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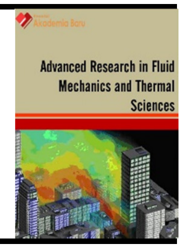

Slip Effects on Peristaltic Transport of Casson Fluid in an Inclined Elastic Tube with Porous Walls

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ABSTRACT

The present study investigates the combined effects of slip and inclination on peristaltic transport of Casson fluid in an elastic tube with porous walls. The modeled governing equations are solved analytically by considering the long wavelength and small Reynolds number approximations. A parametric analysis has been presented to study the effects of Darcy number, the angle of inclination, elastic parameters, velocity slip, yield stress, amplitude ratio, inlet and outlet radius on volumetric flow rate. The study reveals that an increase in the angle of inclination has a proportional increase in the pressure rise. Also, an increase in the porosity has a significant reduction in the pressure rise.

1. Introduction

Peristalsis is a mechanism induced by the progressive wave of area contraction and expansion which travels along the walls of the distensible tube. Over the past decades, numerous researchers have investigated the peristaltic transport due to its broad application in the field of biomedical engineering to design and construct many useful devices such as blood pump machine and dialysis machine [1]. Moreover, it is a neuromuscular property of a biological framework in which biofluids are transported along a tube by the propulsive movements of the tube wall. Many biological fluids in the human body have the peristaltic nature, such as movement of the bolus in the esophagus region, chime movement in the cervical canal, the flow of blood in arteries, transportation of urine through the ureter, etc.

The preliminary study on peristaltic flow was carried out by Latham [2] to investigate the flow of urine through the ureter. Later, Burns and Parkes [3] studied peristaltic transport by taking two cases, in the first instance they considered peristaltic flow without pressure gradient, and for the second case, they considered the flow under pressure along a channel or tube. The initial studies on peristalsis were carried by taking the Newtonian fluid to understand the physiological behavior of

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biological systems. The Newtonian approach may be sufficient to understand the flow of classical fluids through microchannel under various assumptions [4,5] and urine flow through the ureter, but it fails to explain the complex rheological action in the stomach, lymphatic vessels and flow of blood in arteries. This emphasizes to use the non-Newtonian models to investigate the physiological behavior in such systems. The initial attempt on peristaltic transport of non-Newtonian fluid was carried out by Raju and Devanathan [6] by using Power law model. Later, numerous researchers have investigated the peristaltic mechanism by using different non-Newtonian models flowing through different geometries [7-9].

The examination of non-Newtonian nature of bloodstream has been of most significance to scientists lately because of their application in exploring the conduct of blood in small arteries. In such circumstances, the presence of slip on the boundary due to the porosity of the walls has a fundamental impact in reviewing the non-Newtonian nature of blood. Hence, slip implications are more verbalized for liquids traveling through geometries which have elastic property, like blood vessels. This slip flow of fluids is used in polishing of the internal cavities and artificial heart valves. Studies on the use of porous boundaries on peristaltic transport have been initially investigated by Elshehawey *et al*., [10]. Later, numerous researchers studied peristaltic flow using Maxwell fluid by taking porous walls into account [11-13]. Most of the work done in the literature are carried out by taking Newtonian, Maxwell or Herschel-Bulkley model. This approach neglects to clarify the physiological conduct of bloodstream in supply routes because of the presence of yield stress. Even though, Herschel-Bulkley flow contains yield stress parameter, it is found that at low shear rates Casson model fits the blood flow better than that of Herschel-Bulkley fluid [14]. In this way, various researchers did the investigation of Casson and Herschel-Bulkley fluid under different physiological situations in recent times [15–20].

The nonlinearity in vascular beds of warm-blooded creatures is ascribed to the flexible idea of veins and their immense distensibility. This elastic property of veins was first perceived by Young [21]. Afterwards, Rubinow and Keller [22] exhibited that the scope of the tube could be controlled by the strain in the dividers and the transmural weight contrast by accepting that the Poiseuille law holds locally. Consequently, there is a necessity for the subjective speculation of blood flow through tubes which are elastic in nature. The stream designs acquired by the models with rigid tube can't clarify the conduct of the stream of blood through contracted supply routes completely. Henceforth, it becomes important to consider the elasticity in the present model.

To the best of authors knowledge, no attempts have been made in the literature to investigate the combined effects of slip and inclination on peristaltic transport of Casson fluid in an axisymmetric porous tube. The present investigation is helpful in filling the gap in this direction. The obtained results are in good agreement with that of Vajravelu *et al*., [16] and Sumalatha and Sreenadh [23] in the absence of velocity slip and porous walls. The resulting equations are solved analytically under the appropriate slip boundary conditions. The influence of yield stress, amplitude ratio, Darcy number, slip parameter and elastic parameters on flux are represented graphically. The outcomes of the present model help in understanding the complex physiological response of blood in the circumstances mentioned above, which intern helps medical people to investigate the blood flow in arteries much better way than the earlier and helps in modeling the heart-lung, dialysis machines, etc.

