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INTERNATIONAL JOURNAL OF ENGINEERING SCIENCES & RESEARCH TECHNOLOGY

Hydromagnetic Peristaltic Flow of Blood with Effect of Porous Medium Through Coaxial Vertical Channel: A Theoretical Study

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Abstract

The present paper investigates the hydromagnetic peristaltic flow of blood with effect of porous medium though coaxial vertical channel: A theoretical study. A perturbation method of solution is obtained in terms of wall slope parameter and closed form of expressions has been derived for axial velocity and pressure rise. The effects of various physical parameters on axial velocity and pressure rise s have been computed numerically. It is noted axial velocity increases with increase in magnetic parameter (M) and Porous parameter (D) for the two cases $\frac{dp}{dx} = -0.5$, $\frac{dp}{dx} = -1$ .The velocity in decreases with increase in Magnetic parameter (M) and Porous parameter (D) for the two cases $\frac{dp}{dx} = 0.5$, $\frac{dp}{dx} = 1$.

Keywords: Peristaltic fluid flow, Reynolds number, Magnetic field, Porous medium and coaxial vertical channel.

Introduction

A peristaltic pump is a device for pumping fluids, generally from a region of lower to higher pressure, by means of a contraction wave traveling along a tube-like structure. This traveling-wave phenomenon is referred to as (peristaltic). This phenomenon is now well known to physiologists to be one of the major mechanisms for fluid transport in many biological systems. The study of the mechanism of peristalsis, in both mechanical and physiological situations, has recently become the object of scientific research. Since the first investigation of Latham [1], several theoretical and experimental attempts have been made to understand peristaltic action in different situations. A review of much of the early literature is presented in an article by Jaffrin and Shapiro [2]. A summary of most of the experimental and theoretical investigations reported, with details of the geometry, fluid, Reynolds number, wavelength parameter, wave amplitude parameter, and wave shape has been given by Srivastava and Srivastava [3].

A porous medium is a medium which contains a number of small holes distributed throughout the matter. Flow through porous media has been of considerable interest in the recent years due to the potential application in all fields of Engineering, Geo-fluid dynamics and Biomechanics. For example study of flow through porous media is immense use to understand transport process in lungs, in kidneys, gallbladder with stones, movement of small blood vessels and tissues cartilage and bones etc. Most of the tissues in the body (e.g. bone, cartilage, muscle) are deformable porous media. The proper functioning of such materials depends crucially on the flow of blood, nutrients and so forth through them. Porous- medium models are used to understand various medical conditions (such as tumor growth) and treatments (such as injections). Interesting investigations on peristaltic flow of Newtonian and non-Newtonian fluids through porous media are described in Refs [26-34].

In the recent past, engineers and scientists became interested in the influence of magnetic field on blood flows with a view to utilizing MHD (Magnetohydrodynamic) in controlling blood flow during surgery and also establishing the effects of magnetic field on blood flows in astronauts, citizens living in the vicinity of electromagnetic towers etc. Since blood consists of a suspension of red blood cells containing hemoglobin, which contains iron oxide, it is quite apparent that blood is electrically conducting and exhibits Magnetohydrodynamic flow characteristics.

Bhargava et al [13] numerically studied the pulsatile flow and mass transfer of an electrically conducting Newtonian biofluids via a channel with porous medium. The flows of blood through arteries in the presence of magnetic field under different physiological conditions were reported in ([14], [15]). Steady laminar flow of blood through a porous medium in an arterial segment having double stenoses under the influence of externally applied magnetic field have been carried out by [16], [17] using numerically as well as analytically by means of Frobenius Method. The potential use of such MHD principles in various arteries have explored by [18], [19], who showed that for unsteady flow of blood in an artery of circular cross-section, a uniform magnetic fields alters the flow rate of blood. A. R. Rao and K. S. Desikachar [20] have investigated using a vorticity formulation of the MHD oscillatory flows in variable cross-sectional channels, reporting a distinct reduction in velocity with a strong applied magnetic field. Many biological tissues such as bones and vascular tissues, the renal system as well as the blood vessels containing fatty deposits are assumed to be porous by nature. A. R. Khaled and K. Vafai [21] have presented a detailed review on heat and fluid flow in a porous media having physiological applications. Pulsatile flow of blood through a stenosed porous medium porous medium under periodic body acceleration has been studied by M. El-Shahed [22]. Magnetohydrodynamic peristaltic flow of a couple stress fluid through coaxial channel containing a porous medium has been studied by Tripathi [23]. Peristaltic transportation of a conducting fluid through a porous medium in an asymmetric vertical channel has been studied by Ramireddy et al [24]. Peristaltic Magnetohydrodynamic (MHD) flows have also received extensive attention in recent years. Many fluids possess an electrically-conducting nature in the presence of a transverse magnetic field. The so-called Magnetohydrodynamic body force effect can therefore be strategically exploited in a diverse range of applications. The magnetic properties of haemotological suspensions have been established for many decades. An important study in this regards has been the article by Roberts [25]. In that study it was established that a Left Ventricular Assist Device (LVAD) utilizing MHD principles was feasible: a high density magnetic field directed electrified blood within the artery close to an extracorporeal electrode along the length of the electrode, effectively generating a fluid pumping force and pressure commensurate with the magnetic field strength and electrode current in accordance with MHD theory and practice. More recent applications of peristaltic magneto-fluid dynamic flows have emerged in the last decade or so. These include magnetohydrodynamic capsule endoscopy, [26-28] cerebrospinal endoscopic magnetic aqueductoplasty, [29] and cardiovascular-assist hydromagnetic medical devices, [30] largely mobilized by the attractive operational characteristics of magnetic peristaltic pumps, namely cleanliness and avoidance of damage to fragile blood cells.

