Computational fluid dynamics analysis of Effects of housing expansion angle, stroke volume and path length of the fiber bundles on function of the artificial lung

Zahra Mollahoseini, Bahman Vahidi[*]

Faculty of New Sciences and Technologies, University of Tehran, Tehran, Iran

ABSTRACT

An artificial lung can help patients waiting in line for a lung transplant or for heart bypass surgery as a respiratory aid. In this study, the incompressible and pulsatile Newtonian blood flow within a complete artificial lung model was investigated including inlet manifold, porous homogeneous medium, and outlet manifold. In this scale, the effect of variation of the expansion angle (15, 45 and 90 degrees), the stroke volume, the path length of the fibers on the artificial lung impedance was studied using computational fluid dynamics. The governing equations are discretized for numerical solution by finite volume method. Also, the turbulence model was selected by measuring the system impedance. In addition to the impedence, the shear stress distribution on the housing walls was investigated. The results showed that reducing the expansion angle, reducing the stroke volume and increasing the path length of the fibers will reduce the impedance of the system. The 45-degree model has been chosen as the appropriate model. Because not only its impedance is low, but also areas with low speed flow, which can lead to clot formation, are less than the 15-degree model. In order to have lower clot formation, it is better to have the artificial lung with the natural one in series.

KEYWORDS

Shear Stress, Pulsatile Flow, Oxygen Exchange, Stroke Volume, System Impedance.

[*] Corresponding author: Email: bahman.vahidi@ut.ac.ir

1. Introduction

More than 235 million people in the world suffer from lung diseases [1]. The mortality rate from acute respiratory distress syndrome (ARDS) is 64% [2]. Several research groups have been developing a total artificial lung (TAL) to treat chronic respiratory failure [3]. The right ventricle can pump blood to a low-resistance TAL. In this case, the need for a mechanical pump is eliminated in TALs [4].

In the present study, after examining a microscopic model of the artificial lung [5] and to continue the research of Schewe [6], designs of an artificial lung system were done with a macroscopic view. In these designs, according to the Reynolds number at the inlet of the device, the flow is turbulent at the inlet and outlet of the chamber, which causes vortex and thus increases the system impedance. The innovation of this research in comparison with previous researches is presented as follows: In the present analysis, by measuring the impedance of the system, the turbulence model was selected. In addition to the system impedance, shear stress on the walls is also a criterion for evaluating geometry to ensure that blood cells are not damaged. In this analysis, the impedance change was studied with parameters such as the length of the fiber bundle path and the stroke volume.

2. Methodology

The effect of the expansion angle on the blood flow through the TAL has been investigated for three values of Θ equal to 15, 45 and 90°. Three geometric models for the three angles were created as shown in Figure 1 using SOLIDWORKS software and the numerical solution was performed by Ansys software. In these models, blood flows through the inlet, enters the inlet manifold, passes through the fiber bundle located in the middle of the device, then goes to the outlet manifold and exits the outlet of the device. Inlet/outlet diameter, height and width of each TAL model are 0.016 m, 0.103 m and 0.102 m, respectively. Fiber bundle length, path length (distance that blood flow passes through the fiber bundle), frontal area (cross-section perpendicular to the bloodstream passing through the fiber bundle) and length of the inlet and outlet manifolds are 0.127 m, 0.038 and 0.016 0 square meters and 0.051 meters, respectively, for all models. Since the inlet/outlet diameter and the geometry of the chamber are constant, the length of the expansion/contraction section increases with decreasing Θ [6].

Transient and incompressible blood flow passes through TAL models. In this study, according to a previous similar study [6], blood was assumed to be Newtonian fluid. Also, for the inlet and outlet part, a turbulent flow with model k-ω is considered. The average Reynolds number for volumetric flow rates of 2, 4 and 6 l/min are 920, 1800 and 2760, respectively. The maximum Reynolds for these volumetric flow rates are 3000, 6100 and 9200, respectively. Blood viscosity and density in this section were considered 0.003 Pa and 1040 kg/m^3, respectively [6]. A homogeneous model has been used to model the environment occupied by the fibers. Thus, the whole area of the fiber bundles is modeled as porous periphery, with porosity of 0.75, permeability of 2.81 e-9 m^2 and with viscosity and fluid density of 0.003 Pa.s and 1040 kg/m^3, respectively. The amount of permeability (k) was obtained using the Darcy relation and compatible artificial lung information [7] (Equation (1)):

$$k = \frac{Q\,\mu L}{\Delta p} \qquad (1)$$

where Q is the volumetric flow rate, μ is the viscosity of the fluid, L is the path length of the fibers, and Δp is the pressure drop that occurs along the fibers. The motion of the flow outside the porous medium is governed by the equations of continuity and momentum.

