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Development and analysis of an integral fluidodynamic model in hollow fibre for different operational modes

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ABSTRACT

An integral fluidodynamic model for hollow fibre (HF) has been developed and analysed. The model explains the pressure and flow profiles in three operation modes: open- and full-shell mode with forced circulation in the shell in cocurrent or countercurrent (OSFC); closed- and full-shell mode with forced recirculation in the shell in cocurrent or countercurrent (CSFR) and open- and empty-shell mode (OES). A methodology has been proposed to determine the parameters of the system and to verify the different operational systems proposed. Simple expressions have been developed to evaluate the pressures and the flows in the lumen fibres and in the shell, the transmembrane pressure, and the permeate flow for each operation mode assayed. The behaviour of the HF depends on the geometry of the module, on the operation mode chosen, and on the flows circulating through the lumen fibres and through the shell. The experimental results found with a commercial HF verified the model developed. The use of these expressions led us to choose the HF, the operation mode and the adequate flows that optimise the objective desired.

1. Introduction

The use of HF devices is growing for sea-water desalination, water clarification, artificial kidneys or hemodiafiltration [1,2], biotechnology applications [3] or catalytic bioreactor [4] due to the advantages of these membrane systems, such as high membrane surface:volume ratio; high density of the catalyst that these reactor systems allow in a small reactor volume [5]; the recovery and reuse of the catalyst from the reaction mixture and the possibility of integrating a catalytic conversion, product separation or concentration and catalyst recovery in a single operation [6].

For an optimal use of the HF for these applications, detailed knowledge of the real functioning of these systems is indispensable, making it necessary to know the flow and pressure profile in the HF. Thus, Berman [7] developed solutions to the Navier–Stokes equations for fluid flow in a rectangular slit with porous walls. The solution was achieved with such assumptions as a steady state, incompressible fluid, laminar flow or the velocity of the fluid leaving the walls of the channel not being position dependent. These assumptions were adopted by later researchers such as Karode [8], who considered the wall velocity proportional to the local transmembrane pressure and compared the model developed, considering the pressure in the fibre to be an independent variable, with the numerical computational fluidodynamics simulation. Also, numerous works have developed theoretic models (many not experimentally verified), which simulate the flow and/or pressure distribution in the different zones of the HF [9–11]. The extreme radius:length ratio of the fibres suggested that a 1-D model could be suitable to simulate the hydrodynamics. Thus, with the use of parameters characteristic of fibres made with Cuprophan, Patkar et al. [12] and Labecki et al. [13] developed a theoretic model to analyse the flow in several different configurations (closed-shell mode, dead-end, cross-flow filtration, countercurrent, and cocurrent contacting). The authors systematically considered the lumen and the shell sides as two interpenetrating porous regions and combined Darcy's law and fluid continuity to give a set of 2-dimensional partial differential equations governing the hydrodynamics. The theoretical simulations made show that both the cocurrent and countercurrent configurations may have implications for high-flux dialysis, because the lumen half-length velocity can be higher in the cocurrent case, thus reducing the concentration polarization at the membrane surface. However, these authors did not analyse the influence of the configuration on the transmembrane flow. Smart et al. [14,15] proposed theoretic models for cocurrent and countercurrent flow configurations to evaluate productivity and selectivity, considering the influence of the pressure, packing density, and fibre diameter. More recently, Kostoglou and

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Karabelas [16] developed complex mathematical models that simulate the fluidodynamic of the HF. Also, different works have proposed models to explain the HF bioreactor process dynamics. Most of these are theoretical analysis that offer analytical expressions for the velocities and pressure profiles in the HF bioreactors by solving the coupled momentum and continuity equations in the fibre-lumen matrix and shell; expressions for the flow and pressure profiles in the HF operated in closed-shell or open-shell; mathematical models to predict the coupled hydrodynamics and mass transport in HF [11,12] or models based on the numerical solution of the dimensionless balance equations governing mass transfer within the regions of the reacting system [5].

