Effects of Changes in the Apparent Viscosity of Blood with Vessel Size on Retinal Microcirculation: Significance of the Fåhraeus-Lindqvist Effect

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Abstract  The purpose of this study was to quantitatively assess the influence of the Fåhraeus-Lindqvist effect on the microcirculation in the arteriovenous network of the human retina. A mathematical model was used to simulate the arteriovenous distributions of hemodynamic parameters within a microvascular network of successive, symmetric bifurcating branches that were constructed based on both flow conservation and a modified Murray's law with a diameter exponent of 2.85. The vessel calibers ranged from a 108-μm arteriole and a 147-μm venule down to the 5-μm capillaries. The distributions of vascular resistance, pressure drop, and wall shear stress as a function of vessel diameter within the retinal microcirculatory network with the Fåhraeus-Lindqvist effect were lower than those without the Fåhraeus-Lindqvist effect. The efficiency of blood transport to tissues in the microvascular bed, which was evaluated in terms of the inverse of the mechanical energy cost of the product of the driving pressure and blood flow, was 44% greater with the Fåhraeus-Lindqvist effect than without the Fåhraeus-Lindqvist effect. These results quantitatively demonstrated that the Fåhraeus-Lindqvist effect plays an important role in reducing the physical energy required to transport blood that flows through the microcirculatory network. The integrated and interactive relationships between shear stress, circumferential wall stress, vessel radius, and wall thickness in response to acute and chronic increases in perfusion pressure are discussed with regard to their coordinating roles in the control of blood flow and pressure in microcirculation.

Keywords : circumferential wall stress, diabetes mellitus, hypertension, retinal blood flow, wall shear stress.

1. Introduction

One of the many important roles of the circulation is to transport oxygenated blood and nutrients to, and deoxygenated blood and metabolites from, the microvascular beds. The oxygen is mainly attached to hemoglobin within red blood cells (erythrocytes). An increase in the number of erythrocytes per unit volume of blood (hematocrit, Hct) promotes oxygen transport. However, this increase in Hct also causes the viscosity of blood to increase dramatically. Importantly, a normal Hct level in humans achieves oxygen transport at the most efficient level with respect to the hemodynamic resistance due to blood viscosity intrinsic to flow (Fig. 1); accordingly, the Hct in normal subjects is approximately 40%~45% [1]. This rheological property of blood in the macrocirculation with respect to the maximum transport of oxygen is evaluated in terms of the ratio of oxygen transport to blood viscosity [2]; by definition, the rate of oxygen transport is expressed as the product of oxygen content and blood flow in the arterial blood.

It is generally considered that blood behaves like a Newtonian fluid in large vessels macroscopically. In contrast, blood in smaller vessels (less than 400 μm in diameter) exhibits the flow of a non-Newtonian fluid; i.e., the apparent viscosity of blood differs according to changes in the fluid shear rate [1]. The apparent viscosity of blood that flows into such minute vessels effectively decreases due to the separate streams of plasma and red blood cells with different velocities as the vessel size decreases; this decrease in the apparent viscosity of blood is referred to as the Fåhraeus-Lindqvist effect [1]. In fact, the apparent viscosity of blood in dog hind limbs was observed to be significantly lower than the values obtained with an in vitro viscometer by Whittaker and Winton [3]. Moreover, an in vivo comparative study [4] demonstrated that a circulatory system perfused with a hemoglobin solution of comparable oxygen-carrying capacity (without the Fåhraeus-Lindqvist effect) required a 37% greater workload on the heart to maintain a given cardiac output, compared with a circulatory system with the Fåhraeus-Lindqvist effect. At present, it is still unclear how a reduction in the apparent viscosity of blood flowing throughout the hierarchy of arterioles, capillaries, and venules influences the peripheral circulation and hemodynamic parameters within the microvascular network.

