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A STUDY ON THE APPLICATIONS OF VISCO FLOWS WITH A REFERENCE TO ELASTIC FLUID IN TUBES

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ABSTRACT

The numerical study of this problem focuses on how waves go through a distensible tube that is filled with a viscous fluid. The most important use is in biofluid mechanics, which places a significant emphasis on the interaction between fluids and solids. A quasi-1D differential model was used to represent the data since it was assumed that pressure waves would have a long wavelength and a modest amplitude. The model took into consideration the viscoelastic properties of the the vessel wall showed deformations along both the radial and axial axes. The nonlinear problem was solved using a finite difference method on a staggered grid. The response was provided without the use of any dimensions. The boundary conditions took into consideration both persistent oscillations caused by a periodic pushing pressure and spontaneous oscillations in a deformable tube fixed at the ends. Natural frequency (denoted by the symbol S t) was found to be independent of viscosity and to approach equality with the square root of the elasticity coefficient. It has been shown that the damping time is inversely proportional to the wall viscosity coefficient, and that the viscosity of the fluid provides an even bigger damping factor. An important factor in these findings.

Keywords: *Visco Flows, Tubes*

INTRODUCTION

Viscous fluid flows in elastic tubes are well understood prevalent in various applications, such as modeling blood flow difficulties. One such use is the simulation of blood flow. Experiments have shown that when an uneven force disturbs a continuous flow in a tube that can expand and contract, which results in the creation of waves that go in the opposite direction. These waves were generated when there was a disruption in the flow. In spite of the fact that a number of hypotheses have been put forward to explain this phenomenon, the mechanism behind its spread is not yet completely understood. This is as a result of the complicated nature of the system as well as the nonlinear interaction between the fluid and the structure. Some of the most fundamental aspects of the damping effect can be understood with a reasonable amount of effort, and simpler models can provide some insightful clues about the behaviour of wave transmission. As a consequence of this, we are able to carry out an exhaustive investigation over a wide variety of dimensions.

Although multi-dimensional models provide a better description of physiological flows, simpler models provide some valuable information on the wave propagation. In point of fact, the unidirectional character of blood flow motivates the effort to apply the approximation to lengthy arterial conduits because of how accurate it is likely to be. In attempt to get some understanding of the mechanical contact that occurs between the blood and the arterial wall, a number of different investigations have been conducted on the process of pulse propagation in arteries. A significant portion of these models concentrate on an incompressible Newtonian fluid that is encased in a tube that is compliant. In addition to this, the majority rely on linearized equations of motion and stress-strain ratios in your analysis.

It is common knowledge that such linear models lead the propagating pulses to take on flow patterns resembling shock waves, which are not seen under physiological conditions. The reason for this is that the models wrongly assume that the vessel wall is completely elastic, despite the fact that this is not the case. Blood vessel walls, on the other hand, have been found to be nonlinear, viscoelastic, and anisotropic by a number of different writers. If an appropriate mathematical model that takes into account the viscoelastic features of the wall is added into the one-dimensional theory, then it is possible to generalise the theory to further contexts. The damping that is produced as a result of viscoelasticity helps to prevent the high peaks of pressure and flow pulses and also helps to smooth out the rapid ascent of wave fronts. Both of these effects are due to viscoelasticity. When compared to the findings of experiments, the predictions that these models produce are consequently more accurate. The vast majority of these articles, on the other hand, model artery wall distensibility by making use of nonlinear algebraic relationships between tube cross section and transmural pressure. Sometimes it is feasible to integrate along the characteristics while the wave rapidity is expressed openly, and these relations sometimes are a function of the frequency. Sometimes it is possible to integrate along the characteristics while the wave rapidity is represented clearly. In contrast to this method, which is devoid of any true mechanical explanation, many constitutive strainstress equations have been developed in order to define the mechanical properties of the artery wall. Recent research has focused on analysing two-dimensional flow inside of a rigid channel using a flexible viscoelastic membrane in place of a portion of the wall. This article demonstrates that a similar but simpler one-dimensional model is sufficient for characterising propagative processes, and it does so by demonstrating the model's applicability to the problem.

