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Original Article

# Flow in Catheterized Tube Induced by Peristaltic Waves: A Theoretical Study

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## Abstract

The study examines peristaltically driven flow in a catheterized tube based on a two-layer fluid model comprising a particle-laden core and a particle-free peripheral layer. The propagation of finite-amplitude sinusoidal waves along the tube wall induces fluid transport, which closely represents the physiological characteristics of blood flow. The problem is formulated under the assumptions of axisymmetric flow and the long-wavelength approximation, with the catheter forming an inner boundary that creates an annular flow region. Sinusoidal peristaltic waves of finite amplitude propagate along the tube wall, driving fluid transport, which is closely related to blood movement in arteries and peristaltic transport in the gastrointestinal system. Closed-form expressions for the pressure drop, volumetric flow rate, and frictional forces at both the tube wall and catheter surface are obtained. The pressure drop increases with both the flow rate and the particle concentration within the core region for any given set of parameters. Frictional forces at the tube and catheter surfaces vary in a manner consistent with the pressure drop across all governing parameters. Nevertheless, the magnitude of friction at the catheter surface remains significantly lower than that at the tube wall.

**Categories:** Biotechnology and Engineering, Thermal and Fluid Systems, Advanced Manufacturing Technologies

**Keywords:** pressure drop, catheter, particle concentration, peripheral layer, friction force

## Introduction

Peristalsis is responsible for transport in many biological organs including the vasomotion of small blood vessels such as arterioles, venules, and capillaries, and is a well-known phenomenon to physiologists and engineers [1]. Since the pioneering work of Latham [2], it has remained a subject of extensive scientific and engineering investigation for more than four and a half decades. Many body passages contain smooth muscles that contract sequentially, producing waves of contraction along the passage wall. These progressive waves propel the contents forward in the direction of wave propagation. Peristalsis is therefore a mechanism of fluid transport in which a traveling wave of luminal area contraction or expansion moves along a distensible tube containing a liquid or mixture. Several bio-mechanical devices, such as heart-lung machines and finger and roller pumps, have been developed based on this principle, and certain aquatic organisms employ peristalsis as a means of locomotion. The fundamental principles of peristaltic pumping in circular tubes and in two-dimensional geometries were described by Shapiro et al. [3] and Jaffrin et al. [4], who clearly established the significance of the governing flow parameters. Comprehensive reviews of the major theoretical and experimental investigations up to 1995 may be found in research studies [1-5], and subsequent and more recent contributions include the works of [6-11] among others. Srivastava et al. [6] examined the influence of an imposed Poiseuille flow on the peristaltic transport of a particle-fluid mixture in a two-dimensional channel whose walls undergo sinusoidal oscillations of small amplitude. The driving mechanism of the muscle was modeled by treating the channel walls as elastic or viscoelastic. Mekheimer et al. [7] investigated the peristaltic motion of a particle-fluid suspension in a planar channel. The theory of particulate suspensions has played an important role in the understanding of many diverse physical phenomena, including powder technology, fluidization, sedimentation, combustion, aerosol filtration, atmospheric fallout, lunar ash flow, and environmental pollution. More recently, increasing interest has arisen in applying the theory of particle-fluid suspensions to physiological transport processes, particularly in relation to vasomotion in small vessels such as arterioles, venules, and capillaries. Peristaltic pumping of particle-fluid mixtures has been examined by [6-13] among others. The flow induced by the sinusoidal peristaltic motion of the tube wall for a non-Newtonian fluid obeying the Herschel-Bulkley model [10] and [11] a particle-fluid suspension in a catheterized circular tube has been investigated under the assumptions of long wavelength and low Reynolds number. Through the details of an experimental work, Hung et al. [12] studied the mechanism of solid-particle transport by peristalsis in a two-dimensional channel. Takabatake et al. [13] studied the problem of peristaltic flow in a two-dimensional channel. In view of the significance of the peripheral layer [9], two-layered peristaltic transport problems have been addressed by [14-17] and others. However, when the viscosity of one layer is varied relative to the other, the interface shape is only marginally affected, and the radii ratio remains approximately constant [9]. Srivastava et al. [15] studied a two-layered analysis in non-uniform channels and tubes, and applied their model to compare theoretical predictions with the experimentally observed flow rates of spermatic fluid (semen) in the vas deferens of rhesus monkeys, as well as with other available experimental results. It is important to note that the interface shape generally depends on the viscosity ratio of the two layers and does not maintain a constant ratio of inner to outer radii if mass in each layer is to be conserved separately [16]. Misra et al. [9] extended the work of Rao et al. [17] by generalizing the analysis from a Newtonian fluid to a non-Newtonian Casson fluid. They applied a two-layered Casson fluid model to investigate the peristaltic pumping of blood in small vessels, giving due consideration to the separate conservation of mass in the peripheral and central layers.

Catheterization plays a crucial role in many biomedical procedures and has become a standard diagnostic and therapeutic tool in cardiovascular medicine [18-21]. The mathematical analysis [21] examines blood flow through a model of a composite stenosis, catheterized artery with a permeable wall, in order to investigate the associated flow characteristics. The mathematical formulation relevant to catheterized flow corresponds to transport in the annular region between two concentric tubes. The analogous bio-mechanical problem of peristaltic transport in the presence of an inserted catheter, particularly with application to ureteral flow, was first studied by Roos et al. [22]. Several researchers including [8,23] and more recently [24] have examined the influence of an endoscope or similar device on the peristaltic transport of chyme in the gastrointestinal tract. Hayat et al. [23] investigated the peristaltic flow of a Jeffrey fluid through the annular space of two concentric circular cylindrical tubes with particular reference to the endoscope effects. The effects of an inserted endoscope and a Carreau fluid on peristaltic flow were studied under zero Reynolds number and infinitely long wavelength approximations by Hakeem et al. [8]. In most physiological situations, the wavelength of the peristaltic wave is long compared with the tube radius. Under this long-wavelength approximation, Shapiro et al. [3] examined inertia-free peristaltic transport with infinitely long waves for a Newtonian fluid. Jaffrin and Shapiro [4] further demonstrated that flow may be treated as inertia-free even when the Reynolds number and wave number are small but finite. Under this commonly employed approximation, studies [1,5] have examined two-layered peristaltic pumping of a Casson fluid, with particular reference to blood. A large number of investigators have used the inertia-free flow approximation to study peristaltic transport under various flow conditions among others.

The present investigation addresses the peristaltic pumping of a two-layer (two-fluid) particulate suspension in a catheterized circular tube. The fluid system consists of a core region comprising a particle-fluid mixture and a peripheral layer of particle-free Newtonian fluid with constant viscosity, identical to that of the suspending medium in the core region. The analysis is performed under the long-wavelength approximation, which is appropriate for many physiological flow conditions. In light of the theoretical framework adopted by Srivastava [25], the results of this study are expected to provide meaningful insights into the flow behavior of blood in narrow arteries during catheterization.

