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CFD ANALYSIS OF BLOOD FLOW WITHIN AORTA OF PATIENT WITH COARCTATION OF AORTA[1]

Summary. Blood flow in a real geometry of aorta was analysed. The CFD analysis was performed using commercial ANSYS/Fluent code. Velocity profile was used to mimic inlet flow conditions during human cardiac cycle. Outlet pressure boundary condition was coupled with lumped parameter model (electrical analogy) of circulatory system.

Keywords: blood flow, aorta, CoA, circulatory system, electrical analogy, CFD, LPM

ANALIZA NUMERYCZNA PRZEPŁYWU KRWI W AORCIE U PACJENTA Z KOARKTACJĄ AORTY

Streszczenie. Przeanalizowano jednofazowy przepływ krwi dla modelu aorty odwzorowującego rzeczywistą geometrię. Symulacja CFD przeprowadzona była za pomocą komercyjnego oprogramowania ANSYS/Fluent. Na wlocie do geometrii zadano profil prędkości, odpowiadający cyklowi pracy serca. Ciśnienie krwi na wylotach określono przez sprzężenie z parametrycznym modelem skupionym (zbudowanym w analogii elektrycznej).

Słowa kluczowe: przepływ krwi, aorta, koarktacja aorty, układ krążenia, analogia skupiona, CFD, LPM

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1. Introduction

CFD (*Computational Fluid Dynamics*) is branch of fluid mechanics that uses numerical analysis and algorithms to solve and analyze problems that involve fluid flows. As such, it is a technique that is widely used in many areas for design and analysis - in wide range of engineering problems and for (bio)medical devices design as well. It is a primary method to perform easy and low-cost verification of assumptions that allows to avoid high costs of laboratory tests. CFD is also very promising tool for minimally-invasive patient assessment [1], where a lot of problems could not be validated against measurements due to ethical aspect. In addition, it should be stressed that CFD model, similarly to other numerical models, can be used as a direct problem solver within inverse analysis. Such analysis is used when direct measurements are not available, e.g. for tumor recognition in a different localizations in a human body [2-5]. CFD technics can also support pre-surgery activities, for better problem understanding helping to choose better treatment [6].

In area of biomedical applications of CFD, a cardiovascular medicine is a field, where usage of CFD can significantly enhance diagnosis, device design and clinical trials. The recent review of possible applications of CFD modelling in cardiovascular medicine is in [1]. Numerical analysis can also help with understanding atherogenesis problem with unexplainable before creation hard and soft atherosclerotic plaque in the specific places [7].

Unfortunately, except of many advantages, the CFD methods are still not ready for clinical use. This is mainly caused by time demanding pre-processing of medical images, geometry model creation, time consuming simulations due to very unstable flow phenomena which require using of complex unsteady solvers which despite very effective and fast computers.

The work presents modelling a blood flow within aorta of patient with CoA (*coarctation of aorta*) - narrowing of the descending aorta opposite to the site of the ductus arteriosus insertion. The untreated CoA causes premature death of the patients. The negative effect on the cardiovascular system comes from exposure of the upper part of the human body to hypertension and blood flow disturbances with all consequences. According to the American College of Cardiology and American Heart Association guidelines for treatment patients with congenital heart disease indications for the therapeutic intervention in CoA require invasive blood pressure measurement before and after the narrowed site.

Blood consist of approx. 55% of blood plasma and 45% different types of blood cells. Over 90% of blood plasma is water, remaining 10% are another substances, mostly proteins. In current work the simplified model of blood is presented: single phase medium with rheological character of average density and viscosity for all parts [8, 9].

Research works usually describe detailed blood flow modelling within selected parts of human circulatory system [10-13]. However, due to the complexity of human circulatory system, none of them analyze the whole system. Instead, isolated sections of the circulatory system (vessels) and/or internal organs are modelled in detail using CFD technique, while the rest of circulatory system, namely vascular tree, are usually modelled as lumped parameter models. Such models, coupled to CFD solution as boundary conditions, are using blood flow analogy to the electrical current in basic circuits, allow to mimic the response of the whole circulatory system [6, 14, 15].

