegedűs: Experimental Validation of a Novel Blood Flow Modelling Method. In *Gépészet konferencia 2006* (konferenciaanyag). 2006.