Therefore, the purpose of the present study was to examine the differences in the arteriovenous distributions of hemodynamic parameters within the micro-
circulatory network between with and without the Fåhraeus-Lindqvist effect and to clarify the hemodynamic and physiological significance of the Fåhraeus-Lindqvist effect. To compare the effects of the two different hemorheological factors on the microcirculation, we used a theoretical model of the microcirculatory network of the human retina and simulated the arteriovenous distributions of hemodynamic parameters such as blood velocity, intravascular pressure, wall shear rate, shear stress, and circumferential wall stress [5, 6].

2. Methods

2.1 Theoretical retinal microvascular network
To quantify the arteriovenous distributions of hemodynamic variables within the microcirculatory network of the human retina, we used a fractal-based microvascular model consisting of a dichotomous symmetric branching system [5, 6]. Briefly, following the central retinal artery, every bifurcation was recursively developed at a distance of an individual branch length \( L(r) = 7.4r^{1.15} \) by a centrifugal scheme in which each of the vessel segments was assigned a generation number from upstream to downstream. The relation between the radius of the mother vessel and the radii of the two daughter vessels at each of the dichotomous branches was defined by a modified Murray’s law with a diameter exponent of 2.85 (Fig. 2), which was developed by Takahashi et al. [5]. The network model consisted of 14 generations that branched dichotomously to the ends of arterioles or to the precapillaries with a diameter of 5.1 \( \mu \)m. The venous system was formed in the same way as the arterial system. In practice, the arterial and venous networks were constructed from 1st-generation vessels with diameters of 108 and 147 \( \mu \)m, respectively. The arteriolar ends and the venular ends were connected by sets of four capillaries [7] with a diameter of 5.0 \( \mu \)m [8] and a length of 500 \( \mu \)m [9].

2.2 Equations in the hemodynamic simulation
Based on the input data (\( r_1 = 2.055 \text{ cm s}^{-1}; r_1 = 5.4 \times 10^{-3} \) cm in radius) for the 1st generation branch, the volumetric flow rate (\( Q_v \)) and mean flow velocity (\( \bar{v}_v \)) in a cylindrical vessel of the 4th generation were calculated as

\[
Q_v(r_4) = Q_v(r_1)/2^{1-4} = \pi r_4^2 \bar{v}_4/2^{1-4}
\]

The mean blood pressure in the proximal arteriole of the 1st generation was 38.9 mmHg [5]. A decrease in pressure (\( \Delta P \)) along a length \( L(r) \) of any vessel with a radius \( r \) was estimated by Hagen-Poiseuille’s equation

\[
\Delta P = 8\mu(r) \cdot L(r) \cdot Q(r) / (\pi r^5),
\]

where \( \mu(r) \) is the apparent viscosity of blood in the vessel and can be expressed approximately as \( \mu(r) = 0.043/(1+4.29/r)^2 \) for arterioles and \( \mu(r) = 0.046/(1+4.29/r)^2 \) for venules [5]. Vascular resistance to blood flow through a vessel segment was also computed from the Poiseuille resistance, \( R = 8\mu(r) \cdot L(r)/(\pi r^5) \).

The tangential force of the bloodstream acts on the luminal surface of the blood vessel as wall shear stress \( \tau_w(r) \), which can be given as the product of blood viscosity and wall shear rate: \( \tau_w(r) = \mu(r) \cdot \gamma_w(r) \), where the shear rate \( \gamma_w(r) \) at the wall surface is calculated as \( 4\bar{v}/r \). The
vessel wall is also exposed to perpendicular forces from inside and outside the wall as circumferential wall stress: \( \sigma_w(r) = P_i(r) \cdot \frac{r}{w(r)} \), where \( P_i(r) \) is the transmural blood pressure, \( r \) is the inner radius of the vessel, and \( w(r) \) is the thickness of the vessel wall. The wall thickness is given by empirical expressions \( w_w(r) = 4.62 \times 10^{-7} r^{1.12} \) for arterioles and \( w_w(r) = 2.31 \times 10^{-7} r^{0.311} \) for venules [6].

Hemodynamic parameters in the venular network were recursively determined using blood flow from the capillary vessels supplied by the arteriolar network, rather than blood flow measured in the large venule [5]. The relevant details about the numerical parameters used for the calculation were given in our previous study [6].

