1. INTRODUCTION

Peristalsis involves the propulsion of physiological fluids via rhythmic contraction of the walls of a vessel. It arises in pleural trans-location in belays, embryonic heart development, blood transport in narrow vessels and intestinal dynamics.‡ Most physiological fluids including blood exhibit non-Newtonian properties which manifest as deformation-rate dependence, yield stress, viscoelasticity, stiffness. 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 likely to contribute strongly in blood micro-pump dynamics even at low deformation rates. Modern pumps are increasingly embracing biological mechanisms[27]. In this presentation, dimensional simulations of both non-Newtonian and non-Newtonian fluids in a peristaltic micropump are described. The Carreau and power-law models are deployed to characterize rheological behavior. These formulations can correctly model the non-linear viscoelastic 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 (excess units)) and pseudoplastic (shear-thinning, i.e. viscosity is reduced under increasing shear strain and power-law index (less than unity) behavior). Carreau and power-law model are relevant to providing more meaningful 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 length 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 in the inlet layer of 0.2 mm width to represent the pump inlet surface. Laminar, viscous dominated peristaltic flow is considered in the geometrical micro-pump model domain.

3. ANSYS FLUID SIMULATION & VALIDATION

A body-fitting meshing approach, comprising mainly of tetrahedral cells to accommodate the anomalous structure of the model, was used to produce an adequate mesh that satisfies the problem requirements. Refinement concentrations at the edges of the inlet and outlet where placed to resolve the complex flow in these regions. The smallest elements were situated at the pump outlet to accommodate boundary layer conditions. The cell distribution around the model can be seen in below. Additionally, refined results 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 footprint stabilizes around 300,000 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 latter offers a deeper degree of convergence and requiring more memory. Convergence is critical to achieving fast, accurate solutions. Monitoring of the regulated memory usage was closely monitored and a coupled algorithm was chosen. This algorithm choice does not guarantee the effective validity of results. Some of the cases might not fulfill the required residual criteria regardless of the validity of the mesh and the number of iterations. The initial solutions were even with low residuals. Thus, the monitoring of the mass, conservation and output pressure is required. The total fractional difference between the inward and outward mass flow rate of the fluid was less than 1%.

5. DISCUSSION AND CONCLUSIONS

All cases follow the same trend as the fluid flux from the inlet towards the outlet. They exhibit a sigmoidal velocity growth which is characteristic of micro-pumps As inlet velocity induces a small spike in the velocity profile across the four pump interfaces that mean at a density rate towards the outlet. The curves indicate that the pressure in the outlet of the pump where the fluid leaves the pump from the pump’s mouth. For higher velocity, the pressure decreases gradually due to thinner viscous nature. A constant pressure difference across the three micro-pumps was observed. Blood is assumed to have constant density of 1055 kg/m³. A steady state pressure solver, coupled with a viscous laminar model, is deployed. All current simulations have been performed using a Lenovo Y50 laptop machine with a 6GB RAM and an Intel Core i7-4700MQ CPU of 2.4 GHz processor with a Windows 764-bit OS CPU running on a Windows 10 platform. The solver was set to include double precision option to allow 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 limited by the length of the microchannel due to the symmetrical nature of the results on both sides.

REFERENCES