A Mathematical Analysis of Steady Blood Flow inside a Catheterized Artery with Asymmetric Stenosis

4.1 Introduction

Living things became an object of scientific enquiry and increasing interest, presumably in the 17th century. In fact, the quest for knowing physiological systems had its initiation in the long past (Fung, 1984). Perhaps, it might be the art of healing that had led to the study of anatomy, physiology and biological sciences. In recent years, considerable attention and keen interest have been evinced in the study of real life situations that may be due to the fundamental understanding of complex situations and probable applications in the better functioning of diseased arteries or crippled organs in human systems (Guyton, 1970). Indeed, theoretical results and experimental observations on such real life situations, are motivated toward getting an insight into the complex systems in living things in general and human beings in particular (Young, 1968; Chaturani and Biswas, 1984; Bugliarello and Sevilla, 1970; Young and Tsai, 1973). Among different systems present in a human body, one important is called the cardiovascular system (cvs), comprising of body fluid blood, blood vessels and the prime mover of blood i.e., our heart. Human blood is regarded as a suspension of red cells, white cells and platelets or thrombocytes which are suspended in an aqueous phase called plasma (Fung, 1981; Guyton, 1970). Study of blood flow through human circulatory system, has been the subject of scientific research for about a couple of years. Like most of the problems of living world, it is a complex one due to the complicated structure of blood, its pumping mechanism, the circulatory network of vessels and their constituent materials (Fung, 1981; Guyton, 1970).

It is now known that under diseased conditions, an abnormal and unfamiliar growth that develops at different locations of an inner arterial wall is generally called stenosis (Guyton, 1970). This stenosis or atherosclerosis is considered as one of the most wide
spread diseases that not only reduces the effective diameter of an artery but also disturbs the normal blood supply through the human circulatory systems. There is no exact information about this unwanted growth at the lumen of an artery. However, it may result from the deposits of cholesterol, lipid, fat etc., on the vessel wall that grow inward and restrict the blood flow (Chaturani and Swamy, 1985). Further, stenosis formation in coronary arteries can result in myocardial infarction, leading to heart failure, thrombosis etc., and that arising in arteries carrying blood to brain, can lead to cerebral strokes, haemorrhage etc. It is said that hydrodynamic factors could take a significant role in the formation, development and progression of an arterial stenosis (Young, 1968).

Many authors have proposed theoretical models for investigating blood flow through stenosed arteries, by considering blood behaving like a Newtonian fluid and stenosis as axially symmetric (Young, 1968; Biswas and Chakraborty, 2009; Biswas and Laskar, 2011). Singh et al. (2010) have considered blood flow through radially non-symmetric stenosed artery with taking blood as a Power law fluid. Recently, Chakraborty et al. (2011) have presented a suspension model blood flow through an inclined tube with an axially non-symmetrical stenosis.

Recently, studies of blood flow through catheterized arteries have drawn the attention of many researchers. Catheters are of extensive use in modern medicine. At times, artificial catheters are inserted for various clinical purposes. Pressure-flow relationship alters considerably due to such insertion in an obstructed artery. The flow of blood in uniform, non-uniform, curved or constricted artery with blood acting as Newtonian or non-Newtonian fluid and flow as steady or pulsatile, is studied by many authors (MacDonald, 1986; Shankar, 2009). The effect of catheterization and its movement on various physiologically flow characteristics, are addressed by Sarkar and Jayaraman (1995). In their blood flow model for curved stenosed artery with an inserted catheter, Dash et al. (1996) have estimated the increase in wall shear stress due to the presence of a catheter. It is also reported, due to catheterization in presence of stenosis, important physiological characteristics like, pressure drop, impedance to flow and wall shear stress, vary markedly throughout the stenosis. There may be the possibility that in keeping the ratio
of catheter radius to artery radius small, the failure rate of balloon angioplasty could be reduced.

