**Title**  
Numerical simulation of a peristaltic biorheological smart micro-pump

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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 phlem translocation in bronchi, embroyic heart development, blood transport in narrow vessels and intestinal dynamics. Peristalsis' physiological effects include blood non-equilibrium properties which manifest as deformation-rate dependency, yield stress, viscoelasticity, thrombosis. 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 (micro-circulation). Non-Newtonian effects are therefore also likely to contribute strongly in micro blood-micro-pump dynamics even at low deformation rates. Modern pumps are increasingly employing biomaterials engineering 3D printed devices. In the present study, 3-dimensional simulations of both human and non-human fluids in a peristaltic micro-pump are presented. The Carreau and power-law models are depicted to characterize rheological behavior. These formulations can correctly model the Newtonian viscous behavior -vis 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 (excess units)) and pseudo-plastic (shear-thinning, Newtonian-like fluid). The shear-thinning behavior is of paramount importance for the Newtonian fluids. 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 2mm in width. The outlet of the tube leading to the pump outlet measured 0.2 mm. 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 flow of 0.2 mm width to represent the pump inlet surface. Laminar, viscous dominated peristaltic flow is considered in the geometrical micropump domain.

3. ANSYS FLUENT SIMULATION & VALIDATION

A body-fitting meshing approach, compromising mainly of tetrahedral cells to accommodate the anisotropic structure of the model, was used to produce an adequate meshing that satisfies the problem specifications. Refinements concentrated at the edges of the inlet and outlet were placed to resolve the complex flow in these areas, as the smallest elements were situated at the tube's end to accommodate boundary layer conditions. The cell distribution along the model can be seen in below. Additionally, refined mesh 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 outlet velocity extrema are stable at around 380,000 elements, with a slight difference in values after that, thus making the original mesh the ideal choice in this simulation. 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 and coupled one. The pressure projection algorithm is used, a decay of convergence and requiring more memory. Convergence is critical to achieving fast, accurate solutions. Monitoring of the regulated maximum and minimum relative residuals and pressure gradient were performed. The current choice algorithm does not guarantee the effective validity of results. Some of the cases might not fulfill the required residual criteria regardless of the accuracy of the mesh and the numerical results. The method of solution was run in an implicit method. All isotropic solutions were with low residuals. Thus, the modeling of the mass, conservation and output pressure is required. The total fractional difference between the inward and outward mass flow of the flow should be less than 0.001. It is necessary that the integral of the difference between the inflow and the outflow should be zero constant for a number of iterations before setting for convergence. Furthermore validation of the ANSYS Fluent simulations confirm the accuracy of the computations. In the present study, results of the modified model were computed and compared with the experimental work of Biki et al. on Table 1. Table 2 below shows the experimental results for the flow rate versus velocity pressure drop in the optimized design examination of the pump. The results were converted to accumulate with ANSYS Fluent pump parameter requirements and the simulation was conducted for each actuating pressure to verify the simulation results. Successful verification of solutions is demonstrated in Table 2. Consistency in the present ANSYS Fluent results is therefore justifiedly high. Table 3 shows the boundary conditions used in the CFD simulations.

4. SIMULATION & VALIDATION

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:

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

\[
\rho \left( \frac{\partial u}{\partial t} + u \frac{\partial u}{\partial x} + v \frac{\partial u}{\partial y} + w \frac{\partial u}{\partial z} \right) = -\frac{\partial p}{\partial x} + \mu \left( \frac{\partial^2 u}{\partial x^2} + \frac{\partial^2 u}{\partial y^2} + \frac{\partial^2 u}{\partial z^2} \right) + \rho f_x
\]

\[
\rho \left( \frac{\partial v}{\partial t} + u \frac{\partial v}{\partial x} + v \frac{\partial v}{\partial y} + w \frac{\partial v}{\partial z} \right) = -\frac{\partial p}{\partial y} + \mu \left( \frac{\partial^2 v}{\partial x^2} + \frac{\partial^2 v}{\partial y^2} + \frac{\partial^2 v}{\partial z^2} \right) + \rho f_y
\]

\[
\rho \left( \frac{\partial w}{\partial t} + u \frac{\partial w}{\partial x} + v \frac{\partial w}{\partial y} + w \frac{\partial w}{\partial z} \right) = -\frac{\partial p}{\partial z} + \mu \left( \frac{\partial^2 w}{\partial x^2} + \frac{\partial^2 w}{\partial y^2} + \frac{\partial^2 w}{\partial z^2} \right) + \rho f_z
\]

5. DISCUSSION AND CONCLUSIONS

All cases follow the same trend as the fluid flows from the inlet to the outlet. They exhibit a sigmoidal velocity growth which is characteristic of micro-pumps. Air inlet induces a sudden spike in the velocity profile across the four micro-pumps that waves at a density rate towards the outlet. All micro-pumps later on present a gradual increase in the velocity profile. The main reason of this is due to the volumetric flow rate of the fluid, which increases from the inlet to the outlet. For peristaltic pumps, the fluid flows through the outlet. The higher spike is associated with the non-Newtonian blood flow simulated by the Carreau model in the experimental work. The validation results in the present study are at par with the asymptotic results, which can be further inferred from the Carreau model. However, the spike settled at a lower velocity of approximately 1.5 m/s. Carreau model was found to present the best performance in terms of the power efficiency and the stability of the numerical results. The current data are well within the range of experimental results. The boundary layer thickness is increased as the flow is moved from a Newtonian fluid to a non-Newtonian fluid. Different fluid model with magnetic field and non-magnetic field are subjected for the same flow rate. Magnetic field reduces the flow rate. In addition, it reduces the average time in order to pass through the system for the non-Newtonian cases. The nozzle is a good visual representation of the fluid behavior in 1D peristaltic pumping. The Newtonian blood model predicted pressure drops due to higher viscous nature with a starting pressure of 8.6 Pa decreasing at a higher rate to 7.5 Pa. The shear-thinning behavior, common in non-Newtonian fluids, results in magnitudes of pressure drop which are significantly lower than those generated for Newtonian fluids. The starting pressure for the non-Newtonian model and the Carreau model remains at 8.6 Pa and increased to 8.80 Pa for the power-law model and 8.80 Pa for the Carreau model. This again proved the efficiency of the Carreau model over the power-law model for peristaltic pumping. Varieties showed an appearance of the tendency of a fluid to saturate or circulate at a particular point. It contributes strongly to improving phenomena in peristaltic fluid transportation. ANSYS Fluent allows direct parameter manipulation and fluid behavior. It can be shown that the flow behavior can be controlled directly by manipulating flow parameters such as pressure, velocity, mass, and temperature. This can be useful for the study of blood flow in the complex micro-geometric domain of the micro-pump.

REFERENCES