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Effectiveness of microbubble removal in an airtrap with a free surface interface

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ABSTRACT

An end stage renal disease patient will undergo haemodialysis (HD) three or four times a week for four to five hours per session. Because of the chronic nature of the treatment, any minor imperfection in the extracorporeal system may become significant over time. Clinical studies have raised concerns relating to small microbubbles entering HD patients. These bubbles lead to further pathophysiological complications with the size of the bubble being a major factor. Microbubbles of different sizes can be generated throughout the extra-corporeal HD circuit. It is important to understand the possibility of these bubbles passing through the air trap or successfully being removed which indicates the performance of the air trap, the only mechanics of removing air bubbles. Chronic exposure to various sizes of microbubbles was analysed in detail for haemodialysis patients. However, smaller microbubbles are shown to be able to pass our modelled air trap. While studies have reported the presence of bubbles before and after the air trap, because these bubbles are only counted and not tracked, the performance of the air trap for removing different bubble sizes is not understood. Here, the performance of the air trap in filtering bubbles and the possibility of different bubble sizes passing through the air trap with the presence of the free surface interface have been evaluated. The modelled air trap is shown to be ineffective for filtering small micro bubbles.

1. Introduction

During haemodialysis blood is pumped through an extracorporeal circuit which uses a dialyser to filter out toxins from the blood. The system has a number of safety measures, including pressure detectors, and air traps. All HD systems have an air trap located on the venous side while occasionally an additional air trap is added on the arterial side. The venous air trap is located just before the blood re-enters the patient and is the last stage and only mechanism for eliminating air from the system. The air traps are cylindrical in shape with a vertical axis having the inlet at the top and the outlet at the bottom. The inlet and outlet are off-axis and so the flow is asymmetrical with the off-axis jet impacting a blood–air interface. An air trap works by making use of two opposing forces acting on a bubble. Buoyancy floats the bubble upwards in the air trap and the blood flow drags the bubbles along the direction of the flow. Several cases of small air emboli have been reported in the haemodialysis circuit after the air traps (Rollé et al., 2000; Woltmann et al., 2000; Jonsson et al., 2007; Stegmayr et al., 2007; Forsberg et al., 2010, 2012; George et al., 2012; Stegmayr et al., 2012). Hypothetical sources of micro bubbles in the extracorporeal line have been identified (Polaschegg, 2007), but are yet to be confirmed. The air trap itself is a potential source with studies showing it is unable to totally eliminate microbubbles, but also could be a source of new micro bubbles (Jonsson et al., 2007). Higher air trap blood levels significantly reduce the number of micro bubbles in the system (Forsberg et al., 2012). However, manufacturer specifications indicate that the purpose of the air trap is to prevent massive air embolism rather than removing all possibilities of an infusion of micro bubbles (Polaschegg, 2007).

Haemodialysis patients require frequent and long-term treatment. Therefore any small anonymity in the system can form significant pathophysiological complications over years. One such factor, first noted in the 1970s is that of micro embolization (Bischel et al., 1975). Platelet aggregation were thought to be the cause (Ishii et al., 1996), but recent studies suggest that these micro emboli are microbubbles (Barak and Katz, 2005). The clinical significance of micro bubbles resulting from long-term usage of HD is likely to be primarily seen in the lungs and brain. Post mortem examination of HD patients has shown a very high percentage of acute and chronic lung disease (Fairshter et al., 1982). While chronic exposure of pulmonary vessels to micro bubbles is likely to contribute to developing lung injury (Droste et al., 2002; Forsberg et al., 2010; Stegmayr et al., 2012), the

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high prevalence of pulmonary hypertension found in chronic HD patients has also been described (Barak et al., 2008). Additionally, it has been shown that these small microbubbles may cross from the venous to the arterial circulation and cause varying degrees of neurological damage. cerebral atrophy and deterioration of neurocognitive functions in chronic HD patients is a recognized problem (Savazzi et al., 1995, Pliskin et al., 1996; Barak et al., 2008; George et al., 2012), and cardiovascular morbidity and mortality rates are higher in the HD population (Schon et al., 2004). The presence of micro bubbles may be a contributor.

It follows that a higher blood flow rate is more likely to carry bubbles into the bloodstream. Analysis of the blood flow and the movement of the bubbles inside an air trap are required for the improvement of the efficiency of the air trap as well as understanding the range of microbubble sizes being filtered or sent back to the patient. This can be best achieved through computational study on the micro bubbles inside an air trap. Clinical studies have highlighted concerns, and preliminary 2D computational analysis have shown the dispersion of bubbles inside a single phase venous air trap reservoir (Keshavarzi et al., 2013). However, a full model of an air trap and its performance has not been investigated to understand the behaviour of these bubbles and the probability of different sizes entering the body. This analysis is vital to the understanding of the performance of the device in preventing microbubbles passing the system and also understanding the range of sizes of which will pass through the device into the body for further clinical studies. The study will provide detailed information on the probability of different micro bubbles passing through the air trap to the venous line. This information can also be used for future clinical practice and design improvements.