2. Mathematical Formulation

The flow of blood is modelled to be laminar, steady, incompressible, two-dimensional, fully-developed, axisymmetric, inclined at an angle β with the horizontal and exhibiting peristaltic motion in an elastic tube (Fig. 1). The flow of blood is modelled by the Casson fluid and facilitates the choice of the cylindrical coordinate system to study the problem. The wall deformation, due to an infinite sinusoidal propagation of peristaltic waves, is represented by

$$h(r,z)=1+\varepsilon\sin\left[\frac{2\pi}{\lambda}(z-ct)\right]. \tag{1}$$

where ε is the amplitude ratio, λ is the wavelength, c is the wave speed, t is the time and z is the axial direction.

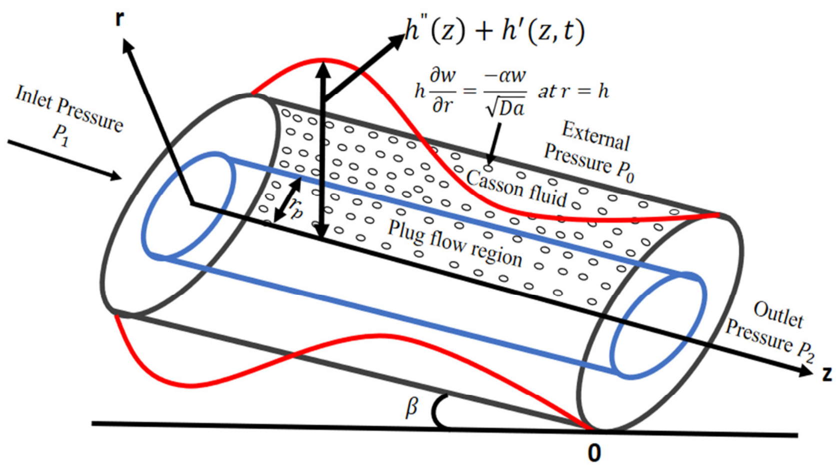

Fig. 1. Geometrical representation of Peristaltic waves in an elastic tube

3. Mathematical Modelling

The equations of motion and energy in the wave frame of reference, moving with speed *c*, under the lubrication approach (Nadeem and Akbar [24]) is as follows:

$$R_e\left(u\frac{\partial}{\partial r}+w\frac{\partial}{\partial z}\right)w=-\frac{\partial p}{\partial z}+\frac{1}{r}\frac{\partial}{\partial r}(r\tau_{rz})+\delta\frac{\partial}{\partial r}(\tau_{zz}), \tag{2}$$

$$R_e\delta^3\left(u\frac{\partial}{\partial r}+w\frac{\partial}{\partial z}\right)u=-\frac{\partial p}{\partial z}+\frac{\delta}{r}\frac{\partial}{\partial r}(r\tau_{rr})+\delta^2\frac{\partial}{\partial r}(\tau_{rz}), \tag{3}$$

where *u* and *w* are the axial and radial velocities, R_e is the Reynolds number, δ is wave number, *r* is radial coordinate, τ_{rr} is shear stress in radial coordinates τ_{zr} is shear stress in axial and radial coordinates, τ_{zz} is shear stress in axial coordinate and τ_{rz} is the shear stress along radial and axial coordinates.

The following nondimensional variables are introduced.

$$\bar{r}=\frac{r}{a},\ \bar{z}=\frac{z}{\lambda},\ \bar{t}=\frac{ct}{\lambda},\ \bar{\tau_0}=\frac{\tau_0}{\mu_0\left(\frac{c}{a}\right)},\bar{\tau_{rz}}=\frac{\tau_{rz}}{\mu_0\left(\frac{c}{a}\right)},\bar{P}=\frac{Pa^2}{\lambda c\mu_0},R_e=\frac{\rho ca\delta}{\mu_0},$$

$$\bar{r_p}=\frac{r_p}{c},\ \varepsilon=\frac{b}{a},,\ \bar{u}=\frac{u}{c},\ \bar{w}=\frac{w}{c},\ \bar{\tau_a}=\frac{\tau_a}{\mu_0\left(\frac{c}{a}\right)},\ \ F_1=\frac{\mu_0 c}{\rho g a^2}. \tag{4}$$

