Methodology for Predicting Oxygen Transport on an Intravenous Membrane Oxygenator Combining Computational and Analytical Models

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1 Introduction

Respiratory conditions, such as acute respiratory distress syndrome (ARDS), are the cause of morbidity and mortality in severely ill patients. ARDS pathology involves severe and rapidly progressive deterioration of the lung gas exchange capability due to fluid accumulation in interstitial and alveolar spaces. Associated fatality rates can be as high as 50% among the 150,000 cases reported yearly in the U.S. [1]. Temporary support of the lung function can be provided by extracorporeal membrane oxygenation or mechanical ventilation [2,3]. Intra-venous oxygenation appears to be a valid alternative to the therapies mentioned above, as it provides the metabolic oxygen and carbon dioxide exchange requirements directly to the blood before it reaches the lungs [4–12]. An intravenous membrane oxygenator (IMO) has been developed by the University of Pittsburgh Artificial Lung program with the design goal of providing 50% of the metabolic gas exchange requirements [13–25]. The IMO device is composed of thousands of porous microfibers surrounding a pulsatile balloon and is intended to be positioned inside the inferior vena cava as seen in Federspiel et al. [16]. The microfibers carry high-pressure oxygen that diffuses from the fiber surface to the blood stream.

The rationale behind the design of the IMO device is that the crosswise secondary blood flow generated by the balloon pulsation enhances the oxygen transport from the fibers by means of convective flow mixing enhancement. In vitro experiments have shown that the IMO device presents a small pressure drop associated with its placement within the inferior vena cava [1,14–16]. These studies indicate that initially the balloon pulsation mode enhanced gas exchange capabilities up to a critical frequency of about 180 beats per minute (bpm). Gas exchange rates for frequencies above 180 rpm dropped due to incomplete inflation-deflation of the balloon caused by limitations on the pneumatic systems [16]. Mathematical modeling and experimental trials demonstrated that testing the device with water as a working fluid, rather than blood, presents fewer complications, as it does not have to take into account either hemoglobin binding or blood clotting over the fiber surfaces [16], and the oxygen transfer rate results can be directly scaled to predict transfer rates in blood [6,7,16,17,19].

The mathematical model reported by Hewitt et al. [17] and Federspiel et al. [19], based on a lumped compartment approach and empirical correlations for a mass transfer coefficient, proved to capture the main characteristics of the gas exchange process,
but also displayed strong dependence on the empirical correlation choice. Experimental trials performed to validate the model using steady flow perfusion over the fibers reported a higher mass transfer coefficient than the model proposed by Refs. [6,7], but good agreement (within 10%) between the predicted and measured oxygen exchange rates. However, it was also found that an oxygen partial pressure buildup within the fiber region (oxygen tension according to Hewitt et al. [17]), translates directly into reduced gas exchange efficiency. Ex vivo testing of an IMO device using animal blood shows that gas exchange rates are within the design target range of 50% of the basal metabolic requirements, and that balloon pulsation decreased the gas exchange dependence of the blood flow rate. These results, however, were generated using a balloon-generated flow rate several times greater than the longitudinal flow rate, thus creating severe flow reversal within the test section, which was damped by the use of compliance bags [1]. It is not clear what effects the flow reversal will have on cardiac function if the device is operated under similar in vivo conditions, or if the pressure loss associated will be higher than the physiological maximum allowable of 10–15 mm Hg. With the purpose of obtaining a better understanding and enhancing the IMO performance, recent experimental investigations have been carried out to evaluate characteristics such as gas permeability of hollow fiber membranes, vessel compliance and plasma resistant, as well as carbon dioxide exchange, in vitro hemolysis, and balloon volume sensors on the oxygen transfer rates and IMO performance [20–25].

Numerical simulations of simple three-dimensional (3D), but feasible, IMO models have been performed for both steady and balloon-pulsation modes for relatively low blood flow rates and small balloon-generated flow rates in order to determine the flow characteristics and evaluate oxygen transport rates [26–30]. Assuming fully developed flow over the central part of the IMO, a simplified 3D model of an angular periodic section with one fiber was constructed. It was found that for a stationary balloon, the velocity profile remained parabolic throughout the domain, and that the Sherwood number, the mass transfer nondimensional parameter, reached an asymptotic value at low flow rates [26]. Balloon pulsation simulations, performed with a time-dependent sinusoidal balloon motion, showed the existence of a time-dependent secondary flow around the fibers, which remained laminar and without separation zones in the fibers region, for the small balloon-generated flows [26,30]. Mass transfer evaluations for balloon pulsations show an increase in the gas exchange rates, with respect to the stationary balloon condition [30,31]. Multifiber computational domains assuming fully developed streamwise flow for a stationary balloon were also studied by Loyola et al. [29], who found that the pressure drop increases linearly with flow rate, and the slope of the linear relationship depends on the number of fibers considered. It was found that increasing the number of fibers actually reduces the gas exchange rates due to oxygen tension buildup, which is consistent with the experimental findings of Hewitt et al. [17]. The previous computational 3D models used fully developed flow boundary conditions and inlet and balloon-generated flow rates which are not those developed during the in vitro and ex vivo experiments with the actual device. Thus, the generated simulation results, although meaningful, are not directly comparable with experimental data [26,29,31], which are performed for higher flow rates [16–21,23]. Therefore, further numerical investigations with improved computational models must be developed and used to generate mass transfer rates that can be compared to the existing experimental data, predict mass transfer performance of the IMO device for several combinations of diameter and fiber length, number of fibers, and frequency and amplitude of balloon pulsations, and, improve the IMO design for optimal performance.

The main objective of this paper is to present a computational methodology to accurately determine and predict the flow and oxygen transport characteristics and performance of an IMO device. This methodology is based on numerical simulations of three-dimensional (3D) computational models to determine the flow mixing characteristics and oxygen transfer rates, and analytical models to indirectly compare and validate the numerical predictions with the experimental data. The specific objectives of this investigation are (1) to create and test improved computational models of the IMO device; (2) to develop and carry out a research methodology, based on numerical and analytical models, that can allow us to validate computational model and numerical simulations with experimental results; and (3) to evaluate the flow mixing characteristics and oxygen transport performance under both, stationary and pulsatile balloon regimes.

To achieve these objectives, a series of suitable three-dimensional computational models that allow us to capture the essential physical processes that describe the oxygen exchange inside of the IMO device for both steady-state and balloon pulsation operational modes are defined, tested and validated. Direct numerical simulations of the conservation of mass, momentum, and species governing equations are performed with the 3D models to determine and predict the flow field and oxygen transport performance in response to changes on the device length, number of fibers, and balloon pulsation frequency. An analytical lumped model based on the conservation of mass principle in control volume formulation (similar to that developed by Hewitt et al. [17]), serves as the link to compare and validate numerical and experimental results reported in the literature.

Multi fiber models are used to investigate interfiber interference and length effects for a stationary balloon. A single fiber model is used to analyze the effect of the balloon pulsation in the velocity and oxygen concentration fields and to evaluate the oxygen transfer rates. Numerical simulations are performed with a finite element computational program, where balloon pulsations are imposed by a periodic displacement of a flat moving boundary. Considering that the device is ultimately intended to work in vivo and it has been tested in water, we simulate both blood and water as working fluids in order to characterize potential differences in the fluid dynamics that could afterward lead to variations in the device oxygen-exchange performance.