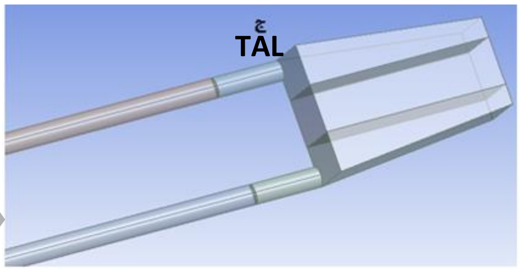

Figure 1. TAL model.

On all three-dimensional models of artificial lungs, the condition of symmetry is applied to reduce the computational cost. Therefore, on the plane of symmetry, the velocity perpendicular to it is zero. Since there is no second-order derivative in the momentum equations used to model the porous medium, the effects of viscosity are negligible and the slip condition can be considered for the walls of the porous medium. However, the non-slip condition was used on the entrance and exit walls of the models' compartments. Time-dependent velocity profiles are applied to the inlet and a constant pressure of 10 mmHg is applied to the outlet to simulate the left artery pressure conditions [6]. For this analysis, the governing equations are discretized for numerical solution by finite volume method. Mesh sensitivity analysis is also investigated based on the average value of inlet pressure. The fine mesh with 370191 cells is selected for the 45-degree model.

3. Results and Discussion

The shear stress distributions for the three geometries with angles of 15, 45 and 90 degrees were analyzed for the three times of systole start, end and diastole end. According to these results, as time passes, the stress areas of the models move towards the distal end of the inlet section. At the end of diastole, the maximum values of stress in the models defined with 45 and 90 degrees, occurred at the distal end. However, the 15 degree model still has the maximum magnitudes of stress at the proximal end. Shear rates in the three models were also analyzed at the three systolic start and end and diastolic end times. It can be reported that the shear rate is higher in the inlet section. At the beginning of the systole, in all three models, the highest shear rate is seen in the developed inlet area. Over time, for all three models, areas with maximum magnitudes of shear rate move toward the end of the system at the inlet. For the 15-degree model, this happens more slowly. As at the end of diastole, this area is still at the proximal end of the inlet portion (Figure 2). Also, the shear stress distribution on the chamber walls was investigated in all three models to ensure that the shear stress on the chamber walls did not exceed the extent that it could damage blood cells (150 Pa) [8]. The results showed that this does not happen in any of the models.

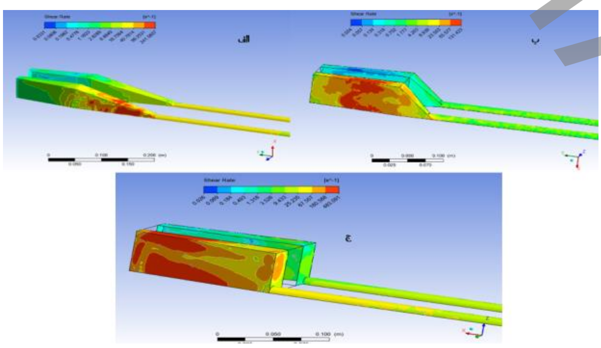

Figure 2. Shear stress distribution for the 15˚ model (Left), 45˚ model (Right) and 90˚ model (Down) at the end of diastole.

4. Conclusions

In this study, three total artificial lung models were created with three expansion angles of 15, 45 and 90 degrees. In the present analysis, the porous medium was modeled as a homogeneous environment. Although the results of this analysis are highly consistent with the experimental results, the shear stress within the fibers may be such that it causes hemolysis of blood cells. Considering the homogeneity of the porous medium prevents the study of the flow between the fibers. But this requires high computational time and cost. In addition, accurate geometry of the artificial lung should be available for more realistic results. Also, according to the results (average mean shear rate, for different expansion angles at the beginning of systole, end and end of diastole), the Newtonian fluid assumption is acceptable for both 15 and 90 degree models. Suggestions can be made in relation to the restrictions. First, if there is an accurate geometry of the artificial lung with a certain number of fibers and specified distances between them and it is possible to model it, it is better to examine the pulsatile flow inside the fibers. Or the inlet and outlet conditions of the fiber bundles should be obtained over time, assuming a homogeneous model, and then they could be applied as the boundary conditions of the fiber bundles. Thus, the analysis will be multiscale. Also, for the 45 ° model and any model with a low shear rate, non-Newtonian blood flow should be tested.

5. References

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