## Materials And Methods

### Formulation of the problem

We consider an axisymmetric flow through a catheterized tube of radius $a$, modeled as a circular cylindrical annulus consisting of two fluid layers. The inner core region, of radius, $a_1$ contains a particle-laden Newtonian fluid, while the outer peripheral layer, of thickness $(a - a_1)$ consists of a particle-free Newtonian fluid. The flow within the tube is induced by peristaltic waves propagating along its flexible walls. The tube wall is assumed to be sufficiently elastic to support sinusoidal peristaltic waves of finite amplitude traveling along its surface. Figure *1* depicts the geometric features of the wall surface.

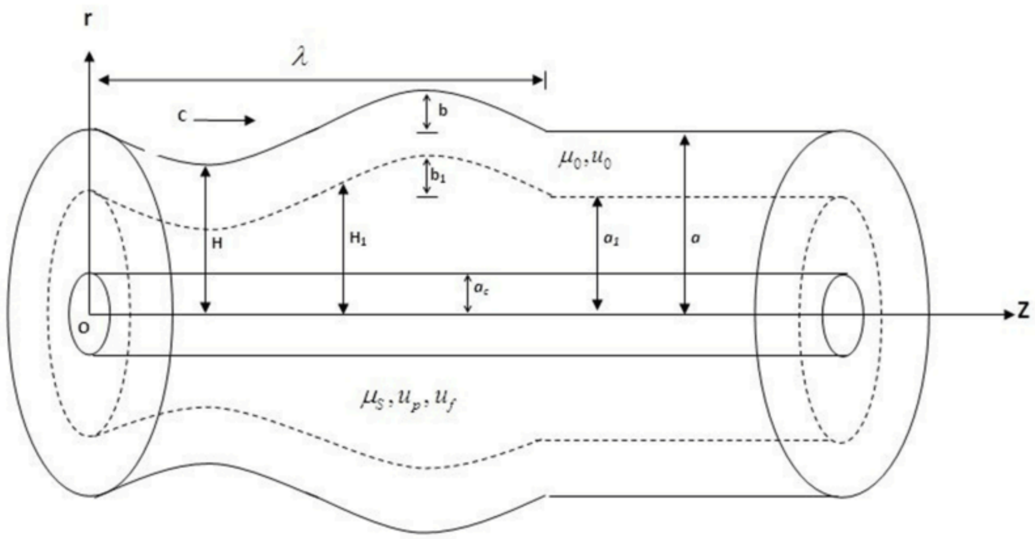


**FIGURE 1: Flow geometry of peristaltic pumping in a catheterized tube**

The surface of tube is defined [1] as:

$$H(z,t) = a + b \sin \frac{2\pi}{\lambda}(z - ct) \tag{1}$$

where $b$ is the amplitude of the wave, $\lambda$ is the wavelength, $c$ is wave propagation speed, $t$ is time and $z$ is the axial variable on z-axis.
The governing equations for two-layered flow have been discussed in detail by [9,26], and [27], including the dimensional and non-dimensional forms of the boundary conditions. Additional studies addressing flow behavior in the presence of a catheter are provided in [9,11,21]. For a comprehensive description of the notation and parameters used, the reader is referred to [10,11,15,25-27].

Accordingly, the mathematical model governing the present problem is formulated as follows:

$$(1-C)\frac{dp}{dz} = (1-C)\frac{\mu}{r}\frac{\partial}{\partial r}(r\frac{\partial}{\partial r})u_f + CS(u_p - u_f), \qquad \epsilon \le r \le h_1, \tag{2}$$

$$C\frac{dp}{dz}=CS(u_f-u_p),\epsilon\leq r\leq h_1, \tag{3}$$

$$\frac{dp}{dz}=\frac{1}{r}\frac{\partial}{\partial r}(r\frac{\partial}{\partial r})u_o,h_1\leq r\leq h, \tag{4}$$

$$u_o=-1,\ \ at\ \ r=h=1+\phi\sin 2\pi z, \tag{5}$$

$$u_f=u_o\ \ and\ \ \tau_f=\tau_o,\ \ at\ \ r=h_1=\alpha+\phi_1\sin 2\pi z, \tag{6}$$

$$u_f=0,\ \ at\ \ r=\epsilon \tag{7}$$

with $(h,h_1)=(1,\alpha)+(\phi,\phi_1)\sin 2\pi z$; $\tau_f=(1-C)\mu_s\partial u_f/\partial r$ and $\tau_o=\mu_o\partial u_o/\partial r;\tau_f,\tau_o$ are shearing stress of the core and peripheral regions, respectively.

**Analysis**

Solving equations (2)-(4) subject to boundary conditions of equations (5)-(7), the expression for the velocity profiles, $u_o,u_f,u_p$ attained as

$$u_o=-1-\frac{1}{4}\frac{dp}{dz}\left\{h^2-r^2+\left(\frac{\epsilon^2-h_1^2-\mu(1-C)(h^2-h_1^2)}{\mu(1-C)log\frac{h_1}{h}-\log\frac{h_1}{\epsilon}}\right)log\frac{r}{h}\right\}, \tag{8}$$

$$u_f=-1-\frac{1}{4\mu(1-C)}\frac{dp}{dz}\left\{\epsilon^2-r^2+\left(\frac{\epsilon^2-h_1^2-\mu(1-C)(h^2-h_1^2)}{\mu(1-C)log\frac{h_1}{h}-\log\frac{h_1}{\epsilon}}\right)log\frac{r}{\epsilon}\right\}, \tag{9}$$

$$u_p=-1-\frac{1}{4\mu(1-C)}\frac{dp}{dz}\left\{\epsilon^2-r^2+\left(\frac{\epsilon^2-h_1^2-\mu(1-C)(h^2-h_1^2)}{\mu(1-C)log\frac{h_1}{h}-\log\frac{h_1}{\epsilon}}\right)log\frac{r}{\epsilon}+\frac{4\mu(1-C)}{S}\right\} \tag{10}$$

The volume flow rate $q=q'/\pi a^2c$, $q'$ is the flow rate in the moving system which is same as in laboratory frame of reference, in non-dimensional form is obtained as:

$$q=2\left[\int_{h_1}^{h}ru_odr+\int_{\epsilon}^{h_1}r\{(1-C)u_f+Cu_p\}\right]dr$$

$$q=\epsilon^2-h^2-\frac{1}{8\mu(1-C)}\frac{dp}{dz}\Bigg[\mu(1-C)\{h^4+h_1^4-2h_1^2h^2\}-\epsilon^4-h_1^4+2h_1^2\epsilon^2+\beta(h_1^2-\epsilon^2)\Bigg)$$

$$+\Bigg(\frac{\epsilon^2-h_1^2-\mu(1-C)(h^2-h_1^2)}{\mu(1-C)log\frac{h_1}{h}-\log\frac{h_1}{\epsilon}}\Bigg\{\mu(1-C)\Bigg(h_1^2-h^2-2h_1^2\log\frac{h_1}{h}\Bigg)+\epsilon^2-h_1^2+2h_1^2\log\frac{h_1}{\epsilon}\Bigg\}\Bigg], \quad (11)$$