2 Mathematical and 3D Computational Models

2.1 Geometry and Computational Domain. The large aspect ratio between the actual IMO device total length and the fiber diameter (in addition to the large Schmidt number for oxygen transport in blood and water) determines that computational simulations of the whole device are highly demanding and, thus, either impractical or unfeasible. It is possible, however, to assume angular symmetry and to divide the IMO device in periodic angular sections, as done similarly by Guzmán et al. [31], each section consisting of physical boundaries (vena cava, balloon, and manifold walls) and symmetric boundaries, as shown in Fig. 1. The computational domain extends from the distal manifold to the proximal manifold, with an extended outlet length to allow the flow to fully develop.

Balloon pulsation computational simulations with many fibers, with the above-defined domain, are intense CPU time consuming, even with current computer platforms. Balloon pulsation simulations are performed with a single fiber model because this investigation intends first to understand and isolate the characteristics of periodic flow convection over one fiber while limiting the computational time expenses to an affordable point, and at the same time, evaluate the mass transfer rate per fiber surface area, so that meaningful comparisons can be done with existing experimental and numerical data. Multifiber models of several lengths are used in this investigation to study both the effects of the fiber length and the influence of fiber-to-fiber interference on the gas exchange performance of the device under steady state operation. Table 1 shows a summary of the dimensions for each model. Schematic representations of the multifiber models for three, four, and nine
fibers are shown in Fig. 2.

As computational requirements to solve the flow in the single fiber model are still demanding, moving-boundary simulations are restricted to the operational conditions in which the IMO device shows the most promising mass transfer characteristics, as predicted by the lumped-compartment approach analytical model developed in Sec. 4, based on Hewitt and co-workers [17,19], and the available experimental data. These conditions of maximum oxygen transfer are given by the combination of highest balloon pulsation frequency and inlet flow rate, which are 180 bpm and 3 L/min. At that point, balloon-generated convective flow is about 14 times the inlet flow rate, which creates strong flow suction towards the device for half of each pulsation cycle. As it is rather difficult to find suitable boundary conditions for that flow regime, the controlled balloon pulsation amplitude was set to limit the balloon-generated maximum flow rate to 95% of the inlet volumetric flow rate.

### Table 1 Computational models domain dimensions

<table>
<thead>
<tr>
<th>Dimensions</th>
<th>One fiber</th>
<th>Three fibers</th>
<th>Four fibers</th>
<th>Nine fibers</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of fibers</td>
<td>133</td>
<td>237</td>
<td>316</td>
<td>450</td>
</tr>
<tr>
<td>Height, (d) (cm)</td>
<td>1.15</td>
<td>1.15</td>
<td>1.15</td>
<td>1.15</td>
</tr>
<tr>
<td>Length, (l) (cm)</td>
<td>5.0</td>
<td>10.0</td>
<td>10.0</td>
<td>5.0</td>
</tr>
<tr>
<td>Width, (w) (cm)</td>
<td>0.09</td>
<td>0.17</td>
<td>0.17</td>
<td>0.27</td>
</tr>
</tbody>
</table>

### Table 2 Transport properties of blood and water

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>(\alpha_0): solubility of oxygen in water</td>
<td>(3.16 \times 10^{-5}) ml (O_2) ml(^{-1}) mmHg(^{-1})</td>
</tr>
<tr>
<td>(\alpha_b): solubility of oxygen in blood</td>
<td>(3.0 \times 10^{-5}) ml (O_2) ml(^{-1}) mmHg(^{-1})</td>
</tr>
<tr>
<td>(D_w): diffusivity of oxygen in water</td>
<td>(2.8 \times 10^{-5}) cm(^2) s(^{-1})</td>
</tr>
<tr>
<td>(D_b): diffusivity of oxygen in blood</td>
<td>(1.8 \times 10^{-5}) cm(^2) s(^{-1})</td>
</tr>
<tr>
<td>(\nu_w): kinematic viscosity of water</td>
<td>(7 \times 10^{-3}) cm(^2) s(^{-1})</td>
</tr>
<tr>
<td>(\nu_b): kinematic viscosity of blood</td>
<td>(2.35 \times 10^{-3}) cm(^2) s(^{-1})</td>
</tr>
<tr>
<td>(C_T): binding capacity of hemoglobin</td>
<td>0.167 ml (O_2) ml(^{-1})</td>
</tr>
</tbody>
</table>

### 2.2 Governing Equations and Boundary Conditions.

Considering both blood and water flow in the IMO device as Newtonian, incompressible, and time-dependent with constant fluid properties, the governing equations in a Cartesian coordinate system are the conservation of mass and linear momentum (Navier-Stokes) equations, Eqs. (1) and (2), respectively. In large vessels and with no external pulsatile flows, like the vena cava where the IMO device is placed, blood can be modeled as a Newtonian fluid with a constant viscosity. For predicting oxygen transport, we employ the conservation of species equation, written in terms of oxygen partial pressure, in which the oxygen partial pressure \(P_{O_2}\) is obtained by dividing the oxygen concentration \(C_{O_2}\) by the oxygen solubility \(\alpha\), which leads to Eq. (3)

\[
\nabla \cdot \vec{v} = 0,
\]

\[
\rho \left( \frac{\partial \vec{v}}{\partial t} + \vec{v} \cdot \nabla \vec{v} \right) = -\nabla P + \mu \nabla^2 \vec{v},
\]

\[
\frac{\partial P_{O_2}}{\partial t} + \vec{v} \cdot \nabla P_{O_2} = \nabla \cdot (D_{eff} \nabla P_{O_2}).
\]

The reaction term due to hemoglobin binding is considered in the numerical simulations by means of an effective oxygen diffusivity in blood, as seen in Eq. (3). Here, the effective diffusivity of oxygen in blood is given by the same expression as in the analytical model of Sec. 4.1.2. Table 2 shows the physical properties of blood and water used in this investigation.

The boundary conditions for the three-dimensional computational models are defined as follows: a velocity nonslip condition at the manifold walls and vena cava wall; a sinusoidal motion with a specified frequency and amplitude, where its velocity is given by the first derivative of the wall position, at the balloon wall; a symmetry condition for the lateral surfaces, on which both the streamwise-\(x\) and crosswise-\(y\) components of velocity are unknown and the spanwise-\(z\) component is set to 0; a fully developed parabolic velocity profile at the inlet, which assumes that the flow fully develops after passing through the manifold; and an outflow condition at the outlet, in which the velocity gradient equals zero and a pressure reference of zero. The oxygen transport model considers the balloon and vena cava walls as nonpermeable, and by symmetry, the lateral surfaces are restricted to a zero oxygen flux. The inlet has an imposed oxygen concentration similar to the venous blood level at an oxygen partial pressure of 40 mm Hg, the fibers have a constant concentration corresponding to 750 mm Hg of oxygen partial pressure, and the outlet has an outflow condition in which the oxygen concentration gradient equals zero [17].