It is reported that although, blood exhibits a Newtonian character at high shear rates, however at low shear rates it shows a non-Newtonian nature (Merrill, 1965; Charm and Kurland, 1965). The elementary rheological properties of normal blood were put forward by many researchers. The refinement of this knowledge and its applications to illness, disease and trauma as well as the study of complex rheological or particulate nature of erythrocytes which control their behaviour in microcirculation drew the attention of many investigators (Lingard, 1981). It is pointed out that under certain flow situations; blood possesses a finite yield stress (Fung, 1981; Kapur et al., 1982). In theoretical models, the traditional zero-slip condition (Schlichting and Gerstein, 2004) in velocity is used. A good number of studies of suspensions, in general and blood flow, in particular both theoretical (Vand, 1948; Nubar, 1967; Chaturani and Biswas, 1984) and experimental (Bugliarello and Hayden, 1962; Bennet, 1967), have suggested the likely presence of slip at vessel wall. It is reported that apparent viscosity gets reduced as a result of employing a wall slip (Biswas, 2000). Recently, (Mishra and Shit, 2007; Ponalgusamy, 2007; Biswas and Chakraborty, 2009; Biswas and Laskar, 2011) have introduced a slip condition in theoretical models. Thus, it seems to be rationale, in using a slip condition in blood flow modelling.

In the present work, an effort is taken to study the effect of an axial velocity of slip at the constricted wall for blood flow in the annular region of a catheterized asymmetric stenosed artery. Body fluid blood is assumed to behave as a Bingham plastic fluid and, flow is steady, laminar and unidirectional.

### 4.2 Formulation of the problem

We assume an axially symmetric, laminar, steady and fully developed flow of blood through a stenosed catheterized artery with mild stenosis as shown in Figure 4.1. Blood is assumed to act like a non-Newtonian (Bingham Plastic) fluid and the artery is assumed to be a rigid circular tube.
Figure 4.1: Schematic diagram of a catheterized artery with axially non-symmetric stenosis.

The geometry of non-symmetrical stenoses in Figure 4.1, as expressed mathematically (Chakraborty et. al, 2011), is given by

$$R(z) = R_0 - A \left[ L_0^{-1} (z-d) - (z-d)^m \right], \quad d \leq z \leq d + L_0,$$

$$= R_0, \quad \text{otherwise,} \quad (4.2.1)$$

where $R(z)$ and $R_0$ are the respective radii of the artery with the presence and absence of a stenosis, $L_0$ is the stenosis length and $d$ indicates its location, $m \geq 2$ is a parameter, usually used in determining the stenosis shape and termed as stenosis shape parameter. However, an axially symmetric stenosis occurs when $m = 2$ and the parameter $A$ is given by

$$A = \frac{\delta}{R_0 L_0^{-1}} \left( \frac{m}{m-1} \right)$$

where $\delta$ denotes the maximum height of the stenosis at the position $z = d + L_0 / m^{2.0-1}$, such that

$$\delta / R_0 < < 1.$$
4.3 Method of Solution

We consider one-dimensional steady, laminar and fully developed flow of blood (assumed to be incompressible), inside a catheterized circular tube with an axially asymmetric but radially symmetric mild stenosis as shown in Figure 4.1. The wall of the straight uniform tube is assumed rigid and the body fluid blood is represented here by a Bingham plastic fluid. The consideration of a rigid artery seems to be rationale as due to the formation of a stenosis, the flexibility of an arterial wall becomes reduced. It is already reported that the radial velocity is negligibly small and hence, it can be neglected for a low Reynolds number flow in a vessel with mild stenosis (Biswas and Chakraborty, 2009). The Navier-Stokes equations for uni-directional steady flow in cylindrical polar coordinates \((r,\theta,z)\) (Schlichting and Gerstein, 2004) in absence of body acceleration and inertia terms, may result in the following governing equation of the fluid flow

\[
\frac{1}{r} \frac{\partial}{\partial r} \left( -r \tau_{rz} \right) = \frac{dp}{dz},
\]  

where \(r\) and \(z\) denote radial and axial coordinates respectively, \(p\) is the pressure of the fluid and \(\tau_{rz}\) is the shear stress at a radial distance \(r\) from the axis.

