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Reducing blood viscosity with magnetic fields

R. Tao[1,*] and K. Huang[2]
[1]*Department of Physics, Temple University, Philadelphia, Pennsylvania 19122, USA*
[2]*Department of Radiation Oncology, University of Michigan, Ann Arbor, Michigan 48109, USA*
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Blood viscosity is a major factor in heart disease. When blood viscosity increases, it damages blood vessels and increases the risk of heart attacks. Currently, the only method of treatment is to take drugs such as aspirin, which has, however, several unwanted side effects. Here we report our finding that blood viscosity can be reduced with magnetic fields of 1 T or above in the blood flow direction. One magnetic field pulse of 1.3 T lasting ∼1 min can reduce the blood viscosity by 20%-30%. After the exposure, in the absence of magnetic field, the blood viscosity slowly moves up, but takes a couple of hours to return to the original value. The process is repeatable. Reapplying the magnetic field reduces the blood viscosity again. By selecting the magnetic field strength and duration, we can keep the blood viscosity within the normal range. In addition, such viscosity reduction does not affect the red blood cells' normal function. This technology has much potential for physical therapy.

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I. INTRODUCTION

Strokes and heart attacks, the leading causes of death in the industrialized world, happen to people with high or low blood pressure or cholesterol, people who are overweight or thin, smokers, nonsmokers, men, women, and so on. Research indicates that there is one common thread that links all vascular disease: high blood *viscosity*. For example, a recent clinical research from the University of Chicago demonstrates that lupus patients with a history of heart attack or stroke have significantly elevated blood viscosity [1]. Studies have also shown that the more viscous the blood is, the more injurious it is to blood vessels [2,3]. The cells that line damaged arteries and veins will attempt to adapt to the assault and offset the impact by actually building up plaque, a condition called atheroma. A safe and reliable method to reduce high blood viscosity is thus important and may be valuable.

At present the only method to reduce the blood viscosity is to take medicine such as aspirin. Here we report our finding that blood viscosity can be reduced by the use of high magnetic fields of 1 T or above parallel to the blood flow direction. One magnetic field pulse of 1.3 Tesla lasting ∼1 min can reduce the blood viscosity by 20%-30%. The strong magnetic field aggregates red cells along the field direction to form short chains. In addition to increasing polydispersity, etc., for the blood, this change also makes the blood flow similar to a nematic liquid crystal flow with the molecule alignment parallel to the flow direction: The viscosity along the flow direction is significantly reduced.

After the treatment, in the absence of magnetic field, the blood viscosity slowly increases, taking a couple of hours to return to the original value. The process is repeatable. Reapplying the magnetic field reduces the blood viscosity again. Therefore, by selecting magnetic field strength and the duration of field exposure, we can keep the blood viscosity within the normal range. The technology has, in our opinion, much potential for physical therapy.

*Corresponding author: rtao@temple.edu

II. THEORY AND EXPERIMENT

Blood is a liquid suspension of red blood cells (erythrocytes), white blood cells (leukocytes), and platelets in the base liquid, plasma, a complex water solution of gases, salts, proteins, carbohydrates, and lipids. Plasma has low viscosity, $\eta_0 \approx 1.0$ cP at 37 °C. The overall viscosity of whole blood, η, increases as the percentage of cells in the plasma increases, mainly due to the red blood cells. The volume fraction of red blood cells, i.e., the hematocrit, is the main factor affecting the viscosity of blood. With a normal hematocrit of about 40%, the relative viscosity of whole blood to the plasma's viscosity, η/η_0, is slightly above 4. When the hematocrit rises to 53%, which happens in patients with polycythemia, or abnormally high red blood cell counts, the relative viscosity is ∼8 (Fig. 1).

The increase of blood viscosity with the volume fraction of red cells can be explained by Einstein's theory. For a dilute liquid suspension of noninteracting uniform spheres in a base liquid of viscosity η_0, Einsten found the effective viscosity η increased [4]:

$$\eta = \eta_0(1 + 2.5\phi), \tag{1}$$

which is correct when the volume fraction is small, i.e., $\phi < 0.01$.