Mathematical formulation and Solution

Consider the unsteady hydromagnetic flow of a viscous, incompressible and electrically conducting couple-stress fluid through a two-dimensional channel of non- uniform thickness with a sinusoidal wave travelling down its wall. The plates of the channel are assumed to be electrically insulated. We choose a rectangular coordinate system for the channel with x along centerline in the direction of wave propagation and y transverse to it.

The geometry of the wall surface is defined as

$$H_1(X,t) = a_0 + b_1 Cos\frac{2\pi}{\lambda}(X-ct) \qquad (1)$$

$$H_2(X,t) = -a_1 - b_2 Cos\frac{2\pi}{\lambda}((X-ct)+\theta \qquad (2)$$

where b_1, b_2 are amplitudes of the waves, $a_0 + a_1$ is the width of the channel , λ is the wave length, θ is the phase differences $(0 \le \theta \le \pi)$, c is the propagation velocity and t is the time.

The constitutive equations and equations of motion for a couple stress fluid are

$$T_{ji,j} + \rho f_i = \rho\frac{\partial v_i}{\partial t} \qquad (3)$$

$$e_{ijk} + T^A_{jk} + M_{ji,j} + \rho C_i = 0 \qquad (4)$$

$$\tau_{ij} = -p^{'} + 2\mu\, d_{ij} \qquad (5)$$

$$\mu_{ij} = 4\eta\,\omega_{j,i} + \eta^{'}\omega_{j,i} \qquad (6)$$

Where f_i is the body force vector per unit mass, C_i is the body moment per unit mass, v_i is the velocity vector, τ_{ij} and T^A_{jk} are the symmetric and antisymmetric parts of the stress tensor T_{jk} respectively, M_{ij} is the couple stress tensor, μ_{ij} is the deviatoric part of M_{ij} , ω_i is the vorticity vector , d_{ij} is the symmetric part of the velocity gradient , η and $\eta^{'}$ are constants associated with the couple stress , $p^{'}$ is the pressure, and the other terms have their usual meaning from tensor analysis

We introduce a wave frame of reference (x, y) moving with velocity *c* in which the motion becomes independent of time when the channel length is an integral multiple of the wavelength and the pressure

ISSN: 2277-9655
Impact Factor: 1.852

difference at the ends of the channel is a constant (Shapiro et al., (1969)). The transformation from the fixed frame of reference (X,Y) to the wave frame of reference (x, y) is given by

$x = X - ct$, $y = Y$, $u = U - c$, $v = V$ and $p(x) = P(X, t)$

where (u, v) and (U, V) are the velocity components, p and P are pressures in the wave and fixed frames of reference, respectively.

The equations governing the flow in wave frame of reference are given by

$$\frac{\partial u}{\partial x} + \frac{\partial v}{\partial y} = 0 \qquad (7)$$

$$\rho\left[\frac{\partial u}{\partial t} + u\frac{\partial u}{\partial x} + v\frac{\partial u}{\partial y}\right] = -\frac{\partial p}{\partial x} + \mu\left[\frac{\partial^2 u}{\partial x^2} + \frac{\partial^2 u}{\partial y^2}\right] - [\sigma B_0{}^2][u+c] - \left[\frac{\mu}{k_1}\right][u+c] \qquad (8)$$

$$\rho\left[\frac{\partial v}{\partial t} + u\frac{\partial v}{\partial x} + v\frac{\partial v}{\partial y}\right] = -\frac{\partial p}{\partial y} + \mu\left[\frac{\partial^2 v}{\partial x^2} + \frac{\partial^2 v}{\partial y^2}\right] - [\sigma B_0{}^2]v - \left[\frac{\mu}{k_1}\right]v \qquad (9)$$