Under the assumption of long wavelength $\delta << 1$ and small Reynolds number $(R_e = 0)$, equations (2) and (3) takes the form as,

$$\frac{1}{r}\frac{\partial}{\partial r}(r\tau\ )=-\frac{\partial p}{\partial z}+\frac{\sin\beta}{F_1}, \tag{5}$$

$$0=\frac{\partial p}{\partial r}. \tag{6}$$

The constitutive equation for Casson's fluid in the non-dimensional form is given by

$$\frac{\partial w}{\partial r}=-\left[\sqrt{\tau_{rz}}-\sqrt{\tau_0}\right]^2,\quad \tau_{rz}\geq\tau_0, \tag{7}$$

$$\frac{\partial w}{\partial r}=0,\quad \tau_{rz}\leq\tau_0. \tag{8}$$

The corresponding non-dimensional boundary conditions are

$$h\frac{\partial w}{\partial r}=\frac{-\alpha w}{\sqrt{Da}}\ \text{ at } r=h \tag{9}$$

τ_{rz} is finite at $r=0$. (10)

The slip boundary condition (equation (9)) is given by Saffman [25]. Further, Da represents the Darcy number (porous parameter) and α represents the slip parameter.

Solving equations (5) and (6) subjected to the boundary conditions (9) and (10) we obtain the velocity w as,

$$w\ =\frac{P+f}{2}\left[\frac{4}{3}r_p^{1/2}\left(r^{3/2}-h^{3/2}\right)-\frac{1}{2}\left(r^2-h^2\right)-r_p(r-h)\right]+\frac{h\sqrt{Da}(P+f)}{\alpha}\left[h+r_p-2\sqrt{hr_p}\right], \tag{11}$$

where $P=-\frac{\partial P}{\partial z}$.

Using the condition $\tau_0=\frac{Pr_p}{2}$ at $r=r_p$, the upper limit of plug flow region is obtained as $r_p=\frac{2\tau_0}{P}$. Also by using the condition $\tau_{rz}=\tau_h$ at $r=h$ (Bird et al., [26]), we obtain

$$P=\frac{2\tau_h}{h}. \tag{12}$$

Hence,

$$\frac{r_p}{h} = \frac{\tau_0}{\tau_h} = \tau. \tag{13}$$

Using relation (13) and taking $r = r_p$ in equation (11), we obtain the plug flow velocity w_p as,

$$w_p = \frac{P+f}{2}\left[r_p h - \frac{4}{3} r_p^{1/2} h^{3/2} - \frac{r_p^2}{6} + \frac{h^2}{2} \right] + \frac{h\sqrt{Da}(P+f)}{\alpha}\left[h + r_p - 2\sqrt{h r_p} \right]. \tag{14}$$

The instantaneous flow rate Q across any cross section of the artery is defined as given below:

$$Q = 2\left[\int_0^{r_p} w_p \, r \, dr + \int_{r_p}^{h} w \, r \, dr \right]. \tag{15}$$

$$Q = \frac{(P+f)Fh^4}{8}, \tag{16}$$

where, $F = \left[1 - \frac{16}{7}\tau^{1/2} + \frac{4}{3}\tau - \frac{\tau^4}{21} + \frac{2\sqrt{Da}}{\alpha}\left(1 - \tau^{1/2}\right) \right].$

It is worth mentioning that, the results of Vajravelu *et al.*, [16] can be obtained as a special case by taking $Da = 0$ and $\beta = 0$ in equation (16).

4. Theoretical Determination of Flux with an Application to Flow through an Artery

A theoretical calculation of the flux Q is carried out for an incompressible Casson fluid through an elastic tube of radius $h(z, t) = h'(z, t) + h''(z)$. The fluid is assumed to enter the tube with a pressure P_1 and leave the tube with pressure P_2, while the pressure outside the tube is P_0. If z denotes the distance along the tube from the inlet end, then the pressure $P(z)$ in the fluid at z diminishes from $P(0) = P_1$ to $P(\lambda) = P_2$. The tube may contract or expand due to the difference in pressure of the fluid $P(z) - P_0$. Subsequently, the cross section of the tube may have a deformation due to the elastic property of the walls. Thus, the difference in pressure influences the conductivity σ_1 of the tube at z. We consider the conductivity $\sigma_1 = \sigma_1[P(z) - P_0]$ to be a known function of the pressure difference $(P(z) - P_0)$. This conductivity is assumed to be the same as that of a uniform tube having an identical cross section at z. The relation between Q and the pressure gradient is given by

$$Q = \sigma_1 (P - P_0)(P + f). \tag{17}$$

Under the considerations of peristaltic motion and the elastic property of the tube wall, equation (17) can be written as,

$$\sigma_1(P-P_0)=\frac{F}{8}(h'+h'')^4, \tag{18}$$

where h'' is the change in radius of the tube due to elasticity and is a function of pressure $P-P_0$ at each cross section due to the Poiseuille flow *i.e.,* $[h''(P-P_0)]$. Equation (17) with the inlet condition $P(0)=P_1$ gives

$$Q \ = \int_{P(z)-P_0}^{P_1-P_0}\sigma_1(P')dp'+\int_0^1\frac{F}{8}f(h'+h'')^4, \tag{19}$$

where $P'=P(z)-P_0$. This equation gives $P(z)$ in terms of z and Q. Setting $z=1$ and $P(1)=P_2$ in equation (19), we get Q as,

$$Q=\int_{P_2-P_0}^{P_1-P_0}\sigma_1(P')dp'+\int_0^1\frac{F}{8}f(h'+h'')^4. \tag{20}$$