For high ϕ, we must consider the maximum volume fraction ϕ_m available for adding particles. Following the idea of Mooney [5], let us consider adding $d\phi$ volume fraction of spheres to a liquid suspension of volume fraction ϕ. As the net available volume fraction to add spheres is $1 - \phi/\phi_m$, the increase of viscosity would be

$$d\eta/\eta = 2.5d\phi/(1 - \phi/\phi_m). \tag{2}$$

Integrating this equation gives us an expression to estimate the viscosity at high ϕ:

$$\eta = \eta_0(1 - \phi/\phi_m)^{-2.5\phi_m}. \tag{3}$$

Krieger and Dougherty introduced the intrinsic viscosity $[\eta]$,

$$\eta = \eta_0(1 - \phi/\phi_m)^{-[\eta]\phi_m}, \tag{4}$$

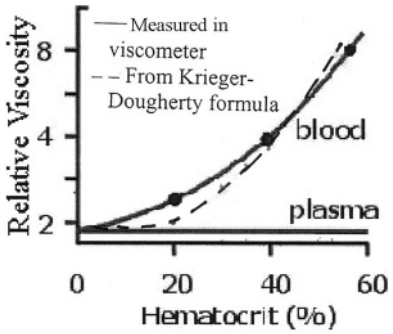

FIG. 1. The relationship between blood viscosity and hematocrit.

enabling us to estimate the viscosity for particles of any shape by choosing a suitable $[\eta]$ and ϕ_m [6–9]. In fact, as shown in Fig. 1, the curve for measured blood viscosity is approximated by Eq. (4) by selecting $\phi_m = 0.72$ and $[\eta]$ 2.3.

When ϕ is unchanged, the most widely used method to reduce viscosity η is to reduce η_0, for example, by raising the temperature or using medicine. Raising the temperature is not applicable to blood in humans. On the other hand, it is clear from Eq. (4) that there is another method: If we change the rheology of the suspension to increase the value of ϕ_m and reduce $[\eta]$, we will reduce the viscosity η. The effective viscosity depends on how much freedom the suspended particles have in the suspension. A high ϕ_m and low $[\eta]$ mean high freedom for the suspended particles, which leads to lower dissipation of energy and lower viscosity.

The following three mechanisms contribute to the viscosity reduction [10]: (1) increasing the average size of suspended particles, (2) aggregating the particles into clusters with a streamlined shape, and (3) increasing the suspension polydispersity.

At a high ϕ, there is substantial evidence with monodispersed suspensions of particles in the nanometer, submicron, or micrometer range showing that at constant ϕ, the viscosity goes down as the spherical particle size increases [11–14]. The explanation is as follows [15]: In a liquid suspension, there is a short-range repulsive force between the suspended particles, which only plays a role when the particles get very close, preventing them from touching each other [16]. Therefore, this short-range force must be considered when the packing becomes closer. Let us denote the repulsive force range as λ. If the maximum volume fraction for random packing spheres

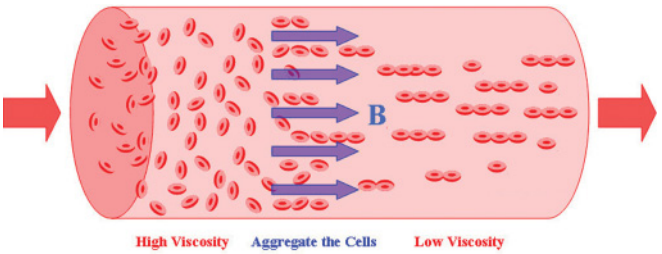

FIG. 2. (Color) Aggregated red-cell clusters have the shape that favors the flow dynamics, leading to further viscosity reduction.

