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Longitudinal Impedance Tomography for Blood Pressure Characterization of Valve Deformation

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ABSTRACT

Aorta is formed in a dynamic environment which gives rise to imbalances between many forces that tend to extend the diameter and length. Furthermore, internal forces tend to resist this extension. Impedance tomography can show this imbalance to stimulate the stenosis of aortic valve, growth of the elastic, collagen and to effectively reduce the stresses in the underlying tissue. In blood flow, auscultation noises occurred and in the echocardiography decrease in left ventricular ejection speed can be observed. In this paper, we have modeled an aorta based on anatomical studies to simulate natural, 20% and 30% stenosis as usual heart disease to early diagnosis. Valve deformation causes different impedance tomography in 3D mesh of aorta as blood pressure. Remodeling of aorta and its flow is found when a cylindrical slice of the fully retracted blood aorta is cut longitudinally through the wall.

Key words: *Aorta, Aortic Valve, Electric Impedance, Blood Pressure, Tomography, Hemodynamics*

INTRODUCTION

The forces needed for the motion of the blood in the body are provided by the heart that serves as a four-chambered pump which propels the blood around the circulatory system. The flow in curved tubes exhibits secondary flow patterns perpendicular to the axis of the tube. The strength of this secondary flow depends on the curvature of the tube which is expressed in another dimensionless parameter. At last it will be shown that the flow in branched tubes shows a strong resemblance to the flow in curved tubes.

The mean pressure in the systemic circulation is approximately 13 KPa, which is more than three times the pressure in the pulmonary system, and also the thickness of the left ventricular muscle is much larger than of the right ventricle.

Arterial contraction, induced by a stimulus for muscle contraction of the sinoatrial node, causes filling of the ventricles. In the left heart when the mitral valve is opened it offers very low resistance. The aortic valve is closed. In a short time after this, at the onset of systole the two ventricles contract simultaneously controlled by a stimulus generated by the atrioventricular node. The mitral valve closes and a sharp pressure rise in the left ventricle occurs. This ventricular pressure exceeds the pressure in the aorta, the aortic valve opens and then blood is ejected into the aorta. The ventricular and aortic pressure rise at first and then fall as a result of a combined action of ventricular contraction forces. The resistance and compliance of the systemic circulation fall too. Due to this pressure fall, the aortic valve closes and rapidly the pressure in the ventricle drops, the mitral valve opens while the heart muscle relaxes (sounds diastole). Since both the blood flow velocities as well as the geometrical length scales are relatively large, the fluid mechanics of the heart is determined by inertial forces which are in equilibrium with pressure forces. An extended overview of physiological processes that are enabled by heart and virtue of the cardiovascular system can be found in standard textbooks on physiology like Guyton.[1]

The other studies on cardiac output can be observed in impedance cardiography papers based on simple geometries.[2] The other way to analysis dynamic flow in the heart has been done to introduce basic noninvasive methods to diagnosis heart behavior in its cycle.[3-6]

Pressure, in the aorta, changes with increasing distance from the heart. The peak of the pressure pulse delays downstream with a certain wave speed indicating wave propagation along the aorta.

This wave phenomenon is a direct consequence of the dispensability of the arterial wall, allowing a partial storage

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of the blood injected from the heart due to an increase of the pressure and the elastic response of the vessel walls. The cross-sectional area of the vessels depends on the pressure difference over its wall.[7,8]

This pressure consists of several parts. First, there exists a hydrostatic part proportional to the density of the blood inside ρ the gravity force g and the height h. The pressure is composed of a time-independent part p_0 and a periodic time-dependent part p. Hence, the transmural pressure can be written as:

$$p_{tr} = \rho g h + p_0 + p \tag{1}$$

Due to the complex nonlinear anisotropic and viscose elastic properties of the arterial wall, the relation between the pressure and the cross-sectional area of the vessel is mostly nonlinear and can be complicated. Important quantities with respect to this relation, used in physiology, are the impedance (compliance) or alternatively the dispensability of the vessel.

