Influence of blood rheology and vessel wall motion on arterial fluid mechanics

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Most model studies of arterial fluid mechanics have assumed that blood is a Newtonian fluid and that vessel wall motion driven by the pressure pulse has a small influence on the local velocity and pressure distributions. This paper provides a brief historical review of arterial flow modeling which emphasizes recent developments in non-Newtonian blood analog fluids and studies of the influence of vessel wall motion on local flow fields. It is pointed out that vessel wall motion can have a dominant effect on mean pressure gradient and a significant effect on mean wall shear stress in the aorta.

INTRODUCTION

The classical model of blood flow in the circulatory system derives from experiments of the French physician J. L. Poiseuille (1799-1869). Poiseuille's law describes the relationship between pressure drop (ΔP_0) and flow rate (Q_0) for steady flow of a Newtonian fluid (viscosity μ) in a rigid, straight vessel of length L and uniform diameter ($D_0 = 2R_0$),

$$\Delta P_0 = \left(\frac{8\mu L}{\pi R_0^4}\right) Q_0 \qquad (1)$$

The axial velocity profile associated with this flow has the well known parabolic shape, and the shear stress imposed by this flow on the wall of the vessel is given by

$$\tau_0 = \left(\frac{4\mu}{\pi R_0^3}\right) Q_0 \qquad (2)$$

Equation (1) is the standard model employed in Physiology textbooks to describe blood flow in individual vessels and to explain the distribution of flow in the series / parallel network of vessels which distributes blood to tissues of the body. Because the shear stress of flowing blood on vessel walls has been shown in recent years to have a profound influence on the biology of the arterial wall (Frangos, 1993), physicians and physiologists have used equation (2) to estimate τ_0 from measurements of flow rate and vessel dimension.

Of course Poiseuille's law does not account for the characteristic pulsatile nature of blood flow or for the elasticity of blood vessel walls which allow pressure and flow waves to propagate from the heart. The one-dimensional, inviscid theory of wave propagation in elastic tubes dates back to the first half of the 19th century and is associated with the names T. Young and E. H. Weber (see Noordergraaf, 1969). However, the two-dimensional, viscous, linear theory was not well developed until much later in the work of Morgan and Kiely (1954) and Womersley (1955). D. A. McDonald, in his classic monograph "Blood Flow in Arteries" (First edition, 1960) described Womersley's theory in great detail and reviewed available experimental data from the arteries of dogs which showed a very satisfying agreement between Womersley's theory and measurements of pulsatile pressure gradient and flow.

Womersley's theory reveals that the unsteadiness of oscillatory flow, as characterized by the unsteadiness parameter α ($\equiv R_0\sqrt{\omega/\nu}$; ω is the angular frequency of the fundamental flow harmonic and υ is the kinematic viscosity of the fluid) can have a major influence on the magnitude of the

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longitudinal impedance (pressure gradient / flow rate) of the flow and its associated wall shear stress. At high values of α ($\alpha > 10$) which are characteristic of pulsatile flow in the largest arteries, the modulus of the longitudinal impedance normalized by its Poiseuille flow value $(8\mu/\pi R_0^4)$ approaches $\alpha^2/8$ and the modulus of the wall shear stress divided by the flow rate normalized by its Poiseuille flow value $(4\mu/\pi R_0^3)$ approaches $\alpha/4$. Thus, in a large artery like the aorta where $\alpha \sim 20$ for the fundamental flow oscillation, the peak pressure gradient would be some 50 times that predicted by Poiseuille's law and the peak wall shear stress could be 5 times the Poiseuille flow value. These are very significant effects associated with flow unsteadiness which are described by Womersley's theory but not accounted for by Poiseuille's law.

Interest in the effects of complex vessel geometry on arterial fluid mechanics came into focus in the late 1960's and early 1970's when attention was drawn to the fact that atherosclerotic lesions on blood vessel walls tended to be localized around branch points and in regions of vessel curvature and that the wall shear stress, either through a mechanical injury mechanism or a mass transport mechanism, was influential (Fry, 1969; Caro et al, 1971). Many studies, both experimental and theoretical, followed this initial impetus and demonstrated, using rigid wall models and Newtonian blood analog fluids, that wall shear stress is strongly dependent on spatial position in complex geometries (Friedman et al, 1981; Ku et al, 1985; Chang and Tarbell, 1988). These studies and others have provided circumstantial evidence that the localization of atherosclerotic plaques in arteries is associated with the wall shear stress distribution (Nerem, 1992).

However, Moravec and Liepsch (1983) and Liepsch and Moravec (1984) conducted flow visualization and laser-Doppler anemometry (LDA) studies of pulsatile flow through an elastic arterial branch model using a non-Newtonian blood analog fluid and observed large differences in the velocity profiles relative to those measured with a Newtonian fluid in a rigid model. These studies stimulated considerable interest in the effects of non-Newtonian blood rheology and elastic vessel wall motion on velocity and wall shear stress distributions in arteries.