$$-\frac{dp}{dz}=8\mu(1-C)\frac{q+h^2-\epsilon^2}{\Theta h^4+\eta h^2+\gamma} \quad (12)$$

where $\Theta=\mu(1-C)(1+\alpha^4-2\alpha^2)-\alpha^4$,
$\eta=(\beta+2\epsilon^2)\alpha^2+\Omega\Bigg[\mu(1-C)(1-\alpha^2)+\alpha^2+2\Bigg(\mu(1-C)\log\alpha-\log\frac{\alpha h}{\epsilon}\Bigg)\alpha^2\Bigg]$,
$\gamma=-(\Omega+\beta+\epsilon^2)\epsilon^2$,
$\Omega=\frac{[\mu(1-C)(1-\alpha^2)+\alpha^2]h^2-\epsilon^2}{\mu(1-C)\log\alpha-\log\frac{\alpha h}{\epsilon}}$.
The mean volumetric flow rate, over a period of the wave [3], as:

$$Q=1+q+\frac{\phi^2}{2}-\epsilon^2$$

Across one wavelength the pressure drop, $\Delta p$ is obtained as:

$$\Delta p=\int_0^1\Bigg(-\frac{dp}{dz}\Bigg)dz$$

$$\Delta p=2\mu(1-C)\Bigg\{\Bigg(Q-1-\frac{\phi^2}{2}\Bigg)I_1+I_2\Bigg\} \quad (13)$$

where $I_1=4\int_0^1\frac{dz}{\Theta h^4+\eta h^2+\gamma}$ and $I_2=4\int_0^1\frac{dz}{\Theta h^2+\eta+\gamma/h^2}$.
The relation between the pressure drop, $\Delta p$ and the flow rate, $Q$ is obtained as:

$$Q=1+\frac{\phi^2}{2}-\frac{I_2}{I_1}+\frac{\Delta p}{2\mu(1-C)I_1} \quad (14)$$

The dimensionless friction force at the tube wall is defined as, $F_a=F'_a/\pi\lambda c\mu_0$; where $F'_a$, represents the dimensional frictional force acting at the upper (tube) surface, measured in either the laboratory or the wave frame of reference. Similarly, the dimensionless friction force at the catheter surface is given by, $F_c=F'_c/\pi\lambda c\mu_0$, where $F'_c$, denotes the dimensional frictional force at the catheter surface, also measured in either frame of reference. The expressions for $F_a$ and $F_c$ are thus obtained as follows:

$$F_a=\int_0^1 h^2\Bigg(-\frac{dp}{dz}\Bigg)dz$$

$$F_a = 2\mu(1-C)\left[I_3 - \frac{I_2^2}{I_1} + \frac{I_2}{2\mu(1-C)I_1}\Delta p\right], \tag{15}$$

where $I_3 = 4\int_0^1 \frac{dz}{\Theta + \eta/h^2 + \gamma/h^4}$.

$$F_c = \int_0^1 \epsilon^2\left(-\frac{dp}{dz}\right)dz = \epsilon^2\Delta p \tag{16}$$

The pressure rise, $-\Delta p$, for zero flow rate, $Q$, and the flow rate, $Q$, for zero pressure rise, $-\Delta p$, are obtained as:

$$(-\Delta p)_{Q=0} = 2\mu(1-C)\left[\left(1+\frac{\phi^2}{2}\right)I_1 - I_2\right], \tag{17}$$

$$(-Q)_{\Delta p=0} = 1 + \frac{\phi^2}{2} - \frac{I_2}{I_1} \tag{18}$$

It is worth noting that, in the limiting case $\epsilon \to 0$, i.e. the absence of the catheter, the results of the present investigation reduce to those obtained for a two-layered particulate suspension flow. When the core mixture behaves as a Newtonian fluid of constant viscosity, $\mu_1 (= \mu_s \neq \mu_0)$, the present analysis yields results identical to those reported by Medhavi et al. [26]. Similarly, the results of Shukla et al. [14] are recovered from the present formulation under the limits $\epsilon \to 0$ and $\mu_1 = \mu_s$. Furthermore, when $\alpha = 1$, the findings of the current study coincide with those of Medhavi [11]. Interestingly, in the absence of the particulate phase in the core region, the core fluid becomes identical to the peripheral fluid, causing the interface between the two layers to vanish. In this case, the model simplifies to that of a single-layer Newtonian fluid in a catheterized tube, which, in the additional limit $\epsilon \to 0$, reproduces the classical results of Shapiro et al. [3].

## Results And Discussion

From equations (13)-(16), the quantitative results of various parameters at a temperature of 37°C and within a tube of radius 0.01 $cm$ are graphically represented in Figures *2*-*12*. The particle concentration, which determines the suspension viscosity, influences the thickness of the peripheral layer, as reported in previous studies by Srivastava [24]. Considering a particle diameter of 2$a_0$= 8 $\mu m$, the corresponding peripheral layer thicknesses $\in (\mu m) \approx \in (C)$ = 6.18, 4.67, 3.60, 3.12, 2.58, and 2.18 for particle concentrations C= 0.1, 0.2, 0.3, 0.4, 0.5, and 0.6, respectively, Haynes [28], and $\epsilon$ (non-dimensional catheter radius) = 0, 0.1, 0.2, 0.3, 0.4, 0.5, 0.6, in the present analysis. Present study corresponds to the flow of a two-layered particulate suspension; to the flow of a single-layered two-phase fluid in uncatheterized tube [11]; to the flow of a single-layered Newtonian viscous fluid in a catheterized tube and to the flow of a single-layered Newtonian viscous fluid in an uncatheterized tube [3] for the parameter values, $\epsilon \to 0, \alpha = 1$, C= 0, respectively.

The pressure drop, $\Delta p$, increases with the flow rate, $Q$, for any given particle concentration, $C$, and amplitude ratio, $\phi$, and the pressure-flow rate relationship exhibits linear behavior for the selected set of parameters in both catheterized and uncatheterized tubes, as shown in Figure *2*. For a given catheter size, $\epsilon$, the flow characteristic, $\Delta p$, also increases with the particle concentration, $C$, within the core region, Figure *3*. The pressure drop, $\Delta p$, increases with the particle concentration, $C$, for any given catheter size, $\epsilon$, and amplitude ratio, $\phi$, as illustrated in Figure *4*, for any non-zero flow rate, $Q$, the pressure drop, $\Delta p$, increases with particle concentration, $C$, for given values of $\epsilon$ and $\phi$. However, for zero flow rate, $Q$, the pressure drop decreases with increasing particle concentration, $C$.