The impedance C is defined as:

$$C = \frac{\partial A}{\partial P} \tag{2}$$

Where, A is sectional artery area and *P* is blood pressure. The dispensability D is defined by the ratio of the impedance C and the cross-sectional area and hereby is given by:

$$\mathrm{D} = \frac{\partial A}{\partial p} \times \frac{1}{\mathrm{A}} = \frac{\mathrm{C}}{\mathrm{A}} \tag{3}$$

These quantities will be related to the material properties of the arterial wall. For thin walled tubes, with wall thickness h and radius a, without longitudinal strain, e.g. it can be derived that:

$$D = \frac{2a}{\mathrm{h}} \times \frac{1-\mu^2}{\mathrm{E}} \tag{4}$$

Here E Young's modulus and μ denotes Poisson's ratio. We can see that besides the properties of the material of the vessel (E, μ) also geometrical properties (a, h) play an important role.

The value of a/h varies strongly along the arterial tree. The veins are more distensible than the arteries. Mostly the pressure-area relationship that is, the compliance or dispensability, of the arteries or veins that are considered have to be determined from experimental data. The compliance changes with the pressure load since at relatively high pressure, the collagen fibers in the vessel wall become stretched and prevent the artery from further increase of the circumferential strain. Flow is driven by the gradient of the pressure and hereby determined by the propagation of the pressure wave.[9-11] The pressure wave will have a pulsating periodic character. The flow phenomena can be described between steady and unsteady part. Often it is assumed that the unsteady part can be described by means of a linear theory, so we can introduce the concept of pressure and flow waves which are super positions of several harmonics:

As mentioned the flow of blood is driven by the force acting on the blood induced by the gradient of the pressure. The relation of these forces to the resulting motion of blood is expressed in the longitudinal impedance:

$$Z_L = \frac{\partial \hat{p}}{\partial z} / \hat{q} \tag{5}$$

The longitudinal impedance is a complex number defined by complex pressures and flows. It can be calculated by numerical analysis of the pressure. As it expresses the flow induced by a local pressure gradient, it is a property of a small segment of the vascular system and depends on local properties of the vessel. The impedance plays an important role in the characterization of vascular segments.[7]

It can be measured by a simultaneous determination of the pressure in the vessel with a simulation of flow type in some routine stenosis in aorta.

The longitudinal impedance as first important quantity will be derived mathematically using a finite element numerical theory for aorta in rigid and distensible tubes. A second important quantity is the input impedance defined as the ratio of the pressure and the flow at a specific cross-section of the vessel:

$$Z_i = \frac{\hat{p}}{\hat{q}} \tag{6}$$

The input impedance is not a local property of the vessel itself but a property of a specific site in the vascular system. If some input condition is imposed on a certain site in the system, then the input impedance depends on the properties of the entire vascular tree distal to the cross-section where it is measured. In general the input impedance at a certain site depends on both the distal and proximal vascular tree.

The compliance of an arterial segment is characterized by the transverse impedance defined as:

$$Z_T = \frac{\hat{p}}{\frac{\hat{q}}{z}} \approx -\frac{\hat{p}}{i\omega \hat{A}} \tag{7}$$

Where, P is pressure, q is flow and z are special impedance of tissue. This paper expresses impedance tomography of different situations of aorta valve stenosis natural 20% and 30%. The flow quantity decreased due to the storage of the vessel caused by the radial motion of its wall at a given pressure (note that $i\omega\hat{A}$ represents the partial time derivative $\frac{\partial A}{\partial t}$). Impedance tomography as defined here can be very useful in the analysis of propagation of pressure.

