Mathematical Modeling of Blood Flow through Arteries in the Presence of Magnetic Field with Porosity

Dr. Mahesh Chandra; Jharna Goyal
Assistant Professor
Department of Mathematics Department of Mathematics
FS University, Shikohabad (U.P.) India

Abstract:
The present paper explores a comparative study of two non-Newtonian mathematical models with porosity. The first mathematical model is based on the Power law model with porosity, and the second mathematical model is developed on the Maxwell principle having porosity. An attempt has been made to study how the shear-thinning viscoelastic affects the theology of blood under the magnetic field and porosity, when porosity through the vessel wall is taken into account with magnetic field in the blood arteries. The solution has been obtained by solving the differential equations occurring during the mathematical modeling by Finite Difference Method.

Keywords: shear thinning viscosity, Rheology, Porosity

1. Introduction

In human physiology, we come across the vessels where tissues are present, and tissues function for secreting nutrients. This creates an effect of porosity on blood flow. In the present paper, we are studying the effects of a magnetic field with porosity.

A large number of researchers have already worked on a similar topic. Here we will list some important researchers' work which is closely related to our work.

Cribber (1965) has developed the model taking into account pressure gradient. Rabbis and Kaferen (1982) have studied a benchmark work related to the topic by developing a mathematical model taking into account the pressure gradients. Lipsch and Moravee (1984) developed a mathematical model for pulsatile blood flow in various branches of arteries in the presence of a magnetic field. Chakra arty (1986) developed two different mathematical models which were majorly based on limitations of oscillatory conditions of blood flows. Rind et al. (1987) had studied the computational study of pulsatile flow of blood taking into account blood as a composition of RBC, WBC, and plasma. Rodkiewicz et al. (1990) compared various mathematical models of blood. They have studied various mathematical models by solving various numerical techniques. Duttal et al. (1992) studied taking into account blood as a particle-fluid suspension. Brookshier and Tarwell (1993) developed a model for wall shear stress of blood vessels. Sharma and Kapur (1995) formulated the blood flow problem by using finite element method. Tarbel (1996) compared two different rheological models. Korenga et al. (1998) studied the bio-chemistry of blood flowing in human arteries. Rachev et al. (1998) showed how blood vessels take adaptation during transverse magnetic field.

Naduvinamani et al. (2002) studied the squeezing lubrication of film by taking anisotropic rectangular plate taking porosity. Farah et al. (2002) studied the two models by taking into account of porosity and showed their results by solving different computational methods. Chein Chao (2003) studied the heated hosted cylinder. Michel et al. (2004) studied the transportation of reactive solutions in porous media. Nicholson and Petropoulous (2006) have made an investigation of branches of arterial flows. Banyal and Devnath (2006) studied the effects of flow of blood

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through arteries by taking into account of periodic body acceleration. Raddaich and Prasada (2007) solved the convective diffusion problem by the finite element method and also compared their results by solving the finite volume method.
Tarchie (2008) studied the effects of porous media of different layers of flows. Nakayama and Kashwon (2008) developed a general heat transfer model. Valochi (2008) studied the effects of porous media on mixing-controlled reactions. Thomas and Houle (2009) developed a mathematical model applicable to a surface sand filter. Michel et al. (2010) developed a two-phase model for reactive solute transport in porous media. Mansourish and Shoki (2011) studied the effects of salt concentration on evaporation from porous media. Khalid and Vafai (2012) studied how patterns in human blood flow. Dabiri and Mar (2013) simulated the behaviour and background of blood flow. Dabiri et al. (2014) studied quantitative measurement of ostraca swimming. Dabiri (2015) studied vortex-enhanced propulsion. Dabiri et al. (2016) predicted the basis of ecological and invasive.

Tenophor and Witlessly (2017) used turbine machines for fish schooling as a basis of vertical axis. Gallegos et al. (2018) studied habitat alteration by species ranging from microbes to jellyfish. Nawroth (2019) developed a mathematical phenomenon for effective fluid interaction. Costello et al. (2020) studied a comparative study of swimming performance. Farewall and Dabiri (2021) studied a Lagrangian approach to identifying vote vortex. Pinch et al. (2022) developed a mathematical model for the behavior of seven co-occurring jellyfish.