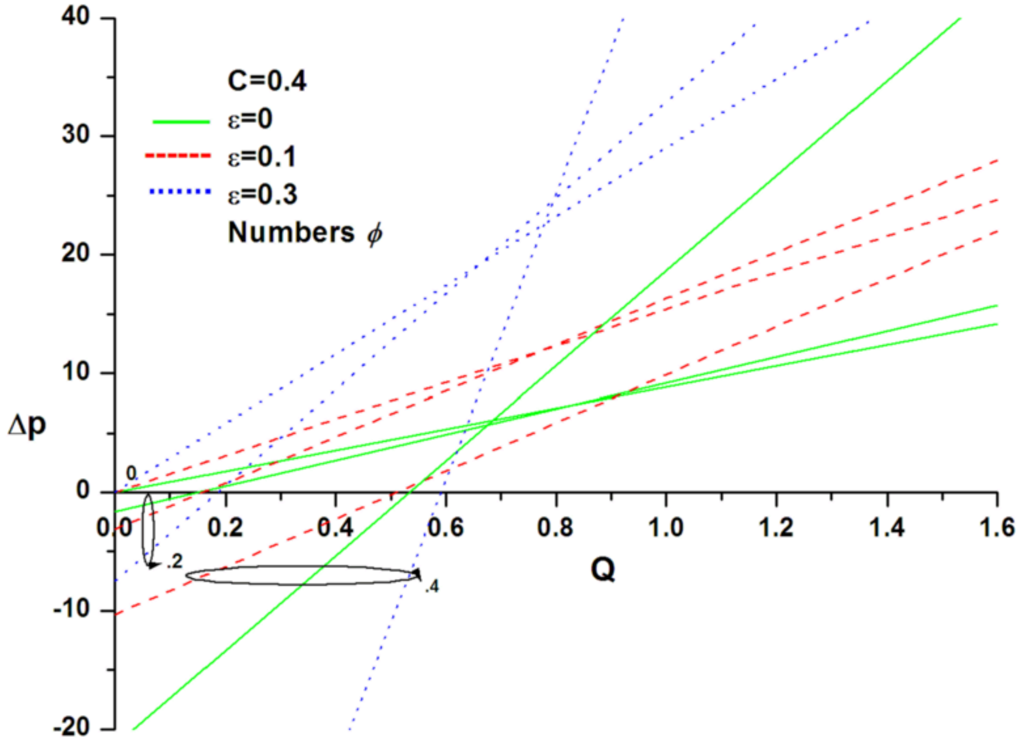


**FIGURE 2: The pressure-flow rate relationship for different ε and φ**

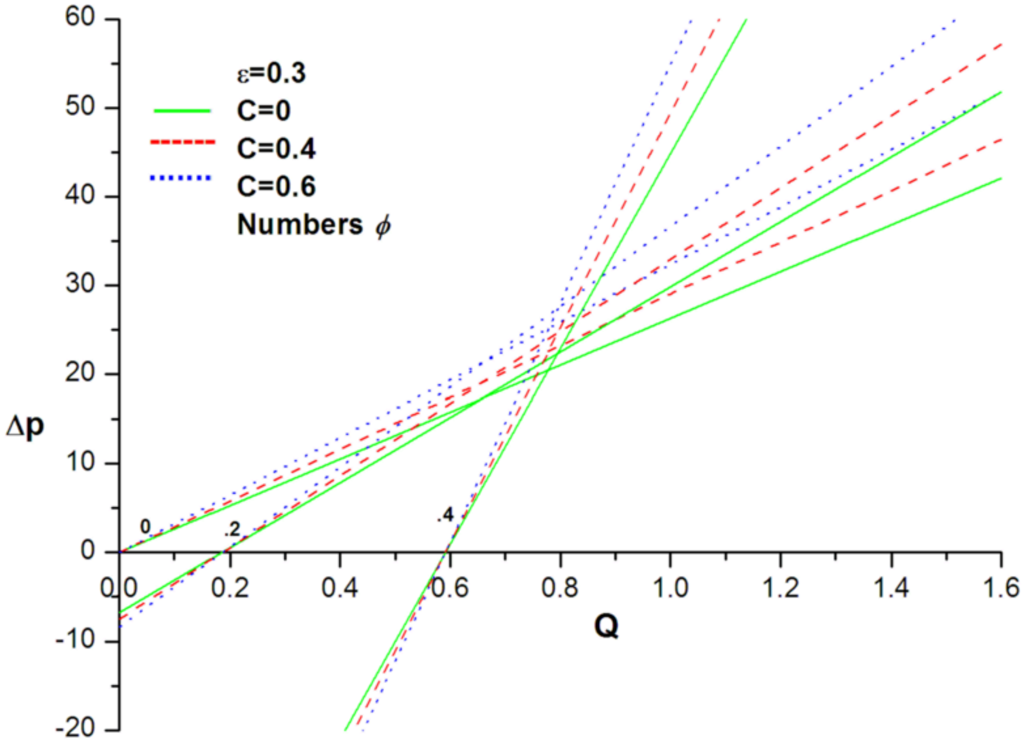


**FIGURE 3: The pressure-flow rate relationship for different C and φ**

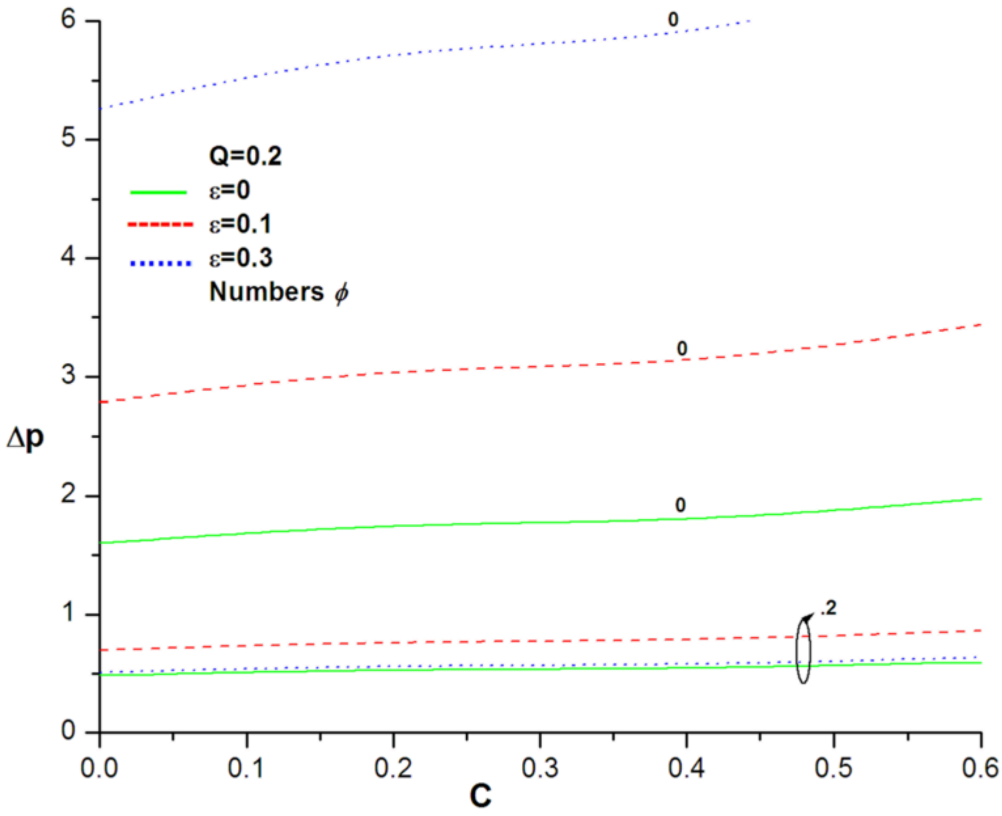


**FIGURE 4: The pressure drop, Δp versus particle concentration, C for different ε and φ**

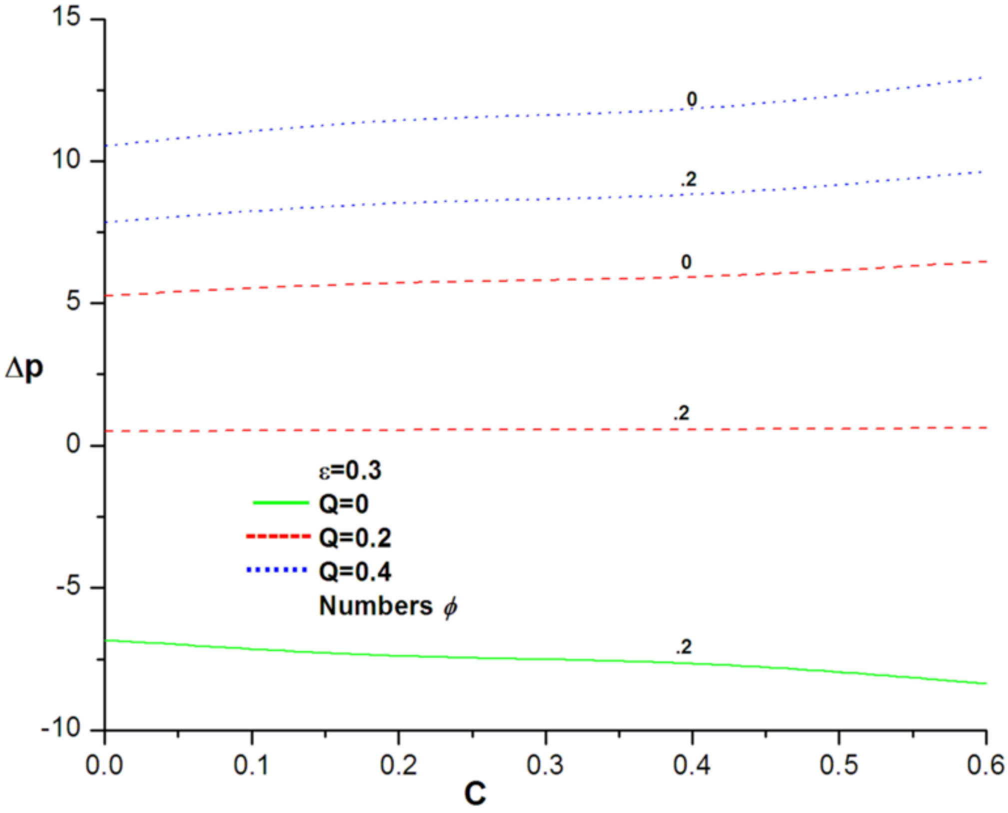


**FIGURE 5: The pressure drop, Δp versus particle concentration, C for different Q and φ**

Furthermore, for any non-zero flow rate, $Q$, the pressure drop, $\Delta p$, continues to rise with increasing particle concentration, $C$, under the same catheter size, $\epsilon$, and amplitude ratio, $\phi$, as shown in Figure 5. The pressure drop, $\Delta p$, decreases indefinitely with increasing catheter size, $\epsilon$, and assumes very low asymptotic magnitude depending on the value of the flow rate, $Q$, for a given set of parameters. It is noticed that the pressure drop, $\Delta p$, decreases indefinitely with increasing catheter size, $\epsilon$, Figure 6. It is also observed that, for a given particle concentration, $C$, the pressure drop, $\Delta p$, decreases continuously with increasing amplitude ratio, $\phi$, for any specified set of parameters, Figure 7. The pressure drop, $\Delta p$, decreases indefinitely for a given value of flow rate, Q with increasing amplitude ratio, $\phi$, and assumes an asymptotic low magnitude depending on the value of the catheter size,$\epsilon$, Figure 8.