2. CFD modelling of aorta blood flow

Complete CFD analysis can be divided into several independent tasks, showing the typical work flow for cardiovascular application:

- medical imagining; most often used imaging modalities are CT (*Computer Tomography*) i.e. AngioCT (*Angiography CT*) or MRI (*Magnetic Resonance Angiography* - MRA),
- image segmentation; i.e. identification of a vessel’s lumen for each image,
    - reconstruction of real, patient-specific 3D vessel (artery) geometry; i.e. creation of STL (*STereoLithography*) model (the raw unstructured triangulated surface geometry of a three-dimensional object) and post-segmentation activities, like: smoothing, ROI (*region of interest*) selection, etc.
- discretization of the model; i.e. creation of numerical mesh in analyzed domain,
- definition of materials; i.e. blood rheological model,
- definition of boundary conditions; i.e. inflow/outflow, coupling with lumped models of the rest of the circulatory system,
- solving; i.e. performing numerical simulations, to achieve speedup parallel processing is widely used,
- results analysis; i.e. post-processing.

2.1. Geometrical model

The paper presents CFD analysis of blood flow in a within aorta with CoA. The geometry was adopted from real 8-year old patient MRA imagining (*Gadolinium-enhanced Magnetic Resonance Angiography*). The numerical case, with geometry, in- & out-flow conditions and other necessary data was based on a problem defined within CFD challenge problem[2] for development numerical models capable to model blood flow (model ID: OSMSC0111[3], OSMSC Corp., USA). The ready STL model of aorta was used. Prior to CFD analysis, the STL

[2] http://www.vascularmodel.org/miccai2012 [Accessed 30 April 2016]
[3] http://www.vascularmodel.com/sandbox/doku.php?id=repository_aorta [Accessed 30 April 2016]

model was processed using Geomagic DesignX (3D Systems, Inc., USA) software to smooth the surfaces and cut ROI.

The geometry – namely the volume of blood inside the vessel – used in current analysis is presented in the Fig. 1. Within presented work the vessel walls were treated as rigid, with no interaction on fluid-solid interface.