In this section, we address the importance that the vessel's constitutive equation plays in arterial flow difficulties, as well as the wall-fluid interaction, which contains a viscoelastic factor, are both investigated in this study. The potential to comprehend the development of minor flow disruptions caused either by the insertion of a local vessel or by a diseased condition is what prompted this research in the first place. The investigation of models has shown to be quite helpful in shedding light on the process involved, as well as the changes to the pulse propagation. The quasi flow equations paired with the massless membrane equilibrium equation provide the foundation of the model that is used. In addition to this, a

We offer a strain-stress constitutive equation for the wall that allows for bending in both the radial and longitudinal axes. Even though inertia is the principal factor responsible for blood flow in large arteries, circumstances involving very tiny Reynolds number limits as well as very large Reynolds number limits are also taken into account. This is done in order to have a better understanding of the dissipative mechanism that is caused by viscous friction. This mechanism is responsible for producing damped waves as the system gets closer to its steady state and is an essential component of the process. The crucial parameters that are responsible for the probable numerical instability are highlighted below. These factors are involved in the mechanics.

OBJEACTIVES

1. The Study Visco Flows with A Reference to Elastic Fluid in Tubes.
2. The Study Many practical situations involve the flow of viscous fluids through flexible tubes.

RESEARCH METHODOLOGY

We look at the flow of an incompressible viscoelastic fluid that is axi-symmetric through a circular cylindrical diverging tube. It is possible to provide the governing equations for a viscoelastic fluid which is characterized by the Maxwell model.

$$\left(1+t_m\frac{\partial}{\partial t}\right)\left(\frac{\partial}{\partial t}+u\frac{\partial}{\partial r}+w\frac{\partial}{\partial z}\right)u=-\frac{1}{\rho}\left(1+t_m\frac{\partial}{\partial t}\right)\frac{\partial p}{\partial r}+\nu\left(\frac{\partial^2}{\partial r^2}+\frac{\partial^2}{\partial z^2}+\frac{1}{r}\frac{\partial}{\partial r}-\frac{1}{r^2}\right)u$$

$$\left(1+t_m\frac{\partial}{\partial t}\right)\left(\frac{\partial}{\partial t}+u\frac{\partial}{\partial r}+w\frac{\partial}{\partial z}\right)w=-\frac{1}{\rho}\left(1+t_m\frac{\partial}{\partial t}\right)\frac{\partial p}{\partial z}+\nu\left(\frac{\partial^2}{\partial r^2}+\frac{\partial^2}{\partial z^2}+\frac{1}{r}\frac{\partial}{\partial r}\right)w$$

$$\frac{\partial u}{\partial r}+\frac{\partial w}{\partial z}+\frac{u}{r}=0,$$

where tm is the relaxation time, t is the time parameter, u and w are radial and axial velocities, r and z are radial and axial coordinates, and are the density and kinematic viscosity of the fluid, and p is the pressure. In addition, r and z are the radial and axial coordinates. The fluid's density and kinematic viscosity are found in the fluid. It is anticipated that the following series of gradually decreasing contraction waves have a tiny amplitude and will travel down the wall of the tube:

$$H(z,t)=a(z)-b\cos^2\frac{\pi}{\lambda}(z-ct),$$

where H represents the wall displacement, a(z) represents the radius of the tube, b represents the amplitude of the wave, represents the wavelength, and c represents the wave's velocity. Although Eq. (4) allows for expansion and contraction of the wall, it does not allow the wall to move beyond its natural boundaries. This is consistent with the wall motion seen in physiological ducts.

where the particular solution is

$$\phi_{21p}(r) = \frac{A_0}{24} r^2 I_3(\alpha r)\left[6\left(1-k_0 a^2\right)+k_0 r^2\right],$$

and the constants A1 and B1 are given b

$$A_1 = \frac{A_0}{24\Delta} a^2\left[\alpha\left(6-5k_0a^2\right)\left(I_1I_2 - I_0I_3\right)+2k_0 a I_1 I_3\right],$$
$$B_1 = \frac{A_0}{24\Delta} a^2\left[a\alpha\left(6-5k_0a^2\right)\left(I_1I_3 - I_0I_2\right)+12\left(1-k_0a^2\right)I_0I_3\right].$$