u and v are the velocity components in the corresponding coordinates p is the fluid pressure, ρ is the density of the fluid, μ is the coefficient of the viscosity, k_1 is the permeability of the porous medium, σ electrical conductivity and k is the thermal conductivity. Proceeding with the analysis, we introduce the following dimensionless parameters:

$$x^* = \frac{x}{\lambda} \quad y^* = \frac{y}{a_0} \quad u^* = \frac{u}{c} \quad v^* = \frac{\lambda v}{a_0 c}$$

$$p^* = \frac{a_0{}^2 p}{\lambda \mu c} \qquad t^* = \frac{ct}{\lambda}, d = \frac{a_1}{a_0}$$

$$Re = \frac{\rho c a_0}{\mu} \quad M = \sqrt{\frac{\sigma}{\mu}} B_0 a_0 \quad \delta = \frac{a_0}{\lambda}$$

$$h_1{}^* = \frac{H_1}{a_0} = 1 + \phi_1 \operatorname{Cos} 2\pi x$$

$$, h_2{}^* = \frac{H_2}{a_0} = -d - \phi_2 \operatorname{Cos}(2\pi x + \theta), \phi_1 = \frac{b_1}{a_0},$$

$$\phi_1 = \frac{b_2}{a_0}$$

where δ, ε, ϕ ,Re, M designate the wave number, ratio of half width of channels, amplitude ratio, Reynolds number and Hartmann (Magnetohydrodynamic body force) number respectively. Utilizing the long wavelength and low Reynolds number approximation in Eqs. (7)-(9), we obtain:

$$\frac{\partial u}{\partial x} + \frac{\partial v}{\partial y} = 0 \qquad (10)$$

$$Re\delta[u_t + u\,u_x + vu_y] = \left[-\frac{\partial p}{\partial x} + \delta^2 u_{xx} + u_{yy} - \left(M^2 + \frac{1}{D}\right)u - \left(M^2 + \frac{1}{D}\right)\right] \qquad (11)$$

$$Re\delta^3[v_t + u\,v_x + vv_y] = \left[-\frac{\partial p}{\partial y} + \delta^4 v_{xx} + \delta^2 v_{yy} - M^2\delta^2 v - \frac{1}{D}\delta^2 v\right] \qquad (12)$$

Using long wavelength (i.e.,$\delta << 1$) and negligible inertia (i.e.,$Re \to 0$) approximations, we have

$$\frac{\partial^2 u}{\partial y^2} - \left(M^2 + \frac{1}{D}\right)u - \left(M^2 + \frac{1}{D}\right) = -\frac{\partial p}{\partial x} \qquad (13)$$

$$\frac{\partial p}{\partial y} = 0 \qquad (14)$$

With dimensionless boundary conditions

u = -1 at y = h_1, u = -1 at y = h_2 (15)

$\frac{\partial^2 u}{\partial y^2} = 0$ at y = h_1, $\frac{\partial^2 u}{\partial y^2} = 0$ at y = h_2 (16)

Solving equation (13) using the boundary conditions (15 and 16), we get

$$u = [A-1]N_1 Sinh\,[\alpha_1 y] + [A-1]N_2 Cosh\,[\alpha_1 y] - A \qquad (17)$$

Where $\alpha_1 = \sqrt{M^2 + \frac{1}{D}}$,

$$N_1 = \frac{[Cosh\,[\alpha_1 h_1] - Cosh\,[\alpha_1 h_2]]}{[Sinh\,[\alpha_1 h_1]\,Cosh\,[\alpha_1 h_2] - Sinh\,[\alpha_1 h_2]\,Cosh\,[\alpha_1 h_1]]}$$

$$N_2 = \frac{[Sinh\,[\alpha_1 h_1] - Sinh\,[\alpha_1 h_2]]}{[Sinh\,[\alpha_1 h_1]\,Cosh\,[\alpha_1 h_2] - Sinh\,[\alpha_1 h_2]\,Cosh\,[\alpha_1 h_1]]}$$

$$A = \left[1 + \frac{\frac{dp}{dx}}{B}\right] \quad , B = M^2 + \frac{1}{D}$$

The rate of volume flow 'q' through each section is a constant (independent of both x and t).It is given by

$$q = \int_{h_1}^{h_2} u\,dy = [A-1]N_3 + [A-1]N_4 - A[h_2 - h_1] \qquad (18)$$

Where

$$N_3 = \frac{N_1}{\alpha_1}\left[\left[Cosh\,[\alpha_1 h_2] - Cosh\,[\alpha_1 h_1]\right]\right]$$

$$N_4 = \frac{N_2}{\alpha_1}\left[\left[Sinh\,[\alpha_1 h_2] - Sinh\,[\alpha_1 h_1]\right]\right]$$