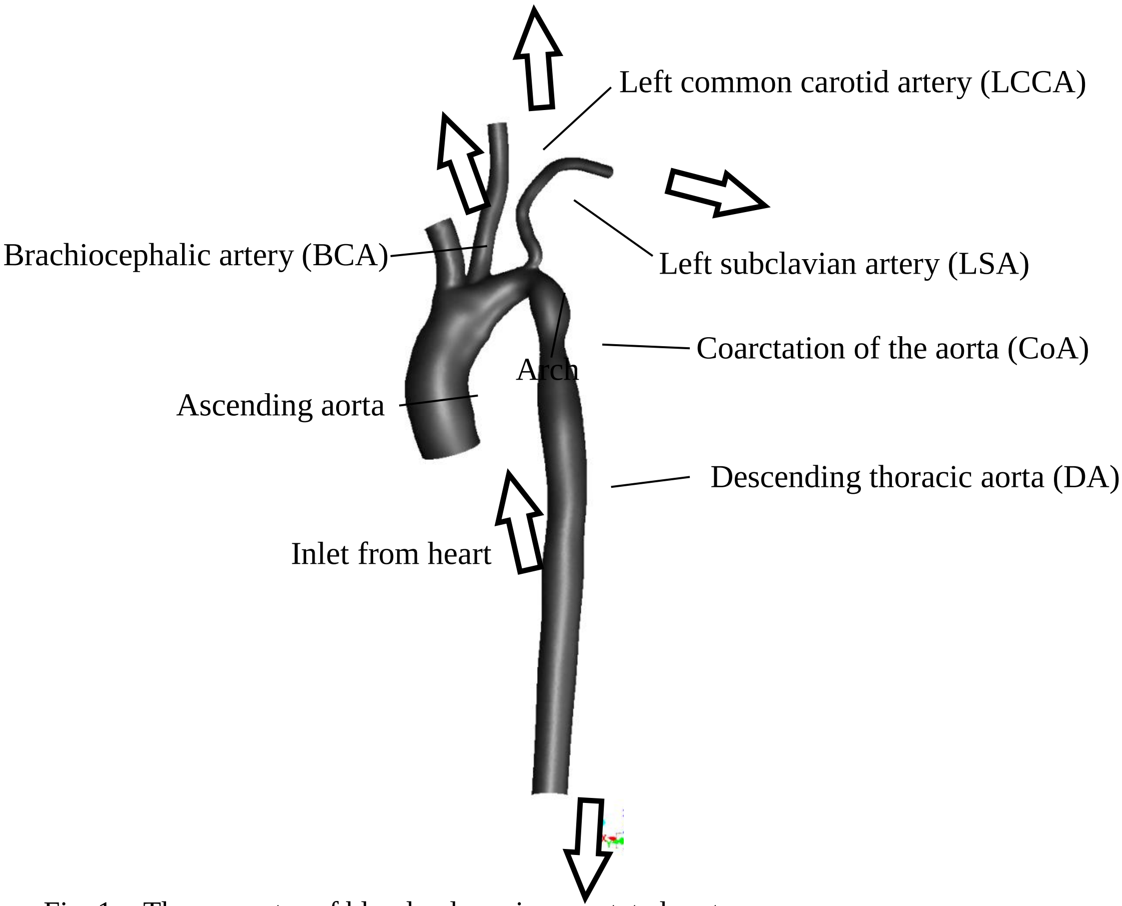


Fig. 1. The geometry of blood volume in coarctated aorta
Rys. 1. Geometria naczynia (krwi) w zwężonej patologicznie aorcie

2.2. Numerical model

2.2.1. Mesh

The generated hybrid mesh consists of about 700,000 elements. It was chosen after mesh sensitivity and provides balanced accuracy of results and the computational time. In the inlet part and DA part hexahedral elements were used in order to stabilize flow. Mesh is presented in the Fig. 2a and 2b.

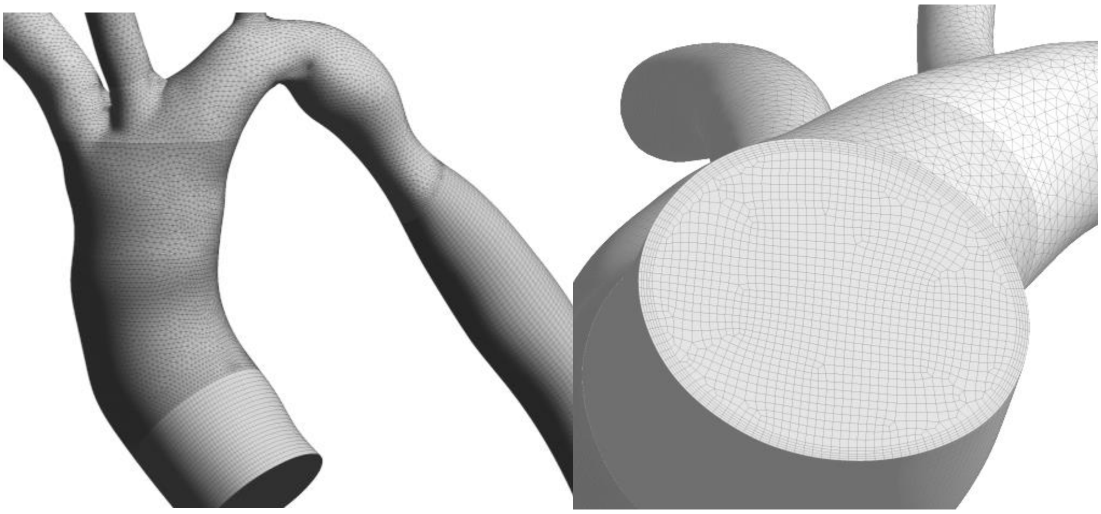


Fig. 2. Generated hybrid mesh with hexahedral and tetrahedral elements
Rys. 2. Hybrydowa siatka podziału numerycznego z elementami heksa-hedralnymi oraz tertrahedralnymi

2.2.2. Boundary conditions

One of the most important point during setting CFD model are boundary conditions (BCs) In the current work one inflow and four outflow BCs are addressed.