Now, using equation (18) in equation (20), we have

$$Q=\frac{F}{8}\left[\int_{P_2-P_0}^{P_1-P_0}(h'+h'')^4dp'+f(h'+h'')^4\right]. \tag{21}$$

Equation (21) can be solved if we explicitly know the function $h''(P-P_0)$. If h'' is known as a function of the tension $T(h'')$ or stress, then $h''(P')$ can be determined from the equilibrium condition (Rubinow and Keller [22]) given by

$$\frac{T(h'')}{h''}=P-P_0. \tag{22}$$

Rubinow and Keller [22] carried out experimental investigations by controlling static pressure volume connection of a 4-cm long piece of a human iliac artery and gave an expression for tension in an elastic tube as:

$$T(h'')=t_1(h''-1)+t_2(h''-1)^5. \tag{23}$$

Using equation (23) with $t_1=13$ and $t_2=300$, equation (21) takes the following form

$$dp'=\left[\frac{t_1}{h''^2}+t_2\left(4h''^3-15h''^2+20h''-10+\frac{1}{h''^2}\right)\right]dh''. \tag{24}$$

Using equation (24), equation (21) can be written as

$$Q=\frac{F}{8}\left[\int_{P_2-P_0}^{P_1-P_0}(h'+h'')\left[\frac{t_1}{h''^2}+t_2\left(4h''^3-15h''^2+20h''-10+\frac{1}{h''^2}\right)\right]dh''+f(h'+h'')^4\right] \tag{25}$$

Letting $P=P_1$ and $P=P_2$ in equation (22) the solutions are obtained for h_1'' and h_2'' respectively.

Equation (25) can be rewritten as

$$Q=\frac{F}{8}\left[\left(g(h_1'')-g(h_2'')\right)+f\,h_2''^4\right], \tag{26}$$

where,

$$
\begin{aligned}
g(h) = t_1 & \left(\frac{h''^3}{3} + 2h'h'' + 6h'^2 h'' + 4h'^3 \log\log h'' - \frac{h'^4}{h''} \right) + t_2 \left(\frac{h''^8}{2} + \frac{h''^7}{7}(16h' - 15) + \right. \\
& \frac{h''^6}{6}\left(24h'^2 - 60h' + 20\right) + \frac{h''^5}{5}\left(16h'^3 - 90h'^2 + 80h' - 10\right) \\
& + \frac{h''^4}{4}\left(4h'^4 - 60h'^3 + 120h'^2 - 40h'\right) + \frac{h''^3}{3}\left(-15h'^4 + 80h'^3 - 60h'^2 + 1\right) \\
& \left. + \frac{h''^2}{2}\left(20h'^4 - 40h'^3 + 4h'\right) + h''\left(-10h'^4 + 6h'^2\right) + 4h'^3 \log\log h'' - \frac{h'^4}{h''} \right).
\end{aligned}
\tag{27}
$$

5. Results and Discussion

The present paper emphasizes the flow behavior of peristaltic transport of blood in an elastic tube, modeled as a Casson fluid. The present model is the extension of the work carried out by Rubinow and Keller [22] and Vajravelu *et al.*, [16] by considering the effects of slip and inclination in a porous tube. The results are obtained for the effects of various physiological parameters on the flow rate are analyzed graphically with the help of MATLAB and are presented in Figures 2 to 11.

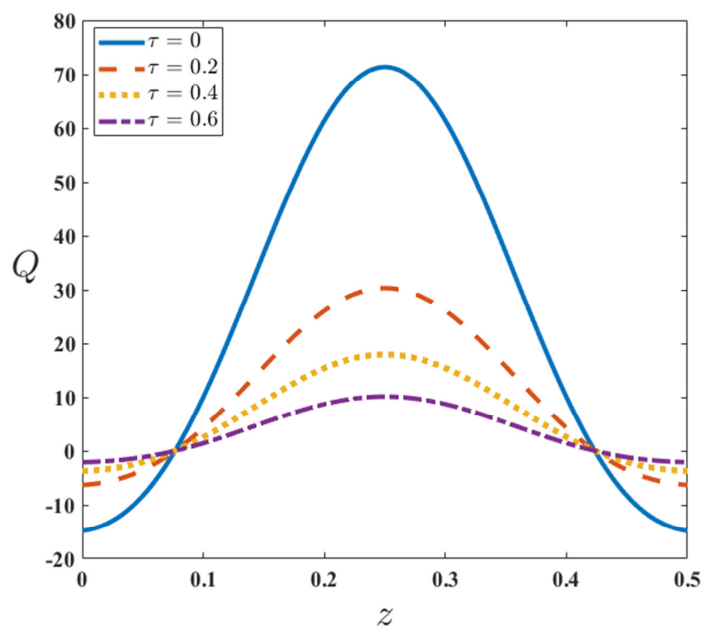

Fig. 2. Q versus z for varying τ with $h_2'' = 0.2,\ t_1 = 13,\ t_2 = 300,$ $\varepsilon = 0.6,\ \beta = \frac{\pi}{4},\ Da = 0.02,\ \alpha = 0.2,\ h_1'' = 0.3.$