### 2-3 Comparative analyses

To examine the effects of decreases in the apparent viscosity of blood flow in the microvessels of the network on the microcirculation, we compared the arteriovenous distributions of hemodynamic parameters within the human retinal microcirculatory network in the presence of the Fåhraeus-Lindqvist effect and in the absence of the Fåhraeus-Lindqvist effect, assuming that a constant viscosity of bloodflowinthemicrovascularsofthenetworkwasconsidered. The blood velocity in the venular system was considerably lower than that in the arteriolarsystem.However,theflowvelocityincreasedalmostlinearlyfromthelargearteriolewithadiameterof108mm through small arterioles to precapillaries, whereas that in the venular network gradually increased as the vessel size increased. The flow velocity in the venular system was considerably lower than that in the arteriolar system in vessels of the corresponding sizes. The blood velocity in the true capillary vessel with a diameter of 5 \( \mu \)m dropped precipitously because of a larger increase in the total cross-sectional area of parallel capillaries. The shear rate reached a maximum in the precapillary vessels and then fell rapidly to a minimum in the capillary vessels. The lower shear rate in the capillary vessel compared with that in the precapillary vessel was caused by the large decrease in mean blood velocity.

As shown in Fig. 3, the network distributions of mean blood flow velocity and wall shear rate versus vessel diameter. The blood flow velocity decreased almost linearly from the large arteriole with a diameter of 108 \( \mu \)m through small arterioles to precapillaries, whereas that in the venular network gradually increased as the vessel size increased. The flow velocity in the venular system was considerably lower than that in the arteriolar system in vessels of the corresponding sizes. The blood velocity in the true capillary vessel with a diameter of 5 \( \mu \)m dropped precipitously because of a larger increase in the total cross-sectional area of parallel capillaries. The shear rate reached a maximum in the precapillary vessels and then fell rapidly to a minimum in the capillary vessels. The lower shear rate in the capillary vessel compared with that in the precapillary vessel was caused by the large decrease in mean blood velocity.

As shown in Fig. 3, the network distributions of mean...
blood velocity and wall shear rate in both the presence and absence of the Fåhraeus-Lindqvist effect were identical. The reason for these identical distributions was that the microcirculatory network of the present model was constructed based on the principle of conservation of flow in the successively branching system. In contrast, the differences in microhemodynamic parameters between with and without the Fåhraeus-Lindqvist effect are described below.

**Figure 4a** shows the arteriovenous distributions of the apparent viscosity of blood within the microcirculatory network with the Fåhraeus-Lindqvist effect and without the Fåhraeus-Lindqvist effect (constant blood viscosity of 4.3 cP) versus vessel diameter. The apparent blood viscosity of 3.7 cP in the proximate arteriole following the central retinal artery with the Fåhraeus-Lindqvist effect reflected a 16% decrease from the blood viscosity of 4.3 cP in a large artery.

Figure 4b shows the arteriovenous distributions of vascular resistance to flow through networks with and without the Fåhraeus-Lindqvist effect versus vessel diameter. The vascular resistance sharply increased from a smaller arteriole with a diameter of 40 μm to the true capillary vessel, regardless of the Fåhraeus-Lindqvist effect. Throughout the network, the vascular resistance to flow with the Fåhraeus-Lindqvist effect was lower than that without the Fåhraeus-Lindqvist effect. The vascular resistance of the venular network was similar to that of the arteriolar network both with and without the Fåhraeus-Lindqvist effect.