Inflow. At the inflow (see *Inlet from heart* shown in Fig. 1) velocity inlet boundary condition is used. Velocity profile (time dependent) based on the volumetric flow taken from patient-specific (i.e. heart rate, cardiac output) conditions in the human cardiac cycle. The velocity flow profile is presented in Fig. 3.In the same figure two characteristic points are highlighted: systole (maximum), when the atrium contracts, and blood flows from atrium to the ventricle, and diastole (minimum), when the semilunar valves (the pulmonary valve and the aortic valve) is close, and the atrioventricular (AV) valves (the mitral valve and the tricuspid valve) is open, and the whole heart is relaxed. All results in the paper are discussed for the highest (systole) specific point of heart cycle.

Periodic and pulsatile flow is prescribed to the inlet using UDF (*User Defined Function*). UDF is an *in house* code written in the C# that is dynamically loaded and used by ANSYS Fluent solver to enhance the standard features of the commercial code.

Outflow. The remaining boundary sections are outflows: BCA (*brachiocephalic artery*), LCCA (*left common carotid artery*), LSA (*left subclavian artery*), DA (*descending thoracic aorta*), cf. Fig. 1. Pressure outlet type of BC was used. Outflow section pressure, being response of the vascular tree attached to each of the outflows, was modelled using LPM (*lumped parameter model*). This condition was implemented using above mentioned UDF functionality.

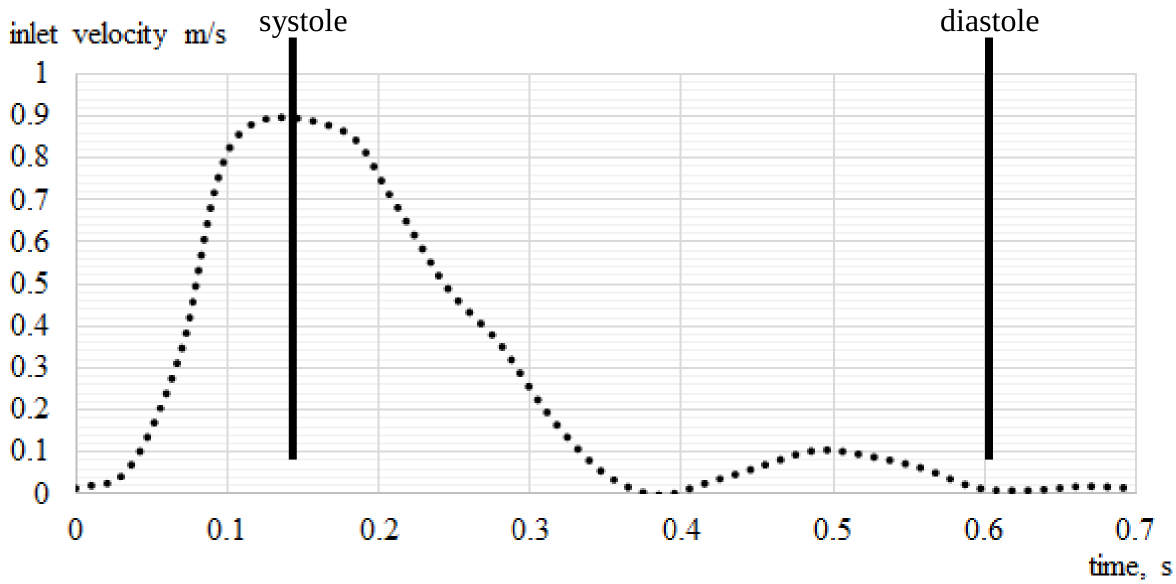


Fig. 3. Velocity flow profile at the aorta inlet
Rys. 3. Prędkość na wpływie do aorty

2.2.3. Lumped Parameter Model of circulatory system

One of the possibility to obtain information about parameters and behavior of the remaining part of cardiovascular system is using LPM [16-19]. LPM is based on electrical analog of blood flow in vascular bed. The idea of LPM is presented in Figs 4 and 5.