Hence the flux at any axial station in the fixed frame is found to be given by

$$Q = \int_{h_1}^{h_2}(u+1)dy = q + (h_2 - h_1) \quad (19)$$

Averaging volume flow rate along one time period, we have

$$\bar{Q} = \int_0^1 Q dt = q + 1 + d \quad (20)$$

The pressure gradient obtained from equation (18) can be expressed as

$$\frac{dp}{dx} = B\left[\frac{\bar{Q} - 1 + \varepsilon + (h_2 - h_1)}{N_3 + N_4 - (h_2 - h_1)}\right] \quad (21)$$

The pressure rise Δ_p (at the wall) in the channel of length L, non-dimensional form is given by

$$\Delta_p = \int_0^1 \left(\frac{dp}{dx}\right)dx = \int_0^1 \left(\frac{B\,(\bar{Q} - 1 + d) + (h_2 - h_1)}{(N_4 - N_3) - (h_2 - h_1)}\right)dx$$

Results and Discussion

In this paper we investigate hydromagnetic peristaltic flow of blood with effect of porous medium though coaxial vertical channel: A theoretical study, we have presented the graphical results of the solutions axial velocity u and pressure rise Δp.

We now discuss the behavior of axial velocity for various in the governing parameters Magnetic parameter (M) and Porous parameter (D)for the phase shipt □□□□□□as depicted in figures (1) to (8).The axial velocity distribution (u) with Magnetic parameter (M)as depicted in figures (1) and (2). We notice that the axial velocity increases with increase in Magnetic parameter (M) for all the two cases $\frac{dp}{dx}$ = - 0.5, $\frac{dp}{dx}$ = -1.It is interesting to note that the opposite behaviour observed for $\frac{dp}{dx}$ = 0.5, $\frac{dp}{dx}$ = 1 as shown in Figures (3) and (4).Figures (5) and (6) reveal the velocity distribution with porous medium (D). It is observed that as the Porous parameter D increases the maximum velocity increased for all the two cases $\frac{dp}{dx}$ = - 0.5, $\frac{dp}{dx}$ = -1. However, opposite effects are noticed for $\frac{dp}{dx}$ = 0.5, $\frac{dp}{dx}$ = 1 as shown in Figures (7) and (8). (Increase in porous parameter D).We now discuss the behavior of axial velocity for various in the governing parameters Magnetic parameter (M) and Porous parameter (D) for phase shipt □□□□□/ 4.(see figures (9) to (16)).The axial velocity distribution (u) with Magnetic parameter (M)for phase shipt □□□□□/ 4as depicted in figures (9) and (10). It is observed the maximum velocity (u) increases with increase in Magnetic parameter (M) for all the two cases $\frac{dp}{dx}$ = - 0.5, $\frac{dp}{dx}$ = -1.It is interesting to note that the opposite behaviour observed for $\frac{dp}{dx}$ = 0.5, $\frac{dp}{dx}$ = 1 as shown in Figures (11) and (12) even if for the phase shipt $\theta = \pi/4$. The maximum velocity increases with increase in porous parameter (D) as depicted in figures (13) and (14) for all the two cases $\frac{dp}{dx}$ = - 0.5, $\frac{dp}{dx}$ = -1. It is observed the opposite behaviour observed for $\frac{dp}{dx}$ = 0.5, $\frac{dp}{dx}$ = 1 as shown in Figures (15) and (16) with increase in porous parameter (D) for the phase shipt $\theta = \pi/4$.

In Figure (17), the effect of Magnetic field on pressure is illustrated. Inspection of this figure indicates that the pressure decreases with increase in magnitude of Magnetic parameter. In Figure (18), the effect porous medium on pressure is illustrated. Inspection of this figure indicates that the pressure decreases with increase in magnitude of porous parameter.

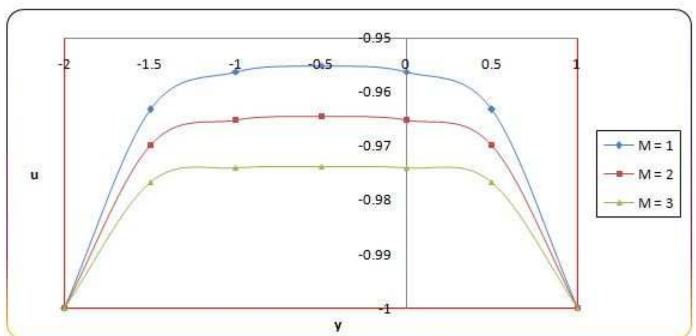


Figure (1): u with M when D = 0.1, $\frac{dp}{dx}$ = -0.5, $\phi_1 = 0.7$, $\phi_2 = 1.2$, x = 0.25, d= 2, $\theta = 0$