Figure 4c shows the arteriovenous distributions of intravascular blood pressure with and without the Fåhraeus-Lindqvist effect versus vessel diameter. The mean blood pressure in the proximal large arteriole of the 1st generation following the central retinal artery was 38.9 mmHg in the presence of the Fåhraeus-Lindqvist effect, and 38.7 mmHg in the absence of the Fåhraeus-Lindqvist effect. With the Fåhraeus-Lindqvist effect, intravascular pressure decreased gradually from the large vessel to small arteriolar vessels and decreased rapidly from the precapillary vessels to the true capillary vessels. Without the Fåhraeus-Lindqvist effect, intravascular pressure decreased rapidly from a smaller arteriole with a diameter of 40 μm to the true capillary compared to that with the Fåhraeus-Lindqvist effect. The difference in blood pressure in the true capillaries between the two conditions was 7.1 mmHg (22.9 mmHg with the Fåhraeus-Lindqvist effect vs. 15.8 mmHg without the Fåhraeus-Lindqvist effect). While intravascular pressure in the venular network decreased much more gradually under both conditions, intravascular pressure with the Fåhraeus-Lindqvist effect was higher by about 9 mmHg than that without the Fåhraeus-Lindqvist effect. Consequently, the difference in blood pressure (the net driving pressure) between the proximal arteriole and the corresponding venule was 21.0 mmHg with the Fåhraeus-Lindqvist effect and 30.6 mmHg without the Fåhraeus-Lindqvist effect. The greater driving pressure required for blood flow without the Fåhraeus-Lindqvist effect was due to large frictional losses encountered in moving the blood across the microcirculatory network.

**Figures 5a** and 5b show the arteriovenous distributions of wall shear stress and circumferential wall stress with and without the Fåhraeus-Lindqvist effect versus vessel diameter. The wall shear stress with the Fåhraeus-Lindqvist effect was almost constant from the large arteriole to small arterioles with a diameter of 60 μm, and thereafter greatly decreased toward the precapillary vessels so that wall shear stress roughly followed the contour of the change in apparent blood viscosity. However, wall shear stress at the precapillary vessels markedly increased associated with a large increase in apparent blood viscosity, since narrow channels impeded the passage of red blood cells through them, according to the inverse Fåhraeus-Lindqvist effect [13]. In contrast, wall shear stress without the Fåhraeus-Lindqvist effect increased from the large arteriole to the precapillaries and decreased from the postcapillaries to the proximal large venule. The arteriovenous distribution of wall shear stress with the Fåhraeus-Lindqvist effect was consistently lower than that without the Fåhraeus-Lindqvist effect, except that wall shear stress in the true capillary vessels was slightly higher due to the inverse Fåhraeus-Lindqvist effect. The changes in wall shear stress in the network without the Fåhraeus-Lindqvist effect directly depended on the changes in wall shear rate.

Overall, circumferential wall stress with the Fåhraeus-Lindqvist effect gradually decreased. However, there were slight increases in circumferential wall stress in the true capillaries and smaller venules. The circumferential wall stress without the Fåhraeus-Lindqvist effect decreased steadily throughout the arteriovenous vascular network. The relatively large decrease in circumferential stress in the true capillary wall was ascribable to the large decrease in intravascular pressure. The circumferential stress in the venous vascular walls without the Fåhraeus-Lindqvist effect decreased to negative levels associated with negative transmural pressure, since the large decreased intravascular pressure was lower than the intraocular pressure; i.e., the negative circumferential wall stress was compressive, which imposed a constraint on the vessel wall from radial dilation.

**Figure 6** shows the arteriovenous distributions of cost functions for individual vessel segments and total vessel segments classified according to the generations of vessels within networks with and without the Fåhraeus-Lindqvist effect versus vessel diameter. The cost function for individual vessel segments, which decreased curvilinearly with a reduction in vessel size, in the arteriolar and venular systems was lower with the Fåhraeus-Lindqvist effect than without the Fåhraeus-Lindqvist effect. Accordingly, the cost function for the total vessel segments was lower with the Fåhraeus-Lindqvist effect, except that the cost function in the true capillary network was slightly larger due to the inverse Fåhraeus-Lindqvist effect, than without the Fåhraeus-Lindqvist effect. Both with and without the Fåhraeus-Lindqvist effect, the large increase in the cost function in the true capillary vessels...
depended on the metabolic factor of vascular volume associated with the large number of capillary vessels. However, the contribution of the mechanical factor to the cost function decreased because of the large decrease in intravascular pressure. In practice, the differences in these cost functions for capillary networks between with and without the Fåhraeus-Lindqvist effect depended on the pressure drop in the vessel segments, since both blood flow and vessel dimension were the same in the two conditions. The total energy costs for the microcircula-

