Experimental study of the flow in the elastic model of the abdominal aortic bifurcation

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Abstract. This paper reflected preliminary results of physical modeling of pulsating flow in a model of abdominal aortic bifurcation with taking into account the physiological elasticity of the vessel walls. Elastic vessel models were made via molding from a silicone mixture based on Lasil-T4 silicone rubber. The auxiliary study was performed to assess the elastic properties of the silicone mixture and select a necessary composition. The experiment on the pulsating flow in the rigid and elastic models of the abdominal aortic bifurcation was carried out using a blood flow simulator with circulation of blood-emulating fluid. It was revealed that interaction between the elastic model and closed rigid circuit of the blood flow simulator resulted in generation of intense parasite flow oscillations and prevented from getting similar flow conditions for rigid and elastic models. A way to solve the problem is to include dampers with liquid in the hydraulic circuit of the blood flow simulator at the inlet and the outlets of the elastic model.

1. Introduction
An experimental study of the flow structure in vessel models of cardio-vascular system is an essential part of the forming vortex structure research. An important component of the physical modeling in the vascular bed models is the formation of experimental conditions similar to real physiological processes. In order to achieve this, many factors have to be considered: pulsating flow regime, input and output values of flow rates and pressures, extensibility of the vessel models wall, operation modes of the heart. These factors affect the developing structure of the flow in the vessels.

The study of the flow structure in abdominal aortic bifurcation area began in the 1980s [1], often in the literature there are computational and experimental studies in various configurations of simplified rigid models that do not take into account the elasticity of the wall [2,3]. In this work, first stage of physical modeling of a pulsating flow in a model of a simplified average statistical configuration, including the bifurcation region of the abdominal aorta and common iliac arteries, are presented. The main attention is paid to the comparative analysis of the emerging input and output conditions considering the elasticity of the wall.

2. Model of the abdominal aortic bifurcation and iliac arteries
The study was conducted using two identical-geometry models of an average configuration vessel [4-6]: rigid and elastic. The model includes an outlet region of the abdominal aorta with a diameter of 18
mm that divides into the right and left common iliac arteries with a diameter of 11 mm; the iliac arteries deviation angle from the abdominal aorta axis is 25 °. The model does not take into account the spatial bends of the considered part of the vascular bed. The model was designed using SolidWorks 2016 software package. The rigid model (figure 1, a) was created via 3D prototyping from the Tough2000 v1 photopolymer that is suitable to study the flow structure by the ultrasonic Doppler method; the wall thickness of the completed model was 1 mm.

The elastic model was made by means of casting a mixture based on Lasil-T4 silicone rubber into a detachable 3D printed mold (figure 1, b). To form the inner surface of the model, a sectional rod consisting of three parts was installed into the mold; the rod geometry and dimensions corresponded to the rigid model. Vacuum chamber was used to remove air bubbles from the silicone mixture that arose when mixing the components. Then this mixture was poured into the mold through the holes. After silicone solidification (24 hours), the model with the rod inside was detached from the mold, and then the rod parts were extracted from the elastic model. The wall thickness of the model was 1 mm that corresponded to the clinical data on the abdominal aorta wall thickness.

![Figure 1](image_url)

**Figure 1.** Models of abdominal aortic bifurcation: (a) – scheme and dimensions of the rigid model, (b) – the casting mold, the rod, and the finished elastic model photo

### 3. Selection of the model extensibility

According to clinical data [7, 8], the abdominal aorta diameter changes by 5-20% during the cardiac cycle. In the experiment, the model extension depends on the pressure pulsations magnitude and the elastic properties of the wall material that, in turn, depend on the silicone mixture composition. In addition to Lasil-T4 silicone rubber, the mixture contains Lasil-T4 curing agent and DC-200 diluent for silicones. Since the amount of the curing agent in the mixture is prescribed by the manufacturer, only the diluent proportion affects the material extensibility. An auxiliary experiment was performed to assess the silicone mixture elasticity and select a composition that would provide a reasonable extensibility of the model.