The constitutive equation for Bingham plastic fluid (Kapur et al., 1982) is given by

\[
\tau_{rz} = \tau_0 - \mu \frac{\partial u}{\partial r},
\]  

where \(\tau_0\) denotes the yield stress and \(\mu\) is the viscosity of blood. The boundary conditions applied for integrating the equations (4.2) and (4.3) are

\[
u(r) = u_s \quad \text{at} \quad r = R(z),
\]  

\[
u(r) = 0 \quad \text{at} \quad r = R_1,
\]  

where \(u_s\) is the slip velocity at the stenotic wall (Biswas and Chakraborty, 2009) and \(R_1 (<< R_0)\) is the radius of the catheter.
Using analytical methods and employing the conditions (4.3.3-4.3.4) in equation (4.3.1) and (4.3.2), velocity function becomes

\[
u(r) = \frac{C}{4\mu} \left[ (R_1^2 - r^2) - 2r_0 (R_1 - r) \right] + \left[ \frac{u_s + \frac{C}{4\mu} \left\{ R^2(z) - R_1^2 - 2r_0 (R(z) - R_1) \right\}}{\ln \left( \frac{R(z)}{R_1} \right) - \left( \frac{R(z) - R_1}{r_0} \right)} \right],
\]

(4.3.5)

Putting \( u = u_0 \) and \( r = r_0 \) in equation (4.3.5), we get the expression for core velocity as

\[
u_0 = \frac{C}{4\mu} \left\{ (R_1 - r_0)^2 \right\} + \frac{\left\{ u_s + \frac{C}{4\mu} \left\{ (R(z) - R_1)(R(z) + R_1 - 2r_0) \right\} \right\}}{\ln \left( \frac{R(z)}{R_1} \right) - \left( \frac{R(z) - R_1}{r_0} \right)},
\]

(4.3.6)

The volumetric flow rate is defined by as

\[
Q = 2\pi \left[ \int_{R_1}^{R_2} \rho u_0(r) dr + \int_{r_0}^{R(z)} \rho u(r) dr \right],
\]

(4.3.7)

This after using equations (4.3.5) and (4.3.6) becomes

\[
Q = \frac{\pi C}{8\mu} \left[ M + 2\psi F_z N \right] + \pi u_s \psi N,
\]

(4.3.8)

where
Wall shear stress is defined as

\[ \tau_{\eta(z)} = -\mu \frac{\partial u(r,z)}{\partial r} \bigg|_{r=R(z)} \]

using the expression for \( u(r,z) \) from equation (4.6), we have

\[ \tau_{\eta(z)} = \frac{C}{2} \left( R(z) - r_0 \right) - \mu \psi \left( \frac{1}{R(z)} - \frac{1}{r_0} \right) \left\{ u_s + \frac{C}{4\mu} F_2 \right\}, \]

Effective viscosity is defined by

\[ \mu_e = \frac{\pi CR^4(z)}{8Q}, \]

Where \( C = -\frac{dp}{dz} \) and \( Q \) is defined in equation (4.3.8), using these in above equation, we get the effective viscosity as
Pressure gradient can be found from equation (4.9) as

\[
\frac{dp}{dz} = 8\mu \left[ \frac{Q}{\pi} - u_x, \psi N \right] (M + 2\psi F_2 N)^{-1}.
\] (4.3.14)

Flow resistance is defined by

\[
\lambda = \frac{1}{Q} \int_0^L \left( -\frac{dp}{dz} \right) dz,
\] (4.3.15)

where \( Q \) and \( -\frac{dp}{dz} \) is defined in equations (4.3.8) and (4.3.14) respectively.

Introducing the following non-dimensional scheme for the foregoing mathematical analysis

\[
\bar{R}(z) = \frac{R(z)}{R_0}, \bar{\delta} = \frac{\delta}{R_0}, u_x, u(r, z) = \frac{u(r, z)}{U_0}, U_0 = \frac{C_0 R_0^2}{4\mu}, \bar{Q} = \frac{Q}{Q_0},
\]

\[
Q_0 = \frac{\pi C_0 R_0^4}{8\mu}, \bar{r} = \frac{r}{R_0}, \tau_{\bar{r}} = \frac{\tau_{\bar{r}(z)}}{\tau_0}, \tau_0 = \frac{C_0 R_0}{2},
\]

\[
\bar{\mu}_a = \frac{\mu_a}{\mu_0}, \bar{R} = \frac{R}{R_0}, \bar{\lambda} = \frac{\lambda}{\lambda_0}, \left( \frac{dp}{dz} \right)_0 = \frac{\bar{\mu}_a}{\pi R_0^4}, \frac{dp}{dz}
\]

\[
\bar{\lambda}_0 = \frac{8\mu L}{\pi R_0^4} \left( \frac{dp}{dz} \right)_0 = \frac{8\mu Q_0}{\pi R_0^4}, z = \frac{z}{L}.
\] (4.3.16)
The flow geometry (equation (4.2.1)) in dimensionless form, is written as

\[
\bar{R}(\bar{z}) = 1 - \bar{A} \left[ L_0 \left( \bar{z} - \bar{d} \right) - \left( \bar{z} - \bar{d} \right)^m \right], \bar{d} \leq \bar{z} \leq \bar{d} + L_0, \quad \text{otherwise.}
\]  