In this paper, an impedance tomography introduction of pressure in cardiovascular fluid mechanics is given. Based on cardiovascular anatomical knowledge and medical studies of this disease, a simple model has been derived that the system is characterized by a finite element in aorta and flow impedance. In this model, it is ignored that the fluid mechanics and impedance tomography of the cardiovascular system is characterized by complex geometries and complex constitutive behavior of the blood and the vessel wall. The vascular system is strongly time-dependent. In the aorta, the flow will be determined by blood characteristics.[10-12]

PHYSICAL MODEL

In the early exit of the aortic from left ventricle, the aortic valve with a length of 18 mm and a diameter of 25 mm is located. After leaving the left ventricle, aorta is divided into three parts: The ascending aorta, aortic arch, and the descending aorta. Coronary arteries are derived from the beginning of the aorta (major ascending artery) and, therefore, are the first arteries that receive a large amount of oxygenated blood. The brachia cephalic arteries with a diameter of 8/8 mm are separated from the aortic arch. The second main artery that is separated from the aorta is the left carotid artery with a diameter of the 8/8 mm, and the third one is the arteries under the collarbone with a diameter of 8/8 mm. The thoracic aorta (Thoracic) in the chest area with a diameter of 25 mm and a length of 75 mm forms the rest of the descending aorta, and it is called an abdominal aorta in the abdominal area. The right and left intercostal arteries, bronchus, and esophagus get separated from the thoracic aorta and the celiac artery, renal artery, mesenteric (intestinal) and … get separated from the abdominal aortic.[1,5,6]

COMPUTATIONAL MODELING AND GOVERNING EQUATIONS

We survey the properties of fluid and analysis of laminar or turbulent of flow with continuity with Navier-Stokes equations used to solve this problem.

Flow Behavior

Fluid is said to substance that deforms continuously under the shear stress although how small the shear stress may be.

When Newtonian fluid which shear stress is directly proportional to the rate of strain, it following Newton's law.[3] For example water, gas, air and oil are as this method.

$$\tau_{xy} = \tau_{yx} = \mu\left(\frac{\partial v}{\partial x} + \frac{\partial u}{\partial y}\right) \tag{8}$$

$$\tau_{zx} = \tau_{xz} = \mu\left(\frac{\partial w}{\partial x} + \frac{\partial u}{\partial z}\right) \tag{9}$$

$$\tau_{yz} = \tau_{zy} = \mu\left(\frac{\partial v}{\partial z} + \frac{\partial w}{\partial y}\right) \tag{10}$$

In upper equations, *x, y, z* subscripts are directions in Cartesian-Coordinate-System, τ is shear stress, μ is dynamic viscosity and *v, u, and w* are velocity vectors in *x, y, z* directions.[8]

Newtonian behavior assumption is accepted for blood from many researchers.[9,10]

In fluid mechanics, a dimensionless quantity namely Reynolds number is used to determine laminar or turbulent flow. Reynolds number is the ratio of internal forces to viscous forces and introduce as follow:

$$Re = \frac{\rho v D_H}{\mu} = \frac{v D_H}{\vartheta} = \frac{Q D_H}{\vartheta A} \tag{11}$$

$$D_H = \frac{4A}{P} \tag{12}$$

D_H is hydraulic diameter, P is wetted perimeter, Q is volumetric flow rate, A is cross-sectional area, v is mean flow velocity, μ is dynamic viscosity, v is kinematic viscosity and ρ are density.

If the flow consider in part of a pipe as seen in Figure 1, the passer flow can have three situations that specify by Reynolds number:

Flow is laminar if

$$Re_{pipe} < 2300$$

Flow is in a transitional situation if

$$Re_{pipe} = 2300 - 4000$$

Flow is turbulent if

$$Re_{pipe} > 4000$$

The blood flow in this paper is turbulent in natural situation and transitional in 20% and 30% aortic stenosis inlet valve as respects to value of blood dynamic viscosity, blood density, blood velocity of flow in aortic inlet as $\mu = 0.0035(Pa.sec)$, $\rho = 1050\left(\frac{Kg}{m^3}\right)$ and $V_i = 0.55\left(\frac{m}{sec}\right)$ an aortic inlet diameter as Natural 25 mm, 20% aortic stenosis inlet valve 20 mm and for 30% aortic stenosis inlet valve 17.5 mm.

$$D_H = \frac{4A}{P} = \frac{4 \times \pi \times r^2}{2 \times \pi \times r} = 2r = D$$

Reynolds number in the natural situation is:

$$Re = \frac{\rho v D_H}{\mu} = \frac{\rho v D}{\mu} = \frac{1050 \times 0.55 \times 25 \times 10^{-3}}{0.0035} = 4125 > 4000$$

So, the flow is turbulent.