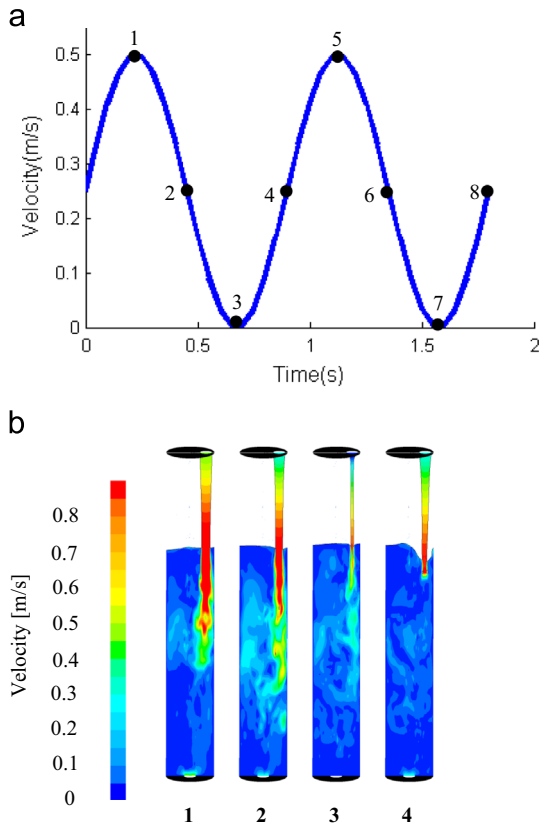

Fig. 1. (a) Inlet velocity of the airtrap for two cycles. (b) Instantaneous velocity profile of a middle plane in the blood flow in the venous airtrap at different stages.

2. Methods

Computational analysis allows the ease of visual and detailed analysis (micro bubbles in the blood are not readily visible to the naked eye (Barak et al., 2008)). In the present analysis, blood was modelled as an incompressible, homogeneous and Newtonian fluid with a density of 1050 kg/m^3 and viscosity of 0.003 kg/ms. ANSYS Fluent 14.5 with a complied user defined code was used to solve the finite volume approximation of the Navier–Stokes equations. The VOF (Volume of Fluid) model was used to capture the multiphase flow of blood and air in the airtrap. The second order upwind scheme is applied for the spatial momentum discretization, and pressure staggering option (PRESTO) scheme was used for pressure discretization. The PISO (Pressure Implicit Solution by Split Order) algorithm is employed to couple the pressure with the velocity in the incompressible limit. The appropriate evaluation of the fluxes across the element faces of the advection terms has been achieved by the geometric reconstruction based on the PLIC (Piecewise-Linear Interface Calculation) scheme. A detail of the equations along with the validation of the numerical model used in this study has been described (Keshavarzi et al., 2014).

The DPM (Discrete Phase Model) model, which calculates the trajectory of a bubble by integrating the forces on each bubble in a Lagrangian frame of reference, has been implemented to track the bubbles. This is suitable for the low concentration (a volume fraction less than 10%) of bubbles inside the air trap (Jonsson et al., 2007). The main forces on the bubble are the body force (the buoyancy), steady drag, lift force (Saffman lift), virtual mass, Basset history, and the pressure gradient force. The steady drag on the bubble depends on the size and relative velocity of the bubble to the fluid. The lift force term or the Saffman lift is the lift from the viscous shear and is only significant for very small bubbles and high shear rates. For the case of an air trap it is negligible because of the low shear rates. The added mass force (or virtual mass) and the Basset history force are the unsteady drag forces acting on the bubble. The added mass term is due to the kinetic energy gain of the fluid from the moving bubble, and can be represented as an added mass to the bubble. The basset history force is the rate of velocity change between the bubble and the surrounding fluid and appears only in unsteady drag motions, and is insignificant for such bubble sizes and velocity rate Finally the pressure gradient force is the force acting on the bubble from local pressure gradients in the flow.

A user defined code was written and compiled with the software to apply the forces and allow the analysis of the discrete bubble phases inside the multiphase (blood and air) environment inside the air trap. Micro bubbles have a high internal pressure and are very stable, and it is a reasonable assumption to consider such small bubbles as spheres.

A three-dimensional analysis of a representative dialysis system air trap, with a range of flow rates corresponding to those investigated by Jonsson et al. (2007), was modelled (see Fig. 1(a)). Venous air traps usually have a varying and adjustable blood level of 80–130 mm. At 100 mm, the level of blood is known as a high level of blood in the air trap, which is preferred over the low level of blood and has been used in this study (Forsberg et al., 2012). A fully 3D structured mesh consisting of 1.46 million cell elements was developed for the region of the air trap. A mesh refinement study showed mesh convergence. A time step of 0.0001 s was used to solve for the transient flow field and provide stability in the solution.

The venous air trap shown in Fig. 1 has been selected for this study. The height of the air trap is 150 mm with 100 mm of it being filled with blood in the case of high blood level which is recommended both clinically and in previous studies (Forsberg et al., 2012). The venous air trap has a circular cross section of 20 mm, and the inlet and outlet are off axis by 2.5 mm as seen in Fig. 1 and both have an inner diameter of 5 mm. The dialysis pump is a peristaltic roller pump which operates at 60–70 cycles per minute with a varying flow rate leading to blood velocities between 0 and 0.5 m/s. To approximate this, a top hat velocity distribution is assumed at the inlet with the velocity varying between 0 and 0.5 m/s as a sine wave in time with a period of 0.9 s as shown in Fig. 1(a).

Initially, 15 full cycles of blood flow were monitored to allow the flow to settle and become repeatable. Microbubbles of 5, 50,100, and 200 μm in diameter were then injected at each time step for one full cycle (0.9 s) to allow analysis on the bubble behaviour and distribution in the flow. The injection of micro bubbles was then stopped after one cycle of injection. The simulation was continued for 10 cycles (9 s after injection) to the point where a negligible number of bubbles were present in the air trap flow.

3. Results

The flow inside the air trap dictates the behaviour of the microbubbles as well as the residence time (time which a bubble will remain) inside the air trap. Generally, very small bubbles and large bubbles will have low residence times. Larger bubbles will rise to the top of the blood surface and will merge with the air on the surface in line with the purpose of the air trap. On the other hand, very small bubbles will follow the blood flow and may or