NON-NEWTONIAN RHEOLOGY INFLUENCE

Human blood is a non-Newtonian fluid which displays marked shear thinning behavior at low shear rates (below 100 s^{-1}) in steady viscometric flows (Cokelet, 1987). At higher shear rates, blood viscosity approaches an asymptotic value of 3.5-4.0 cp at normal hematocrits (volume fraction of red blood cells) and appears to be Newtonian. If one uses the Poiseuille flow formula (eqn 2) to estimate mean (time-averaged) wall shear rates in arteries based on physiological measurements of Q_0 and R_0 (McDonald, 1974), one typically finds values above 100 s^{-1}; peak values are much higher. Thus, on the surface, it may seem reasonable to treat blood as a Newtonian fluid in modeling flow in arteries. However, regions of curvature and branching in the circulation may display flow separation, flow reversal and secondary flow, all of which can lead to non-uniform distribution of shear rate around the periphery of the vessel including low and high shear areas (Chang and Tarbell, 1985).

To address the question of the influence of non-Newtonian rheology on pulsatile flow in complex geometries Moravec and Liepsch (1983) introduced aqueous polyacrylamide as a transparent blood analog fluid which would be useful for flow visualization and LDA studies of complex flow fields. Mann and Tarbell (1990) questioned the use of aqueous polyacrylamide as a blood analog fluid because the solutions are highly elastic (Bird et al, 1987), displaying significant normal stresses which are not characteristic of human blood (Copley and King, 1975). To test this hypothesis, Mann and Tarbell (1990) prepared aqueous polyacrylamicle solutions which accurately matched the shear-thinning power law behavior of blood in steady viscometric flows. They measured large normal stress values for these solutions in the same shear rate range in which Copley and King (1975) were unable to detect normal stresses in human blood. Measurements of wall shear rate waveforms in oscillatory through a curved artery model were then conducted using flush-mounted hot-film anemometry, and significantly different wall shear rate values were measured for the aqueous polyacrylamide blood analog fluid and bovine blood under nearly identical sinusoidal flow conditions. In fact, at the inner wall of the curved artery (nearest the center of curvature) the wall shear rate waveforms were qualitatively different, and the peak values differed by a factor of four (see Fig 8 of Mann and Tarbell, 1990). This study indicated that aqueous polyacrylamide is not a good blood analog fluid.

In a subsequent study, Brookshier and Tarbell (1993) developed a transparent blood analog fluid using aqueous solutions of the natural polymer Xanthan gum and glycerin. These solutions matched both the viscous and elastic components of the complex viscosity of low, medium and high hematocrit blood in the shear rate range 1-1000 s^{-1} at 2Hz.

The normal stresses in this blood analog fluid are relatively low, as demonstrated previously for aqueous Xanthan gum solutions without glycerin (Mann and Tarbell, 1990). Measurements of wall shear rate in pulsatile flow were conducted in straight and curved artery models using both the blood analog fluid and porcine blood to compare their flow behavior. Peak wall shear rates determined in these models under nearly identical flow conditions with the two fluids were not statistically different ($p>0.05$). These results indicate that aqueous Xanthan gum / glycerin solutions provide a good blood analog fluid for flow conditions characteristic of large arteries. Because the fluid is transparent and made from readily available and fairly inexpensive materials, it may be generally useful for model hemodynamic studies, particularly those employing optical techniques.

In addition to providing a useful blood analog fluid for experimental investigations, these studies suggest that a purely viscous, shear thinning, rheological constitutive equation should provide a good model for theoretical studies of blood flow in arteries. Simple examples would be the two-parameter Casson model and the two-parameter power law model. In fact, three-dimensional pulsatile flow simulations in rigid arterial bifurcations using these rheological models have been reported recently (Perktold et al, 1991; Xu et al, 1992). By making comparisons with Newtonian flow simulations under the same flow conditions these studies conclude that the non-Newtonian effects on velocity fields are relatively small. This is consistent with the experimental studies of Brookshier and Tarbell (1991, 1993).

ELASTIC WALL MOTION INFLUENCE

Blood vessel walls are elastic, and the diameter of large arteries can vary by ± 5 percent over the cardiac cycle (McDonald, 1974). Traditional interest in elastic vessels has focused on the problem of propagation of pressure and flow pulses in the cardiovascular system (Noordergraff, 1969). Less attention has been paid to the influence of wall motion on the local flow field at a particular axial position in an artery (Ling and Atabek, 1972), and the potential influence of wave propagation / reflection on local flow fields seems to have been clearly recognized only recently (Klanchar et al, 1990).