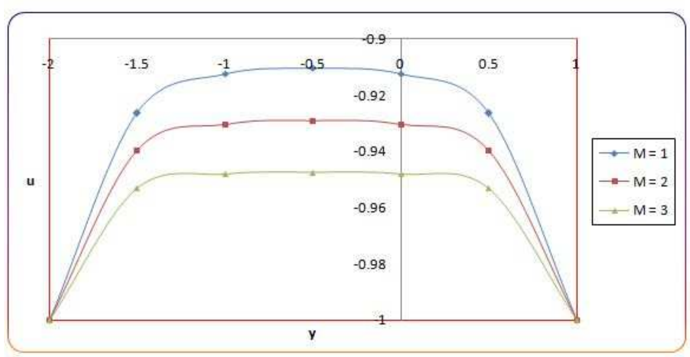


Figure (2): u with M when D = 0.1, $\frac{dp}{dx}$ = -1, $\phi_1 = 0.7$, $\phi_2 = 1.2$, x = 0.25, d= 2, $\theta = 0$

ISSN: 2277-9655
Impact Factor: 1.852

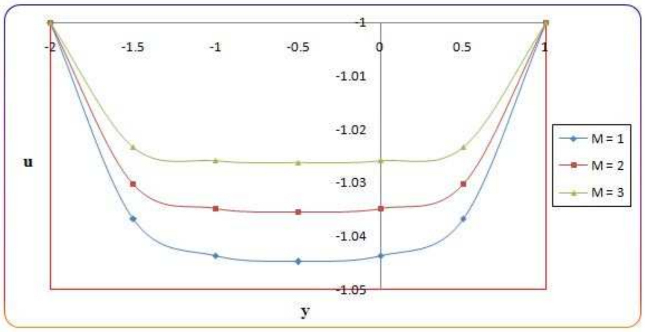


Figure (3): u with M when D = 0.1, $\frac{dp}{dx}$ = 0.5, $\phi_1 = 0.7$, $\phi_2 = 1.2$, x = 0.25, d= 2, $\theta = 0$

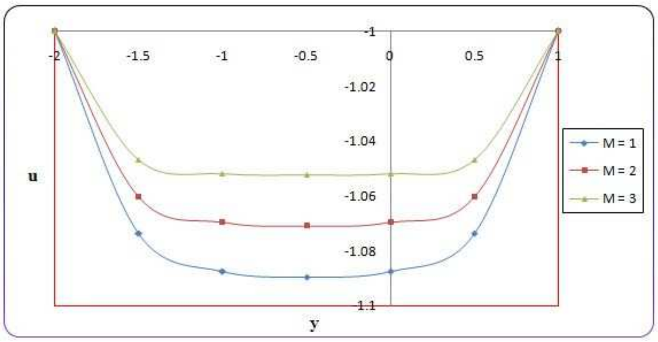


Figure (4): u with M when D = 0.1, $\frac{dp}{dx}$ = 1, $\phi_1 = 0.7$, $\phi_2 = 1.2$, x = 0.25, d= 2, $\theta = 0$

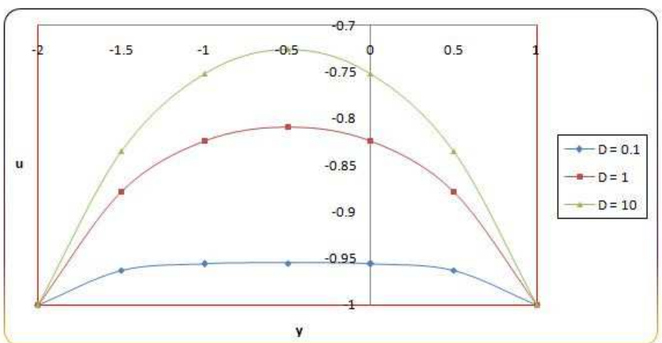


Figure (5): u with D when M = 1, $\frac{dp}{dx}$ = -0.5, $\phi_1 = 0.7$, $\phi_2 = 1.2$, x = 0.25, d= 2, $\theta = 0$