2. **Governing Equations for the Blood**

Casson (1959) developed the equation of the from

$$\tau^{1/2} = n_0\gamma^{1/2} + \tau_1^{1/2} \quad \ldots\ldots\ldots\ldots\ldots\ldots \quad (1)$$

$$\gamma = 0, |\tau| < \tau_1 \quad \ldots\ldots\ldots\ldots\ldots\ldots \quad (2)$$

Where τ is the Shear Stress γ represents the stress and n is the casson viscosity.

$$\tau = m\gamma^{n} \quad \ldots\ldots\ldots\ldots\ldots \quad (3)$$

$$\tau + \tau_P = \frac{n_0}{1+(\gamma T)^{v}} \quad \ldots\ldots\ldots\ldots \quad (4)$$

$$\tau = \sum_{P=1}^{n} \tau_P + n\gamma \quad \ldots\ldots\ldots\ldots\ldots \quad (5)$$

3. Mathematical Modeling of the Problem

Here we present a mathematical model by treating blood as a homogeneous incompressible fluid by taking the electrically charged and Porosity inconsideration, and flow takes Place in an Anisotropic mode.

The Equations are

$$\frac{dw}{dt} + u\frac{dw}{dr} + w\frac{dw}{dz} = -\frac{1}{P}\frac{dp}{dz} - \frac{1}{Pr}\frac{d}{dr}(\gamma\tau) - \frac{6}{e}B^2w = \frac{\mu}{ex}w \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots (6)$$

$$\frac{du}{dr} + \frac{dw}{dz} + \frac{\mu}{y} = 0 \quad \ldots\ldots\ldots \quad (7)$$

where u and ware velocity components in the rand z directions, respectively. P is pressure and μ is density, e is the electrical conductivity and B is the magnetic field intensity. The magnetic field is such that induced magnetic field is neglected.
The boundary conditions are

$$\frac{d\omega}{dr} = 0 \quad \mu = 0 \quad \alpha t\gamma = 0$$

$$W=0 \quad u = \frac{dr}{dt} \quad \alpha t\,\gamma = \gamma = R(t,z) \quad \ldots\ldots\ldots\ldots\ldots\ldots(9)$$

Facing the difficulties due to the moving boundary a transformation
z = r/(R(l, z)is introduced. The transformed governing equations become

$$\frac{dw}{dr} + \left(\frac{u}{r} - \frac{\epsilon_1}{R}\right)\frac{dw}{dz} - \frac{w}{r}\left(\frac{du}{ds} + \frac{u}{s}\right) = -\frac{1}{P}\frac{dP}{dZ} - \frac{1}{PR}\frac{d\tau}{ds} - \frac{\tau}{PRs} - \frac{\tau}{PRs} - \frac{6}{P}B^2w - \frac{u}{k}\omega \text{ Or } \frac{dw}{dr} = -\frac{1}{P}\frac{dP}{dZ} - \frac{1}{Pr}\frac{d\tau}{ds} - \frac{d\tau}{ds} - \frac{\tau}{PRs} + \left(\frac{s}{R}\frac{dr}{dt} - \frac{\mu}{y}\right)\frac{d\omega}{ds} + \frac{\omega}{R}\left(\frac{d\mu}{ds} + \frac{\mu}{s}\right) - \frac{6}{P}B^2w - \frac{\mu}{K}\omega \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots(10)$$

$$\mu = \frac{dR}{dP}\frac{dP}{dZ}\left[\varepsilon\omega - \frac{Z}{s}\{\textstyle\int_0^s \varepsilon\omega d\varepsilon - \frac{\int_0^1 s\omega ds}{\int_0^1 s\omega^2 ds}\int_0^s \varepsilon\omega d\varepsilon\} + \frac{1}{s}\frac{dr}{dt}\frac{\int_0^1 s\omega ds}{\int_0^1 s\omega ds}\right] \ldots\ldots\ldots\ldots (11)$$

Where $\frac{dR}{dP}$ represents the elastic response of the artery and experimental values for this are

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available [White(1991)]. The transformed boundary conditions are

$$\frac{d\omega}{ds} = 0 \quad \mu = 0 \quad \alpha t\varepsilon = 0$$

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$$\omega = \mathbf{0} \quad \boldsymbol{\mu} = \frac{dr}{dt} \quad \boldsymbol{at}s = \mathbf{1}$$

………….13

For the Solution of flow problem of a power law fluid equation (3) can be written as

$$r = -\boldsymbol{m}\left(\frac{d\text{m}}{dr}\right)^{n-1}\frac{d\text{m}}{dr}$$

…..……….(14)

Where $\left(-\frac{dw}{dr}\right)$ is the shear rate for one-dimensional tube flow. The equations (9) and (14) are transformed into new Co-ordinate system such that.