Several cylindrical vessel models with the inner diameter of 6 mm and the wall thickness of 1 mm were made by means of casting into a mold with different compositions of the silicone mixture. The model extension due to hydrostatic pressure was recorded by a camera. The model outer diameter was obtained from the photo using Photoshop facilities; the statistical uncertainty of the results did not exceed 4%. The measurement results are shown in figure 2, a. One can see that (a) the considered range of pressure provides reasonable values of the model extension (up to 20%), and (b) the model becomes more extensible along with an increase of the diluent fraction.
In figure 2, b, the measurement data are presented taking into account the well-known analytical solution for extension of the thin-walled tube under the inner pressure load:

\[ p = \frac{4}{3} \frac{Eh_0}{R_0} \left( \frac{R_0^2}{R} - \frac{R_0^2}{R^2} \right) \]

where \( R \) is the tube radius, \( R_0 \) and \( h_0 \) are the radius and the wall thickness of the unloaded tube \((p = 0)\). One can conclude that, within the apparent uncertainty of experimental data, the results for all tested models are well approximated by straight lines that correspond to linear deformation with a constant value of the Young's modulus, \( E \). As expected, the model with the highest amount of diluent (with 10:1:10 mass fractions of the silicone rubber, the curing agent, and the diluent) was most extensible, but its strength was found to be insufficient. Taking into account the data obtained, the 10:1:7.5 mixture composition was selected for the making of the abdominal aorta bifurcation model, as it has a sufficient strength to withstand pulsating pressure of 40 - 110 mmHg (that was observed in experiments with the rigid model) and yields the desired level of expansion (10-25%) of the vessel model with diameter of 18 mm.

Additionally, the expansion of the finished abdominal aortic bifurcation model was assessed by a pulsating pressure on a blood flow simulator, shown in Figure 3, using a high-speed camera (50 frames / sec). As a result of assessments, in the range of Reynolds numbers at the inlet of the abdominal aorta \((Re = 500 – 2500)\) during the pulsation cycle the level of expansion of the vessel model within 5 - 25% was recorded.

![Figure 2](image-url)  

**Figure 2.** Determination of model extensibility: (a) the pressure effect on the pipe outer radius and (b) the pressure against the pipe expansion complex parameter at different compositions of the silicone mixture

4. Flow measurements in elastic and rigid vessel models

The study of the flow in the developed models of the abdominal aortic bifurcation was performed using the blood flow simulator (see figure 3) with a closed circuit of the blood-emulating fluid (36% water-glycerol solution, \( \rho = 1100 \text{ kg/m}^3 \), \( \mu = 0.00375 \text{ Pa s} \)). The acoustic tray 5 with the elastic model was filled with agar-agar gel (1 g per 100 ml of water) as a tissue-emulating material; in the rigid model experiments, the acoustic tray was filled with water. A centrifugal pump with a timer 2 was used to create a pulsating flow; the pulsation cycle was 1 s (0.3 s the pump actuated the liquid; the remaining 0.7 s the pump was off). The electromagnetic flow meter sensors 3 provided synchronous
registration of the liquid flow rate at the inlet and at both outlets of the model. Using the flow regulator 7, the flow regime was adjusted so that to provide the peak flow rate at the model outlets about 1 l/min. The maximum Reynolds number at the inlet of the rigid abdominal aorta was \( \text{Re}_{\text{max}} = \frac{V_{\text{max}} \rho}{\mu} = 692 \), Womersley number - \( \text{Wo} = D \sqrt{\frac{2 \pi \rho}{\mu T}} = 24.4 \), Strouhal number - \( \text{St}_{\text{max}} = \frac{f D}{V_{\text{max}}} = 0.14 \), where \( V_{\text{max}} \) – maximum velocity, \( D \) – diameter of vessel, \( \rho, \mu \) – density and dynamic viscosity of liquid, \( T \) – period of pulsation, \( f \) - frequency of vortex shedding. The simulated maximum of flow rate in the common iliac arteries in the rigid model was at the lower level of the physiological range.