Reynolds number in 20% aortic stenosis inlet valve is:

$$Re = \frac{\rho v D_H}{\mu} = \frac{\rho v D}{\mu} = \frac{1050 \times 0.55 \times 20 \times 10^{-3}}{0.0035} = 3300$$

$$2300 < Re < 4000$$

So, flow is in a transitional situation.

Reynolds number in 30% aortic stenosis inlet valve is:

$$Re = \frac{\rho v D_H}{\mu} = \frac{\rho v D}{\mu} = \frac{1050 \times 0.55 \times 17.5 \times 10^{-3}}{0.0035} = 2887.5$$

$$2300 < Re < 4000$$

So, flow is in a transitional situation.

Boundary, Flow and Fluid Conditions

The boundary condition of governing equations is:

Flow is in steady state if:

$$\frac{\partial \rho}{\partial t} = \frac{\partial v_r}{\partial t} = \frac{\partial v_\theta}{\partial t} = \frac{\partial v_z}{\partial t} = 0 \qquad (13)$$

Because the fluid is incompressible Hence, continuity equation can be simplify to the following:

$$\frac{\partial \rho}{\partial t} + \frac{1}{r}\frac{\partial}{\partial r}(r v_r) + \frac{1}{r}\frac{\partial}{\partial \theta}(v_\theta) + \frac{\partial}{\partial z}(v_z) = 0 \qquad (14)$$

Velocity in aortic inlet is:

$$U_{inlet} = 0.55\, m/s$$

Furthermore, we have assumed that the flow is continuum and also Newtonian based on its condition. Flow is turbulent for natural situation of aortic inlet valve, Flow is in a transitional situation for 20% and 30% aortic stenosis inlet valve, Blood properties are constant and gravity effect is considered.

Numerical Procedure

The figures of aorta are plotted in 3D for each natural and 20% and 30% aortic stenosis inlet valve situations. Governing equations are discredited by finite volume approach. It is necessary to controlling the location of surfaces in finite volume approach after gridding.

RESULTS OF SIMULATION

Aorta in Natural Situation

Analysis of pressure

According to Figure 2 3D pressure contour, maximum pressure is referring to inlet and primary of first three outlets