### 2.3 Solution Methodology and Parameter Calculations.

A commercial computational code (FIDAP) based on the finite element method is used to solve the governing equations in the computational domain. The governing equations are first transformed into a set of discrete algebraic equations in which velocity, pressure, and oxygen concentration are calculated at predetermined nodal points. For balloon pulsation simulations, the finite element mesh is composed of two main regions, one with elements of
constant size and nodes with time-independent position, and the second, with deformable elements and nodes with time-dependent positions in the balloon vicinity. For example, a single fiber model of about 41,000 nodal points contains approximately 8000 mobile nodal points. The multifiber models, which do not require time-dependent deformable elements consider between 150,000 and 250,000 nodal points, depending on the number of fibers. Notice that the oxygen concentration field is given in terms of the oxygen partial pressure field $P_{O_2}$, since this unknown field is solved in Eq. (3).

2.3.1 Oxygen Flux Calculations. The oxygen flow from the fiber can be determined by an oxygen flow balance in a control volume that involves the computational domain; it contains the fiber, the inlet and the outlet surfaces. Since every boundary of the domain except the inlet, outlet and fiber, is defined as nonpermeable, the mass conservation principle gives

$$\dot{V}_{O_2}(\text{fiber}) = V_{O_2}(\text{out}) - V_{O_2}(\text{in})$$

$$V_{O_2}(\text{fiber}) = - \int_{\text{fiber}} D_b \cdot \frac{\partial P_{O_2}}{\partial \hat{n}} \cdot dA_{\text{fiber}}$$

$$= \int_{\text{out}} \alpha_b \cdot P_{O_2} \cdot \hat{V} \cdot \hat{n} \cdot dA - \int_{\text{in}} \alpha_b \cdot P_{O_2} \cdot \hat{V} \cdot \hat{n} \cdot dA,$$

where $V_{O_2}$ is the oxygen flow in ml O$_2$/s; $P_{O_2}$ is the oxygen partial pressure in mm Hg; $\alpha_b$ is the oxygen solubility in ml O$_2$/ml mmHg; $\hat{n}$ is a unit vector going out of the surface; and $\alpha_b(\partial P_{O_2}/\partial n)$ is diffusion of oxygen concentration. The direct integration over the fiber surface allows us to determine the oxygen flow trespassing the fiber surface due only to the diffusive oxygen transfer process, while for both the outlet and inlet surfaces, the main transport mechanism corresponds to oxygen convection. The oxygen flux $\dot{V}_{O_2}$ is then obtained by dividing the oxygen flow $V_{O_2}$ by the fiber total surface area $A_{\text{Fiber}}$.

3 Numerical Simulation Results and Discussion

3.1 Flow and Oxygen Transport Characteristics: Stationary Balloon Simulations

3.1.1 Multifiber Model. Multifiber analysis with a stationary balloon is performed using blood physical properties in a computational domain with nine fibers (shown in Fig. 2(b)) of dimensions similar to those of the idealized IMO device described in Hewitt et al. [17]. Figures 3(a)–3(c) show the velocity fields in a center plane-cut view of the device for 1, 2, and 3 L/min of inlet volumetric flow rate, respectively. For all cases, a high velocity stream is injected into an almost-stagnant deeper compartment, where the fiber bundle essentially prevents the flow from reaching the near-balloon wall zone. Recirculation zones exist immediately downstream of the inlet, increasing in size as the inlet volumetric flow rate approaches the physiological level of 3 L/min. For all volumetric flow rates, the flow above the fiber bundle is mainly in the streamwise $x$ direction. The fiber bundle is completely submerged into recirculating flow upstream from a point located close to half the domain length. Downstream of that point, blood flows slowly within the fiber bundle and in the near-wall zone. Below the fiber bundle, the flow changes its characteristics from a low intensity one directional parabolic velocity profile to a recirculating and almost stagnant flow, as the inlet volumetric flow rate increases from 1 to 3 L/min. The high-velocity streamwise flow above the fiber bundle does not mix with either the interfiber or the below-fiber regions flow.

The oxygen concentration distribution resembles the velocity field for all inlet volumetric flow rates of the stationary balloon simulations as shown in Fig. 4. For inlet volumetric flow rates of 1 L/min, the oxygen concentration remains mostly near the fiber.
bundle region, without being transported either above or below the fiber bundle. Above the fiber bundle, the flow is steady and mainly parallel to the fiber bundle. Its high streamwise velocity develops thin oxygen boundary layers above, below and between the fibers. Within the bundle, the fibers are submerging into slow flow of highly oxygenated blood, which causes reduced oxygen transfer rates. This behavior is shown in Fig. 5, where the plane cuts perpendicular to the fibers length show great interference from neighbor fibers on each fiber-oxygen boundary layer. Under this condition, the oxygenated blood flow that leaves the domain comes from downstream of the fiber bundle, where the slow-moving flow passes through the fiber bundle and exits the domain. Thus, it is seen that most of the outlet blood flow leaves the domain without actually being in contact with the highly oxygenated blood zones and with almost the same inlet oxygen concentration level. The oxygen flux behavior of the IMO device, as a function of the inlet volumetric blood flow rate for the previous stationary balloon conditions, is shown in Fig. 6. It is observed that the oxygen flux continuously increases as a function of the inlet volumetric blood flow rate, showing a nonlinear trend within the range of these simulations.

3.1.2 Three Fiber Model With Blood and Water. This section describes and compares the behavior of multifiber models obtained to understand the differences that exist between the originated flow and oxygen concentration patterns for both blood and water. The specific purposes for performing simulations with water are (a) to compare directly with available experimental information and (b) to scale the numerical results obtained with water to blood behavior. The motivation for doing this is that Schmidt numbers are large, thus convective effects become more important than diffusive effects. This in turn translates into a strong mass transfer dependence on the flow pattern and its convective effects. Additionally, it is important to understand the mechanisms of
mass transfer enhancement using both fluids, either to validate the use of water to test a device ultimately intended to operate in a blood flow, or to find differences that will potentially decrease the performance of the device. Numerical simulations are performed with computational models that contain three and four fibers within the domain, as shown in Fig. 2(a).

The velocity and oxygen concentration fields, with blood inlet flow rates from 1 to 3 L/min, are shown in Figs. 7 and 8, respectively. The velocity fields for increasing inlet flow rates from 1 to 3 L/min, are very similar. For a given flow rate, they present distinctive characteristics: above the three fiber bundle, there is a streamwise flow that tends to curve toward the fibers and remains in contact with them; whereas below the three fibers, a recirculation flow develops, which increases in size as the inlet flow rate increases. The viscous nature of the blood slows the flow, preventing a jet-type inlet flow, which remains longer in the fiber region, forcing to carry more of the oxygenated flow from the fiber region in its journey to the device exit. The effect of this flow pattern on the oxygen concentration field for increasing inlet flow rates can be seen in Fig. 8. Notice that the oxygen concentration fields closely resemble the velocity fields. The oxygen concentration is not uniform throughout the entire region, particularly among and above the three fiber bundle. The nonuniform oxygen concentration above the fibers indicates that a small amount of oxygenated flow from below and among the fibers moves upward and mixes with the streamwise flow from above the fibers. Whether this nonuniform oxygen concentration field causes an increase in the oxygen transfer rate or an invariant behavior for increasing inlet blood flow rates, will depend on the size of the recirculation region below the fibers and the size of the oxygen boundary layer thickness among the fibers. Experimental results reported by Federspiel and co-workers [16,19] demonstrate that for a given IMO physical configuration and fluid, mass transfer rates do not increase, but remain almost constant for increasing inlet flow rates.