Figure 2 shows the variation of yield stress (τ) on flow rate (Q). It is noticed that an increase in the values of τ decreases the flow rate in an inclined elastic tube. This behavior is expected due to the presence of yield stress in the model. Figure 3 compares the results of present and Vajravelu et al. [16] model for a fixed value of yield stress $(\tau = 0.2)$. The results of the present model are plotted by taking $Da = 0$ and $\beta = 0$ to show the comparison with their model. It is clear from Figure 3 that the results of the present model exactly match with their results, which also validate our results. Further, the slight variation is observed due to the nature of sinusoidal wave considered in the models. Figures 4 and 5 represent the behavior of amplitude ratio (ε) and porous parameter (Da) on Q respectively. It is observed that an increase in the values of ε and Da increases the flow rate Q. The effect of velocity slip parameter shows the opposite trend as that of ε and Da (Figure 6). The variation of angle of inclination (β) on flow rate Q is plotted in Figure 7. It is noticed that Q increase with an increase in the angle β. The observation of β is in concurrence with the results of Sumalatha and Sreenadh [23]. Figures 8 and 9 are plotted to examine the variation of elastic parameters t_1 and t_2 on flow rate Q respectively. Flux in an elastic tube increases with an increase in the values of elastic parameter t_1 (Figure 8). The same trend holds good for the other parameter, namely t_2 (Figure 9). The flux profiles with inlet and outlet elastic radius variations are shown graphically in Figures 10 and 11. For a fixed value of outlet radius, the effect of increasing values of inlet elastic radius makes the flux to decrease and hence flow rate decreases as the elastic radius increases (Figure 10). However, the opposite behavior is observed when we fix inlet elastic radius and vary outlet elastic radius (Figure 11).

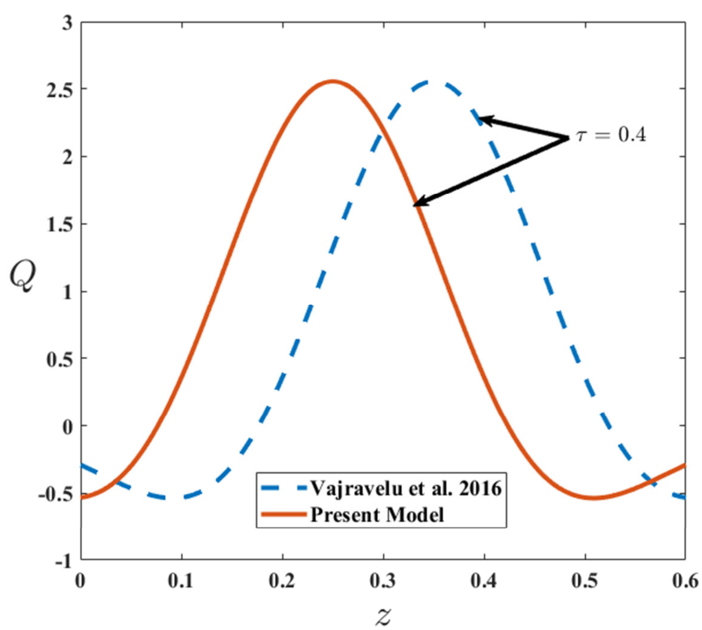

Fig. 3. Q versus z with $h_2^{"} = 0.2,\ t_1 = 13,\ t_2 = 300,$ $\varepsilon = 0.6,\ \beta = \dfrac{\pi}{4},\ Da = 0.02,\ \alpha = 0.2,\ h_1^{"} = 0.3.$