. The solution of (48) may be writte

$$\begin{aligned}\phi_{12}(r) = {} & A_2 r I_0(\alpha r) + B_2 I_1(\alpha r) - \frac{A_1}{24} r^2 I_3(\alpha r)\left\{6\left(1-k_0a^2\right)+k_0r^2\right\}\\ & -\frac{A_0}{12}\left[\left(1-k_0a^2\right) r^3 I_4(\alpha r)\left(\frac{1-k_0a^2}{2}+\frac{k_0r^2}{5}\right)+\frac{k_0^2}{5\alpha}a^2 r^6 I_5(\alpha r)+\frac{k_0^2}{42} r^7 I_6(\alpha r)\right]\\ & -\frac{A_0}{24}\frac{t_m\alpha}{R} r^2 I_3(\alpha r)\left[6\left(1-k_0a^2\right)-k_0r^2\right]\end{aligned}$$

Where

$$\begin{aligned}A_2 = {} & -\frac{A_1^2}{A_0} - \frac{A_0 a^3}{120\alpha\Delta}\Bigg[\left(1-k_0a^2\right)\alpha^2\left(5-3k_0a^2\right)\left(I_1I_3 - I_0I_4\right)\\ & + a\alpha\left\{2k_0^2a^4+4k_0\left(1-k_0a^2\right)\right\} I_1I_4 - 2k_0^2a^5\alpha I_0I_5 + \left(4+\frac{5}{21}\alpha^2\right)k_0^2a^4 I_1I_5 + \frac{5}{21}\alpha\, k_0^2 a^3\left(2I_1I_6 - a\alpha I_0 I_6\right)\Bigg]\\ & -\frac{A_0}{24\Delta}\frac{t_m\alpha}{R}a^2\left[\alpha\left(6-5k_0a^2\right)\left(I_1I_2 - I_0I_3\right)+2k_0 a I_1I_3\right]\\ B_2 = {} & -\frac{A_1B_1}{A_0} - \frac{A_0a^3}{120\alpha\Delta}\Bigg[a\alpha^2\left(1-k_0a^2\right)\left(5-3k_0a^2\right)\left(I_1I_4 - I_0I_3\right)\\ & + \left\{10\left(1-k_0a^2\right)^2 - 2k_0^2a^6\right\}\alpha I_0I_4 + \alpha\, k_0^2a^5\left\{2aI_1I_5 + \frac{5}{21}\alpha\left(I_1I_6 - I_0I_5\right)\right\}\Bigg]\\ & + \frac{A_0}{24}\frac{t_m\alpha}{R\Delta}a^2\left[a\alpha\left(6-5k_0a^2\right)\left(I_0I_2 - I_1I_3\right) - 2I_0I_3\left(6-5k_0a^2\right)\right].\end{aligned}$$

DATA ANALYSIS

The analysis was carried out to explore the effects of relaxation time, the tapering of the tube, and Reynolds number, which is commonly disregarded when studying creeping flows when the fluid in issue possesses a viscoelastic property and the flow pattern is what it is. flows when the viscoelastic property is present. The purpose of this was to get insight into the changes that occur to the flow pattern when a viscoelastic fluid is present. Measurements of the vas deferens are carried out so that we can determine whether or not the parameters utilised in the analysis are appropriate. Because of this, we are able to extrapolate the theoretical model to the flow of the vas deferens. The evidence that was presented by Guha et al. indicated that the rhesus monkey possesses the following characteristics: When k0 equals zero, a flow that is totally peristaltic occurs, and this state of affairs is referred to as free pumping. Take the gradient of the tube to be K, where K is the radius of the tube, and let the gradient be represented by the parameter.

When the amplitude ratio is shown against the average flow rate that is calculated We find that the flow rate decreases with increasing tm values by utilising a range of low Reynolds numbers for various values of the temporal relaxation parameter tm. This enables us to draw this conclusion. The long wave approximation approach can be used to produce graphs that look quite similar to each other and demonstrate findings that are very similar. Whether p is positive, indicating an unfavourable pressure differential, or negative, suggesting a beneficial pressure difference, the discoveries are the same. The only difference that can be seen is that the minimum flow rate is determined by the pressure differential that is imposed at both ends of the tube. This is the only difference that can be seen. It should come as no surprise that favorable and unfavorable pressure differences result in flow rates that are positive and negative, respectively. 5a and 5b show the influence that the time relaxation tm has on the axial velocity as it travels over the radial distance. It has been noticed that there is a decrease in the axial velocity if tm is increased. On the other hand, one should keep in mind that tm is only useful for very big numbers. is predicated on the idea that the Reynolds number is relatively low, while is founded on the concept of approximating long waves.