Numerical results obtained with water for increasing inlet flow rates have indicated that the velocity fields also present two distinctive regions, independent of the flow rate: a region above the fiber bundle where the inlet flow behaves like a jet and remains above and without crossing the fiber bundle, and a recirculation region below the fiber bundle that extends until it fills the entire region below the three fiber bundle. This flow pattern is caused by the water lower viscosity. The numerical results have also indicated that the oxygen concentration field strongly resembles the velocity field characteristics, with oxygen concentration fields that are stratified, particularly around the fiber bundle. Therefore, the low concentration inlet stream passes through the domain without mixing itself with the high concentration region near the fiber bundle. More details on the numerical results and discussion of the flow and oxygen transport characteristics can be found in Escobar [30].

3.1.3 Oxygen Flux Evaluations. Figure 9 shows the oxygen flux as a function of the inlet flow rate for multifiber models of three and four fibers in the domain, from 1 to 3 L/min, using both water and blood. First, for the range of parameters investigated in this research, the oxygen flux remains constant for increasing inlet volumetric flow rates, regardless of the number of fibers or type of fluid, although there are some small local changes in the oxygen flux when the inlet flow rates move around 2 L/min. Second, it is observed that an increase in the number of fibers in the domain translates into a decrease in the net oxygen flux, which is consistent with previous numerical findings reported by Loyola [29] and Guzmán et al. [31], and analytical predictions based on the lumped compartment approach developed by Escobar [30]. Third, the oxygen flux obtained with blood is consistently higher than the flux obtained with water, regardless the number of fibers in the fiber bundle. Fourth, the oxygen flux ratio between blood and water is approximately 2, which is also in good agreement with the experimental findings of Hewitt et al. [17] and Federspiel et al. [19] for a nonpulsating test of idealized devices.

Regardless of the number of fibers, the oxygen flux for blood is greater than the oxygen flux for water. This behavior originates in the different flow patterns within the IMO device when using blood, with a viscosity three times higher than water, which in turn diminishes the inertial effects of the inlet flow. For blood, the oxygen flux reaches a local maximum at 2 L/min, and it is greater than at 1 and 3 L/min, while for water, the oxygen flux reaches a local minimum at 2 L/min. Whether or not local oxygen flux
3.2 Flow and Oxygen Transport Characteristics: Pulsatile Balloon Simulations

3.2.1 Single Fiber Models With Water and Blood. This section discusses numerical simulations with balloon pulsations performed with a computational model of one fiber, for an inlet flow rate of 2 L/min, using water (Fig. 10) and blood (Fig. 12) to transport oxygen from the fiber surface. The computational geometry and domain are the same as Fig. 2(b), but using one fiber of a shorter length than the stationary balloon simulations. The balloon pulsation frequency is 1 Hz (60 bpm).

Single fiber model with water. Figure 10(a) shows two instantaneous representations of the temporal evolution of the velocity field for one oscillation cycle of period T, by displaying mid-plane cuts of the computational domain for an inlet flow rate of 2 L/min, at times \( t = t_0 + T/4 \) and \( t = t_0 + 3T/4 \). From a global perspective, the balloon pulsations and the inlet flow generate a flow dynamic characterized by very active flow mixing in the whole domain, and reversible time-dependent recirculation regions. Balloon pulsations generate a crosswise radial time-dependent up and downward flow, which passes continuously through the fiber and reaches the fiber region above last, where it mixes with the incoming streamwise flow, to ultimately move to the exit region. In a given initial time \( t = t_0 \), the balloon velocity is instantaneously zero and the inlet stream flows longitudinally over the superior part of the fiber. Conservation of mass indicates that the exit and inlet flow rates are the same. A slow-moving recirculation zone holds under the inlet stream, with a three-dimensional vortex near the outlet proximal manifold. At \( t = t_0 + T/4 \), during balloon deflation, the balloon downward velocity reaches maximum value and develops a recirculation region in the outlet manifold with a near-zero flow rate. At this time, all inlet flow passes through the fiber, generating a strong three-dimensional recirculation zone, which appears to fill half of the domain. At \( t = t_0 + T/2 \), the balloon has reached its minimum volume and instantly becomes stationary. Conservation of mass states that the outlet and inlet flow rate are the same. The inlet stream slows and deflects downward towards the end of the fiber section, where it generates time-dependent vortices; it is then accelerated upward near the beginning of the outlet section. The domain now has a larger cross-sectional area than at the beginning, and recirculation is faster in the downstream region. At \( t = t_0 + 3T/4 \), balloon inflation reaches its point of maximum velocity and the outlet volumetric flow rate gets its maximum value. The inlet stream follows a straight path between the fiber and vena cava without being deflected; simultaneously, part of the recirculation zone flow is forced upward to leave the domain. Thus, the outlet flow rate is composed by the inlet flow and the flow generated by balloon pulsations. This cycle repeats itself periodically on time, ending for \( t = t_0 + T \) with a velocity field similar to that of the given initial time.

The oxygen partial pressure field, shown in Fig. 10(b) for the two instantaneous representations, perfectly mimics the time-dependent flow pattern described in the previous paragraph for the

![Fig. 9](H20849) Oxygen flux versus inlet flow rate for the three and four fiber model, from 1 to 3 L/min, for water and blood

![Fig. 10](H20849) Two instantaneous representations at times \( t = t_0 + T/4 \) and \( t = t_0 + 3T/4 \), for one oscillation cycle \( T \), in mid planes for inlet water flow rate of 2 L/min: (a) velocity; (b) oxygen concentration fields
balloon periodic oscillation cycle. These instantaneous representations show that the oxygen is widely spread to the different regions of the computational domain by the action of the generated time-dependent flow. This figure shows that, although there is still a region close to the vena cava with almost no oxygen (which is relatively small), the time-dependent 3D flow resulting from the combination of inlet flow and balloon pulsations, is able to transport oxygen by a convective mechanism to different regions of the domain. The fluid regions below the fiber, rich in oxygen, are transported along and below the fiber in a time-dependent manner. The flow moves freely in the whole domain, improving flow mixing and therefore, enhancing the oxygen transport, because highly oxygenated regions move continuously before exiting the outlet proximal manifold. While this event develops, the fresh, deoxygenated flow contacts with the fiber surface before exiting, repeating the entire flow mixing and evacuation process.

The maximum oxygen transport occurs at the downstream section of the fiber where the deoxygenated flow from the main stream is continuously sucked to the near-balloon region, thus forcing it to contact with the oxygen-rich fiber surface. The temporal evolution of the net oxygen flux from the fiber is depicted in Fig. 11. Starting from the steady state converged simulation, the oxygen flux stabilizes in periodic oscillations after about 20 s. Again, the oxygen flux closely mimics the balloon time-dependent oscillations. The maximum instantaneous oxygen flux occurs immediately after the point of maximum upwards balloon velocity, while the minimum value is at the beginning and end of each cycle. The time average (mean) oxygen flux value is obtained by performing a time integration of the oxygen flux over several complete cycles, and dividing it by the total time of those cycles.

**Single fiber model with blood.** Numerical simulations with time-dependent moving boundaries, corresponding to balloon pulsations, are performed with blood to investigate the effect of higher viscous forces on both the flow field and the mass transfer performance, and the differences between the results with water, if they exist. Simulations are performed with the same model, physical dimensions and frequency and amplitude of balloon pulsations of the previous section.

