Numerical simulation of a peristaltic biorheological smart micro-pump

El Gendy, M, Beg, OA, Beg, TA, Kadir, A and Jouri, W

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1. INTRODUCTION

Peristalsis involves the propulsion of physiological fluids via rhythmic contraction of the walls of a vessel. It arises in planar trans-location in bileway, embroyonic heart development, blood transport in narrow vessels and intestinal dynamics [1]. Most physiological fluids including blood exhibit non-Newtonian properties which manifest as deformation-rate dependence, yield stress, viscoelasticity, thinning. The high concentration of suspended particles (red blood cells, proteins, nutrients, leukocytes) and their inherent elasticity contribute strongly to non-Newtonian behavior, especially in narrow vessels (microcirculation). The non-Newtonian effects are therefore likely to contribute strongly to micro blood-micro pump dynamics at low deformation rates. Modern pumps are increasingly embracing biological mechanisms [2]. In this presentation, 3-dimensional simulations of both Newtonian and non-Newtonian fluids in a peristaltic blood micro-micro pump are described. The Carmou and power-law models are employed to characterize rheological behavior. These formulations can correctly predict the Newtonian viscosity behavior via a power-law index for the shear rate and includes both dilatant (shear-thickening i.e. viscosity is elevated under increasing shear strain and power-law index (exceeds unity)) and pseudo-plastic (shear-thinning i.e. viscosity is reduced under increasing shear strain and power-law index is less than unity) behavior. Thus, these models are more being more appropriate for blood flows. ANSYS Fluent® finite volume software is implemented for the simulations. Contour plots for velocity, pressure and vorticity are constructed. The current work is relevant to providing more numerical simulations of actual peristaltic propulsion mechanisms in bio-inspired micro pumps and it is envisaged that it will provide a useful complement to experimental studies.

2. SMART PERISTALTIC MICRO-PUMP MODEL

A 3-D model was constructed using the design software AutoCAD. The domain of peristaltic pump model was 9 mm in length and 2 mm in width. The inlet of the tube leading to the pump outlets measured 0.2 mm in width. This geometrical model was imported to ANSYS Fluent® Software and the material specifications on the layers were defined. The design was further improved by increasing the inlet area of 0.2 mm width to represent the pump inlet surface. Laminar, viscous dominated peristaltic flow is considered in the geometrical micropump domain.

The fundamental equations for mass and momentum conservation employed in ANSYS FLUENT® are the 3-D unsteady incompressible Navier-Stokes equations which comprise the mass conservation and x, y, and z-momentum conservation equations. These may be stated as follows:

\[
\frac{\partial u}{\partial x} + \frac{\partial v}{\partial y} + \frac{\partial w}{\partial z} = 0
\]

\[
\frac{\partial u}{\partial t} + \frac{\partial}{\partial x}(ux) + \frac{\partial}{\partial y}(vy) + \frac{\partial}{\partial z}(wz) = \frac{\partial P}{\partial x} - \mu \left( \frac{\partial^2 u}{\partial x^2} + \frac{\partial^2 u}{\partial y^2} + \frac{\partial^2 u}{\partial z^2} \right) + \frac{1}{Re} \nabla^2 (ux)
\]

\[
\frac{\partial u}{\partial t} + \frac{\partial}{\partial x}(vx) + \frac{\partial}{\partial y}(vy) + \frac{\partial}{\partial z}(wz) = \frac{\partial P}{\partial y} - \mu \left( \frac{\partial^2 v}{\partial x^2} + \frac{\partial^2 v}{\partial y^2} + \frac{\partial^2 v}{\partial z^2} \right) + \frac{1}{Re} \nabla^2 (vy)
\]

\[
\frac{\partial u}{\partial t} + \frac{\partial}{\partial x}(wx) + \frac{\partial}{\partial y}(vy) + \frac{\partial}{\partial z}(wz) = \frac{\partial P}{\partial z} - \mu \left( \frac{\partial^2 w}{\partial x^2} + \frac{\partial^2 w}{\partial y^2} + \frac{\partial^2 w}{\partial z^2} \right) + \frac{1}{Re} \nabla^2 (wz)
\]

Blood is assumed to have constant density of 1055 kg/m³. A steady state pressure solver, coupled with a viscous laminar model, is employed. All current simulations have been performed using a Lenovo 9600 Laptop machine with a 8 GB RAM and an Intel® Core i7-4770K CPU at 3.50 GHz. The solver was set to include double precision option to achieve a higher rate of accuracy and parallel processing option was enabled to utilize the power of the multi-core system and the double GPU feature within the machine. The effect of gravity was also taken into consideration with g ≈ 9.81 m/s² along the z-axis. First the velocity distribution along the length of the micro channel was computed. Next the velocity was also computed at specified regions of action around the design of the micro pump. All of the distributions presented were the influence of the length of the microchannel due to the symmetrical nature of the results on both sides.

