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THE ROLE OF POROUS MEDIA IN MODELING FLUID FLOW WITHIN HOLLOW FIBER MEMBRANES OF THE TOTAL ARTIFICIAL LUNG

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Abstract

A numerical study was conducted to analyze fluid flow within hollow fiber membranes of the artificial lungs. The hollow fiber bundle was approximated using a porous media model. In addition, the transport equations were solved using the finite-element formulation based on the Galerkin method of weighted residuals. Comparisons with previously published work on the basis of special cases were performed and found to be in excellent agreement. A Newtonian viscous fluid model for the blood was used. Different flow models for porous media, such as the Brinkman-extended Darcy model, Darcy's law model, and the generalized flow model, were considered. Results were obtained in terms of streamlines, velocity vectors, and pressure distribution for various Reynolds numbers and Darcy numbers. The results from this investigation showed that the pressure drop across the artificial lung device increased with an increase in the Reynolds number. In addition, the pressure drop was found to increase significantly for small Darcy numbers.

Keywords

artificial lung; blood oxygenation; hollow fiber membrane

1. INTRODUCTION

Transport phenomena through porous media have been the subject of various studies due to an increasing need for a better understanding of the associated transport processes. Some aspects of transport in porous media were discussed in recent monographs by Nield and Bejan (1995), Vafai (2000, 2005), Hadim and Vafai (2000), and Vafai and Hadim (2000). Significant advances have been accomplished in applying porous media theory to modeling biomedical applications. Examples include computational biology, tissue replacement production, drug delivery, advanced medical imaging, porous scaffolds for tissue engineering and effective tissue replacement to alleviate organ shortages, and transport in biological tissues (Yang and Vafai, 2006; Ai and Vafai, 2006; Khanafer et al., 2003a, 2003b; Khaled and Vafai, 2003). Vafai and Tien (1981, 1982) presented an in-depth analysis of the generalized transport through porous media. They developed a set of governing equations

utilizing the local volume-averaging technique. Recently, Khanafer et al. (2009) developed a numerical model to quantify the reduction in blood velocity and pressure resulting from the placement of endovascular coils within a cerebral aneurysm using physiological velocity waveforms. The flow characteristics within the aneurysm sac were modeled using the volume-averaged porous media equations. Their results showed that the velocity fields were significantly affected by the presence of endovascular coil within the aneurysm sac. Moreover, they estimated that a volume density of 20% platinum coil in the aneurysmal sac was sufficient to cause blood flow arrest in the aneurysm and to allow for thrombus formation.

Another important application of porous media includes modeling flow characteristics through the fiber bundle of the artificial lung due to the difficulty in describing the geometry of the individual hollow fiber membranes and also the density and total number of nodal points required to capture the characteristics of the flow. Lung disease is the third leading cause of death and responsible for one in seven fatalities in the United States according to the American Lung Association, totaling close to 335,000 Americans each year (American Lung Association, 2000). Developing efficient and biocompatible artificial lungs has received considerable attention due to high fatality rates because of acute and chronic diseases of the lung (Zwischenberger et al., 2001; Zwischenberger and Alpard, 2002). The goal of the artificial lungs is to provide oxygen and carbon dioxide transfer sufficient to support patients with acute respiratory insufficiency or end-stage pulmonary disease (Cook et al., 1996, 2005; Cook, 1996; Haft et al., 2003; Lynch et al., 2000; Lick et al., 2001). Artificial lungs (ALs) attached directly to the pulmonary circulation, using the right heart as the blood pump, have been shown to be capable of fully supporting basal O₂ and CO₂ transfer requirements in pigs and sheep for as long as 7 days (Cook et al., 1996; Witt et al., 1999). Thus, the well-designed artificial lung should be small to allow implantation and offer a low pressure drop, capable of providing full gas exchange, and impermeable to plasma.