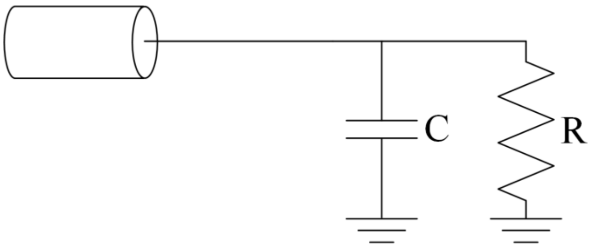


Fig. 4. 2-element Windkessel model
Rys. 4. 2-elementowy model Windkessela

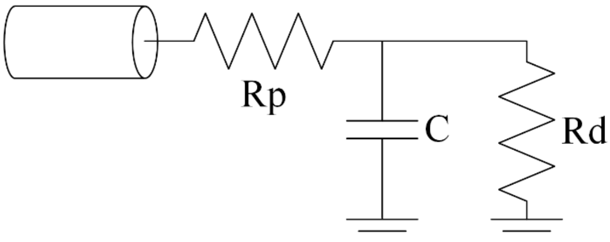


Fig. 5. 3-element Windkessel model
Rys. 5. 3-elementowy model Windkessela

The base of this assumption is that heart and systemic arterial system behaves similar to the closed hydraulic circuit comprised of a water pump connected to the air chamber (ger. *Windkessel*). The basic equation which links the hydraulic flow with the flow resistance was proposed by Poiseulle [20]. This law directly corresponds to the Ohm's law.

Difference of electrical potential correspond to difference of pressure (as driving force), electrical currents correspond to flow rates. The main idea of LPM model is an assumption that particular vessel or group of vessels, can be described as resistors, capacitors and inductors or their combination as simple circuits. Equation 1 presents the idea and correlation between hydraulic and electrical law, given for 2-element Windkessel model (cf. Fig 4):

$$I(t) = \frac{P(t)}{R} + C\frac{dP(t)}{dt} \tag{1}$$

where: $I(t)$ is the water (blood) flow out of the pump (heart) as a function of time, $P(t)$ is water (blood) pressure as a function of time, C is the constant ratio of air pressure to air volume (capacitance of the capacitor – mass of the blood), R is the flow-pressure proportionality constant (resistance of the resistor – resistance of the blood vessels).

Extended model, called 3-element Windkessel model [21], where additional resistive element was added to the 2-element Windkessel model. 3-element Windkessel model differential equation is presented in equation 2 (cf. Fig 5):

$$\left(1 + \frac{R_p}{R_d}\right)I(t) = CR_p\frac{dI(t)}{dt} = \frac{P(t)}{R_d} + C\frac{dP(t)}{dt} \tag{2}$$

where R_p is the resistance encountered by blood as it enters the aortic or pulmonary valve, while R_d is the resistance of blood as it passes from the aorta to the narrower arerioles. The remaining variables are the same as in the 2-element Windkessel model (see description of eq. 1).

Outflow from the model is modeled by setting pressure $P(t)$ in outflow section. This pressure is calculated explicit from equation (2) according volumetric flow of the blood which is read from CFD solution in the outlets. Separate model is used for each of the four outflows, for different LPM model parameters (taken after model ID: OSMSC0111[4], OSMSC Corp., USA). Table 1 summarizes the LPM model data used in current research.

Table 1

LPM model parameters

Artery name	Abbrev.	R_p, $kg{\cdot}m^{-4}{\cdot}s^{-1}$	C, $m^4{\cdot}s^2{\cdot}kg^{-1}$	R_d, $kg{\cdot}m^{-4}{\cdot}s^{-1}$
Right Subclavian Artery*	RSA	$0.885{\cdot}10^8$	$7.13{\cdot}10^{-10}$	$1.3858{\cdot}10^9$
Right Common Carotid Artery*	RCCA	$1.122{\cdot}10^8$	$5.62{\cdot}10^{-10}$	$1.7571{\cdot}10^9$
Left Common Carotid Artery	LCCA	$1.124{\cdot}10^8$	$5.62{\cdot}10^{-10}$	$1.7611{\cdot}10^9$
Left Subclavian Artery	LSA	$2.895{\cdot}10^8$	$2.12{\cdot}10^{-10}$	$4.5361{\cdot}10^9$
Descending Thoracic Aorta	DA	$0.271{\cdot}10^8$	$2.922{\cdot}10^{-9}$	$0.3122{\cdot}10^9$

* - in the relation to Fig. 1 Brachiocephalic artery (BCA) is common branch for RSA and RCCA (before bifurcation)