3. ANSYS FLUID SIMULATION & VALIDATION

A body-sizing approach, comprising mainly of tersheshell cells to accommodate the anomalous structure of the model, was used to produce an adequate mesh that satisfies the problem specifications. Reformation concentrated at the edges of the inlet and outlet where placed to resolve the complete flow in these regions. All the mentioned elements were established at the inlet and outlet to accommodate boundary layer conditions. The cell distribution along the model can be seen in the following. Additionally, refined meshes for the domains were employed to investigate the influence of mesh size on the results for the grid independence study conducted below in Table 1. The results of the grid independence study clearly shows that the values for the overall velocity stability increases at around 332.400 elements, with a slight difference in values after that, thus making the original mesh the ideal choice in this situation. The simulations were conducted with the pressure based solver due to the incompressible assumption for blood. The pressure-based solver relies on two types of algorithms, a segregated algorithm and a coupled algorithm. The segregated algorithm uses sequential steps to solve the governing equations in a more efficient manner by storing the discretized equations of the variables within the memory. The coupled solver solves the geometry problem in case of a steady flow, under increasing shear strain and power-law index is less than unity, but it needs less memory, maintaining simplicity of the coupling procedures and requiring no memory. Convergence is critical to achieving fast, accurate solutions. Monitoring of the regulated simulated flow rate and power delivered by the peristaltic pump through open boundary conditions. The current study does not guarantee the effectiveness of results. Some of the results might not be the required result disregarding the validation of the work presented. Table 2 presents the ANSYS Fluent® results and parameters of the simulation conducted for each actuating pressure to verify the simulation results. Subsequent verification of solutions is demonstrated in Table 2. Evidence to confirm the present ANSYS Fluent® results is therefore justified. Table 3 shows the boundary conditions used in the CFD simulations.

5. DISCUSSION AND CONCLUSIONS

All cases follow the same trend as the fluid flux from the inlet toward the outlet. They exhibit a sigmoidal growth which is characteristic of micro pumps as inlet pressure induces a small spike in the velocity profile across the four micro pumps that remain at a steady rate towards the outlet. Also, the outlet pressure plays a significant role in the pressure drop across the devices. The outlet pressure is greater than the inlet pressure which generates pressure drop as the fluid moves through the outlet pipe. The highest spike is associated with the non-Newtonian blood flow simulated by the Carmou model in the simulation results. The fluid flow is assumed laminar, and the pressure drop across the pump is calculated using the Carmou model. However, the future of blood in a low velocity of approximately 1.3 m/s, Carmou flow was shown to achieve the best performance in terms of the facility design, that is, in terms of the shear stress for blood simulation. The current study suggests that the potential of using peristaltic pump as an efficient mechanism for transportation of both low and high velocity fluids. Bloodflow is accelerated and the fluid pressure profile can previously by the magnitude of the cross-sectional area of the pump outlet. The results demonstrate that the pressure drop increases by a rate of (shear-thinning) and (shear-thickening) efficiency between the power-law model and the Carmou model, with the latter retaining more realism. The non-Newtonian properties of blood (i.e. shear-thinning) lead to a drop in viscosity. This drop in viscosity is due to the breakup of the blood flow into small droplets or bubbles, which results in the formation of a laminar flow. This flow is characterized by an average rate in shear observed during the non-Newtonian cases. The figure’s provide a clear visual representation of the fluid behavior in 11, 2019. The peristaltic pump, non-Newtonian blood model simulated lower pressure loads due to higher viscosity nature with a stirring pressure of 0.48 Pa for a higher rate at 5.75 m/s. The shear-thinning behavior, commonly seen in non-Newtonian fluids, results in magnitudes of pressure which are significantly different than those obtained for Newtonian fluids. The starting pressure for both the power-law model and the Carmou model remains high at 0.00 Pa and is decreased to 0.00 Pa for the power-law model and 8.03 Pa for the Carmou results. This again proved the efficiency of the Carmou model over the power-law model for peristaltic pumping. Primary results show an amplification of the tendency of fluid particles to rotate or circulate at a particular point. This can contribute strongly to improving phenomena in peristaltic fluid mechanics. ANSYS Fluent® software allows to obtain the “flow physics” visualization results for the “flow physics” visualization results, which are important for both the optimization and validation of the design. However, more research is needed to determine the effect of the geometrical domain on the optimization results. The study results are important for the future of cardiac pumps and other pumps which consider magnetic-field-enabled micro pumps, which feature electrically conducting blood properties.

REFERENCES