Figure 12(a) shows a sequence of two instantaneous velocity field representations for a complete balloon pulsation of period $T$, at times $t=t_0+T/4$ and $t=t_0+3T/4$. During the balloon pulsation cycle, the effect of the moving boundary in generating a 3D flow with active mixing is less intense than with water, because the greater viscous force of blood is more effective for diminishing the inertial flow of the inlet stream. Higher blood viscosity means that the flow above the fiber has greater inertia and therefore spends less time following the radial flow generated by the moving wall. This behavior indicates that there is less flow mixing with blood than with water, and that the flow below the fiber stays longer within the recirculation region. The velocity field during the entire time period is characterized by four different zones: (1) the main stream coming from the inlet; (2) a recirculation zone submerging the fiber downstream of the inlet and extending from the inlet to approximately 3/5 of the domain length; (3) a zone of streamwise flow over the fiber, just downwards of the recirculation zone; and, (4) an acceleration zone near the distal manifold wall, where fluid from the inlet stream accelerates or decelerates due to the reduction or increase in the cross-section area and is forced to the outlet extension or it stops completely, depending on whether the balloon wall moves up or down within the cycle. This, higher viscous forces keep the flow more ordered, with small secondary recirculation zones, and with less flow passing into the below-fiber region.

The temporal evolution of the oxygen concentration field, for two instantaneous frames, mimics the corresponding velocity field, as shown in Fig. 12(b). As can be seen, little difference exists between each frame, which display oxygen concentration fields of similar characteristics, with a partially oxygenated flow near the balloon wall, higher oxygen concentration regions where the recirculation zone is, and the deoxygenated inlet stream flowing through the domain between the fiber and the vena cava wall, without mixing with oxygenated zones. Although the balloon pulsations originate a time-dependent up-and-backward radial flow, the oxygenated flow remains mostly below the fiber within the recirculation region and confined to a thin oxygen concentration layer above the fiber. This small layer is carried to the exit region.
by the streamwise flow, but most of the region above the fiber 
(where the streamwise flow is the main flow), contains small 
amounts of oxygen. It is worth noting that, unlike the case with 
water—where the main stream was being forced to mix with oxy-
genated zones—here, due to higher viscous forces, the main 
stream remains between the fiber and the vena cava wall, and thus 
no periodic supply of deoxygenated flow reaches high-
centration zones under the fiber. However, the oxygenated 
flow that leaves the domain in this case does so at a higher 
centration, compensating the poorer flow which mixes with blood.

3.3 Analysis and Discussion of Numerical Results

3.3.1 Balloon Stationary Mode. The main effect of the fiber 
bundle on the steady-state flow pattern is to act as a flow resistor, 
bloking flow into the fiber bundle or the near-balloon region and 
preventing deoxygenated inlet flow from contacting with high 
oxygen-concentration fluid. The diffusion oxygen transport 
mechanism between the fiber region and the main stream is almost 
negligible for transporting oxygen away from the fiber region to 
the outlet manifold, as is clearly shown in Figs. 3 and 4. Only a 
small fraction of the oxygen transport mechanism—oxy-
genated region—only that part of the balloon inflation phase in which 
the flow is ejected back to the main stream and forced to leave the fiber bundle region. Flow mixing characteristics depend on several factors: the amount of fluid forced to con-
tact with the fibers and the characteristics of the flow pattern dur-
ing the pulsation cycle, which depends on the type of fluid, fiber 
length, frequency, and amplitude of balloon pulsations, and inlet 
flow rate.

The net enhancement of the mass transfer rates is due to two 
phemena: the convective effects produced when the inlet stream 
crosses forward and backwards of the fiber bundle; and the 
convective-diffusive effects due to mixing and diffusion during 
the residence time of the inlet stream in both the fiber bundle and 
below the fiber regions. These effects are greatly enhanced with 
increased balloon-generated flow rate. Due to stability constraints 
of the computational simulations, this investigation restricted the 
balloon pulsations to generate only 95% of the inlet flow rate, so 
the transversal/longitudinal flow rates ratio is about 1. Simulations 
performed with water obtained approximately 100% increase in 
mass transfer rates, and about 80% for simulations with blood, 
when compared with the stationary balloon operation of the IMO 
device (Figs. 9 and 10). Experimental results have shown between 
50% and 500% increase in mass transfer rates for transversal/ 
longitudinal flow rates ratio from 5 to 14, proving that the balloon 
pulsation is the single main factor for enhancing the mass transfer 
performance of the device [1,15,16].

It should be recalled that in water and blood, balloon pulsations 
develop different flow patterns, which result in slightly different 
oxygen transfer rates. This is because the lower water viscosity 
generates a flow pattern with developing secondary flows in the 
exit region, thus making easier for the inlet stream to cross the 
fiber and mix with oxygenated flow among and below the fibers. 
When using blood, the more important viscous effects prevent the 
inlet stream from fully crossing through the fiber, but rather de-
celerate and gain oxygen by a more diffusive mechanism, mainly 
from the fiber surface, with less interaction with the oxygenated 
flow below the fibers. However, it is expected that higher balloon-
generated flow rates could increase the mixing effects and further 
increase the oxygen transfer rates. Lastly, comparing the oxygen 
flux obtained with water to that obtained with blood, the ratio is 
about 1.25 and not 2, as in a stationary balloon regime. This ratio 
not only depends on the fluid type, but on other factors, such as 
frequency and amplitude of balloon pulsations, fiber length and 
diameter, fiber numbers, and balloon and vena cava diameters.

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4 Experimental Comparison and Numerical Results Validation

Direct comparisons between numerical simulations results and experimental data are difficult because of differences in the geometric configuration and operating conditions between the computational models and the IMO device. The numerical model, when extrapolated to a full-size device, is an IMO device model of only 133 fibers, which are 5 cm long. Available experimental data reports results for devices with up to 2300 fibers, between 20 and 40 cm in length. Therefore, an indirect comparison methodology is used in this investigation to validate the numerical simulation results. An analytical model is defined, constructed and validated by matching the geometric parameters and operating conditions of the reported experiments. Thus, the numerical predictions of the analytical model are compared with the experimental data. This validation allows us to demonstrate that the analytical model captures the main physics of the oxygen transport process, as well the influence of changing operational and geometrical parameters. Once the analytical model is validated, it is used with the geometrical and operating conditions parameters of the geometry of the idealized IMO device, represented by the numerical simulation model. Last, the results from the analytical and numerical simulation model are compared to determine the accuracy and quality of the predictions. This section first presents the main characteristics of the analytical model. Second, comparisons between the results from the analytical model, and the available experimental data for the geometrical and operating conditions parameters of the reported experiments, are presented and discussed. Third, comparisons are presented between the results from the analytical model and the numerical simulations results for the geometrical and operating conditions of the numerical models used in these investigations. Last, a general discussion regarding the quality of the numerical simulation predictions and the choice of analytical models, both as tools for further parametric analysis, is presented.

