Multiscale Study on Hemodynamics in Patient-Specific Thoracic Aortic Coarctation

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Abstract. In this challenge, we intended to mimic the patient's cardiovascular system by using 0D-3D connected multiscale model. The purpose of the multiscale analysis is to find out the appropriate boundary conditions of the innominate artery (IA), left common carotid artery (LCA) and left subclavian artery (LSA) in the local 3D computational fluid dynamics simulation. Firstly, a lumped parameter model(LPM) of the patient's circulatory system was established which could mimic both the rest and stress conditions by adjusting parameters like elastance function of the heart and the peripheral resistance, since that administering is oprenaline leads to the patient's heart beat rate and peripheral resistance changes. Secondly, the values of parameters in the LPM were slightly revised to match the following conditions: 1. provided pressure and flow rate curves, 2. provided blood distribution ratio of the AcsAo, IA, LCA and LSA. Finally, we got the outlet conditions of the IA, LCA and LSA, and then connecting the 0D model and the 3D model at each time step. As the results, we got the streamlines, pressure drop through the coarctation, pressure gradient, and some other parameters by coupled multiscale simuation.

Keywords: Thoracic aortic coarctation, Hemodynamics, Computational fluid dynamics, Multiscale simulation, Pressure gradient.

1 Introduction

Coarctation of the aorta (CoA) usually occurs in the thoracic segments of the aorta which often leads to hypertension. A lot of researches have been performed to study the hemodynamic parameters such as the flow velocity and wall pressure of the morbid aorta which have shown that those parameters are closely related to vascular geometry. Therefore, hemodynamic simulation can be performed and then applied to

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predict the fluid field through a thoracic aorta in presence of the coarctation. In this paper, a lumped parameter model (LPM)[1] of circulatory system was firstly established in order to evaluate the hemodynamic behaviors inside a thoracic aortic coarctation model which is in lack of enough boundary conditions. In accordance with the provided information, the values of parameters in the LPM were properly adjusted to fit both the rest and stress conditions.

Two schemes were planned to get the pressure gradient through the coarctation. One option is to carry out the stand-alone 3D simulation by setting the LPM results as boundary conditions. The other option is to perform multiscale simulation by coupling the LPM and the local 3D model at each time step[2], which is regarded to be able to better reproduce the boundary conditions and represent the interactions between the local geometry and the global circulatory system. As the first results, we performed the stand-alone 3D simulation by setting the 0D simulation results (the volume flow rate curves at the outlets of IA, LCA and LSA) as the boundary conditions which are not given. Then, we performed multiscale simulation to obtain the coupled solutions.

2 Method

(a) Construction of the lumped parameter model

The LPM of the circulatory system was constructed as shown in Fig. 1. In accordance with the given waveforms and flow distribution ratio of the IA, LCA, LSA and DesAo, the values of parameters in the model were properly adjusted to fit both the rest and stress conditions, so that the waveforms of the ascending aortic flow rate, pressure and blood distribution ratio are similar to the curves and values measured from the clinical practice. The comparisons between the original waveforms and LPM waveforms are depicted in Fig. 2, and Table 1 gives the values of parameters in the LPM for both the rest and the stress conditions.

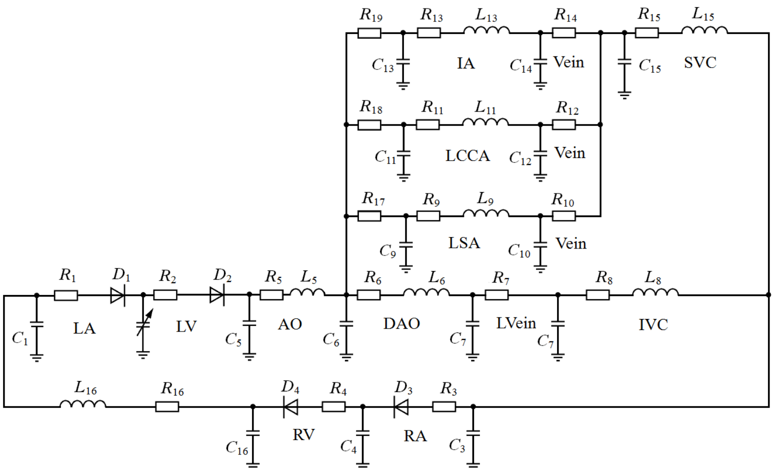

Fig. 1. The LPM of circulatory system (AO: aorta, DAO: descending aorta, IVC: inferior vena cava, SVC: superior vena cava, LA: left atrium, RA: right atrium, LV: left ventricle, RV: right ventricle)