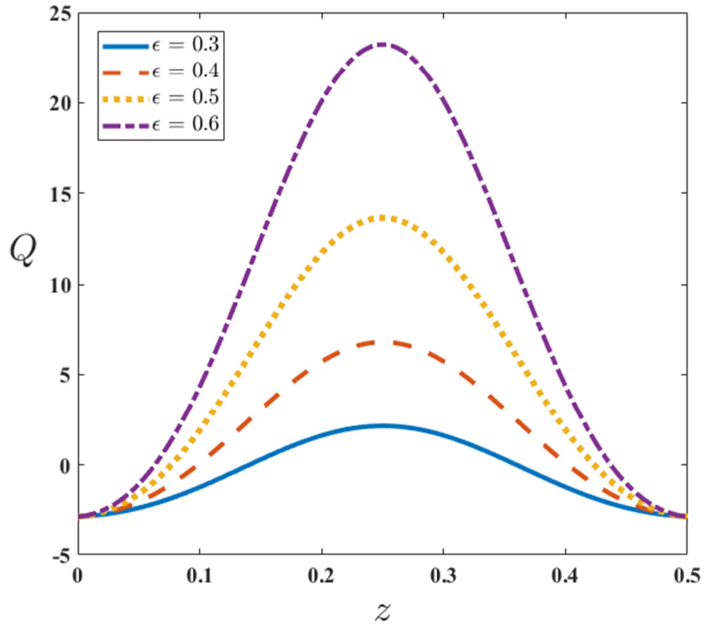

Fig. 4. Q versus z for varying ε with $h_2^{''}=0.2,\ t_1=13,\ t_2=300,$ $\tau=0.5,\ \beta=\dfrac{\pi}{4},\ Da=0.02,\ \alpha=0.2,\ h_1^{''}=0.3.$

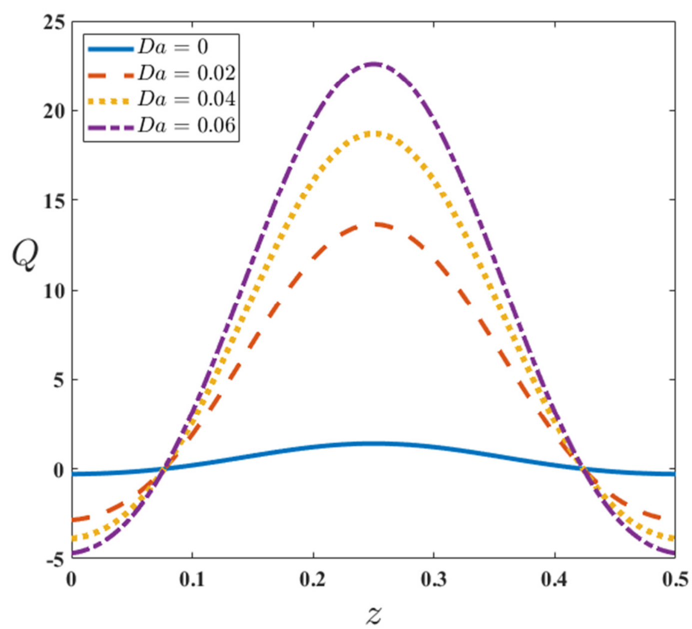

Fig. 5. Q versus z for varying Da with $h_2^{''}=0.2,\ t_1=13,\ t_2=300,$ $\tau=0.5,\ \beta=\dfrac{\pi}{4},\ \varepsilon=0.5,\ \alpha=0.2,\ h_1^{''}=0.3.$

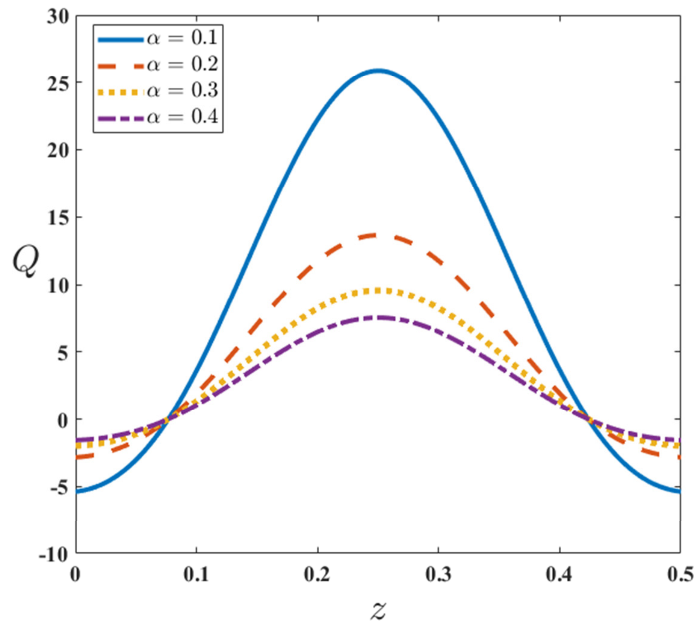

Fig. 6. Q versus z for varying α with $h_2^{"} = 0.2,\ t_1 = 13,\ t_2 = 300,$ $\tau = 0.5,\ \beta = \dfrac{\pi}{4},\ \varepsilon = 0.5,\ Da = 0.02,\ h_1^{"} = 0.3.$