The geometry used in current research was limited to BCA (*Brachiocephalic artery*) section. In result, bifurcation into RSA (*Right Subclavian Artery*) and RCA (*Right Common Carotid Artery*), were not modelled. For such case parameters displayed at the Table 1 cannot be used directly. Therefore, the Laplace transform ($\mathcal{L}$) analysis in *s*-domain (frequency domain) was used to compute equivalent response of two 3-element Windkessel models connected in parallel – namely: RSA and RCCA. Then inverse Laplace ($\mathcal{L}^{-1}$) transform was computed

[4] http://www.vascularmodel.com/sandbox/doku.php?id=repository_aorta [Accessed 30 April 2016]

using Matlab Symbolic Math Toolbox (MathWorks Inc., USA), yielding final equation in time domain. The derived equation was used as equivalent for 2 separate models for each of bifurcated branch of BCA.

2.2.4. Time discretization (time step)

Transient analysis was performed for several subsequent heart actions, each of 0.7 second. Time step sensitivity analysis was performed and the time step was finally taken as 0.01 second. For each time step the outflow BCs were adjusted iteratively. There was a limit of 500 internal iterations per time step, which has never been reached. Iteration procedure was stopped when continuity equation residuals threshold was reached.

3. Results

Mass flow rates: at the outlets and inlet are presented in the Fig. 6.

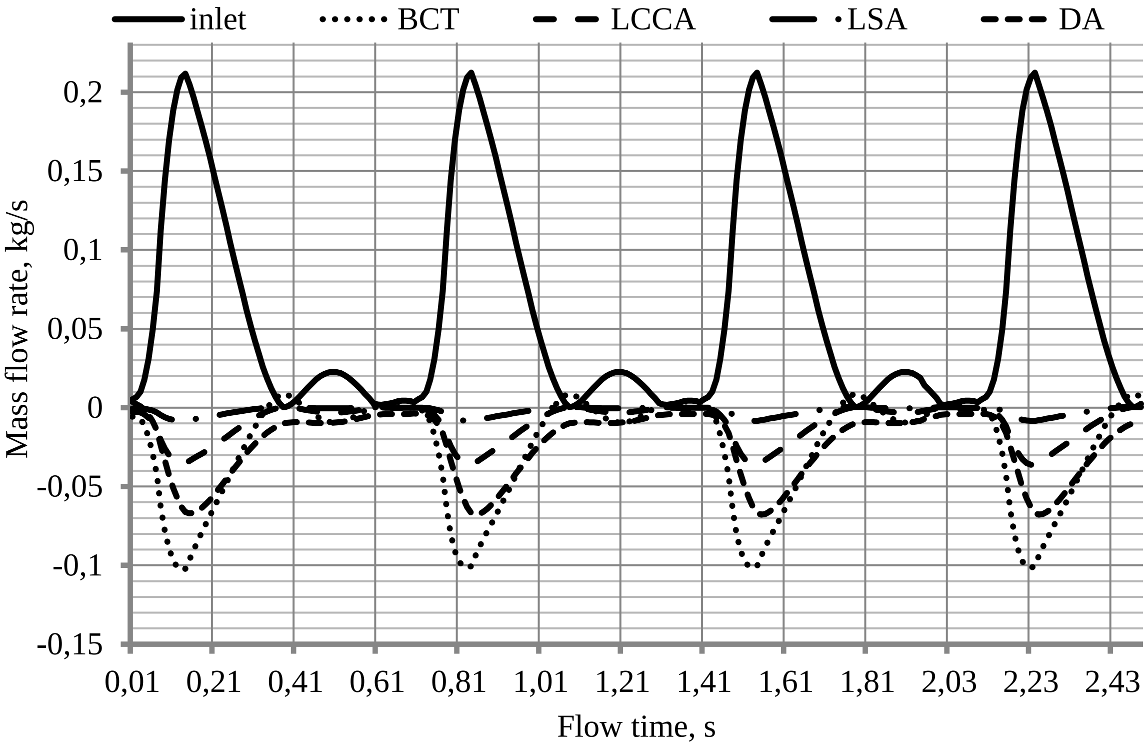
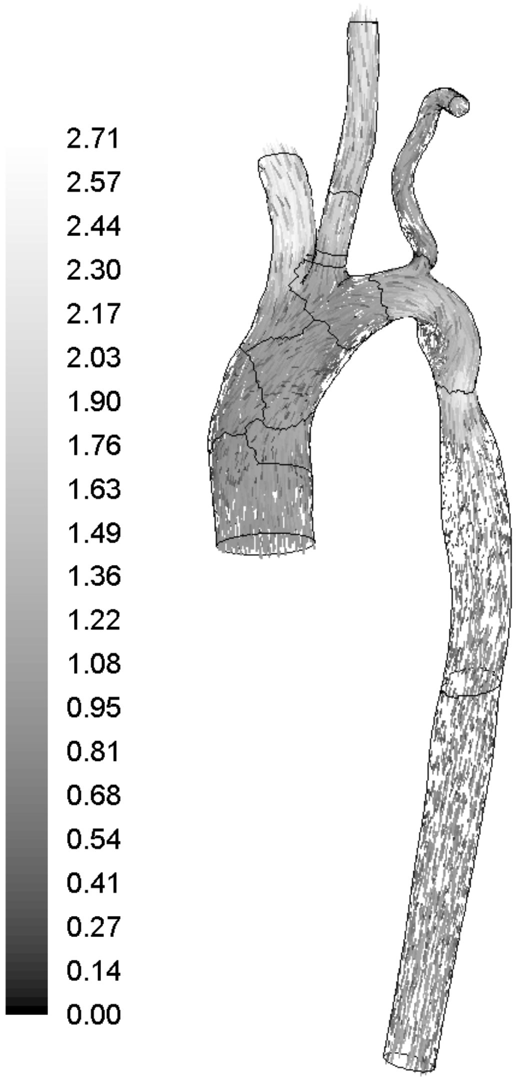