$$r = -\frac{m}{r}\left|\frac{d\text{m}}{\frac{ds}{R}}\right|\frac{d\text{m}}{dz}$$

… .... (15)

The equations (4) and (5) are transformed into the new co-ordinate system as given blow.

$$\frac{dr\boldsymbol{p}}{dr} = \frac{r\boldsymbol{p}}{T\boldsymbol{p}} + \frac{s}{R} + \frac{dr}{dt}\frac{dtp}{ds} + \frac{Tp}{(\mathbf{1}+\gamma r\boldsymbol{p})\gamma}$$

……(16)

$$r = \sum_{P=1}^{0} r\boldsymbol{p} + \boldsymbol{np}\gamma$$

…….(17)

The momentum and continuity equations (14) and (11) can be solved subject to the boundary conditions (12) and (13) to complete the flow field of non-Newtonian fluid in an elastic artery having porosity effect In the initial stage we will concentrate on simple oscillatory function containing a single harmonics following Dutta and Tarbell (1996)

$$\frac{dp(t)}{dt} = \boldsymbol{K} + \boldsymbol{mcos}\text{m}\boldsymbol{t}$$

…….(18)

$$R(t) = \overline{\boldsymbol{R}}(\mathbf{1} + \boldsymbol{krcos}(\text{m}\boldsymbol{t} - \boldsymbol{\theta})$$

…….(19)

Where $\bar{R}$ ***and*** $\bar{k}$ the mean parameters m and K are are the amplitude parameters, ∅ is the phase angle and w is the frequency.

$$Q(t)=Q(1+\boldsymbol{K_0 cos}(\text{m}\boldsymbol{t} - \boldsymbol{\theta})$$ ……(20)

$$T\boldsymbol{T}_{\text{m}}(\boldsymbol{t}) = r(\mathbf{1} + \boldsymbol{kcos}(\text{m}\boldsymbol{t} - \emptyset)$$ ……(21)

where Q(1) and (b) are the low rate and wall shear stress respectively. For our simulation, the amplitude of second harmonics are always less than 8% of that of the first harmonics, indicating very small distortion. For physiological flow simulation, we have used multi-harmonic physiological flow waveforms

z= P/$\boldsymbol{Q}$ and z=P/R

Where is the impedance and $\boldsymbol{z_w}$ is the well impedance.

4. Numerical Method

The above occurring equation is solved by using Finite Difference method by taking appropriate grid scaling.

$$\frac{\omega_i^{n+1}-\omega_i^n}{\Delta t} = \frac{1}{{}_nP_0{}^n}\frac{P_i^{n+1}-P_i^n}{\Delta z} - \frac{1}{e_0\ {}^nR_0{}^n}\frac{\tau_{i+1}-\tau_i}{\Delta s} + \left[\frac{s}{R_0{}^n}\frac{R_i^{n+1}-R_0^n}{\Delta t}\right] - \frac{4i^n}{yi^n}\frac{\omega i^n-\omega u^n}{\Delta s} + \frac{\omega i^n}{R_0{}^n}\left(\frac{4_{i+1}^n-4i^n}{\Delta s} + \frac{4i^n}{s^n}\right) - \frac{-6}{e_0{}^n}B^2\omega i^n - \frac{q}{k}\omega i^n$$

Where $\frac{dv}{dt} = \frac{vi^{n+1}-vi^n}{\Delta t}, \frac{dv}{dx} = \frac{vi^{n+1}}{\Delta x} + 0(\Delta x)^2$……………………………(22)

$$\left(\frac{\partial w}{\partial \xi}\right)_i = \frac{w_i^{n+1} - w_i^n}{\Delta \xi} + o(\Delta\xi)^2 = 0. \qquad at \quad \xi = 0$$

$$w = 0, \quad u = \left(\frac{\partial R}{\partial t}\right)_i = \frac{R_i^{n+1} - R_i^n}{\Delta t} \qquad at \quad \xi = 1$$

This approach is applied for temporary discretization to ensure a second-order accurate solution both in time and space for both convection and diffusion terms. The axial velocity is computed solving momentum equation (22) given by suitable constitute equations and the radial velocity is calculated from equation (12) 1 pressure gradient parameters lawl are chosen initially so that flow wave forms produced for Newtonian fluid in the rigid tube would have an amplitude approximately equal to the mean (4,-1) and mean flow rate characteristic of the theoretic aorta under normal values Q-7.7L/min, a 12. m 18 dynelem', and B-2. The wall impedance has been modeled as a purely elastic element with zero phase angle and constant modulus, including taking porous medium (OR). The effect of porosity is

5. Numerical Results and Discussion

After performing the computation by the finite difference method. we have concluded shat by the

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effects of magnetic field velocity of blood flow through Arteries decreases because of scattering of blood constituents and Porosity also act a major role of decrease in velocity and Pressure Arterial flows as we have developed in our model which are shown graphically to understand the result easily. We hope our study will be useful for medical scientists to diagnose various heart diseases.