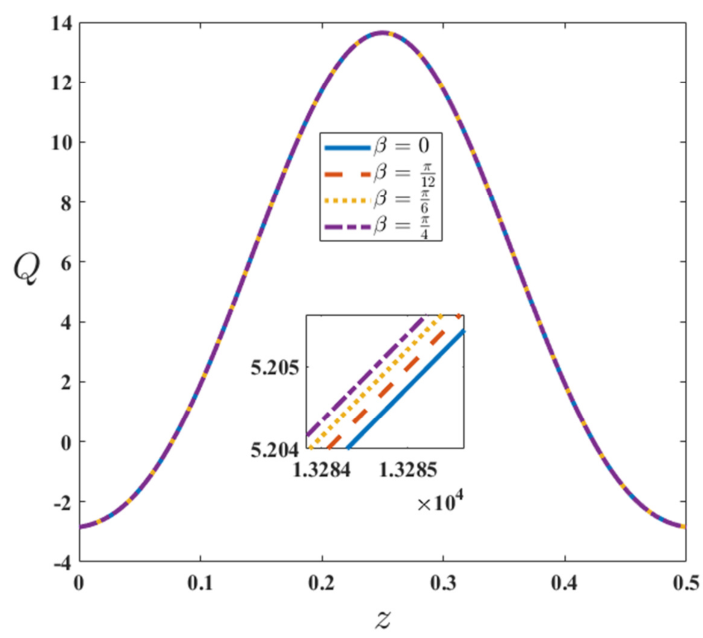

Fig. 7. Q versus z for varying β with $h_2^{"} = 0.2,\ t_1 = 13,\ t_2 = 300,$ $\tau = 0.5,\ \alpha = 0.2,\ \varepsilon = 0.5,\ Da = 0.02,\ h_1^{"} = 0.3.$

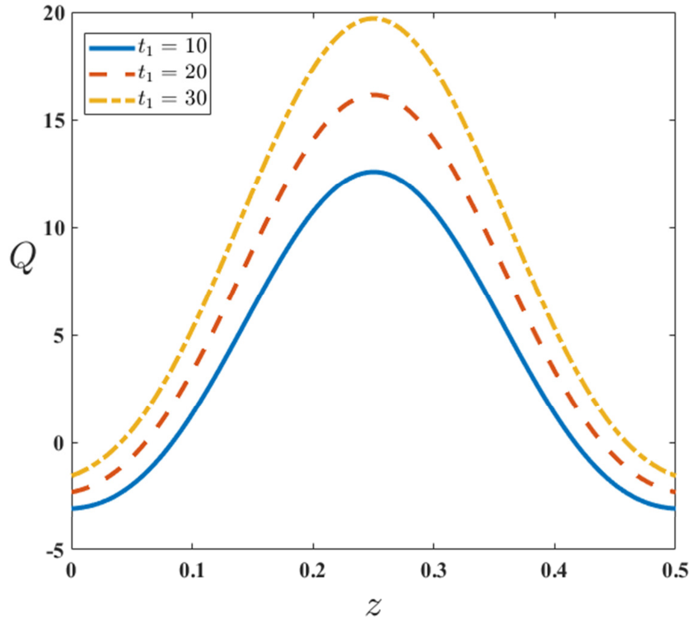

Fig. 8. Q versus z for varying t_1 with $h_2^{''} = 0.2,\ \beta = \dfrac{\pi}{4},\ t_2 = 300,$ $\tau = 0.5,\ \alpha = 0.2,\ \varepsilon = 0.5,\ Da = 0.02,\ h_1^{''} = 0.3.$

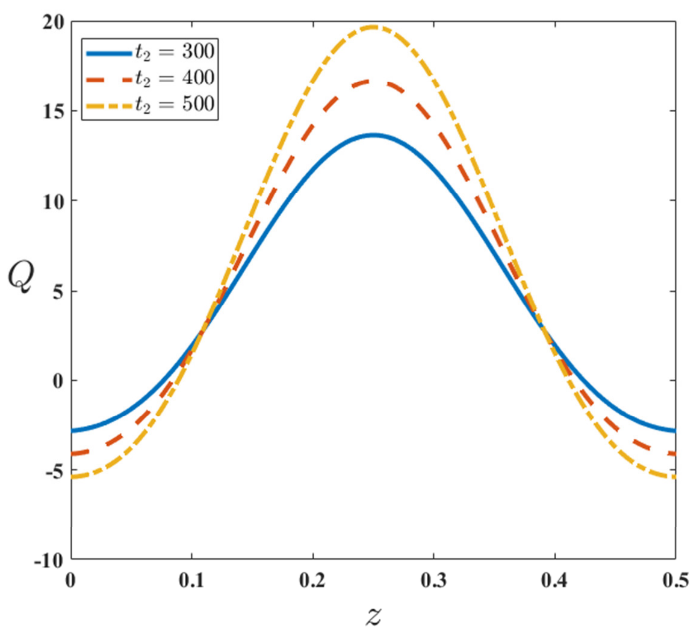

Fig. 9. Q versus z for varying t_2 with $h_2^{''} = 0.2,\ \beta = \dfrac{\pi}{4},\ t_1 = 13,$ $\tau = 0.5,\ \alpha = 0.2,\ \varepsilon = 0.5,\ Da = 0.02,\ h_1^{''} = 0.3.$