Although the current methods such as extracorporeal membrane oxygenation, or ECMO, for supporting patients with lung disease are occasionally successful as a bridge to transplant, ECMO requires multiple transfusions and is complex, labor intensive, time limited, costly, nonambulatory, and prone to infection. Moreover, blood flow through the artificial lung is driven by the right ventricle whereas a mechanical pump is required in an ECMO circuit, which may cause blood cell trauma (Cook et al., 2005). Artificial lungs are usually designed to meet gas exchange requirements for the patient. To achieve efficient gas exchange, artificial lungs use the cross-flow principle and consist of layers of microporous hollow fibers (diameters are around 300 μm) through which air flows. As the oxygen-poor blood flows across the hollow fiber bundle, it generates transverse mixing that is essential for enhanced gas exchange between oxygen inside the fiber and the carbon dioxide from the blood flowing over the bundle. Hollow fiber modules are widely used in bioprocesses (Heath and Belfort, 1990; Ghosh et al., 2000) and in biomedical fields (Fukuda et al., 1998; Kolobow et al., 1986; Haworth, 2003) because of their high mass transfer performance. One of the most essential hollow fiber modules is a membrane oxygenator. Gas exchange membrane technology has advanced significantly, allowing devices with improved efficiency and durability. Flow characteristics within fiber membranes of blood oxygenation devices were modeled using porous medium. Mazaheri and Ahmadi (2006) utilized a finite volume-based computational model to investigate the uniformity of the fluid flow across the hollow fiber membranes in blood oxygenation devices using a porous medium approximation. Gage et al. (2002) developed a three-dimensional computational fluid dynamics (CFD) model of a commercial membrane oxygenator to predict pressure drops throughout the device. Darcy's law was used to account for the viscous drag of the fibers. Funakubo et al. (2003) used CFD software (STAR LT, CD Adapco, Yokohama, Japan) to

calculate flow velocity and pressure drop in the artificial lungs assuming steady flow condition and rigid housing. The inner part (hollow fiber bundle) of the prototype was defined as a porous medium, which acts like a sponge. Darcy's law was assumed in their study. Their results showed that flow vectors were nearly parallel near the inlet port, whereas the flow vectors collided near the outlet port. This area of flow collision corresponded to the area of thrombus formation *in vivo*.

To the best of the authors' knowledge, no attention has been paid to investigating flow characteristics and pressure drop within an artificial lung using various models of porous medium. Therefore, the main objective of the present study was to examine the momentum transport process inside an artificial lung using Darcy's model, Brinkman-extended Darcy, and the generalized model. Another challenge in artificial lung design is minimizing thrombus formation in the blood flowing around the fibers. Therefore, flow field within the simplified model of an artificial lung was analyzed under various pertinent parameters such as the Reynolds number and Darcy number to address the likelihood of platelet activation and thrombosis.

2. MATHEMATICAL FORMULATION

The geometry of the model used for numerical analysis in this investigation is similar to MC3's BioLung® prototype (Michigan Critical Care Consultant, Ann Arbor, MI) which uses radial blood perfusion through a concentrically wound hollow fiber fabric (Fig. 1). A Newtonian viscous fluid model for the blood was used, with a viscosity of 0.00345 Pa s. The Newtonian assumption was considered a good approximation due to the relatively high shear rate of blood flow within the pulmonary artery and anastomoses. The hollow fiber bundle was approximated using a porous medium approximation and the transport equations commonly known as the generalized model were solved to determine flow characteristics. Moreover, the porous medium is viewed as a continuum with the solid and fluid phases in thermal equilibrium, isotropic, homogeneous, and saturated with an incompressible fluid. Hence, the porous medium has a unique porosity ϵ and permeability K values. By incorporating the above points, the system of the governing equations for the fiber bundle can be expressed in vectorial forms based on the volume average technique (Khanafer et al., 2003a; Vafai and Tien, 1981, 1982; Amiri and Vafai, 1994; Khanafer and Vafai, 2006; Khanafer et al., 2007) as **Continuity Equation**

$$\nabla \cdot \langle V \rangle = 0 \quad (1)$$

Momentum Equation

$$\frac{\rho_f}{\epsilon} \left[\frac{\partial \langle V \rangle}{\partial t} + \langle (V \cdot \nabla) V \rangle \right] = -\nabla \langle P \rangle^f + \frac{\mu_f}{\epsilon} \nabla^2 \langle V \rangle - \frac{\mu \langle V \rangle}{K} - \frac{F_\epsilon}{\sqrt{K}} [\langle V \rangle \cdot \langle V \rangle] J \quad (2)$$