4.1 Oxygen Transport Analytical Model. Based on the lumped compartment model described by Hewitt et al. [17] and Federspiel et al. [19], the analytical model proposed in this investigation considers oxygen transport by convection between three separate but connected regions of the model, shown schematically in Fig. 13: the main flow region, the intrafiber region, and the balloon region. The volume in both the main flow and the fiber regions remain constant, whereas the balloon region changes its volume with time to simulate the effect of the balloon pulsation. Oxygen diffuses from the porous fiber to the fiber region and from there to the balloon and main flow regions via the balloon-generated convective flow. A more detailed description of the analytical model and the numerical results and analysis can be found in Hewitt et al. [17] and Escobar [30]. This section presents a brief review of the model for an overall picture of the fundamentals and rationality of the model.

4.1.1 Governing Equations and Initial Conditions. A set of six first-order, governing ordinary differential equations is obtained when the fundamental principle of mass conservation are applied to control volumes describing the three separated regions or compartments. Each of the three compartments shown in Fig. 13 is schematically represented by a control volume which may or may not be time dependent. The conservation of mass principle in each compartment gives place to two first-order ordinary differential equations, one for the balloon inflation, the other for the balloon deflation process. Thus, there are three first-order ordinary differential equations for the inflation process, and three first-order ordinary differential equation for the deflation process. The integral form of the mass conservation principle applied to each compartment is written as follows:

**balloon flow region,**

\[
\frac{d}{dt} \int_{cv} C \cdot d\forall + \int_{S} C \cdot \hat{n} \cdot d\hat{n} = 0,
\]

**fiber flow region,**

\[
\frac{d}{dt} \int_{cv} C \cdot d\forall + \int_{S} C \tilde{v} \cdot d\hat{n} = K \cdot A_{\text{Fiber}} \cdot \Delta P,
\]

**mean flow region,**

\[
\frac{d}{dt} \int_{cv} C \cdot d\forall + \int_{S} C \tilde{v} \cdot d\hat{n} = 0
\]

where \( C \) is oxygen concentration, \( \forall \) is the volume of the control volume (CV) or compartment volume, \( \tilde{v} \) is the velocity vector on the control surface, \( \hat{n} \) is the unit vector perpendicular to the control surface, \( S \) is the control surface, \( K \) is the mass transfer coefficient, \( A_{\text{Fiber}} \) is the total fiber surface area, and \( \Delta P \) is the difference between the average intrafiber oxygen partial pressure and the average oxygen-partial pressure in the fiber region. The oxygen concentration correlates with the oxygen-partial pressure \( P_{O_2} \) through \( C = a(P) \cdot P_{O_2} \), where \( a(P) \) is the solubility of oxygen.

The three previous integral governing equations originate the following sets of first-order differential equations. For the balloon inflation process there are three first-order ordinary differential equations:

**balloon flow region,**

\[
\frac{d}{dt} [a(P) \cdot P_{in}(t) \cdot V_{in}(t)] = -a(P) \cdot P_{in}(t) \cdot Q_{in}(t),
\]

**fiber flow region,**

\[
\frac{d}{dt} [a(P) \cdot P_{f}(t) \cdot V_{f}(t)] = a(P) \cdot P_{in}(t) \cdot Q_{in}(t) - a(P) \cdot P_{f}(t) \cdot Q_{in}(t) + K \cdot A_{\text{Fiber}} \cdot [\bar{P}_{f} - P_{f}(t)],
\]

**main flow region,**

\[
\frac{d}{dt} [a(P) \cdot P_{m}(t) \cdot V_{m}(t)] = a(P) \cdot P_{in}(t) \cdot Q_{in}(t) - a(P) \cdot P_{m}(t) \cdot Q_{m}(t).
\]
mean flow region,
\[
\frac{d}{dt} \left[ \alpha(P) \cdot P_{mf}(t) \cdot \nabla a(t) \right] = \alpha(P) \cdot \dot{P}_f(t) \cdot Q_b(t) + \alpha(P) \cdot P_{mf}(t) \cdot Q_b(t) - \alpha(P) \cdot P_{mf}(t) \cdot \dot{Q}_b(t).
\]

For the balloon deflation process there are also three first-order ordinary differential equations:

balloon flow region,
\[
\frac{d}{dt} \left[ \alpha(P) \cdot P_{b,t}(t) \cdot V_{b,t}(t) \right] = \alpha(P) \cdot \dot{P}_f(t) \cdot Q_b(t),
\]

fiber flow region,
\[
\frac{d}{dt} \left[ \alpha(P) \cdot P_{f,t}(t) \cdot \nabla f(t) \right] = \alpha(P) \cdot P_{mf}(t) \cdot Q_b(t) - \alpha(P) \cdot \dot{P}_f(t) \cdot Q_b(t) + K \cdot A_{Fiber} \cdot \left[ \tilde{P}_b - P_f(t) \right],
\]

mean flow region,
\[
\frac{d}{dt} \left[ \alpha(P) \cdot P_{mf}(t) \cdot \nabla a(t) \right] = \alpha(P) \cdot \dot{P}_f(t) \cdot Q_b(t) - \alpha(P) \cdot P_{mf}(t) \cdot \dot{Q}_b(t) - \alpha(P) \cdot P_{mf}(t) \cdot Q_b(t),
\]

where the unknown \( P_{b,t}(t) \), \( P_{f,t}(t) \), and \( P_{mf}(t) \), are the oxygen partial pressure in the balloon, fiber, and mean flow regions, respectively. These unknowns are calculated by simultaneously solving the set of ordinary differential equations indicated above for each time during the inflation and deflations processes, using a Euler predictor-corrector numerical procedure with proper initial conditions. All the parameters indicated in these equations are known except the previously indicated \( P_{b,t}(t) \), \( P_{f,t}(t) \), and \( P_{mf}(t) \). The simulations are performed until they converge with time-dependent periodic oscillations. Further details can be found in Escobar [30]. Table 2 shows the value of the transport properties used in this section for water and blood.

4.1.2 Balloon Generated Flow Rate and Oxygen Flux Calculations. Balloon pulsations are generated by insufflating helium to the balloon by an external supply [16]. Balloon inflation and deflation is modeled assuming a sinusoidal mode or an impulsive mode [17]. In this investigation a sinusoidal mode is used to model the time-dependent balloon motion [30]. The time-dependent balloon volume is given by \( V(t) = (\nabla f,2/\pi \cdot f \cdot t) \), where \( f \) is the pulsation frequency and \( \nabla f \) is the balloon volume fully inflated. The transversal (radial) volumetric flow rate \( Q_b(t) \) induced by the balloon pulsation is given by the time derivative of the balloon volume
\[
Q_b(t) = \frac{d}{dt} \left( \nabla f \frac{1}{2} \left[ 1 - \cos(2 \cdot \pi \cdot f \cdot t) \right] \right) = \nabla f \cdot \frac{1}{2} \cdot \pi \cdot f \cdot \sin(2 \cdot \pi \cdot f \cdot t).
\]