(4.3.17)

where \( \bar{A} = \frac{\bar{\delta} \bar{m}}{L_0^m (m-1)} \)

The non-dimensional boundary conditions are obtained from equations (4.2.3-4.2.4) as

\[
\bar{u}(\bar{r}) = u_s, \quad \text{at} \quad \bar{r} = \bar{R}(\bar{z}), \quad \text{at} \quad \bar{r} = \bar{R}_1. \]

(4.3.18)  

(4.3.19)

The dimensionless form of the flow variables are:

**Axial Velocity:**

\[
\bar{u}(\bar{r}) = \left[ \frac{\bar{R}_1^2 - \bar{r}^2}{2\tau_0 (\bar{R}_1 - \bar{r})} \right] + \bar{u}_s + \left[ \frac{\bar{R}_1^2 - \bar{R}_1^2 - 2\tau_0 (\bar{R} - \bar{R}_1)}{\ln\left( \frac{\bar{R}}{\bar{R}_1} \right) - \frac{\bar{R}_0 - \bar{R}_1}{\tau_0}} \right] \ln\left( \frac{\bar{r}}{\bar{R}_1} \right) - \frac{\bar{r}_0 - \bar{R}_1}{\tau_0}
\]

(4.3.20)
Plug core velocity:

\[
\tilde{u}_0 = \left\{ \bar{R}_i - \bar{r}_0 \right\}^2 + \left[ \tilde{u}_s + \frac{\tilde{u}_s}{(\bar{R} - \bar{R}_i) (\bar{R} + \bar{R}_i - 2\bar{r}_0)} \right] \cdot \frac{\ln \left( \frac{\bar{R}_i}{\bar{r}_0} \right) - 1 + \left( \frac{\bar{R}_i}{\bar{r}_0} \right)}{\ln \left( \frac{\bar{R}}{\bar{R}_i} \right) - \frac{(\bar{R} - \bar{R}_i)}{\bar{r}_0}}
\]

Volumetric flow rate:

\[
\bar{Q} = \bar{M} + 2\bar{\psi} \bar{N} \left[ \tilde{u}_s + \bar{F}_2 \right],
\]

where

\[
\bar{M} = 2 \left\{ \bar{R}_i - \bar{r}_0 \right\} \left\{ \left( \bar{r}_0 \right)^2 - \left( \bar{R}_i \right)^2 \right\} + \left\{ \bar{R}_i^2 - \bar{r}_0^2 \right\} \left\{ 2 \bar{R}_i^2 - \bar{R}^2 - \bar{r}_0^2 - 2\bar{r}_0 \bar{R}_i \right\} + \frac{8}{3} \bar{r}_0 \left( \bar{R}^3 - \left( \bar{r}_0 \right)^3 \right)
\]

\[
\bar{\psi} = \left[ \ln \left( \frac{\bar{R}}{\bar{R}_i} \right) - \left( \frac{\bar{R} - \bar{R}_i}{\bar{r}_0} \right)^2 \right]^{\frac{1}{2}}
\]

\[
\bar{F}_2 = \left( \bar{R} - \bar{R}_i \right) \left\{ \bar{R} + \bar{R}_i - 2\bar{r}_0 \right\}
\]

\[
\bar{N} = \left\{ \ln \left( \frac{\bar{R}}{\bar{R}_i} \right) - 1 + \left( \frac{\bar{R}}{\bar{r}_0} \right)^2 \right\} \left\{ \left( \bar{r}_0 \right)^2 - \left( \bar{R}_i \right)^2 \right\} + \bar{F}_1
\]

\[
\bar{F}_1 = \bar{R}^2 \ln \left( \frac{\bar{R}}{\bar{R}_i} \right) - \left( \bar{r}_0 \right)^2 \ln \left( \frac{\bar{r}_0}{\bar{R}_i} \right) - 0.5 \left\{ \bar{R}^2 - \left( \bar{r}_0 \right)^2 \right\} - 2 \left( \frac{\bar{r}_0^2}{3} \right) \left( \frac{\bar{R}_i - \bar{R}}{2} \right)
\]
Effective viscosity:

Using the non-dimensional expression for volumetric flow rate from equation (4.3.22), we will get the expression for effective viscosity in dimensionless form as

$$\bar{\mu}_e = \frac{R^4(z)}{Q},$$  \hspace{1cm} (4.3.23)

Wall shear stress:

$$\bar{\tau}_{R(z)} = 2\left(\bar{R} - \bar{\tau}_0\right) - \ddot{\psi} \left(\frac{1}{R} - \frac{1}{\bar{\tau}_0}\right) \left[\bar{u}_s + \bar{F}_2\right]$$ \hspace{1cm} (4.3.24)

where $\ddot{\psi}, \bar{F}_2$ are given in equation (4.3.22)

Flow resistance:

$$\bar{\lambda} = \frac{1}{Q} \int_0^L \left( - \frac{dp}{dz} \right) dz,$$ \hspace{1cm} (4.3.25)

Pressure gradient:

$$-\frac{dp}{dz} = \left[ Q - 2\bar{u}_s \ddot{\psi} \right] \left( \bar{M} + 2\ddot{\psi} \bar{F}_2 \bar{N} \right)^{-1},$$ \hspace{1cm} (4.3.26)

where expressions for $\ddot{\psi}, \bar{M}, \bar{F}_2, \bar{N}$ are given in equation (4.3.22).
4.4 Results and Discussions

Blood flow through a rigid artery with non-symmetrical stenosis, has been investigated by considering blood as a Bingham Plastic fluid. The present investigation deals with the analytic solutions for velocity profile, wall shear stress and effective viscosity. The flow parameters are discussed graphically for better understanding of the problem.

4.4.1 Axial Velocity Profiles

Axial velocity is computed from equation (4.3.20) in dimensionless form and its variation with different parameters is presented graphically.

Variation of axial velocity with radial distance in the constricted region, for different values of slip velocities at the stenotic wall, different stenosis shape parameters, yield stress, different catheter radius, have been represented in Figures 4.2-4.4 respectively.

**Figure 4.2:** Variation of axial velocity with radial distance for different values of slip velocity and catheter radius for \( m=2 \).
From the figures, it is observed that in all cases, axial velocity increases for the employment of a slip condition at the stenotic wall. Its magnitude, calculated with an
axial slip, is greater than that obtained with zero slip. As catheter radius increases, axial velocity decreases for both slip and no-slip cases at the stenotic wall. Axial velocity is maximum at the initiation of the stenosis and minimum at the throat of the stenosis. However, as stenosis shape parameter increases in magnitude, velocity increases throughout the stenotic region. Axial velocity decreases with the increase of yield stress.

4.4.2 Flow Rate

Variation of flow rate can be obtained from equation (4.3.22) against axial distance for the cases of slip and no-slip at the stenotic region, different stenosis size, yield stress and stenosis shape parameter are drawn in Figures 4.5-4.7.

![Flow Rate Variation](image)

**Figure 4.5**: Variation of flow rate with axial distance for different values of slip velocity and yield stress for $m=2$. 
Figure 4.6: Variation of flow rate with axial distance for different values of slip velocity and catheter radius for $n=2$.

Figure 4.7: Variation of flow rate with axial distance for different values of slip velocity and stenosis size for $m=2$. 
The figures show that the flow rate is minimum at the throat of the stenosis and maximum at its either end (i.e., at the initiation and at the termination of the stenosis). As slip increases, magnitude of $Q$ increases rapidly. However, values of $Q$ obtained with velocity slip are greater than those with no-slip at the interface. As height of the stenosis increases from mild to severe form, flow rate decreases significantly. As catheter radius and yield stress increase, flow rate reduces from higher magnitude at the constricted region of an artery.

4.4.3 Wall Shear Stress

Wall shear stress is computed from the equation (4.3.24) and its variations with axial distance for slip and no-slip cases, different catheter radius, stenosis shape parameter $m$, yield stresses and different stenosis size, are shown in Figures 4.8-4.10.

**Figure 4.8:** Variation wall shear stress with catheter radius for different values of stenosis shape parameter and slip velocity for $\delta=1-\sqrt{3}/2$. 
**Figure 4.9:** Variation wall shear stress with axial distance for different values of stenosis shape parameter and slip velocity for $\delta=1-\sqrt{3}/2$.