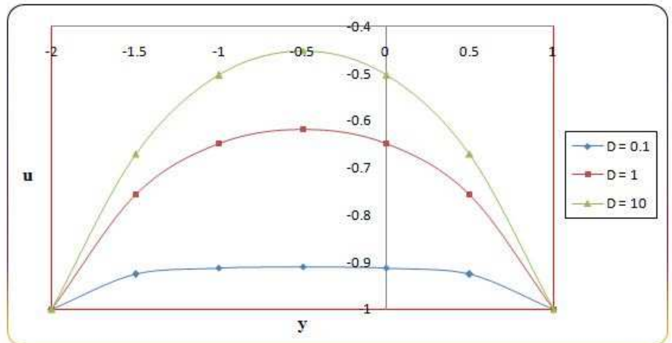


Figure (6): u with D when M = 1, $\frac{dp}{dx}$ = -1, $\phi_1 = 0.7$, $\phi_2 = 1.2$, x = 0.25, d= 2, $\theta = 0$

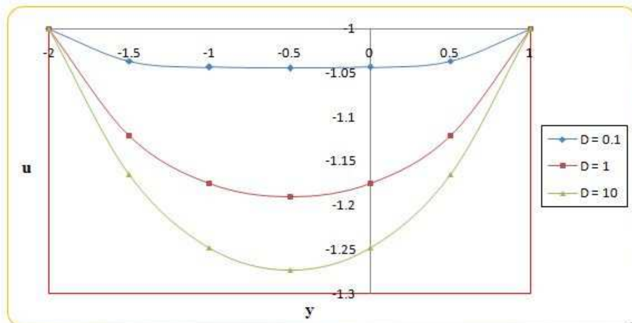
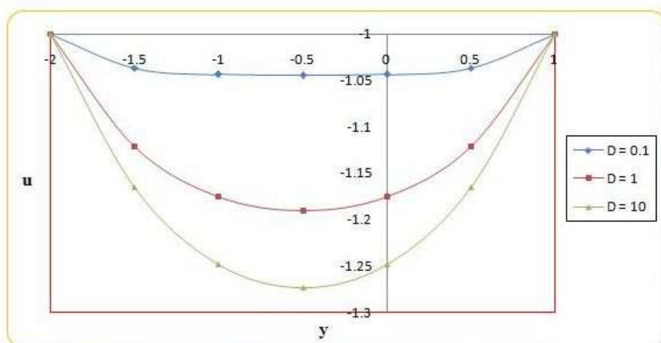


Figure (7): u with D when M = 1, $\frac{dp}{dx}$ = 0.5, $\phi_1 = 0.7$, $\phi_2 = 1.2$, x = 0.25, d= 2, $\theta = 0$



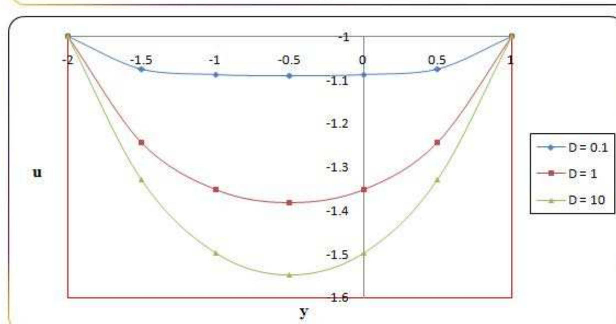


Figure (8): u with D when M = 1, $\frac{dp}{dx}$ = 1, $\phi_1 = 0.7$, $\phi_2 = 1.2$, x = 0.25, d= 2, $\theta = 0$

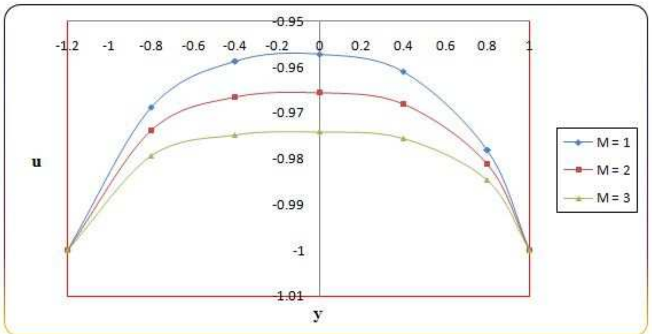


Figure (9): u with M when D = 0.1, $\frac{dp}{dx}$ = -0.5, $\phi_1 = 0.7$, $\phi_2 = 1.2$, x = 0.25, d= 2, $\theta = \frac{\pi}{4}$

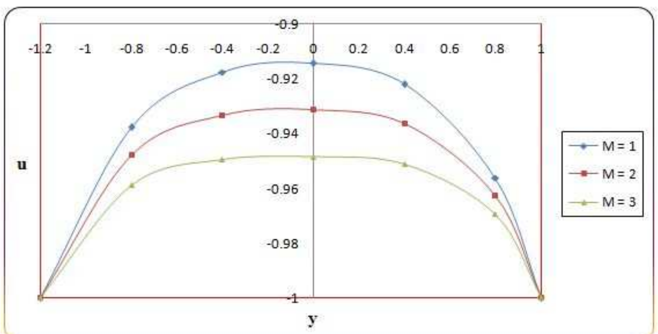


Figure (10): u with M when D = 0.1, $\frac{dp}{dx}$ = -1, $\phi_1 = 0.7$, $\phi_2 = 1.2$, x = 0.25, d= 2, $\theta = \frac{\pi}{4}$