The fluid motion outside the fiber bundle is governed by the Navier–Stokes equations with constant density and fluid properties, together with the continuity equation. In a Cartesian coordinate with a fixed reference frame, the conservation of mass and momentum equations for transient, laminar flow without body forces is given by

$$\nabla \cdot V = 0 \quad (3)$$

$$\rho_f \frac{\partial V}{\partial t} + \rho_f (V \cdot \nabla) V = -\nabla P + \mu_f \nabla^2 V \quad (4)$$

where ρ_f is the blood density, μ_f is the blood viscosity, $\mathbf{P}$ is the pressure $\mathbf{V}$ is the velocity vector, $\mathbf{J} = \mathbf{V}/|\mathbf{V}|$ is a unit vector oriented along the pore velocity vector, and the subscript f refers to the fluid phase. The permeability of the hollow fiber bundle K and the geometric function F can be represented as in Ergun (1952) and Vafai (1984, 1986)

$$K = \frac{\varepsilon^3 d_{\text{HFM}}^2}{150(1 - \varepsilon)^2}, \quad F = \frac{1.75}{\sqrt{150\varepsilon^3}} \quad (5)$$

The void fraction of the fiber bundle region is given as

$$\varepsilon = 1 - \frac{N_{\text{fiber}} d_{\text{HFM}}^2}{D_o^2 - D_i^2} \quad (6)$$

where N_{fiber} is the total number of fibers, d_{HFM} is the hollow fiber membrane outer diameter, D_o and D_i are outer and inner diameter of the fiber bundle, respectively. Employing the hollow fiber membrane diameter of 300 μm and the measured effective blood oxygenator frontal area of about $A_f = 100 \text{ cm}^2$, the fiber bundle properties using Eqs. (5) and (6) were void fraction, $\varepsilon = 0.75$; length, $L = 3.2 \text{ cm}$, and permeability, $K = 3 \times 10^{-9} \text{ m}^2$. Boundary conditions were spatially uniform flow at the inlet of the artificial lung. These boundary conditions can be summarized as

$$\text{Inlet port: } U = -U_o, \quad V = 0 \quad (7a)$$

$$\text{Exit port: } \frac{\partial U}{\partial X} = V = 0 \quad (7b)$$

$$\text{Walls of the artificial lung: } U = V = 0 \quad (7c)$$

At the interface between fluid and porous:

$$V_f = V_{\text{porous}}, \quad \mu_f \frac{\partial V_f}{\partial n} = \mu_{\text{eff}} \frac{\partial V_{\text{porous}}}{\partial n} \quad (7d)$$

where $\mu_{\text{eff}} = \mu_f / \varepsilon$ and Reynolds number (Re) is defined as

$$\text{Re} = \frac{\rho_f U_o H}{\mu_f} \quad (8)$$

where U_o is the inlet velocity and H is the height of the inlet port. The Darcy number is expressed by

$$\text{Da} = \frac{K}{H^2} \quad (9)$$

3. NUMERICAL SCHEME

A finite-element formulation based on the Galerkin method is utilized to solve the governing equations. The application of this technique is well documented by Taylor and Hood (1973) and Gresho et al. (1980). The objective of the finite-element method is to reduce the continuum problem (infinite number of degrees of freedom) to a discrete problem (finite number of degrees of freedom) described by a system of algebraic equations. In the current