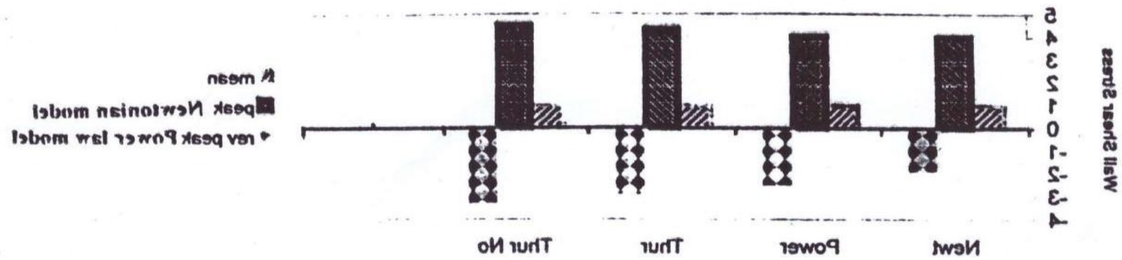

Fig.1: Comparison of wall shear stress for different rheological models in a rigid artery with a fixed pressure gradient wave form in the presence of transverse magnetic field with porosity

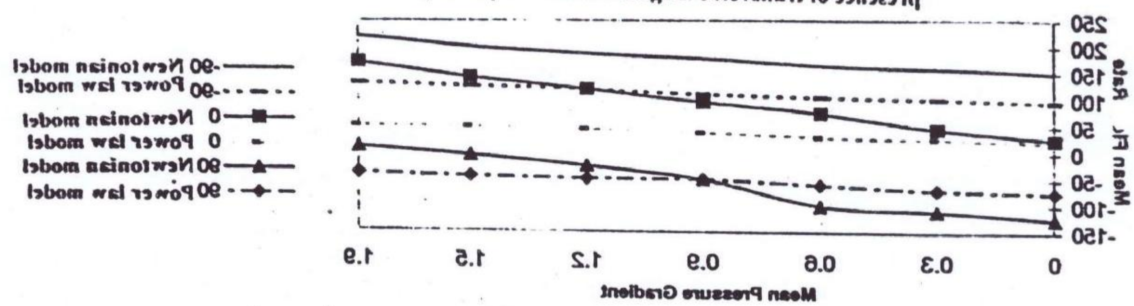

Fig.2: Mean flow rate versus mean pressure gradient in an elastic artery for different parameters, R=1.9 cm. B=2 (-90,0,90) denote the impedance phase angle

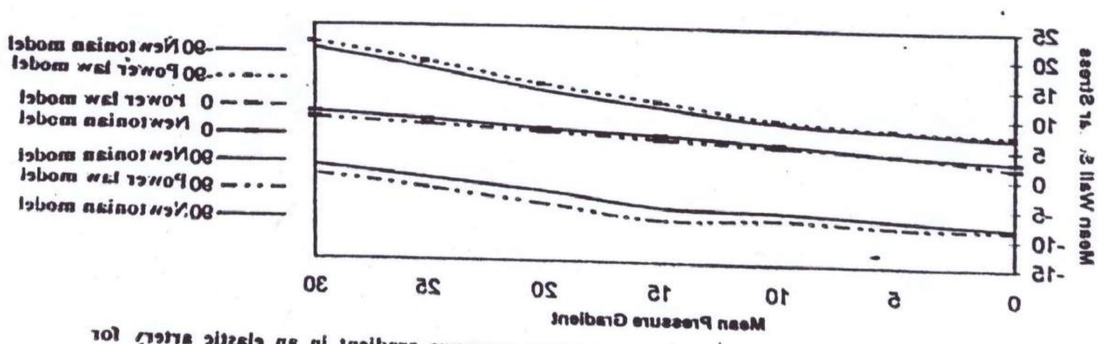

Fig.3: Mean Wall Shear Stress versus mean pressure gradient in an elastic artery for different parameters, (-90,0,90) denotes the impedance angle, k=18dynes/cm², R=1.4. B=2.