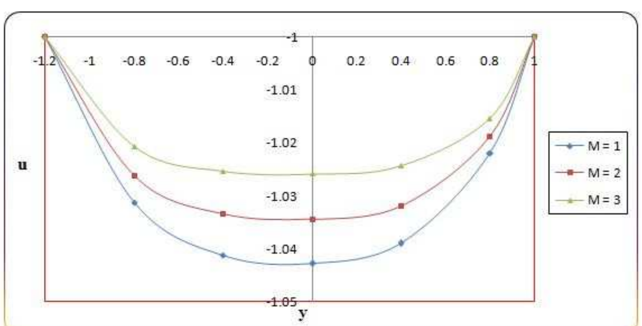


Figure (11): u with M when D = 0.1, $\frac{dp}{dx}$ = 0.5, $\phi_1 = 0.7$, $\phi_2 = 1.2$, x = 0.25, d= 2, $\theta = \frac{\pi}{4}$

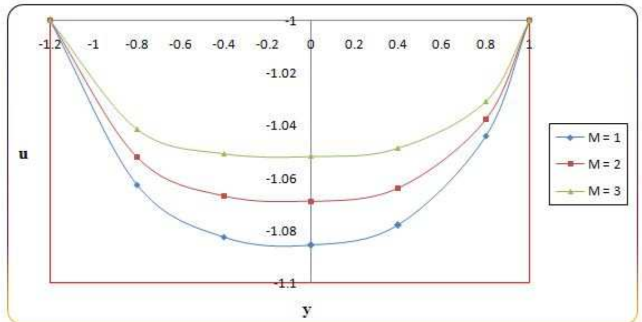


Figure (12): u with M when D = 0.1, $\frac{dp}{dx}$ = 1, $\phi_1 = 0.7$, $\phi_2 = 1.2$, x = 0.25, d= 2, $\theta = \frac{\pi}{4}$

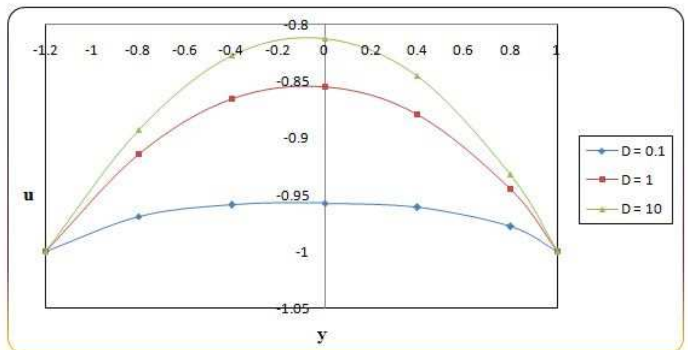


Figure (13): u with D when M = 1, $\frac{dp}{dx}$ = -0.5, $\phi_1 = 0.7$, $\phi_2 = 1.2$, x = 0.25, d= 2, $\theta = \frac{\pi}{4}$

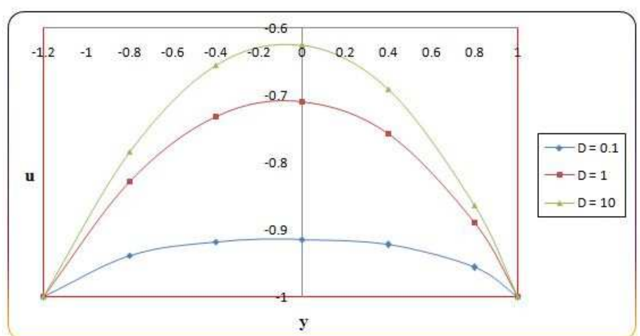


Figure (14): u with D when M = 1, $\frac{dp}{dx}$ = -1, $\phi_1 = 0.7$, $\phi_2 = 1.2$, x = 0.25, d= 2, $\theta = \frac{\pi}{4}$