Fig. 4  (a) Comparison of the distribution of the apparent blood viscosity in the microcirculation with the Fåhraeus-Lindqvist effect (solid line) and without the Fåhraeus-Lindqvist effect (dashed line). (b) Comparison of the distribution of vascular resistance in the microcirculation with the Fåhraeus-Lindqvist effect (solid line) and without the Fåhraeus-Lindqvist effect (dashed line). (c) Comparison of the distribution of intravascular blood pressure in the microcirculation with the Fåhraeus-Lindqvist effect (solid line) and without the Fåhraeus-Lindqvist effect (dashed line).
Fig. 5  (a) Comparison of the distribution of wall shear stress in the microcirculation with the Fåhraeus-Lindqvist effect (solid line) and without the Fåhraeus-Lindqvist effect (dashed line). (b) Comparison of the distribution of circumferential wall stress in the microcirculation with the Fåhraeus-Lindqvist effect (solid line) and without the Fåhraeus-Lindqvist effect (dashed line).

Fig. 6  Comparison of the distributions of the cost function (CF) for individual vessel segments and the total vessel segments categorized according to the size of the vessels in the microcirculation with the Fåhraeus-Lindqvist effect (solid line) and without the Fåhraeus-Lindqvist effect (dashed line).
tory networks with and without the Fåhraeus-Lindqvist effect were 9.42 dyn cm $^{-1}$ and 11.86 dyn cm $^{-1}$, respectively: the cost for the microcirculatory network with the Fåhraeus-Lindqvist effect was about 21% lower than that without the Fåhraeus-Lindqvist effect in terms of the same transport of blood. Moreover, the efficiency of blood transport in the microcirculatory network evaluated in terms of the inverse of the mechanical energy cost of the product of the net driving pressure and blood flow with the Fåhraeus-Lindqvist effect was 41% greater than that without the Fåhraeus-Lindqvist effect.

4. Discussion

The present study compared hemodynamic parameters within the human retinal microcirculatory network in the presence of and in the absence of the Fåhraeus-Lindqvist effect. The results showed that the arteriovenous distributions of vascular resistance and wall shear stress were lower and the arteriovenous distributions of blood pressure and circumferential wall stress were higher with the Fåhraeus-Lindqvist effect than without the Fåhraeus-Lindqvist effect. As a result, the total decrease in pressure across the microcirculatory network with the Fåhraeus-Lindqvist effect was lower than that without the Fåhraeus-Lindqvist effect. The mechanical and metabolic cost of blood flow for the microvascular system with the Fåhraeus-Lindqvist effect was 21% lower than that without the Fåhraeus-Lindqvist effect. In particular, when the work of the system in terms of the transport of blood in the microcirculatory network was evaluated as the inverse of $\Delta P/Q$, the efficiency of blood transport with the Fåhraeus-Lindqvist effect was 44% greater than that without the Fåhraeus-Lindqvist effect. Therefore, these findings suggest that the Fåhraeus-Lindqvist effect plays an important role in reducing the physical energy required to transport the blood that flows through the human retinal microcirculatory network.

The biological improvements in the efficiency of the transport of blood and oxygen to the tissues of the organs in the body that depend on the extent to which structural adaptations are brought about in the cardiovascular and respiratory systems evolved via the principle of natural selection [14, 15]. With regard to the workload on the heart in pumping blood against vascular resistance, there are several supplementary mechanisms, such as the pumping action of muscles to aid venous return [16, 17], the spontaneous contractions of lymphatics to propel lymph and tissue fluid centripetally [18], the reciprocal relations between heart rate and stroke volume in the difficulty of cardiac-filling or cardiac-contracting [19, 20], and the optimum relationship between cerebral blood flow and hematocrit [2]. In addition, we quantitatively evaluated the significance of the Fåhraeus-Lindqvist effect, which plays a substantial role in reducing the energy required for blood transport within the microcirculatory network. In in vivo experiments, it is difficult to examine the distributions of hemodynamic parameters in the microcirculation without the Fåhraeus-Lindqvist effect, since variations in the apparent viscosity of blood are inseparable from blood with red cell suspensions that flow into the microvasculature [1]. Hence, the numerical simulation method used in this study is useful for evaluating the virtual phenomenon without the Fåhraeus-Lindqvist effect, since this condition cannot be produced in in vivo experiments.