As commented above, it is necessary to evaluate the pressure profile in the HF to ascertain the transmembrane flow. Several researchers have used empirical equations to explain the pressure drop in the lumen of the fibres [17] but generally the pressure in the lumen compartment is assumed to drop along the fibre length according to the Hagen–Poiseuille equation [18–20]. Greater controversy exits in explaining the pressure profile in the shell. If there is no forced circulation in the shell, researchers usually consider the pressure on the shell to be constant and equal to the semi-sum of the entry and exit pressure, this being experimentally verified when the exit of the lumen is open to the atmosphere. Nevertheless, the experimental results show that the pressure in the shell varies when a recirculation flux is used in the shell. Thus, different researchers have considered that the flow in the shell could be simulated by empirical equations as Ergun's equation [19,21,22], although other empirical equations have been used [1]. Other authors [18,23,24] have simulated the flux in the shell by considering laminar flow. Also, Ergun's equation is transformed into the Carman–Kozeny equation, the theoretical pressure drops being a hundred times lower than the experimental data [23]. Most of these models consider that the flow circulating through the shell is constant and disregard the transmembrane flow. When the transmembrane flow is low, the simplification would be acceptable, but it would not be correct with high permeabilities. Another aspect to be considered is the distribution of the fibres in the HF shell. Many researchers consider that the fibres distribution is homogenous or that this aspect do not affect significantly on the HF behaviour. Nevertheless, others indicate that the random packing of the fibres affect to the HF fluidodynamic [21,25–28]. Some researchers have used the Voronoi tessellations to model the non-homogenous flow in the shell side due to the randomly packed bundles [29–32]. Bao and Lipscomb [33] used this methodology to analyse the mass transfer for axial flows through randomly packed fibre bundles using a hybrid finite element (FE)–finite difference (FD) method to solve the governing equations. The results obtained by these authors indicate that the mass transfer coefficients can decrease dramatically and that the mass transfer is lower in the low packing regions. Similar conclusions were obtained by Zhen et al. [34] from a theoretical analysis of the shell-side flow distribution made in a randomly packed HF module using the random cell model.

Some authors have used magnetic resonance imaging (MRI) to characterize the velocity profiles within porous media, including hollow fibre modules [35–45]. Frank et al. [46] made experimental measurements of concentration within a hemodialyzer using X-ray computed tomography. They found that although the average values for a cross-section decrease uniformly from lumen inlet to outlet, large variations in concentration occur within a cross-section. They attribute these concentration variations to variations in shell flow finding that the regions of lower shell flow have higher concentrations. The regions of lower shell flow do not always correspond to regions of higher fibre packing.

Also, Eloot and Wachter [47] and Kieffer et al. [48] used Computational Fluid Dynamics (CFD) to visualize the flow in a dialyser. A 3-D finite model of the blood–dialysate interface over the complete length of the dialyser was developed. Also, Ghidossi et al. [17] numerically investigated the flow profile in a HF by CFD as a function of the working conditions and the membrane geometry. Also [8], [49], and [50] developed models with CFD for HF.

On the other hand, different operation modes have been studied by different researchers: continuous open-shell mode [9], closed-shell mode with no recirculation in the shell [9–12], cocurrent operation [13,51], countercurrent operation [13,18,20] or cross-flow operation. Some studies are theoretical without experimental corroboration and, though these configurations are governed by the same physical principles, the flow and pressure distributions and mass transfer depend on the operation mode.

The aim of the present work is to develop a general integrated model to simulate the fluidodynamics of the HF. This general model has been particularized for different operation modes. It has been verified that the models developed reproduce the experimental results found. In addition, simple integrated equations have been developed to calculate the transmembrane pressure and flow in the HF. These equations imply better knowledge of the system that allows the choice of the most appropriate configuration for each application. It has also been observed that both the operation mode and the flows have a strong influence on the transmembrane flow.

2. Experimental set-up

Fig. 1 shows a schematic diagram of the HF. The experimental set-up has two peristaltic pumps, a rotameter and a HF called NT1975 supplied by Sorin Biomedica with membranes made in Cuprophan®, a surface area of 1.95 m^2 and a size pore of 5 kDa. The module has two compartments, one formed by the lumen of the fibres and the other being the space between fibres and the shell cartridge. One peristaltic pump is used to maintain a constant flow of water (20 °C) in the upstream port; the second pump is used to maintain a constant flow of water (20 °C) in the shell in cocurrent or countercurrent flow when these configurations are assayed. The flows assayed are within the range recommended by the manufacturer and the range normally used in these modules. A solution of 1% formaldehyde was used to prevent the bacterial growth in the HF during the experiments.

The experimental system in this work enabled the quick and accurate measurement of pressure in the shell. In the stationary regime, the pressure in different zones of the experimental system was measured by the water column reached in tubes 14 mm in diameter connected to the upstream port and to the different orifices made in the shell of the HF, this allowing the pressure inside the shell cartridge to be measured in the z-direction. The tubes were open and therefore the pressure was measured with respect to the atmospheric pressure. The downstream port was opened to the atmosphere and hence the pressure at the exit of the fibres was considered equal to zero.

When the open- and empty-shell operation mode (OES) (Fig. 2f) was assayed in the HF, the flow was introduced into the lumen of the fibres by the upstream port while the shell was opened and without liquid. The shell was open and, therefore, equal to the atmospheric pressure. The lumen-inlet pressure was measured. The flow range assayed was from 0 to 11.5×10^{-6} m^3/s. When the stationary regime was reached, the lumen-inlet pressure was measured together with the water mass collected in 1.5–4.0 min at the downstream port and at the shell port. The assays were made in triplicate.