Table 1. The values of Parameters in the LPM for the rest and stress conditions

R(mmHg·s·ml^{-1})			C(ml^{-1}mmHg)			L(mmHg·s^{2}·ml^{-1})		
	Rest	*Stress*		*Rest*	*Stress*		*Rest*	*Stress*
R1	0.01	0.0091	C1	0.2	1.6			
R2	0.01	0.0091						
R3	0.01	0.004	C3	4.8	4.74			
R4	0.001	0.004	C4	4.2	4.78			
R5	0.0001	0.1172	C5	0.82	0.028	L5	3.0E-3	2.0E-5
R6	0.06	0.0352	C6	0.014	0.58	L6	1.75E-3	1.5E-5
R7	0.45	0.0105	C7	1.71	0.081			
R8	0.8	0.008	C8	0.505	0.0105	L8	5.0E-4	5.0E-5
R9	1.799	0.192	C9	0.1	0.01	L9	5.0E-4	5.0E-4
R10	0.385	0.0185	C10	0.01	0.001			
R11	2.439	0.514	C11	0.1	0.01	L11	5.0E-4	5.0E-4
R12	0.385	0.0185	C12	0.01	0.001			
R13	0.87	0.027	C13	0.1	0.01	L13	5.0E-4	5.0E-4
R14	0.485	0.0185	C14	0.01	0.001			
R15	1.735	0.001735	C15	0.505	0.105	L15	5.0E-4	5.0E-4
R16	0.0105	0.02	C16	0.2	3.08	L16	5.0E-4	5.0E-4
R17	0.02	0.08						
R18	0.018	0.08						
R19	0.019	0.08						

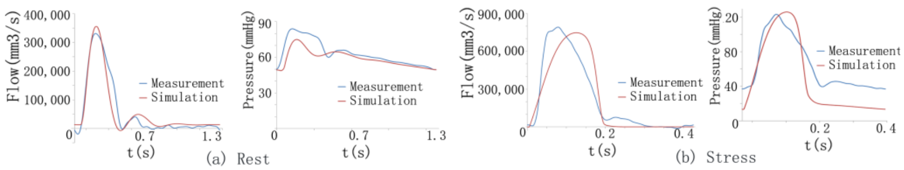

Fig. 2. The comparisons between the original waveforms and the LPM results

The purpose of setting up the LPM is to obtain the unknown boundary conditions at the outlets of IA, LCA and LSA. Once the total flow and flow distribution ratio of each outlet under both rest and stress conditions could match the given values, at the same time the waveforms of ascending aortic flow rate and pressure could be similar to the provided curves, then we assume that the LPM could mimic the patient's cardiovascular system to some extent.

Therefore, the LPM could be coupled with the 3D model. The provided inflow waveform was added as inlet boundary condition. The flow rate waveforms of IA, LCA, LSA and the pressure of descending aorta were calculated from the multiscale simulation.

The structure of the coupled multiscale model is shown in Fig. 3. The flow rate of IA, LCA, LSA and the pressure of descending aorta generated by the LPM are set as boundary conditions, and the pressure of IA, LCA, LSA and the flow rate of the descending aorta calculated by the 3D model are passed to the LPM for the calculation at next time step. The data exchange is executed in every time step.

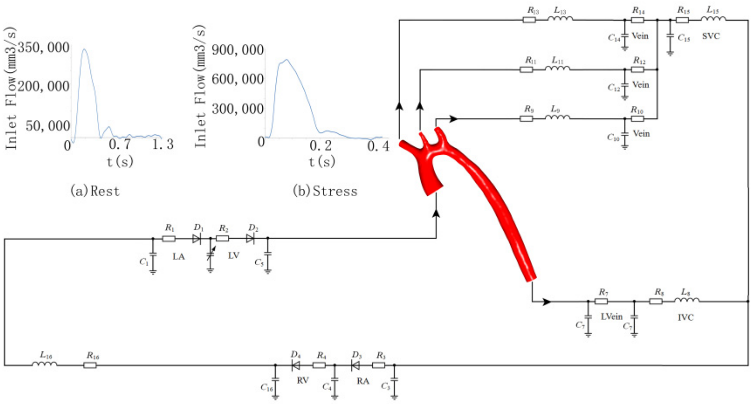

Fig. 3. The structure of the coupled model and the inlet boundary condition

(b) Construction of the finite element model

The provided STL file of arterial model was imported into ANSYS ICEMCFD13.0. Volume meshes were generated by using mesh types of structural hexahedral. The boundary layer was not treated specially. The total numbers of the element and node are 617786 and 480032 respectively. The mesh of a cross-section is shown in Fig. 4.

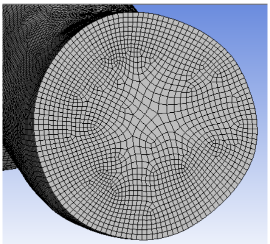

Fig. 4. The mesh in a cross-section

(c) Numerical simulation

Volume mesh file was imported into ANSYS CFX 13.0 to perform the numerical simulation. The following assumptions were employed in this numerical study: non-permeability, rigid wall; incompressible Newtonian fluid; pulsatile and laminar flow. Viscosity and density of blood are 0.004Pa•s and 1000kg/m^3 respectively.