With regard to the microcirculatory system, intravascular pressure levels in exchange vessels, especially the capillary vessels, are crucially important for maintaining the bi-directional movement of materials, such as nutrients and metabolites, and their balance between the blood vessels and the surrounding tissues [7, 21]. This study demonstrated that, without the Fåhraeus-Lindqvist effect, a large reduction in blood pressure occurred along the length of the vessel segments with diameters ranging from 40 to 7 µm, and the pressure gradient without the Fåhraeus-Lindqvist effect was greater than that with the Fåhraeus-Lindqvist effect. The net driving pressure for blood flow from the 1st-generation arteriolar vessel to the capillary vessels without the Fåhraeus-Lindqvist effect was also greater (by 7.1 mmHg) than that with the Fåhraeus-Lindqvist effect. Let us now consider the autoregulation of the microcirculatory system to maintain an optimal perfusion pressure in the capillary vessels. For example, the perfusion pressure in the proximal retinal arteriole without the Fåhraeus-Lindqvist effect was assumed to be increased by 7.1 mmHg so that the pressure in the capillary vessels resulted in the same pressure of 22.9 mmHg as the control value with the Fåhraeus-Lindqvist effect (Fig. 7a). When the arteriolar network has this pressure increase in the upstream feeding arteriole, circumferential wall stress can be maintained by a decrease in caliber (constriction) and a corresponding increase in the thickness of the vessel wall. The relationships between the lack of a difference in circumferential stress in a constricted vessel wall compared to a control vessel and the corresponding changes in the diameter and thickness of the vessel are presented in Fig 7b. When a control vessel with a diameter of 60 µm is decreased by about 13% and the thickness of the vessel wall is increased by about 13%, the circumferential wall stress in the constricted vessel is at the same level as that in the control before constriction; specifically, these calculations maintain a constant volume of the vascular wall during a change in its geometry. In contrast, in the venular network, since blood pressure without the Fåhraeus-Lindqvist effect was lower than that with the Fåhraeus-Lindqvist effect, the same value of circumferential wall stress as that in a control venular vessel must be achieved by both dilation and thinning of the vessel wall against such a decreased pressure. This dilation of the venular vessels, capacitance vessels, contributes to decreasing venous return and cardiac filling, both of which reduce cardiac output and lower excessively increased perfusion pressures. A series of these hemodynamic changes could be involved in the normalization of intravascular blood pressure in patients with high blood pressure. Accordingly, the mechanism by which circumferential wall stress is involved in the
Fig. 7  (a) Comparison of the distribution of intravascular blood pressure in the microcirculation between the control condition with the Fåhraeus-Lindqvist effect (solid line) and the increased pressure condition without the Fåhraeus-Lindqvist effect (dashed line). The blood pressure in the large arteriole of the 1st generation without the Fåhraeus-Lindqvist effect was increased by 7.1 mmHg, to ensure that the capillary pressure would be equal in the two conditions.

(b) Acute relationship between circumferential wall stress, vessel wall thickness, and vessel diameter in response to an increase in blood pressure of 7.1 mmHg. If the circumferential wall stress in a given vessel segment was kept constant with an increase in blood pressure, the radius and thickness of the vessel segment were decreased and increased, respectively, in the arteriolar system. In the venular system, opposite changes are observed.