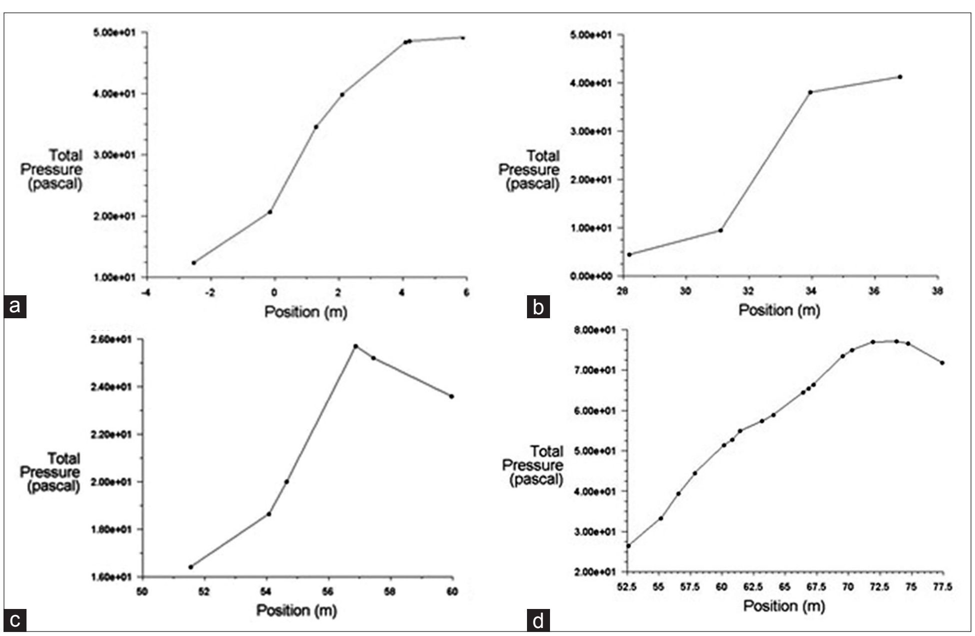

Figure. 1: Pressure distribution of aortic blood flow for 20% stenosis inlet valve situation in outlet cross-sectional. 1,2,3 and 4 in coordinate of $-R \le r \le +R$ (a, b, c and d)

and minimum pressure is referring to walls. High velocity in the main inlet, separating first three outlets from the main flow and aortic arch are cause of this pressure distribution. Furthermore, according to Figure 3, 3D pressure contours in outlets, it is clear that blood flow pressure in proximity of wall is minimum and pressure is increasing with receding from wall that this pressure distribution is emanated from velocity distribution. Based on better showing of images and a good demonstration of blood behavior, codes of images may be differing.

According to Figure 2, pressure diagrams for each four outlets cross-sectional, pressure is minimum in proximity of the wall. Pressure is increasing with receding from wall. Maximum pressure in outlet1 is 138 Pa, in outlet2 is 120 Pa, in outlet3 is 97 Pa and in outlet 4 is 138 Pa. In Figure 3 we have 2D pressure contours of aortic blood flow for natural situation. In Figure 4 pressure distribution of aortic blood flow for natural situation in outlet cross-sectional has been display.

Aorta with 20% Stenosis Inlet Valve

Analysis of pressure

According to Figure 5, 3D pressure contour, maximum pressure is refer to inlet and primary of first three outlets and minimum pressure is refer to walls. High velocity in the main inlet, separating first three outlets from the main flow and aortic arch are cause of this pressure distribution. Also, according to the 3D pressure contours in outlets, it is clear that blood flow pressure in proximity of wall is minimum and pressure is increasing with receding from wall that this pressure distribution is emanated from velocity distribution.

According to the 2D pressure contours and pressure diagrams for each four outlets cross-sectional, pressure is

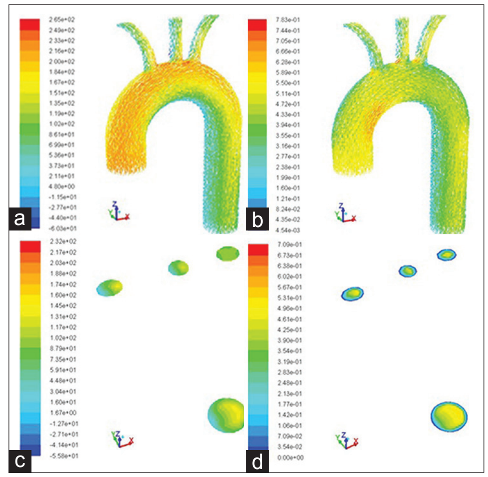

Figure. 2: 3D pressure/velocity simulation of aortic blood flow for natural situation with (Pa) unit (a/b) 3D pressure/velocity contour of aortic blood flow for natural situation in four outlets with (Pa) unit (c/d)

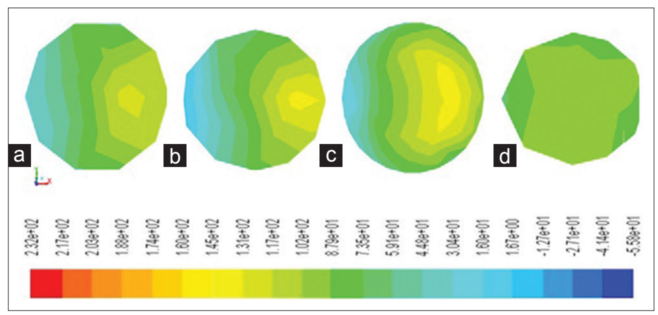

Figure 3: 2D pressure contours of aortic blood flow for natural situation, (a) outlet1, (b) outlet2, (c) outlet3 and (d) outlet4 with (Pa) unit