Fig. 6. Mass flow rate in the outlets as a function of the flow time (cf. Fig. 1.)*
Rys. 6. Strumień masowy na wypływach w funkcji czasu (por. rys. 1.)*
* - the beginning of the graph should be neglected because of the burning period. Negative values represent fluid outflow of the domain.

The highest mass flow rate in the inlet is in the systole point – time 0.15 sec (0.85 sec, 1.55 sec and consecutive time instants, by increment 0.7 sec, i.e. full cardiac action time) as well as for all outlet vessels (negative represent outflow of the domain). The backflow can be observed approx. to the 0.41 sec (and 1.11 sec, 1.81 sec, ...) for the BCT vessel. In the other cases flow is almost zero but reverse flow is not observed.

Exemplary results for the systole (cf. Fig. 3.) are presented in the Fig. 7 which represent velocity vectors and in the Fig. 8 which displays static pressure in the wall.



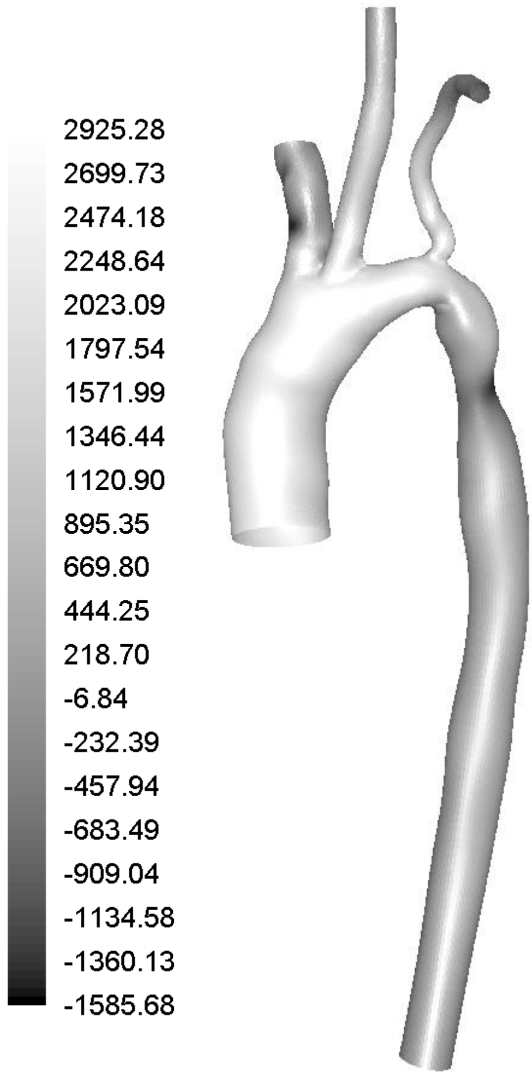


Fig. 7. Velocity vectors during systole (m/s)
Rys. 4. Wektory prędkości w skurczu (m/s)

Fig. 8. Static pressure during systole (Pa)
Rys. 8. Ciśnienie statyczne w skurczu (Pa)

The highest value of the velocity are reported in the two vessels (BCT, LCCA, cf. Fig. 3). Inlet of these arteries are localized near to the main inlet from the heart, additionally the cross section is four and times smaller than the main inlet respectively. Substantial increase of the velocity is also observed in the CoA region.