Once the oxygen partial pressures are calculated within the different compartments, the oxygen flux from the fiber is determined from \( \dot{V}_{O_2}(t) = \dot{V}_{O_2}/A_{Fiber} = K \cdot \left[ \tilde{P}_b - P_f(t) \right] \), where \( K \) is the mass transfer coefficient (or permeability) of oxygen diffusing from the fiber; \( \tilde{P}_b \) is the average intrafiber pressure; \( P_f(t) \) is the average oxygen partial pressure in the fiber flow region, and \( A_{Fiber} \) is the total fiber surface area. Empirical correlations for mass transfer coefficients on cross flow over fiber bundles have been proposed and used by different researchers to find the proper expression for \( K \) [6,7,17]. The Sherwood number, \( Sh = K \cdot d_h^a / \alpha \cdot D \) can be experimentally correlated with the Reynolds and Schmidt numbers as follow: \( Sh = a \cdot Re^{b} \cdot Sc^{c/3} \), where \( d_h \) is the hydraulic diameter, \( a \) is the oxygen solubility, \( D \) is oxygen diffusivity, \( a \) and \( b \) are constants, and \( K \) is the mass transfer coefficient; \( Re = V \cdot d_b / \nu \) is the Reynolds number, \( V \) = \( V(t) \) is the spatial averaged transverse velocity, and \( \nu \) is the kinematics viscosity; and, \( Sc = \nu / D \) is the Schmidt number. An expression for \( K \) can be obtained by properly combining the above terms. This coefficient depends on the fluid: for water the mass transfer coefficient can be written as \( K_w = a \cdot d_h^{b-1} \cdot V_b \cdot \nu_w^{(1/3-b)} \cdot D_w^{2/3} \), with \( a_w \) and \( D_w \) the oxygen solubility and diffusivity in water, respectively, and \( \nu_w \) the water kinematics viscosity. For blood, it has been suggested by Vaslef et al. [7] that the oxygen diffusivity in blood \( D_b \) should be replaced by effective oxygen diffusivity \( D_{eff} \) in the Schmidt number (Sc = \( \nu / D_{eff} \)) to take into account the hemoglobin binding characteristics of the erythrocytes. A proposed expression is \( D_{eff} = D_b/[1 + C_T \cdot [(dSO_2/dPO_2)/\alpha_0]] \), where \( C_T \) is the binding capacity of hemoglobin, and \( SO_2 \) and \( PO_2 \) are the fractional saturation of hemoglobin and the oxygen partial pressure, respectively. Thus, the blood mass transfer coefficient \( K_b \) is given by \( K_b = a \cdot d_h^{b-1} \cdot V_b \cdot \nu_w^{(1/3-b)} \cdot D_w^{2/3} \cdot \left[ 1 + C_T \cdot [(dSO_2/dPO_2)/\alpha_0]] \right] \).

As the total number of fibers is reduced, the cross flow characteristics change, making necessary the use of different empirical correlations for low fiber density devices. Table 3 shows the correlations used to find the mass transfer coefficient as a function of the total number of fibers. The first correlation in Table 3 holds for a number of fibers below 230. This value is considered the critical value at which the fiber arrangement resembles more isolated fibers in cross flow, rather than a fiber bundle. If this is the case, the flow and mass transport behaves as if it were flowing over a single fiber, with no interference on either the hydrodynamic boundary layer or the oxygen boundary layer from the neighboring fibers. If the number of fibers is above 230, then a correlation valid for fiber bundles is used.

4.2 Analytical Model Results and Discussion. Calculations with different combinations of frequency, number of fibers, fiber length and diameter, and inlet volumetric flow rates are carried out to test and validate the analytical model, which are presented in detail by Escobar [30]. These calculations match very well the analytical and experimental results reported by Hewitt et al. [17] and Feder柿piel et al. [19], respectively, providing a solid comparative base for analysis and comparisons with experimental data. Thus, the next section presents those results that are relevant to the purpose of this investigation, which is to compare the numerical simulation results of the 3D models and validate the developed numerical methodology as a tool for obtaining oxygen transfer performance predictions.

4.2.1 Comparison and Experimental Validation. Figure 14(a) shows the oxygen flux as a function of balloon pulsation frequency for the current analytical model, and the experimental data presented in Hewitt et al. [17] for a device 25 cm in length, containing 2300 fibers, and an inlet flow rate of 3 L/min. Because of the high number of fibers a mass transfer correlation for a fiber bundle is used. For this simulation, the analytical model matches all of the relevant dimensions of the experimental device. The lower black dots represent the in vitro experiment in water results obtained by Feder柿piel et al. [16], whereas the upper black dots

<table>
<thead>
<tr>
<th>Number of Fibers</th>
<th>Correlation</th>
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<tbody>
<tr>
<td>Less than 230</td>
<td>( Sh = 0.3 \cdot 0.62 \cdot Re^{0.21} \cdot Sc^{0.12} )</td>
</tr>
<tr>
<td>More than 230</td>
<td>( Sh = 0.71 \cdot Re^{0.51} \cdot Sc^{0.11} )</td>
</tr>
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and numerical data, fibers

\[ L = 25 \text{ cm}, Q_{in} = 3 \text{ L/min}, 2300 \text{ fibers}; \ (b) \text{ analytical model and numerical data, } L = 5 \text{ cm}, f = 60 \text{ bpm}, Q_{in} = 2 \text{ L/min}, 133 \text{ fibers} \]

Fig. 14 Oxygen flux versus balloon pulsation frequency for (a) analytical model and experimental data, \( L = 25 \text{ cm}, f = 60 \text{ bpm}, Q_{in} = 3 \text{ L/min}, \) and 2300 fibers; (b) analytical model and numerical data, \( L = 5 \text{ cm}, f = 60 \text{ bpm}, Q_{in} = 2 \text{ L/min}, \) and 133 fibers

indicate analytical predictions. It is observed that, in agreement with the analytical predictions, the oxygen flux is enhanced with increasing pulsation frequency. For this geometrical configuration and operating condition, the maximum oxygen flux will occur at the combination of highest inlet flow rate and pulsation amplitude and frequency, although this figure shows results for one inlet flow rate. In the frequency range of this simulation, the analytical data trend approaches the experimental data within 10%, with results slightly higher for all the frequencies involved. We believe this is a consequence of the analytical model definition, in which a mass transfer coefficient is imposed to account for oxygen diffusion from the fiber. The mass transfer coefficient for less than 230 fibers, indicated in Table 3, does not take into account the porous membrane character of the fibers. In addition, there are other geometrical and operating condition factors that might play a role in the mass transfer performance, such as the assumption of uniform oxygen partial pressure within the device and a sinusoidal balloon pulsation mode, that could diminish or increase the mass transfer rate predictions of the analytical model. Notice that the analytical simulation results are shown up to a frequency of 180 beats per minute (bpm), because the experimental results reported by Federspiel et al. [16] indicated that the balloon initially failed to inflate completely, due to limitations of the pneumatic circuit, which caused the oxygen flux to remain constant. However, there are no limitations on the analytical model on the balloon pulsation frequency range for further simulations.

The next step in the indirect comparison methodology is to reproduce, with the analytical model, the geometric characteristics of the numerical model with one fiber and balloon pulsations, and compare with the numerical simulation results. The characteristics for this comparison are an IMO device length of 5 cm, containing 133 fibers, a frequency of 60 bpm, and an inlet flow rate of 2 L/min of water. Figure 14(a) shows the comparison between the analytical and numerical data of the oxygen flux as a function of pulsation frequency. Again, the analytical results agree well in general, but are within 5% higher than the numerical results. The trend along with approximation level of this result (less than 5%) is similar to the analytical-experimental comparisons (of about 10%) shown in Fig. 14(a). Thus, the analytical model over predicts both the experimental data and the numerical simulation results, consistently giving higher oxygen transfer rates up to approximately 5% and 10%, respectively.