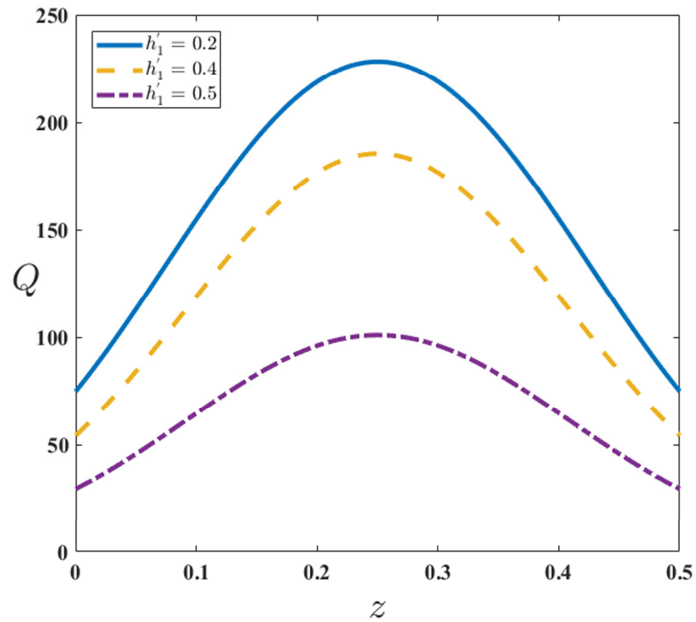

Fig. 10. Q versus z for varying h_1'' with $h_2'' = 0.6,\ \beta = \dfrac{\pi}{4},\ t_1 = 13,$ $\tau = 0.5,\ \alpha = 0.2,\ \varepsilon = 0.5,\ Da = 0.02,\ t_2 = 300.$

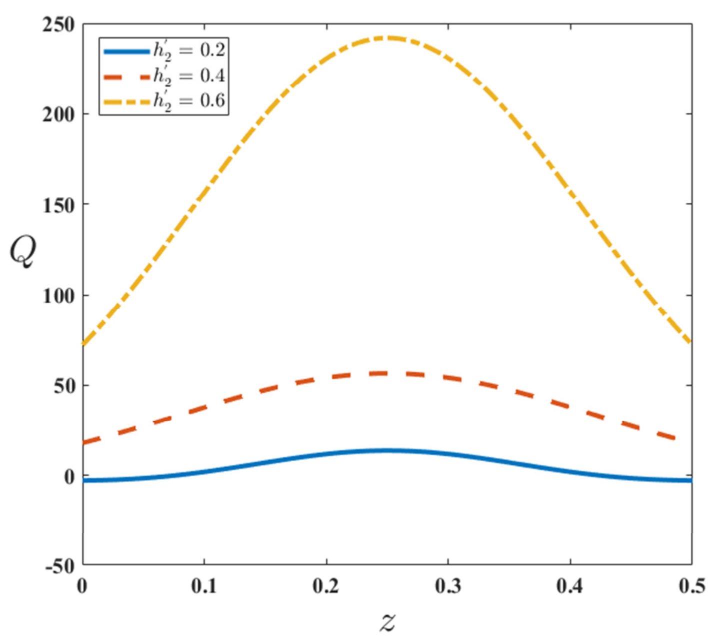

Fig. 11. Q versus z for varying h_2'' with $h_1'' = 0.3,\ \beta = \dfrac{\pi}{4},\ t_1 = 13,$ $\tau = 0.5,\ \alpha = 0.2,\ \varepsilon = 0.5,\ Da = 0.02,\ t_2 = 300.$

6. Summary and Conclusion

The present investigation deals with the study of peristaltic motion of blood in the human circulatory system. The blood flow is modeled by Casson fluid in an elastic tube with porous walls. The present investigation gives an agreeable result that speaks to a portion of the natural phenomena, mainly, the stream of blood in narrow arteries which can be taken care of and prepared if there should be an occurrence of dysfunction. Further, there is a possibility of extending the present model under the influence of external magnetic field which will help the doctors to control the flow of blood during surgeries. Some of the interesting findings are

- The effects of increasing values of yield stress and inlet elastic radius decreases the flux in an elastic tube, whereas, flux increases with increasing values of amplitude ratio, elastic parameters and outlet elastic radius.
- The flux in an elastic tube increases with an increasing value of the Porous parameter (Darcy number) and decreases with slip parameter.
- The effect of angle of inclination has a significant role in increasing the flow rate in an elastic tube.
- The effects of velocity slip, angle of inclination and porous walls have a major role in determining the flux in an elastic tube.

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