The discrete form of differential equations governing the blood flow was upwind scheme. Residual convergence criteria of mass and momentum were set to 10^{-5}. The time step in calculation was 0.005s. Run mode in CFX is "PVM Local Parallel". Convergent solutions were obtained after 3 cycles.

3 Result

(a) The flow rates and pressure of IA, LCA and LSA

The flow rates of IA, LCA and LSA are set as the boundary conditions in the 3D model, and the pressures of them are the input of the LPM. Both of them can only be determined after the multiscale simulation. All of them are shown in Fig. 5.

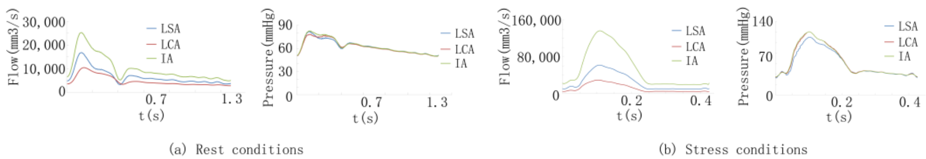

Fig. 5. The flow rates and pressure of IA, LCA and LSA

(b) Pressure drop and pressure gradient

As shown in Table 2, both the peak and time-averaged pressure drop and volume-average pressure gradient through the coarctation at both rest and stress conditions (the planes for calculating the pressure drop and the pressure gradient are depicted in Fig. 6) were obtained.

Table 2. Both the peak and time-averaged pressure drop and volume-averaged pressure gradient

	Pressure Drop(mmHg)		Volume-averaged Pressure Gradient(Pa/m)	
	Peak	Time-averaged	Peak	Time-averaged
Rest conditions	5.69	0.1094	135799	22869.5
Stress conditions	26.034	-0.53663	275898	91339.49

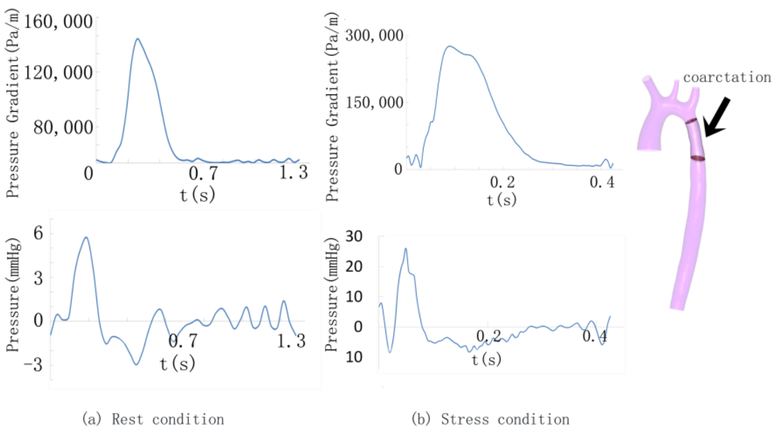

Fig. 6. The pressure drop and volume-averaged pressure gradient through the coarctation and the location of the coarctation

(c) The flow distribution ratio and the pressure proximal to the coarctation
The values of parameters in the LPM were properly adjusted in order to match both the total flow and percentage of ascending aortic flow through the various branches under both rest and stress conditions. As a reference for comparison, table 3 gives the specific distribution values of each opening.

Table 3. The flow distribution ratio and total flow of the various branched in the aortic model under both rest and stress conditions

			AscAo	Inno-minate	LCC	LS	Dia-phAo
Rest Condi-tions	calcul ated	Total Flow(L/min)	3.71	0.607	0.287	0.397	2.419
		% AscAo	100	16	8	11	65
	pro-vided	Total Flow(L/min)	3.71	0.624	0.312	0.364	2.41
		% AscAo	100	17	8	10	65
Stress Condi-tions	calcul ated	Total Flow(L/min)	13.53	3.277	0.6785	1.4901	8.084
		% AscAo	100	24	5	11	60
	pro-vided	Total Flow(L/min)	13.53	3.355	0.6875	1.4575	8.03
		% AscAo	100	25	5	11	59

The systolic, diastolic, and mean pressures of the ascending aorta were measured from the CFD results. Table 4 gives those values which are very close to the value measured clinically.

Table 4. The systolic, diastolic, and mean pressure proximal to the coarctation

Pressure(mmHg)		Systolic	Diastolic	Mean
Rest Conditions	calculated	83.98	49.80	63.36
	provided	83.92	49.68	63.35
Stress Conditions	calculated	118.44	36.38	61.75
	provided	123.35	36.77	64.30

(d) Streamlines, Pressure and Pressure Gradient
The typical moments of 0.255s, 0.83s (under rest condition) and 0.095s, 0.42s (under stress condition) which were the highest and lowest velocity peak point respectively were selected to demonstrate the 3D simulation results.