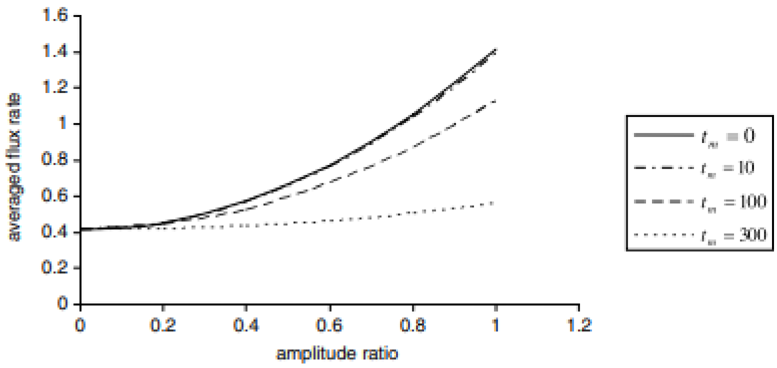

Fig. 1. For various amounts of relaxation time, the diagram shows how the average flux rate depends on the amplitude ratio.

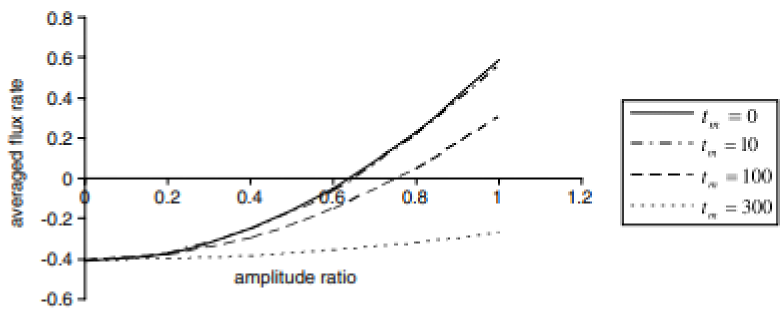

Fig. 2 The diagram that follows demonstrates, for a range of different amounts of relaxation time, how the average flux rate is dependent on the amplitude ratio.

There is a linear connection between the two variables, the pressure differential and the mean flow rate. Between the two variables. When the Reynolds number is increased, there is a significant rise in the flow rate; however, when the time relaxation parameter is increased, there is a decrease in the flow rate for a constant pressure difference. When tm is increased from 0 to 100, there is a possibility of observing a change; however, this is not the case when it is increased to 1000. It is possible to draw the conclusion from this that, assuming all other conditions remain the same, the flow rate of a viscoelastic fluid will be lower than the flow rate of a Newtonian fluid. By generating graphs that show the relationship between the radial distance and the velocity, we are able to investigate the effect that the Reynolds number has on the axial velocity. When the pressure gradient, the wave number, the degree of tapering, and the relaxation time are all held constant, it has been proved that increasing the Reynolds number results in a considerable rise in the axial velocity. Additionally, it has been shown that the Reynolds number has a significant impact on the pressure profile that runs along the tube. Assuming that the wave number, the degree of tapering, and the flow velocity are all kept the same, Figure 3 illustrates a negative relationship between the Reynolds number and the pressure differential. This unequivocally illustrates that the amount of pumping effort that is necessary to move a particular volume of fluid is directly related to the viscosity of the fluid.