The lowest static pressure is observed in CoA region and in the distal parts of the LSA branch. The highest pressure in systole is observed in the main aorta part, right after the heart, what is physical.

4. Conclusions

The goal of the present work was to show an application of computer model for modelling blood flow within aorta of 8-year old patient diagnosed with a coarctation of the descending thoracic aorta (CoA, approx. 65% aortic area reduction). Blood was simulated as a single-phase. Periodic and pulsate flow is prescribed to the inlet while LPM (electrical analogy) are prescribed as pressure outlets. Pressure at the outlets were calculated periodically, according to the volumetric flow, using 3-element Windkessel models – described by equation (3). The analysis shows that LPM (*Lumped Parameter Model*) based on electrical analogy is a valuable and powerful tool, and can be applied in further investigations.

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Omówienie

Modelowanie numeryczne w zakresie zastosowań (bio)medycznych – może być stosowane np. do wczesnego wykrywania nowotworów (inżynieria odwrotna) [2-5], a także np. do działań przedzabiegowych, pozwalających efektywniej leczyć [6]. Numeryczna mechanika płynów (eng. *CFD – Computational Fluid Mechanics*) jest częścią modelowania numerycznego opisującego przepływ płynów. Może zostać wykorzystana w zagadnieniach przepływu krwi [1, 10-15].

Zaprezentowana praca dotyczy analizy numerycznej przepływu krwi w aorcie u 8-letniego pacjenta z koarktacją aorty (zwężenie ok. 65%). Symulacje wykonano przy użyciu komercyjnego oprogramowania ANSYS Fluent (ANSYS Inc., USA).

Przypadek został zaczerpnięty z *CFD challange problem*[2]. Zakupiono model STL analizowanego przypadku, który został przygotowany do analizy CFD (rys. 1). Następnie geometrię zdyskretyzowano i zaproponowano hybrydową siatkę z ok. 700 000 elementów (rys. 2).

Właściwości krwi opisano używając modelu krwi jako ciecz jednorodna z reologicznymi właściwościami średnimi dla gęstości i lepkości składników krwi [8, 9].

Na wlocie zastosowano funkcję naśladującą przebieg rzeczywistego rzutu serca (rys. 2) zaimplementowaną do rozwiązania przy użyciu funkcji użytkownika (*ang. UDF – User Defined Function*).

Warunkami wylotowymi z modelu są ciśnienia w przekrojach wylotowych, obliczone przy użyciu parametrycznego układu skupionego (eng. LPM), który bazuje na analogii elektrycznej [16-19]. W celu uzyskania odpowiednich ciśnień na wypływie rozwiązywane jest równanie (2) zwane modelem Windkessela (rys. 4). Wartości parametrów RLC przedstawiono w tabeli 1.

Wyniki (rys. 6) pokazują strumień masowy na dopływie oraz wypływach w funkcji czasu. Największy wypływ we wszystkich kontrolowanych wylotach zaobserwowano podczas skurczu serca (około 0,15 s; 0,85 s.; 1,55 s, itd. co 0,7 s), co jest fizykalne. Przepływ wsteczny zaobserwowano tylko w najbliższej tętnicy w okolicy napływu krwi (czas ok. 0,41 s; 1,14 s; 1,84 s, itd. co 0.7 s). Wektory prędkości i ciśnienie statyczne na ściance zostały przedstawione w skurczu na rys. 7 i rys. 8. Największe prędkości zaobserwowano w tętnicach najbliższych wlotowi oraz w rejonie koarktacji. Najniższe ciśnienie zaobserwowano na przewężeniu aorty oraz w dystalnym fragmencie aorty LSA.

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