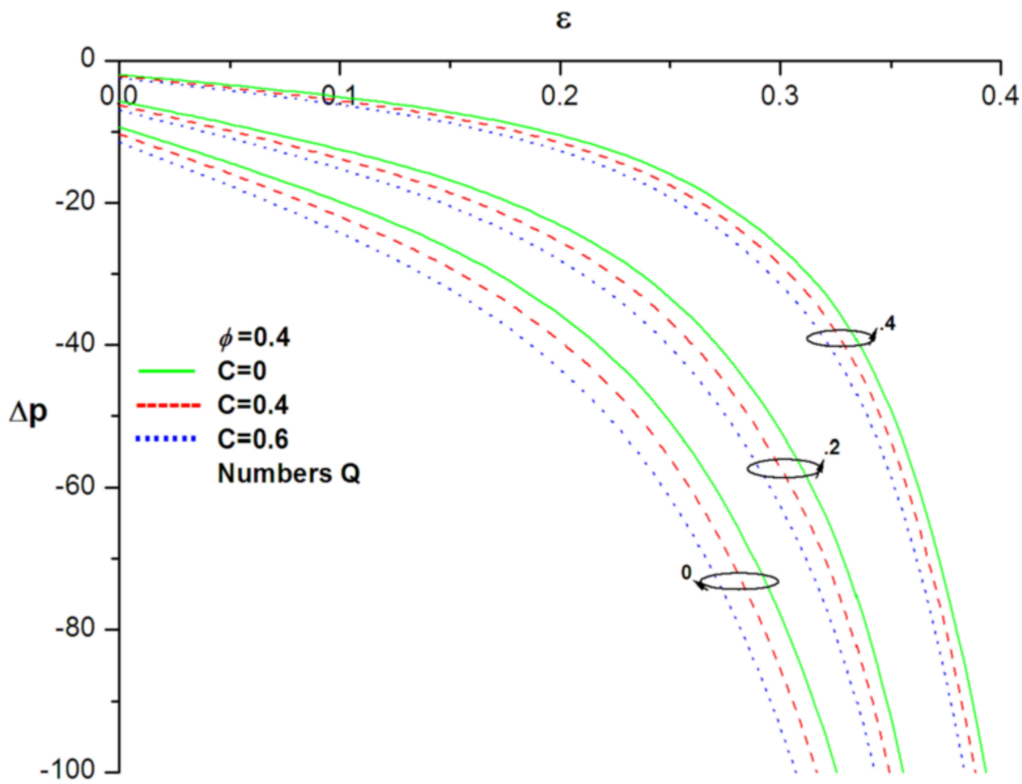


**FIGURE 6: The pressure drop, Δp versus catheter size, ε for different Q and C**

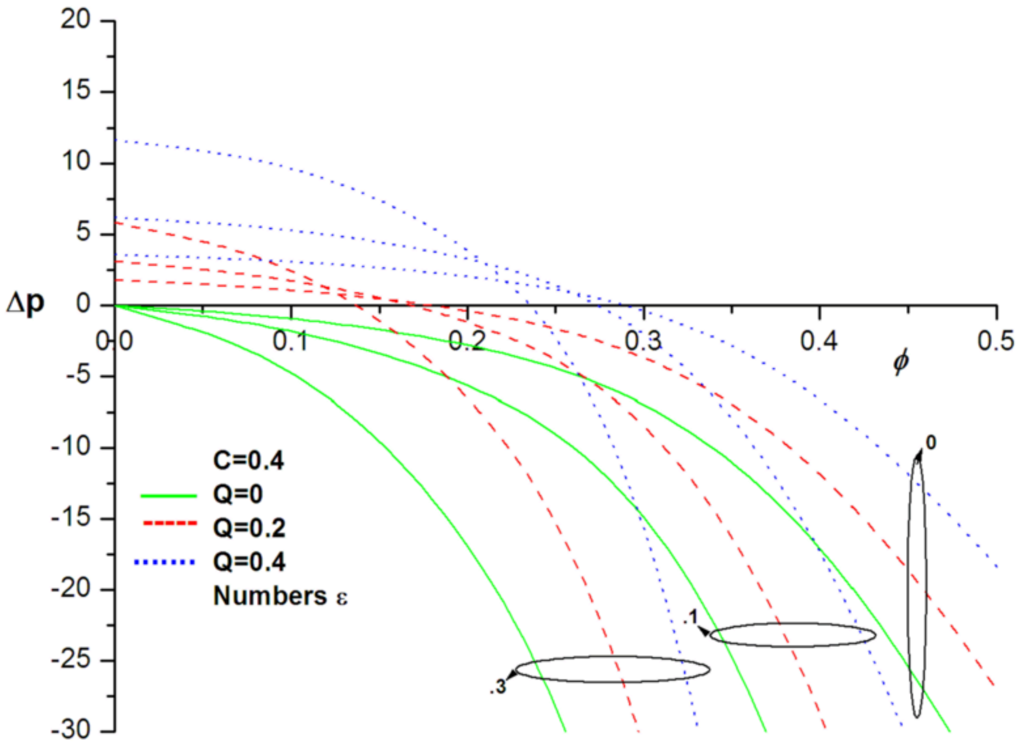


**FIGURE 7: The pressure drop, Δp versus amplitude ratio, φ, for different Q and ε**

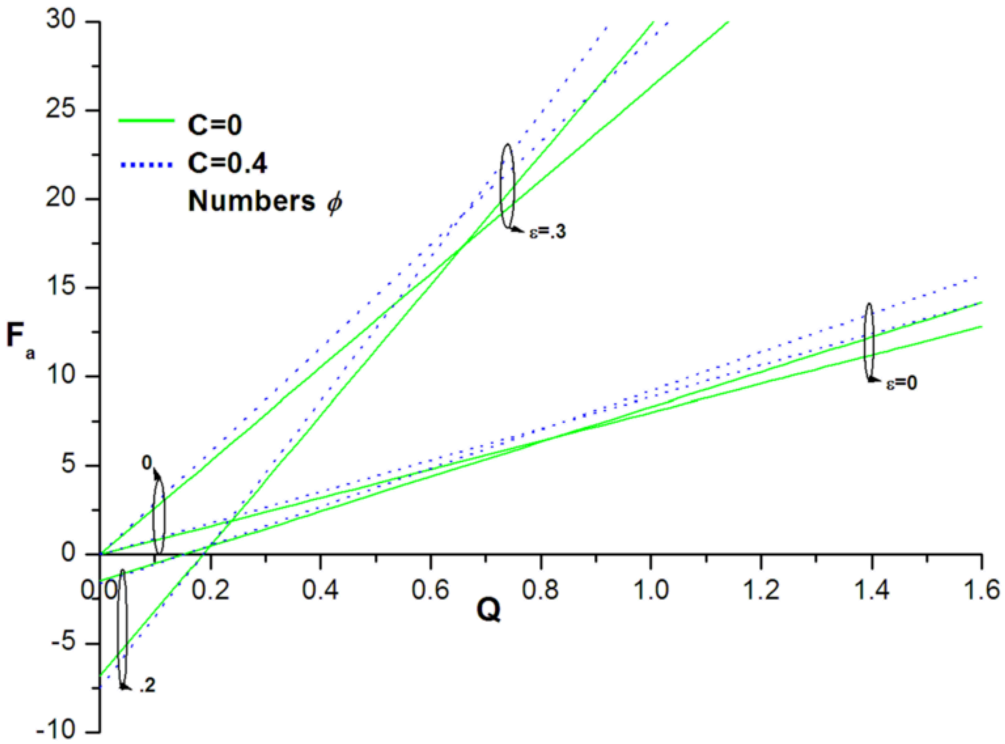


**FIGURE 8: Frictional force, Fa, versus flow rate, Q for different ε and φ**

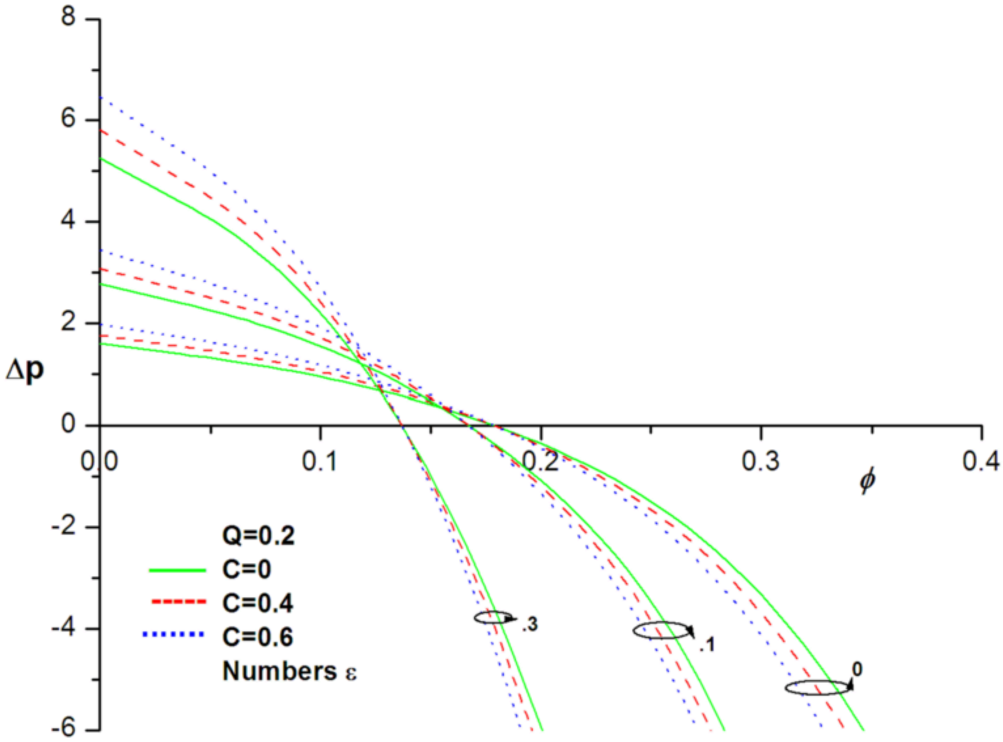


**FIGURE 9: The pressure drop, Δp versus amplitude ratio, φ for different C and ε**

The frictional force at the tube wall, $F_a$, increases with the flow rate , $Q$, for any given set of parameters in both catheterized and uncatheterized tubes, Figure *9*. However, it decreases continuously with increasing amplitude ratio, $\phi$, in the presence as well as in the absence of the catheter and approaches an asymptotically low magnitude that depends on the catheter size, $\epsilon$, for any fixed set of other parameters, Figure *10*.

The flow characteristic friction force, $F_a$, decreases continuously with increasing catheter size, $\epsilon$, for any non-zero amplitude ratio, $\phi$ . However, when $\phi = 0$ in the absence of peristaltic waves, $F_a$, increases with catheter size, $\epsilon$, for all values of particle concentration, $C$, Figure *11*. The frictional force, $F_a$, versus flow rate, $Q$, for different, $\epsilon$ and, $\phi$, Figure *12*. In the theoretical model employed for the present study, the viscosity of the peripheral layer fluid remains constant, $\mu_0$, whereas the mixture viscosity in the core region, $\mu_s$, varies with the particle concentration, $C$.