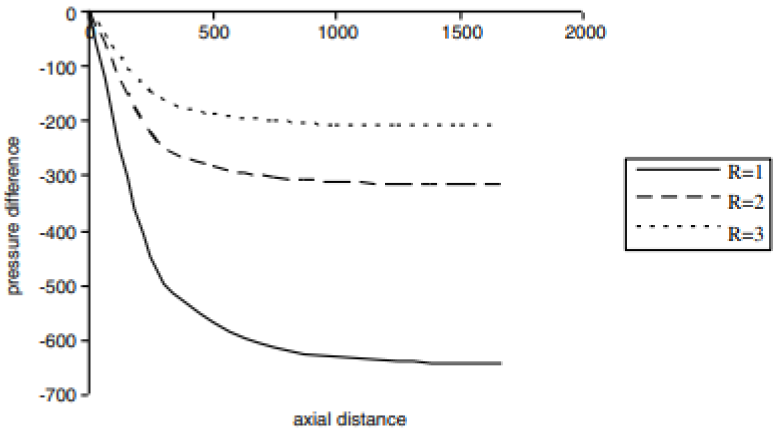

Fig. 3. Based on, This Diagram Displays Pressure Distribution Along Axial Distance For Various Reynolds Number R

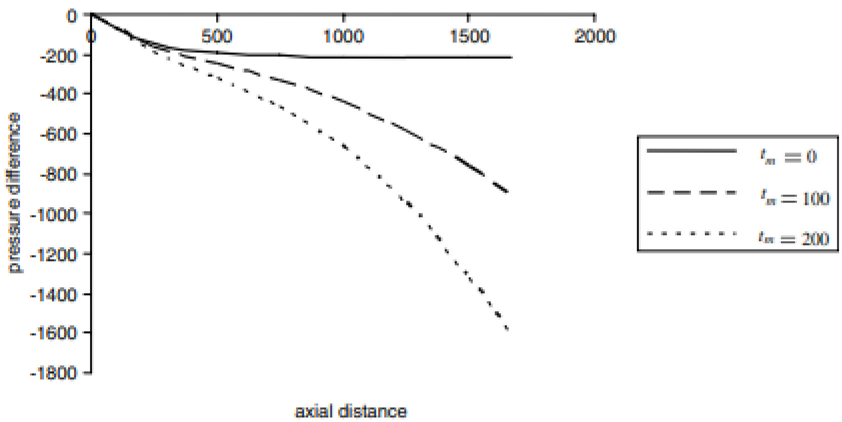

Fig. 4. Diagram Based On Demonstrates How Pressure Changes With Time Relaxation Tm Along The Tube's Length For Q¯ = 4.421, L = 1666.67.

Since seminal fluid is a viscoelastic fluid, one can reason, based on what has been discussed up to this point, that the peristaltic contribution to its flow in the vas culi of rhesus monkeys is slightly less than what Misra and Pandey postulated it to be. This conclusion is supported by the fact that seminal fluid is a viscous fluid came to the conclusion of under a comparable set of conditions. This is the conclusion that was reached by Misra and Pandey.

CONCLUSION

A coupled wall-fluid model has been provided, with application in the study of the irregular motion of a viscous liquid in a viscoelastic tube. This model was developed for the purpose of researching arterial flows. Due to the unidirectional character of blood flow, it is recommended that The observations provide considerable credence to a quasi-mathematical approximation that may be put to use in order to symbolise the waves of flow that are being sent. However, if the wavelength is sufficiently large and the wave amplitude is small in comparison to the mean radius of the tube, then the findings that this method produces may be reasonable. There is a presentation of an article that makes a proposition regarding a linear constitutive relation for the vessel wall. This relation is sensitive to both strain and strain rate. This model takes into account the interaction that occurs between the viscosity of the blood and the viscoelasticity of the solid tube. In a purely elastic wall model, the job of the viscosity parameter is to counteract any instability events and dampen the high-frequency oscillations that are present. Additionally, the viscosity parameter's role is to attenuate oscillations at lower frequencies. The influence of the elasticity parameter has a direct proportional relationship with the frequency of transient oscillations. The model research that has been presented in this article is capable of describing the fundamental physical process of pulse propagation via the nonlinear interaction that takes place between the fluid and the wall, despite the fact that the evaluation of the theory requires specific assumptions that simplify the situation in order to proceed. The determination of the numerical value of the elastic and viscous coefficients that occur in the constitutive equation as well as the boundary conditions requires a comparison of the numerical results with the experimental data observations. This may be done by following the steps outlined in the following paragraph.

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