(c) Chronic relationship between circumferential wall stress, vessel wall thickness, and vessel diameter in response to an increase in blood pressure of 7.1 mmHg. To chronically maintain the circumferential wall stress at an initial level, the wall thickness of arteriolar vessels was increased if there was no change in vessel diameter. At the same time, the wall thickness of venular vessels was decreased if there was no change in vessel diameter.
meditation of vascular tone in response to changes in intravascular pressure has been proposed to account for, at least in part, regulation of the microcirculation [6, 22 – 25]. Naturally, the influences of metabolic and chemical substances of local origin give the microvascular bed the ability to locally mediate the active control of flow and pressure by flow- and pressure-dependent factors to maintain tissue homeostasis [7, 21].

Under such conditions of acute vasoconstriction, the vascular tone in response to an increase in blood pressure may be mediated through maintenance of a constant circumferential wall stress via a change in the ratio of the internal radius to the wall thickness of the vessel [6, 22, 25] as described above. A luminal cross-sectional reduction during the constriction of arteriolar vessels may be associated with an increase in the velocity of blood flow according to conservation of mass flow. This increased blood velocity in the constricted vessel leads in turn to an increase in the vessel diameter through relaxation of the smooth muscle in the vessel wall following release of flow-dependent vasodilators from activated endothelial cells due to increased wall shear stress [24, 26]. If blood pressure is somewhat decreased by the increase in vessel diameter due to increased shear stress following vasoconstriction, a constant circumferential wall stress may be maintained through an optimal balance between the decrease in pressure and an increase in the radius-thickness ratio or is no longer maintained because of a relatively large increase in the radius-thickness ratio compared with the pressure decrease.

Under chronic conditions, in contrast, structural vascular adaptation, including vessel wall thickening or arteriolar narrowing and nicking [27] and venular dilation [28], which probably occur in polycythemia, hyperviscosity syndrome, diabetes mellitus, and hypertension [29, 30], is associated with both induction and maintenance of chronic high-shear-stress or high-pressure states [23, 31]. Figure 7c shows the chronic adaptation of vascular wall thickness after the vessel diameter is restored to its initial control level following the constriction. When the intravascular pressure still deviates from the baseline level, despite the return of wall shear stress to its initial level, circumferential wall stress around a set point may be achieved only through thickening and thinning of the vessel walls in the arteriolar and venular networks, respectively. In this study, in a vessel with a diameter of 60 μm, circumferential wall stress under an increased pressure of about 6.2 mmHg could be sustained by a 30% increase in the thickness of the vessel wall, provided that the diameter of the vessel remained unchanged via negative feedback mechanisms of the vasculature mediating wall shear stress [32]. Similarly, in the corresponding vessel on the venous side, circumferential wall stress under a pressure decrease of about 2.5 mmHg could be sustained by a 46% decrease in the thickness of the vessel wall with little or no change in the vessel diameter.

Although it is still unclear whether there is a causal relationship between higher pressure and blood viscosity in patients with hypertension or diabetes mellitus, patients with hypertension and diabetic retinopathy show significant arteriolar and capillary narrowing and venular dilation with arteriolar, capillary, and venular wall thickening [28, 31]. Vascular remodeling is also associated with changes in the composition of the vascular wall that is exposed to various factors of physical force and chemical erosion. Therefore, further studies are necessary to determine the relationships between the rheological properties of blood, including hematocrit, red cell aggregation, red cell deformability, and white cell adhesion to vessel walls, physiological and pathological vascular alterations, and microhemodynamic parameters, in various diseases as well as in degrees of their disease states.

In conclusion, the efficiency of blood transport in the microcirculatory network of the human retina was 44% greater with the Fåhraeus-Lindqvist effect than without the Fåhraeus-Lindqvist effect. Thus, the Fåhraeus-Lindqvist effect induced by the non-Newtonian fluid behavior of blood with a red blood cell suspension was effective for reducing in the physical energy required for blood to flow through the microcirculatory network. The wall shear stress and the circumferential wall stress in response to changes in intravascular pressure may be interactively regulated to maintain their individual set-points through alterations in the inner radius and wall thickness of vessels.

References


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