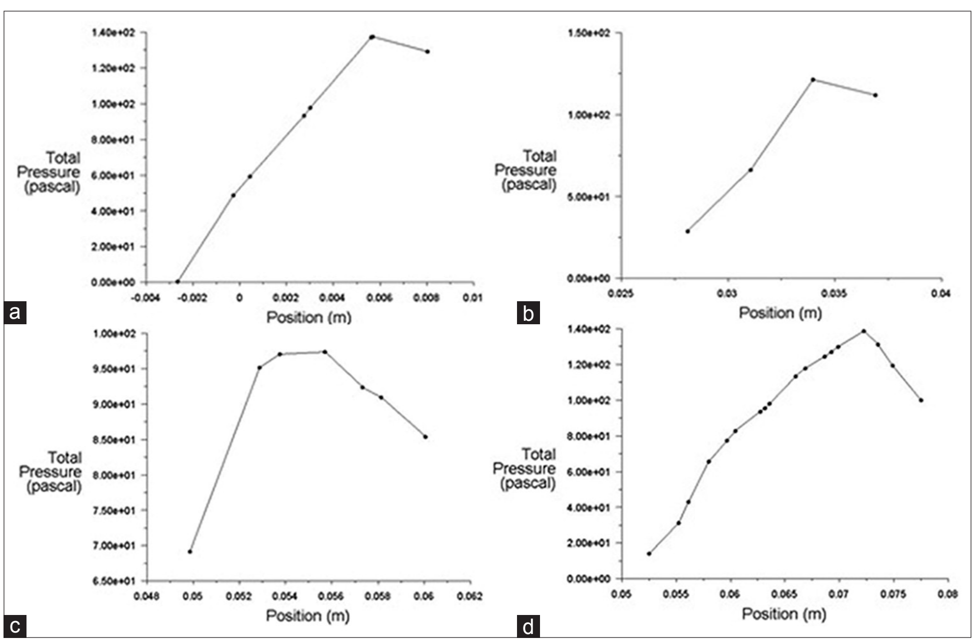

Figure 4: Pressure distribution of aortic blood flow for natural situation in outlet cross-sectional 1, 2, 3 and 4 in coordinate of $-R \leq r \leq +R$ (a, b, c and d)

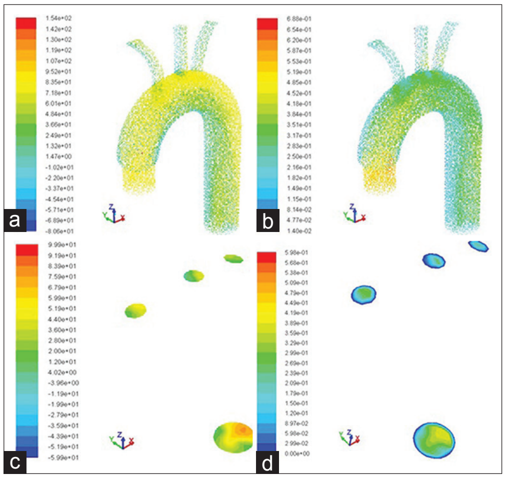

Figure 5: 3D pressure/velocity simulation of aortic blood flow for 20% stenosis inlet valve situation with (Pa) unit (a/b) 3D pressure/velocity contour of aortic blood flow for 20% stenosis inlet valve situation in four outlets with (Pa) unit (c/d)