investigation, the continuum domain is divided into a set of nonoverlapping regions called elements. Nine node quadrilateral elements with biquadratic interpolation functions are utilized to discretize the physical domain. Moreover, interpolation functions in terms of local normalized element coordinates are implemented to approximate the dependent variables within each element. Subsequently, substitution of the approximations into the system of the governing equations and boundary conditions yields a residual for each of the conservation equations. These residuals are then reduced to zero in a weighted sense over each element volume using the Galerkin method. A variable grid-size system is employed in the present investigation to capture the rapid changes in the dependent variables (Fig. 2). Each side of the model was divided into 80 nodes. Extensive numerical experimentation is performed to attain grid-independent results. Steady-state solution was declared when the relative change in the dependent variable between two consecutive iteration was satisfied by the following criterion:

$$\sum |\varphi_{i,j}^{\gamma+1} - \varphi_{i,j}^{\gamma}| / \sum |\varphi_{i,j}^{\gamma+1}| \leq 10^{-4} \quad (10)$$

where $\varphi_{i,j}^{\gamma}$ stands for the dependent variables at iteration γ . The highly coupled and nonlinear algebraic equations resulting from the discretization of the governing equations were solved using an iterative solution scheme called the segregated-solution algorithm. The advantage of using this method lies in that the global system matrix is decomposed into smaller submatrices and then solved in a sequential manner. This technique results in considerably fewer storage requirements. A pressure projection algorithm was utilized to obtain a solution for the velocity field at every iteration step. Furthermore, the pressure projection version of the segregated algorithm was used to solve the nonlinear system. In addition, the conjugate residual scheme was used to solve the symmetric pressure-type equation systems, whereas the conjugate gradient squared method was used for the nonsymmetric advection–diffusion- type equations.

4. MODEL VALIDATION

The present numerical code is first validated against the exact solution given by Vafai and Kim (1989) as shown in Fig. 3, and an excellent agreement was found between both results. As an additional check on the accuracy of our work, we compared the fully developed velocity profile that we obtained in a fully porous channel with the results of Hadim (1994) for various Darcy numbers as depicted in Fig. 4. Figure 4 illustrates an excellent agreement between both results.

5. RESULTS AND DISCUSSION

The characteristics of the flow and pressure fields within an artificial lung were examined by exploring the effects of the Reynolds number and Darcy number. Such field variables were examined by outlaying the steady-state version of the streamline, pressure field, and velocity vectors. In the current numerical investigation, the following parametric domains of the dimensionless groups were considered: $1 \leq Re \leq 200$ and $10^{-6} \leq Da \leq 10^{-2}$. The effect of porosity on the flow characteristics was not considered in the present work because its influence is well documented in the literature. As such, $\epsilon = 0.75$ was assumed in this study.

5.1 Effect of Reynolds Number

Figures 5 and 6 show the streamlines and velocity vectors within a 2D artificial lung model for various Reynolds numbers ranging from $Re = 1$ to $Re = 200$. Plots of streamlines indicate that the streamlines are identical for low Reynolds numbers. As the Reynolds number increases, the intensity of flow circulation within an artificial lung increases. As

such, a small vortex is formed along the inlet port of the artificial lung. The size of this vortex increases with increasing Reynolds number. Thus, thrombus formation occurred preferentially at regions of low flow velocity. To minimize this effect, the artificial lung can be designed by making the inlet and outlet angles more gradual. Figure 7 illustrates that the pressure drop across the artificial lung increases significantly as the Reynolds number increases from 1 to 200. Therefore, the impedance of the total artificial lung device, which is the measure of the opposition to flow presented by a system, increases with an increase in the Reynolds number.