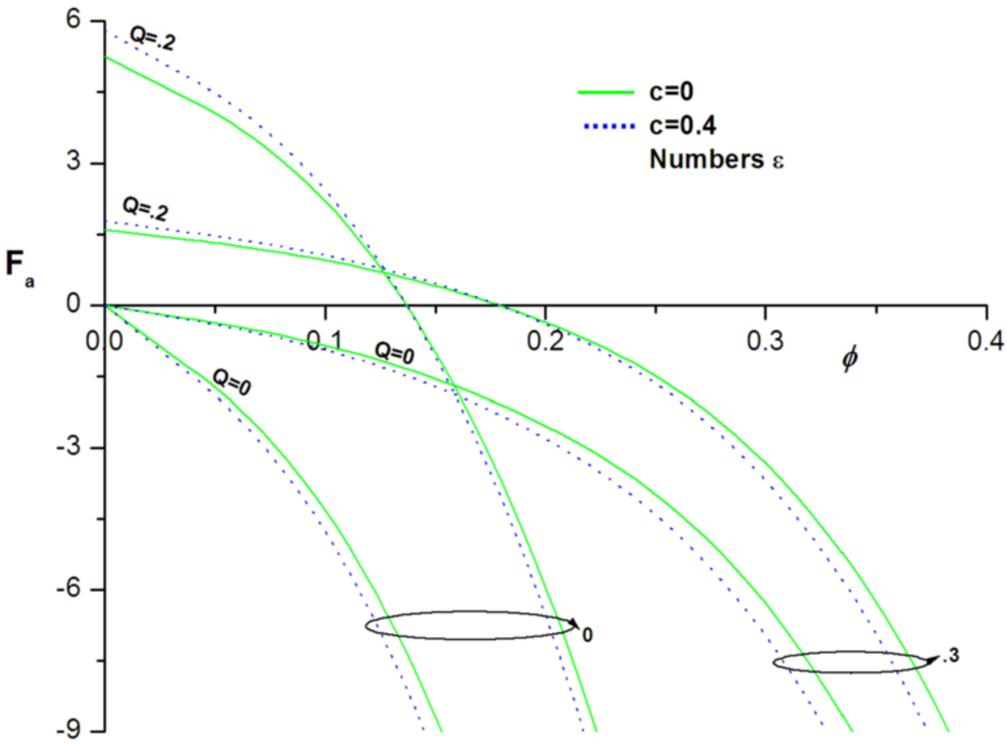


**FIGURE 10: Frictional force, Fa versus amplitude ratio, φ for different C, Q, and ε**

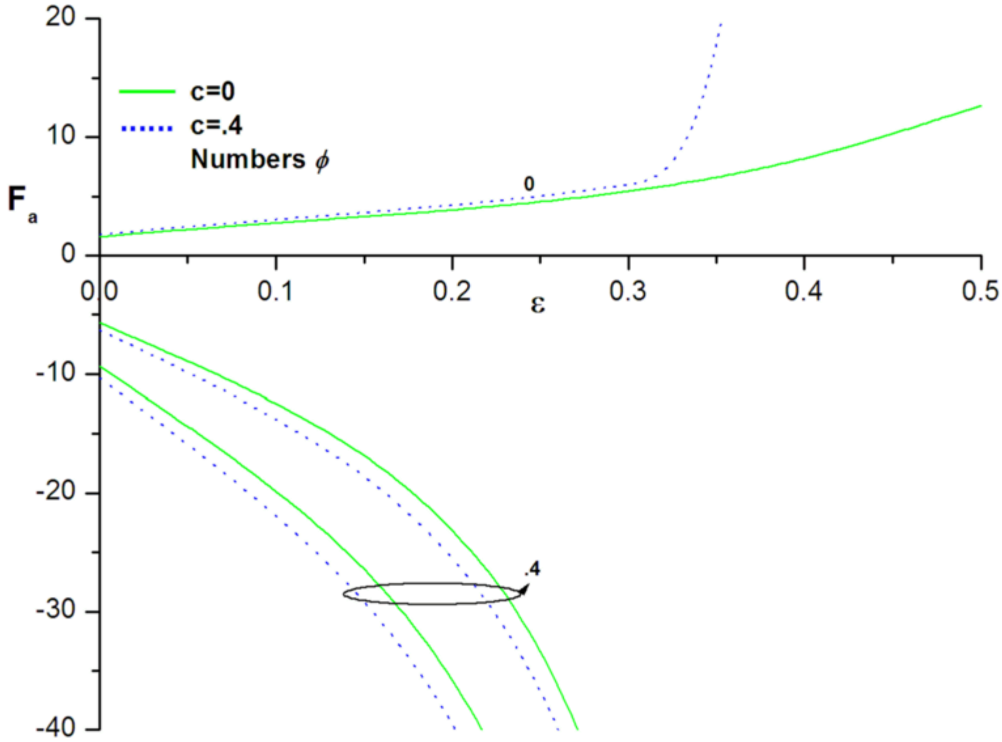


**FIGURE 11: Frictional force, Fa versus catheter size, ε for different C and φ**

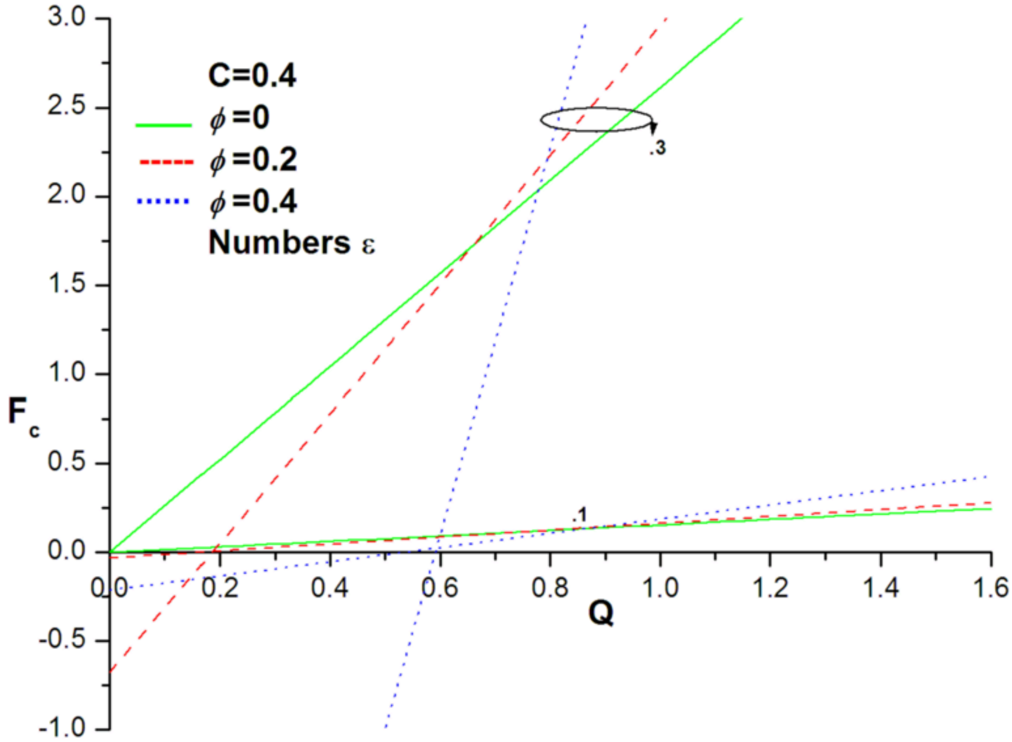


**FIGURE 12: Frictional force, Fc versus flow rate, Q for different ε and φ**

In most physiological situations, the wavelength of the peristaltic wave is significantly larger than the tube radius [5,9], as observed in systems such as the ureter, small intestine, and microcirculatory vessels like arterioles. This supports the applicability of the long-wavelength approximation employed in the present analysis. Under this condition, the Reynolds number becomes very small [3], which justifies the assumption of inertia-free, fully developed flow. The published literature further indicates that the peripheral layer thickness decreases with increasing particle concentration for a fixed tube radius [5,27]. Consequently, the parameter ϵ increases with increasing particle concentration for a given tube radius, but decreases with increasing tube radius for a fixed particle concentration. However, additional remarks are necessary regarding the interface shape. The interface is known to depend on the viscosities of the fluids in the central and peripheral regions and, in general, does not maintain a constant ratio of the radii of the two layers because mass is not conserved separately within each layer [16,17]. Nevertheless, Misra et al. reported that the interface shape is not significantly affected when the viscosity of one layer is varied relative to the other [9]. In the theoretical model adopted in the present study, the peripheral layer viscosity is constant, whereas the mixture viscosity in the core region varies with the particle concentration C. Thus, the condition identified by [9] is satisfied, thereby justifying the use of a constant value of the parameter ϵ for any prescribed particle concentration in the core region.

## Conclusions

The analysis reveals that the pressure drop increases with flow rate and particle concentration, while it decreases with increasing catheter radius and wave amplitude. The presence of the catheter significantly modifies the velocity distribution, pressure gradient, and wall shear stress - parameters that are crucial in minimizing vascular damage during clinical procedures. From a biomedical perspective, the results emphasize the importance of catheter diameter in reducing vascular resistance and the risk of mechanical injury. Frictional forces exhibit trends similar to those of pressure drop; however, their magnitude at the catheter surface is considerably lower than at the tube wall. The inverse relationship between wave amplitude and flow resistance indicates that stronger peristaltic activity enhances transport efficiency, which is relevant not only in