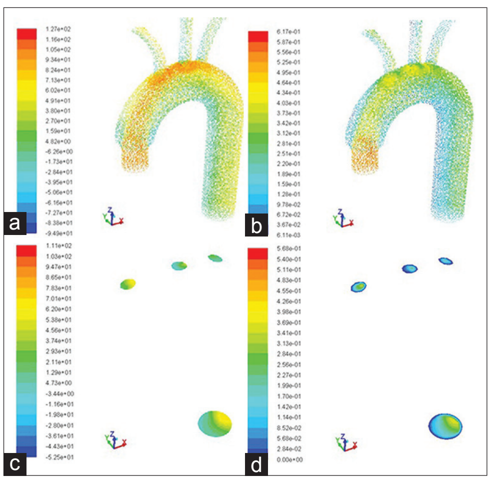

Figure 7: 3D pressure/velocity simulation of aortic blood flow for 30% stenosis inlet valve situation with (Pa) unit (a/b) and 3D pressure/velocity contour of aortic blood flow for 30% stenosis inlet valve situation in four outlets with (Pa) unit (c/d)

minimum in proximity of wall and pressure is increasing with receding from wall. Maximum pressure in outlet1 is 49 Pa, in outlet2 is 42 Pa, in outlet3 is 25.6 Pa and in outlet 4 is 70.75 Pa. In Figure 6 we have 2D pressure contours of aortic blood flow for natural situation. In Figure 1 pressure distribution of aortic blood flow for natural situation in outlet cross-sectional has been display.

Aorta with 30% Stenosis Inlet Valve

Analysis of pressure

According to Figure 5, the 3D pressure contour, maximum pressure refers to inlet and primary of first three outlets and minimum pressure refers to walls. High velocity in the main inlet, separating first three outlets from the main flow and aortic arch are cause of this pressure distribution.

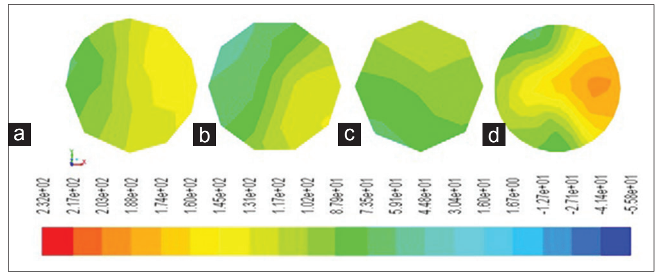

Figure 6: 2D pressure contours of aortic blood flow for 20% stenosis inlet valve situation, (a) outlet1, (b) outlet2, (c) outlet3 and (d) outlet4 with (Pa) unit

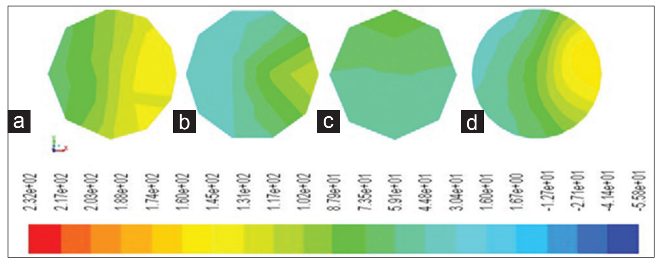

Figure 8: 2D pressure contours of aortic blood flow for 30% stenosis inlet valve situation, (a) outlet1, (b) outlet2, (c) outlet3 and (d) outlet4 with (Pa) unit

Furthermore, according to Figure 7, 3D pressure contours in outlets, it is clear that blood flow pressure in proximity of wall is minimum and pressure is increasing with receding from wall that this pressure distribution is emanated from velocity distribution. According to the 2D pressure contours and pressure diagrams for each four outlets cross-sectional, pressure is minimum in proximity of wall and pressure is increasing with receding from wall. Maximum pressure in outlet1 is 58 Pa, in outlet2 is 56 Pa, in outlet3 is 130 Pa and in outlet 4 is 66 Pa.

Figure 8, we have 2D pressure contours of aortic blood flow for natural situation. In Figure 9 pressure distribution of aortic blood flow for natural situation in outlet cross-sectional has been display.

As a fundamental consideration of blood flow simulation, some computational conditions have been achieved for any stenosis state. These results show that we can insure about this method as an earlier diagnosis of disease. For example in natural aorta we have heart rate 0.87 Hz (52 bpm), cardiac cycle 1.15s, cardiac output $9.17\times10^{-5}\ {m^3}/{s}$ and entrance flow $6.42\times10^{-5}\ {m^3}/{s}$.