(i) Streamlines
Figure 7 shows the streamlines at the time of maximum velocity under both the rest and the stress conditions. Results show that the maximum velocity region appears through the coarctation in both conditions.

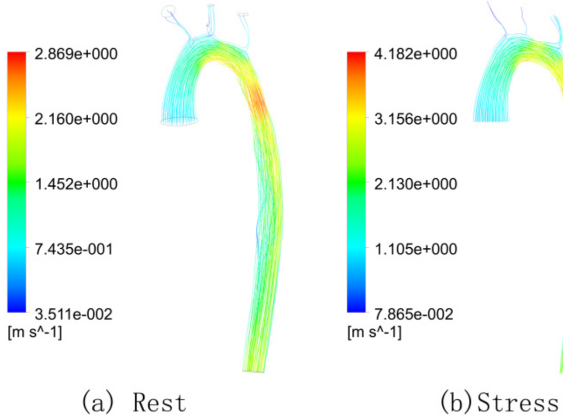

Fig. 7. The streamline at the time of maximum velocity in rest and stress condition

(ii) Pressure and Pressure Gradient

The contours of pressure and pressure gradient are showed in Fig. 8 and Fig. 9.

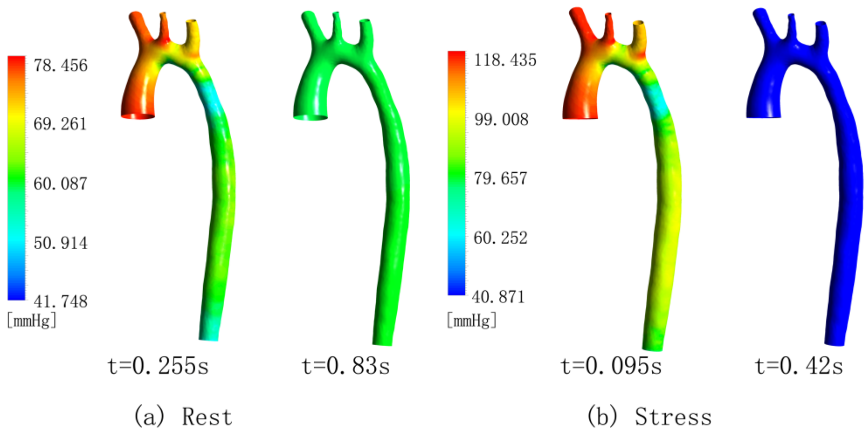

Fig. 8. The pressure at two typical moments under both rest and stress conditions

It can be seen from the pressure contour that, at the time of highest peak velocity, the pressure difference between the coarctation area and the regions before or after the area is dramatic. However in the case of lowest peak velocity, the pressure difference is not that much.

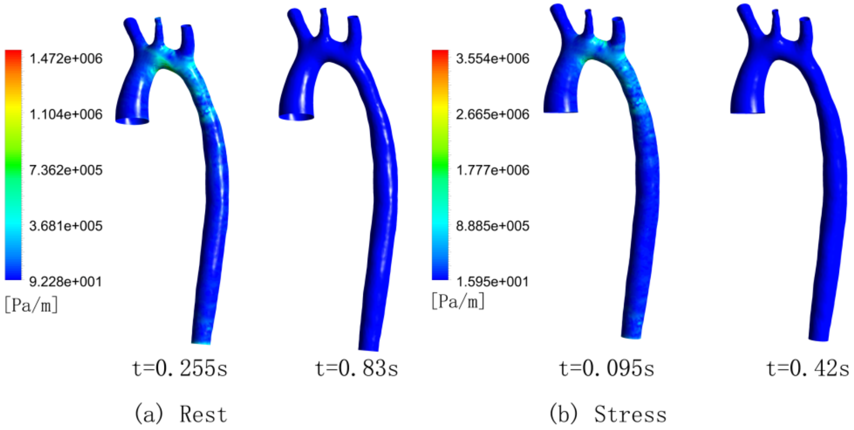

Fig. 9. The pressure gradient at two typical moments under both rest and stress conditions

According to the contour of pressure gradient distribution, it can be found that the pressure gradient through the coarctation is relatively high. However, the highest pressure gradient is located at the bottom of the aortic arch, and the pressure gradient at the time of minimum velocity is much lower than that at the time of maximum velocity.

4 Conclusion

In this paper, the multiscale simulation was presented by coupling the LPM and the 3D model, which is able to better reproduce the boundary conditions and represent the interactions between the local geometry and the global circulatory system.

The results showed that the flow distribution through the various branches and the proximal systolic, diastolic, and mean pressure match the provided data very well.

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