physiological systems such as microcirculation but also in the design of biomedical devices like peristaltic pumps, infusion systems, and artificial circulatory support devices. Furthermore, the increase in flow resistance with particle concentration is consistent with the physiological effects of elevated hematocrit, which increases cardiovascular load. The nearly linear dependence of pressure drop on flow rate offers a simplified yet effective framework for predicting flow behavior in catheterized vessels. Overall, the closed-form analytical solutions for velocity profiles, pressure drop, and frictional forces derived under long-wavelength and low-Reynolds-number assumptions provide a robust and versatile platform for analyzing peristaltic transport of particulate suspensions. These findings are expected to contribute to a better understanding of microcirculation, vasomotion, and hemodynamics in catheterized arteries, and help assess risks during catheterization and support the optimization of biomedical device design and clinical interventions.

## Additional Information

### Author Contributions

All authors have reviewed the final version to be published and agreed to be accountable for all aspects of the work.

**Concept and design:** Rupesh K. Srivastav, Gopi Prasad

**Acquisition, analysis, or interpretation of data:** Rupesh K. Srivastav, Gopi Prasad, Sheetal Deshwal, Shailesh Mishra, Atul Agnihotri, Thakur Vats Singh Somvanshi

**Drafting of the manuscript:** Rupesh K. Srivastav, Gopi Prasad, Sheetal Deshwal, Shailesh Mishra, Atul Agnihotri, Thakur Vats Singh Somvanshi

**Critical review of the manuscript for important intellectual content:** Rupesh K. Srivastav, Gopi Prasad, Sheetal Deshwal, Shailesh Mishra, Atul Agnihotri, Thakur Vats Singh Somvanshi

**Supervision:** Rupesh K. Srivastav, Sheetal Deshwal, Thakur Vats Singh Somvanshi

### Disclosures

**Human subjects:** All authors have confirmed that this study did not involve human participants or tissue. **Animal subjects:** All authors have confirmed that this study did not involve animal subjects or tissue. **Conflicts of interest:** In compliance with the ICMJE uniform disclosure form, all authors declare the following: **Payment/services info:** All authors have declared that no financial support was received from any organization for the submitted work. **Financial relationships:** All authors have declared that they have no financial relationships at present or within the previous three years with any organizations that might have an interest in the submitted work. **Other relationships:** All authors have declared that there are no other relationships or activities that could appear to have influenced the submitted work.

### Data Availability Statements

The datasets (and/or code) supporting this study are available from the corresponding author upon reasonable request.

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