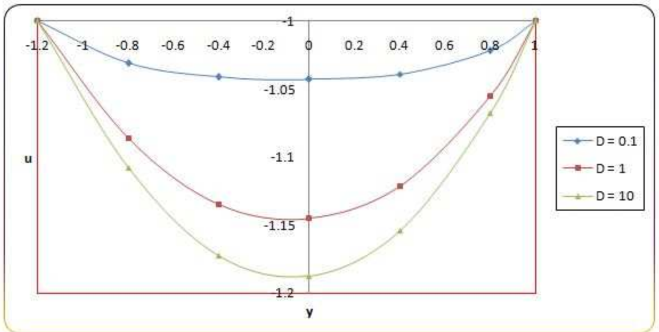


Figure (15): u with D when M = 1, $\frac{dp}{dx}$ = 0.5, $\phi_1 = 0.7, \phi_2 = 1.2$, x = 0.25, d= 2, $\theta = \frac{\pi}{4}$

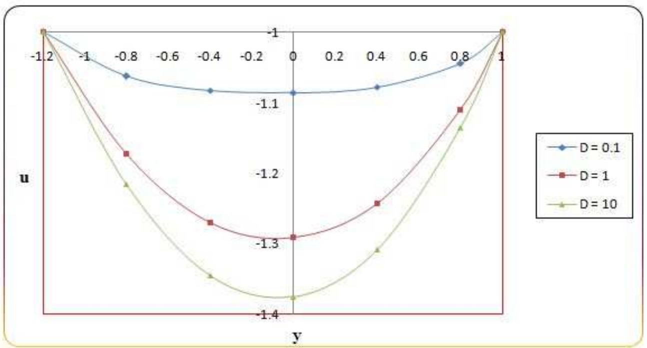


Figure (16): u with D when M = 1, $\frac{dp}{dx}$ = 1, $\phi_1 = 0.7, \phi_2 = 1.2$, x = 0.25, d= 2, $\theta = \frac{\pi}{4}$

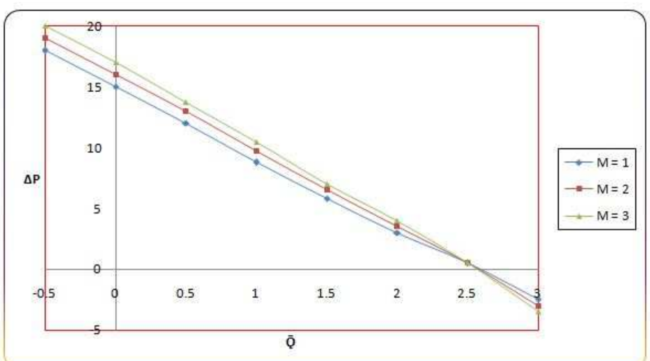


Figure (17). Δ_p With $\bar{Q}$ when D = 0.1, $\phi_1 = 0.7, \phi_2 = 1.2$, x = 0.25, d= 2, $\theta = \frac{\pi}{4}$

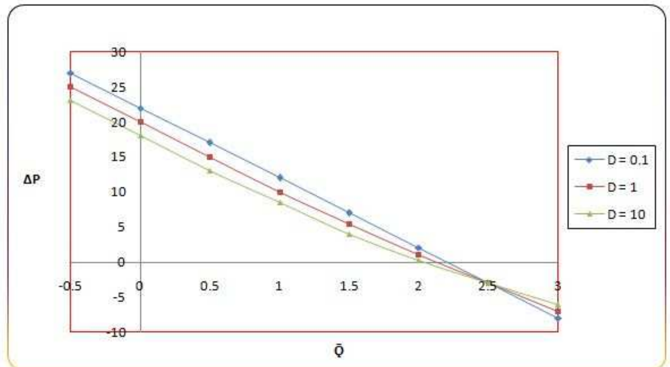


Figure (18). Δ_p With $\bar{Q}$ when D = 0.1, $\phi_1 = 0.7, \phi_2 = 1.2$, x = 0.25, d= 2, $\theta = \frac{\pi}{4}$

Conclusions

In this paper we presented a theoretical approach to study the hydromagnetic peristaltic transportation of couple stress flow of blood in a non-uniform channel congaing porous medium. The governing equations of motion are solved analytically using long wave length approximation. Furthermore, the effect of various values of parameters on axial velocity and pressure rise have been computed numerically and explained graphically. We conclude the following observations:

1. The axial velocity in increases with increase in Magnetic parameter (M) and Porous parameter (D) for the two cases $\frac{dp}{dx}$ = - 0.5, $\frac{dp}{dx}$ = -1(phase shipt □$\theta = 0$ as well as $\theta = \frac{\pi}{4}$).

2. The velocity in decreases with increase in Magnetic parameter (M) and Porous parameter (D) for the two cases $\frac{dp}{dx}$ = 0.5, $\frac{dp}{dx}$ = 1(phase shipt □$\theta = 0$ as well as $\theta = \frac{\pi}{4}$).

3. The pressure decreases with increase in magnitude of Magnetic parameter and porous medium.

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ISSN: 2277-9655
Impact Factor: 1.852

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