**Figure 3.** Blood flow simulator: 1 - hydraulic circuit, 2 - pump with a timer, 3 - electromagnetic flow meter sensors, 4 - model of the abdominal aortic bifurcation, 5 - acoustic tray filled with water or agar-agar gel, 6 - ultrasonic scanner sensor, 7 - flow regulator, 8 - liquid drainage valves, 9 - pump, 10 - reservoir with blood-emulating liquid

Figure 4 shows the flow curves obtained for rigid and elastic models of the abdominal aortic bifurcation. For the rigid model (figure 4, a), the flow rate variation during the cycle was similar to the physiological one, with clearly seen phases of acceleration (0.3 s) and deceleration (0.7 s) of the flow that corresponded to the pump operation and rest periods. The flow rates at the inlet and the outlets of the model varied almost synchronously. The flow regime was stable and repeated cycle-after-cycle.

For the elastic model (figure 4, b), contrary to the expectations, the flow pattern had completely changed. Intense parasite oscillations of the flow rate were observed, the model inlet and outlet flow curves were quite different, successive cycles did not really repeat each other. At the same level of maximum outlet flow in the rigid and elastic models, the maximum flow rate at the inlet of the elastic model was twice as much. Thus, an attempt to form similar flow conditions for a rigid and elastic model (with the intention of further studying the flow structure by the ultrasonic Doppler method) had failed.

Apparently, the cause of the failure was the interaction of the elastic model with a relatively rigid closed hydraulic circuit of the blood flow simulator. In the rigid model experiment, the pump creates an actuating pressure that accelerates the fluid in the entire circuit, and when the pump gets off the
fluid still moves by inertia and gradually decelerates due to friction. In the case of the elastic model, when the pump turns on, an additional pressure is also created at the inlet of the model but, instead of accelerating the fluid throughout the hydraulic circuit, it leads to a rapid expansion of the model (that correlates with a high peak of the inlet flow rate at the beginning of the cycle, see figure 4, b). At that, since the hydraulic circuit of the blood flow simulator is rigid and the amount of fluid in the closed circuit does not change, the expansion of the inlet part of the elastic model is accompanied by narrowing of its outlet branches (that indicates a local depression in this part of the model). In turn, reduction of the model exit area leads to a sharp increase of the inlet pressure and, accordingly, to a decrease in the flow supplied by the pump. Then the process repeats that leads to intense parasite oscillations of the flow rate (especially at the model inlet).

It is worth mentioning that, in experiments with a rigid model, one is able to recognize similar (though less intense) parasitic oscillations due to the presence of relatively rigid but actually flexible pipes in the hydraulic circuit of the blood flow simulator. In particular, in figure 4, a, there is a clear peak in the flow curves at the very beginning of each cycle (that was originally treated as some disturbance of the signal of the electromagnetic flow meter sensor).

It is clear that the described scenario of development of intense parasite flow rate and pressure oscillations in the elastic vessel model has no relation to the real blood flow pulsations. So, we had to postpone the planned study of the flow structure in the elastic model of the abdominal aortic bifurcation using the ultrasonic Doppler method. Currently, we consider the possibility of solving the problem of parasite flow oscillations via installation of dampers upstream of the actuating pump and/or downstream of the model. Installing dampers will compensate for the interaction of a relatively rigid hydraulic circuit with the elastic model and will allow to create a flow curve with clearly phases of acceleration and deceleration of the flow and to suppress parasite flow oscillations at the inlet and at the outlets of the model. After that the study will be resumed.

5. Conclusion
The rigid and elastic models of the average configuration, including the bifurcation region of the abdominal aorta and iliac arteries, were designed and manufactured. The elastic model was made by casting the silicone mixture into the mold. The auxiliary experiments were carried out to study the elastic properties of the silicone mixture and the selection of the composition. As a result of the study,
the composition of the silicone mixture was selected for the manufacture, which provides a change in the vessel diameter (18 mm) within 10 - 25% and corresponds to the physiological range of vessel expansion. The first attempts to simulate the flow in rigid and elastic models were carried out, the technical features of the experimental setup and the ways of eliminating the modeling shortcomings was noted.

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