Wang and Tarbell (1992, 1994) developed a nonlinear analysis of flow in a straight, elastic tube (nonlinear Womersley problem) accounting for convective acceleration and drawing attention to the effects of wave reflection which alters the temporal phase relation between the pressure and flow waves. They modeled blood flow using a homogeneous, incompressible, Newtonian fluid in an isotropic, thin-walled elastic tube with longitudinal constraint. The latter assumption is consistent with observations that longitudinal motion of arteries is highly restricted because the vessel is tethered to the surrounding tissue (McDonald, 1974). The principal motion of the vessel wall is radial. The equations were analyzed by seeking perturbation solutions in the radius variation parameter, ϵ ($\equiv (R_{max}-R_0) / R_0$), where R_{max} and R_0 are the maximum and mean radii of the tube over a pulse cycle, respectively. The case of sinusoidal flow was considered in the perturbation solutions of Wang and Tarbell (1992, 1994), while numerical solutions for multi-harmonic physiological flows were presented by Dutta et al (1992).

An interesting feature of the perturbation solutions is that the velocity field at a particular axial position along the tube (z) can be determined by knowing only the flow rate and radius variation waveforms at that position without a detailed description of the upstream and downstream boundary conditions. This "local flow" concept had been suggested earlier by Ling and Atabek (1972) without theoretical justification. The local flow idea is extremely important in cardiovascular flow modeling because one can avoid the complex axial boundary conditions associated with the branching architecture of the vascular tree. By measuring the flow and radius waveforms locally, it is possible to predict the local shear stress, pressure gradient and other flow features. Local flow results can be found in Wang and Tarbell (1992, 1994).

The mean values of the shear stress and pressure gradient are affected by vessel wall motion to a greater degree than the amplitudes. The mean wall shear stress induced by vessel wall motion, τ_I $(\equiv \bar{\tau} - \tau_0$, where $\bar{\tau}$ is the time-averaged wall shear stress and τ_0 is the Poiseuille value given by eqn 2 with Q_0 and R_0 denoting the mean flow rate and vessel radius), is given by

$$\tau_I/\tau_0 = \frac{\alpha\epsilon}{8}\left(\frac{Q_1}{Q_0}\right)\tau_m(\alpha,\phi) \tag{3}$$

In eqn 3, Q_1 is the amplitude of the sinusoidal flow rate oscillation and τ_m is a function of the phase angle between the pressure and flow oscillations (ϕ) and the unsteadiness (α). τ_m is of order of magnitude 1 and at high α is proportional to $\cos(-\phi+45°)$. The mean pressure gradient induced by vessel wall motion, ΔP_I $(\equiv \overline{\Delta P} - \Delta P_0$, where $\overline{\Delta P}$ is the time-averaged pressure drop and ΔP_0 is the Poiseuille value), is given by

$$\Delta P_I/\Delta P_0 = \frac{\alpha^2\epsilon}{4}\left(\frac{Q_1}{Q_0}\right)\Delta P_m\,(\alpha,\phi) \tag{4}$$

where ΔP_m is a function having order of magnitude 1 which is proportional to $\sin(-\phi)$ at high α.

Equations (3) and (4) indicate that the induced mean shear stress and pressure gradient are proportional to the unsteadiness (α), the diameter variation (ϵ) and the pulsatility (Q_1/Q_0). These quantities are highest in the aorta near the heart. To obtain a sense of the magnitude of the induced quantities, consider a typical aortic flow case: $\alpha = 20$, $\epsilon = .05$ and $Q_1/Q_0 = 3$. For these parameters, the magnitude of τ_I/τ_0 is 0.375 and the magnitude of $\Delta P_I/\Delta P_0$ is 15. Clearly, the induced shear stress is significant and the induced pressure gradient is dominant.

It is also clear that the phase angle (ϕ) has an important influence. At high α and a normal physiologic phase angle between the first harmonic of the pressure and flow pulses of -45°, the induced pressure gradient is negative (opposite the sign of the Poiseuille pressure gradient). This is consistent with physiological flow simulations in the aorta which indicate that the mean pressure actually rises in the direction of flow (Dutta et al 1992). It is interesting to note that a vasoactive drug such as sodium nitroprusside can increase ϕ to 0° and thus reduce the induced pressure gradient nearly to zero. It should also be noted that the sign of the induced wall shear stress changes in the physiologic range at about $\phi = -45°$.

The effects of vessel wall motion on fluid mechanics in complex arterial geometries such as curves and branches are not well understood at present. A study by Duncan et al (1990) compared wall shear rate measurements in rigid and compliant models of a human aortic bifurcation under nearly identical flow conditions. They observed that compliance reduced shear rates at the outer walls, while at the walls of the flow divider the shear rate was increased. The deviations in shear rate between rigid and compliant models at different sites varied from 12% to 124%. However, this study did not use physiologic phase angles and thus there is still uncertainty as to the magnitude of the effects in a true physiologic simulation.

It must also be remembered that the parameters α, ϵ, Q_1/Q_0 and ϕ which are so influential in the straight tube theory, vary widely throughout the circulation, depend on the age and health of individuals, and can be influenced by exercise, vasoactive drugs and environmental factors. Thus it seems that further studies of the influence of vessel wall motion on arterial fluid mechanics would be fruitful.

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