5 Conclusions

A computational methodology for accurately analyzing, determining and predicting the flow and oxygen transport characteristics of an intravenous membrane oxygenator (IMO) device has been developed, tested, and validated with experimental data. Single fiber and multifiber three-dimensional computational models, as well as an analytical model of the IMO device, have been developed to determine flow and oxygen transport for both stationary and pulsatile balloon regimes. This yields precise and consistent predictions that can be compared with experimental information and used as suitable IMO performance predictions.

The developed methodology can be used to optimize the design of intravenous membrane oxygenators, as well as to investigate their main oxygen transport characteristics, and it can be extended to include carbon dioxide removal.

Stationary balloon mode. Numerical results with three and four fibers demonstrate that the oxygen transfer rates decrease with an increased number of fibers. The fiber bundle acts as a flow resistor, prohibiting the main stream from flowing into the fiber bundle or the near-balloon region, thus preventing the deoxygenated inlet flow from contacting the high oxygen concentration fluid. A very small amount of oxygen leaves the near-fiber region and is transported by convection to the main stream. The whole fiber bundle is submerged into a slow moving flow that recirculates without mixing with the inlet stream, minimizing the diffusion process and blocking the convective transport effect of the main stream. The ratio of oxygen transfer rates obtained for a stationary balloon regime with blood and water is approximately two. This ratio is independent of the inlet flow rate within the range of the simulations performed, and it is mainly induced by the blood viscosity, which is about three times higher than the water viscosity. The flow and oxygen concentration fields are strongly affected by the blood’s higher viscosity, independently of the number of fibers.

Pulsation balloon mode. Balloon pulsations promote effective flow mixing within the intravenous membrane oxygenator by creating a time-dependent 3D flow with recirculation around the fibers regions, which increases the oxygen transfer rates from the fiber surface and below the fibers to the main flow. Balloon pulsations induce deoxygenated flow from both the inlet stream and beneath the fibers into the fiber bundle region, to promote mixing with the oxygenated flow that submerges the fiber bundle. The balloon pulsation cycle is characterized by two phases: first, during balloon deflation, in which deoxygenated flow from the inlet stream is forced to cross the fiber bundle; and a second phase, during balloon inflation, where the flow is ejected back to the main stream and forced to leave the fiber bundle region.

Enhancement of the net mass transfer rate is caused by convect-
tive effects when the inlet stream crosses forward and backward across the fiber bundle, and by convective-diffusive effects associated with mixing and diffusion during the residence time of the inlet stream in the fiber bundle and beneath the fiber regions. These effects are greatly enhanced with increased balloon-induced flow rate.

Balloon pulsations induce flow patterns that are slightly different for water and blood, which result in different oxygen transfer rates. The lower water viscosity generates a flow pattern with developing secondary flows in the exit region, making it easier for the inlet stream to cross the fiber bundle and mix with the oxygenated flow among and below the fibers. With blood, viscous effects prevent the inlet stream from fully trespassing the fiber bundle. The flow decelerates and has less interaction with the oxygenated flow below the fibers.

Simulations with water reached approximately a 100% increase in mass transfer rates, and about an 80% for simulations with blood, when compared with the IMO stationary balloon operation. This enhancement demonstrates that balloon pulsations are the single main factor for enhancing mass transfer performance of the device. The oxygen flux ratio between blood and water with balloon pulsations is about 1.25. This ratio also depends on factors such as the frequency and amplitude of balloon pulsations, fiber length and diameter, number of fibers, and balloon and vena cava diameters.

Validation and comparison. Calculations with different combinations of frequency, number of fibers, fiber length and diameter, and inlet volumetric flow rates have tested and validated the analytical model. These calculations match very well the reported analytical and experimental results, which provide a solid comparative base for analysis, predictions, and comparisons with numerical and experimental data. For the simulated frequency range, the analytical model results approach the experimental data within 10%. The slightly higher analytical predictions are caused by the assumptions of the mass transfer coefficient, uniform oxygen partial pressure within the device, and sinusoidal balloon pulsation mode.

The numerical simulation model is able to capture and describe the main characteristics of the oxygen transport process. Comparisons between analytical and numerical data of oxygen flux, as a function of pulsation frequency for similar geometric and operating conditions, demonstrate that the numerical predictions are within 5% lower than the analytical results.

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Nomenclature

- $a$: constant, a length within the computational domain
- $A_{Fiber}$: total fiber surface area
- $b$: constant, a length within the computational domain
- $b$: beats per minute
- $c$: distance between fibers
- $C$: oxygen concentration
- $C_T$: binding capacity of hemoglobin
- $CO_2$: oxygen concentration
- $CV$: control volume
- $d$: fiber diameter, computational domain height
- $d_b$: fully inflated balloon diameter
- $d_{fb}$: fiber bundle diameter
- $d_h$: hydraulic diameter
- $d_{vc}$: vena cava diameter
- $D$: mass diffusivity
- $D_h$: oxygen diffusivity in blood
- $D_{eff}$: effective oxygen diffusivity in blood
- $e$: distance within the computational domain
- $D_w$: oxygen diffusivity in water
- $f$: balloon pulsation frequency, a length within the computational domain
- $IMO$: intravenous membrane oxygenator
- $K$: mass transfer coefficient
- $K_b$: oxygen transfer coefficient in blood
- $K_w$: oxygen transfer coefficient in water
- $L$: computational domain length
- $P$: hydrodynamic pressure
- $P_{in}$: input oxygen partial pressure
- $P_{bol}(t)$: average oxygen partial pressure in the balloon region
- $P_{bmf}(t)$: average oxygen partial pressure in the intrafiber region
- $P_{mfl}(t)$: average oxygen partial pressure in the mean flow region
- $PO_2$: oxygen partial pressure
- $Q_d(t)$: balloon generated volumetric flow rate
- $Q_{l}, Q_{lin}$: inlet volumetric flow rate
- $Re$: Reynolds number=$V_d/b/
u$
- $Sc$: Schmidt number=$\nu/D$
- $Sh$: Sherwood number=$K_d/b/\alpha - D$
- $SO_2$: fractional saturation of hemoglobin
- $t$: time
- $V$: characteristic velocity
- $W$: computational domain width
- $\bar{V}$: unitary vector
- $\bar{v}$: velocity field
- $\forall$: volume
- $\forall_{bol}(t)$: time-dependent balloon region volume
- $\forall_{fl}(t)$: time-dependent intrafiber region volume
- $\forall_{mfl}(t)$: time-dependent main region volume
- $\forall_0$: fully inflated balloon volume
- $VO_2$: oxygen flux
- $\bar{P}_d$: average intrafiber oxygen partial pressure

Greek symbols

- $\alpha, \alpha(P)$: solubility of oxygen
- $\alpha_a$: oxygen solubility in blood
- $\alpha_w$: oxygen solubility in water
- $\mu$: dynamic viscosity
- $\nu$: kinematics viscosity
- $\nu_w$: water kinematics viscosity
- $\nu_b$: blood kinematics viscosity
- $\rho$: density
- $\Delta P$: difference between the average intrafiber oxygen partial pressure and the average oxygen partial pressure

References


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