CONCLUSION

Pressure distribution are delivered for each four outlet and plotted for each outlet cross-sectional in $-R \le r \le R$ with considering boundary conditions and solving governing equations for each situation of aorta (natural and 20% and 30% aortic stenosis inlet valve). Furthermore, 3D pressure contours for four outlets, 2D pressure contours

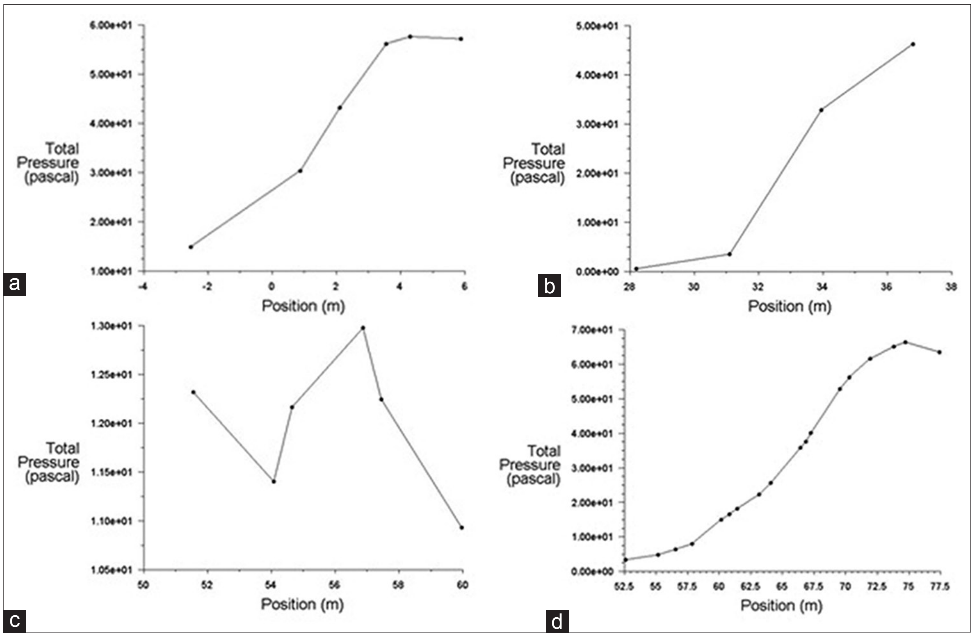

Figure 9: Pressure distribution of aortic blood flow for 30% stenosis inlet valve situation in outlet cross-sectional. 1, 2, 3, and 4, in coordinate of $-R \leq r \leq +R$ (a, b, c, and d)

for each four outlets and 3D pressure contours for whole aorta are plotted. These results could be a measurement of determining internal texture of aortic and its valve situation. In other words, analysis of blood flow pressure is a good approach to better earlier diagnosis of disease. For a person with medical symptoms which occurs because of nonnatural state blood velocity in cardiac output, like decreasing in head blood or decreasing in heart blood and its arrhythmia, one of the first guess is artery's stenosis. Hence, angiography suggested. Based on this modeling and man's symptoms the amount of stenosis forecasted. With this guess and simulation, blood velocity achieved and compared with physiological quantity that cause those symptoms. The algorithm iterates until these two amounts match together, and finally stenosis quantity achieved. Now the doctor can have choices between angiography, stent, balloon or even suggestion to a man to have only rest because the stenosis is not critical perhaps. Therefore, we could avoid of some nonurgency angiography and their restoration period. Furthermore, it could be helpful software to suggest the stent or balloon or even only angiography.

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BIOGRAPHIES

Zahra Vahabi Zahra Vahabi received her B.S. and M.S. in Biomedical Engineering. She is currently pursuating Ph.D. Her main research of interests cover a wide range of topics in digital signal processing, digital image processing and Biomedical signal and image analysis.

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