5.2 Effect of Darcy Number

The effect of Darcy number on the streamlines and pressure distribution is illustrated in Figs. 8 and 9. Figure 8 shows that the Darcy number has a significant effect on the streamlines. As the Darcy number decreases, the activities within the enclosure decrease significantly. For small values of Darcy number, the porous layer is considered less permeable to fluid penetration and consequently the fluid experiences a pronouncedly large resistance as it flows through the porous matrix. This results in hindering flow activities in the porous region as depicted in Fig. 8. This effect is more pronounced in Fig. 9, which shows the effect of Darcy number on the pressure drop within the artificial lung. Figure 9 illustrates that the pressure drop is significantly large when the Darcy number is small.

5.3 Effect of Varying the Flow Model for Porous Media on the Streamlines and Pressure Drop

Figures 10 and 11 illustrate the effect of using different flow models for porous media such as the Darcy's law model, Brinkman's extension, and the generalized model on the streamlines and pressure drop distribution. Note from Figs. 10 and 11 that the Brinkman's extension of the Darcy model and the generalized model are very close for the conditions used. Figure 10 shows that the Darcy's law model exhibits a vortex within the artificial lung and a lower pressure drop (Fig. 11) compared with other models. Because the industrial artificial lungs are characterized by a low Darcy number and a low Reynolds number, Darcy's law model may be sufficient to solve for the flow characteristics within the artificial lungs.

6. CONCLUSION

A computational model of an artificial lung device was developed in this study. The transport equations were solved using the finite-element formulation based on the Galerkin method of weighted residuals. Significant differences in the pressure drop across the artificial lung and the streamlines were found in comparing different models of porous medium. The impedance (or pressure drop) was found to be smaller for small Reynolds number and large Darcy number. The numerical results reported in this work can provide information on the flow properties at low Reynolds number, which is the flow condition inside the human body. Moreover, porous media theory permits the study of fluid motion across small spaces of variable and complex geometry.

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NOMENCLATURE Greek Symbols Subscripts

Da Darcy number, $Da = KH^2$

D_i	inner diameter of the fiber bundle
D_o	outer diameter of the fiber bundle
F	Forchheimer constant
H	height of the inlet port
d_{HFM}	hollow fiber membrane outer diameter
K	permeability
N_{fiber}	total number of fibers
P	pressure
Re	Reynolds number, $(\rho_f U_o H) = \mu_f$
t	time
U_o	inlet velocity
$\mathbf{V}$	dimensional velocity vector
x	x -coordinate
X	dimensionless X -coordinate, x/H
y	y -coordinate
Y	dimensionless Y -coordinate, y/H
ρ	density
e	porosity of the porous medium
μ	dynamic viscosity
f	fluid

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FIG. 1.
Schematic diagram of the 2D artificial lung and boundary conditions.

FIG. 2.
Grid system used in the present study.

FIG. 3.

Comparisons between the numerical results of the present study and the analytical solutions of Vafai and Kim (1989) ($\epsilon = 0.9$, $\Lambda = 100$, $Da = 10^{-3}$, $Re = 100$).

FIG. 4.

Comparison of the fully developed velocity profile in the fully porous channel between the present work and that of Hadim (1994) for various Darcy numbers ($Da = K/H^2$; K = permeability of the porous medium and H is the height of the channel).

FIG. 5.
Effect of varying Re on the streamlines ($Da = 10^{-3}$, $\epsilon = 0.75$).

FIG. 6. Effect of varying Re on the velocity vectors ($Da = 10^{-3}$, $\epsilon = 0.75$).

FIG. 7.
Effect of varying Re on the pressure distribution ($Da = 10^{-3}$, $\epsilon = 0.75$).

FIG. 8.
Effect of varying Da on the streamlines ($Re = 200$, $\epsilon = 0.75$).

FIG. 9.
Effect of varying Da on the pressure distribution ($Re = 200$, $\epsilon = 0.75$).

FIG. 10. Effect of varying porous model on the streamlines ($Re = 100$, $\varepsilon = 0.75$, $Da = 10^{-3}$).

FIG. 11.
Effect of varying porous model on the pressure distribution ($Re = 100$, $\epsilon = 0.75$, $Da = 10^{-3}$).