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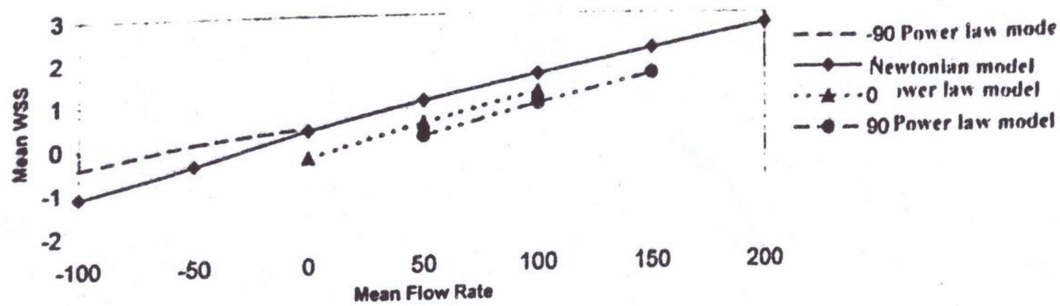

Fig.4: Mean wall shear stress versus mean flow rate for the Power law model for different parameters . (-90,0,90) denotes the impedance angle. k=18dynes/cm^3 , R=1.4, B=2.

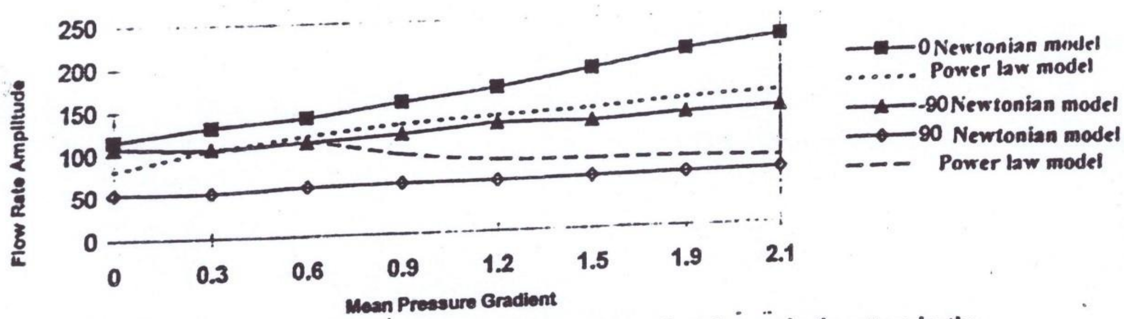

Fig.5: Flow rate amplitude versus mean pressure gradient in an elastic artery in the presence of transverse magnetic field with porosity, k=18dynes/cm^3 , R=1.4, B=2

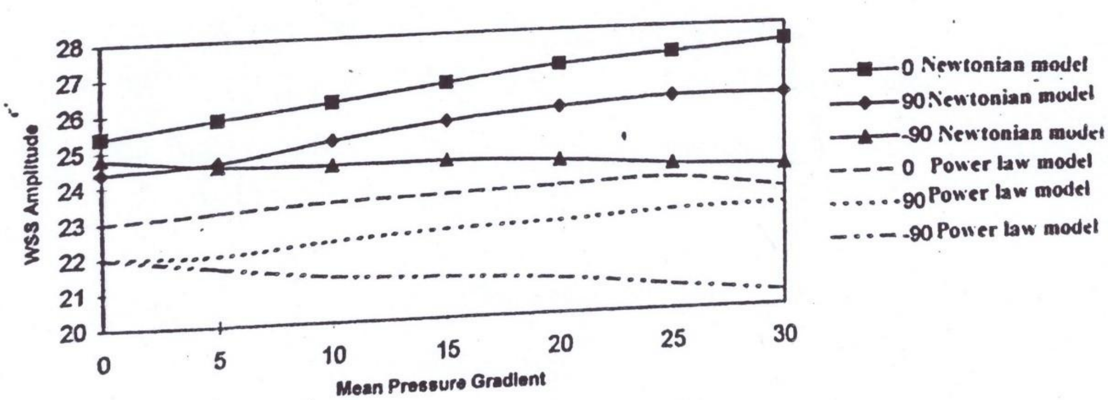

Fig.6: Wall Shear Stress amplitudes versus mean pressure gradient in an elastic artery in the presence of magnetic field with porosity
k= 40.5dynes/cm^3 , R=1.4, B=2

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