**Figure 4.10:** Variation wall shear stress with axial distance for different values of stenosis size and slip velocity for $m=2$. 

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The text seems to be discussing the variation of wall shear stress with axial distance for different values of stenosis parameters. The figures illustrate these variations with different symbols and lines indicating the different conditions.
From the figures, it has been noticed that wall shear stress is the minimum at the either end of stenosis and attains the maximum magnitude at the throat of the stenosis. As stenosis shape parameter m increases, wall shear stress decreases significantly. It is also important to observe that the as catheter radius increases, wall shear stress also increases rapidly. Also, it is observed that increase in stenosis size increase the wall shear stress. In all the cases, wall shear stress is the minimum for a velocity slip at the stenotic wall than that of the the same with no-slip cases.

4.4.4 Effective viscosity

Effective viscosity \( \mu_e \) is computed with the help of the equation (4.3.23) and its variation with axial distance in the stenotic region, for three cases of stenosis size, different values of yield stress, stenosis shape parameter and for slip and no-slip cases, are shown in Figures 4.11-4.13.

![Graph showing effective viscosity variation with catheter radius for different values of yield stress and stenosis size for m=2.]

**Figure 4.11:** Variation effective viscosity with catheter radius for different values of yield stress and stenosis size for m=2.
Figure 4.12: Variation effective viscosity with stenosis size for different values of stenosis shape parameter for $R_I=0.1$.

Figure 4.13: Variation effective viscosity with axial distance for different values of catheter radius and slip velocity for $\delta=1-\sqrt{3}/2$ and $m=2$. 
In the figures, it has been clearly indicated that apparent viscosity decreases with a slip at the boundary and as the slip increases in magnitude, it decreases accordingly. Further, apparent viscosity increases as yield stress increases. Also, as height of the stenosis rises from mild to severe one, apparent viscosity increases from lower to greater magnitude. Magnitude of apparent viscosity is found to be lower as stenosis shape parameter increases from a lower to higher value. As catheter radius increases, apparent viscosity also increases.

4.4.3 Pressure Gradient
Pressure gradient is computed from the equation (4.3.26) in dimensionless form and its variations with different flow parameters, are presented graphically, in Figures (4.14-4.16).

Variation of pressure gradient with axial distance for different stenosis size, slip and no-slip cases and fluid viscosity are represented, in Figures 4.14-4.16 respectively.

**Figure 4.8:** Variation of pressure gradient with axial distance for different values of catheter radius for \( m=2 \) and \( \delta=1-\sqrt{3}/2 \).
Figure 4.9: Variation of pressure gradient with axial distance for different values of yield stress for $m=2$ and $\delta=1-\sqrt{3}/2$.

Figure 4.10: Variation of pressure gradient with axial distance for different values of stenosis size for $m=2$. 
It is clearly indicated in the figures that pressure gradient is lower in magnitude in all the cases for a velocity slip than that of no-slip at the constricted wall. However, as stenosis size increases from mild to severe, pressure gradient significantly increases to a higher magnitude. Pressure gradient is increases as both catheter radius and yield stress increases from a lower magnitude to higher one.

4.5 Conclusion

In this investigation, the steady flow of blood in a uniform artery with axially non-symmetrical stenosis, has been analyzed. A condition for an axial velocity slip, is applied at the flow boundaries and three different stages of stenosis viz., mild, moderate and severe, have been considered. Analytical expressions for axial velocity, flow rate, wall shear stress, apparent viscosity, pressure gradient are obtained and their variations with different flow parameters have been depicted in figures.

It is observed that both axial velocity and the flow rate, will increase with an increase of a velocity slip but decreases with an increase in height of the stenosis. Shear stress attains the maximum magnitude at the throat of the stenosis and lower magnitudes at either end of the stenosis. Further, the wall shear stress is found to be lower, due to the employment of a slip at the wall. An increase in the size of the stenosis increases resistance to blood flow through arteries in the brain, heart and other organs of the body. This may lead to stroke, heart attack and various cardiovascular diseases.

It has been reported that blood shows a Power-law behavior and it exhibits a finite non-zero yield stress. In order to take account of both the characteristics, a non-Newtonian model for blood flow wherein blood behaves as a Herschel-Bulkley (H-B) fluid, is taken in the next chapter.