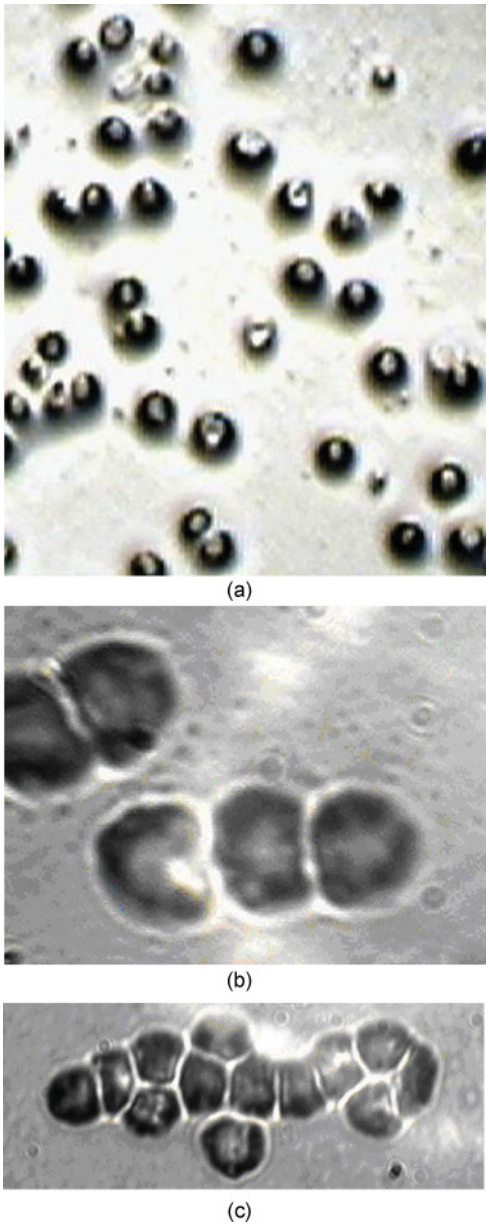

FIG. 3. (Color online) (a) Without magnetic field applied, the red blood cells were randomly distributed in the plasma. (b)After a strong magnetic field of 1.33 T was applied for 1 min, short red-cell chains are formed. (c) After a strong magnetic field of 1.33 T was applied for 12 min, the red blood cells aggregated to form long cluster chains.

of diameter D without repulsive force is ϕ_{m0}, the effective maximum volume fraction for random packing spheres of diameter D with short-range repulsive force is given by

$$\phi_m = \phi_{m0}[D/(D+\lambda)]^3 = \phi_{m0}/(1+\lambda/D)^3. \quad (5)$$

It can now be inferred from Eqs. (4) and (5) that as the diameter D of the particles increases, the effective maximum volume fraction ϕ_m increases and the relative viscosity η/η_0 decreases.

The values of ϕ_m in Eq. (4) also increase with increasing polydispersity [9,10]. This can be understood as

follows: Let us consider a suspension of binary particle-size distribution. For the random close packing, we can allow the big ones to pack first and let the small ones fill the gaps between the large particles. It is clear that ϕ_m for a suspension of binary particle-size distribution is higher than that for a suspension of monodisperse distribution. Following this explanation, we can calculate the viscosity $\eta = \eta_0(1 - \phi_1/\phi_{m1})^{-[\eta_1]\phi_{m1}}(1 - \phi_2/\phi_{m2})^{-[\eta_2]\phi_{m2}}$, which is lower than that of a suspension of uniform small particles with volume fraction $\phi_1 + \phi_2$.

Our method is illustrated in Fig. 2, in which the action of a magnetic field is shown. On the left, the suspension has a high viscosity because of high volume fraction of suspended particles. As the particles flow to the right, passing a region with a strong magnetic field directed parallel to the flow direction, the field polarizes the particles provided that the particles and the base liquid have different magnetic permeability, as is the case of red blood cells and plasma. The induced dipole interaction forces the particles to form short chains or ellipsoids along the field direction. As a result of the induced aggregation, the average size of the suspended particles is increased and the polydispersity is also increased. Moreover, when the applied field is parallel to the flow, the shape of the aggregated short chains or ellipsoids is streamlined, acquiring a reduced $[\eta]$ and the effective viscosity is significantly reduced. This is similar to the flow of nematic liquid crystal with its molecular alignment parallel to the flow direction. Miesowicz found that in such a case, the viscosity is minimized [17]. In fact, our case is even better than nematic liquid crystal because the induced short chains have more polydispersity: They are not uniform; some are longer and some are shorter. This polydispersity further reduces the viscosity.

In the flow through capillary tubes, the viscosity is further reduced because of the tendency for large particles to migrate toward the center of the tubes, where the shear rate is zero [18,19].

It is important to note that application of the magnetic field along the flow direction is important for viscosity reduction. Once the short chains are formed along the field direction, the viscosity is no longer isotropic. Along the chain direction, the viscosity is the minimum, but in the other directions the viscosity is higher.

These three factors all reduce the viscosity significantly. The applied magnetic field is concentrated in a small region, so that the field acts on the flowing particles as a pulsed field. By varying the field strength, we can control the size of aggregated particles, making them optimal for the flow dynamics.

To reduce the blood viscosity, magnetic fields are a logical choice, since the hemoglobin in red blood cell is an iron-containing protein capable of binding oxygen molecules. The molecular structural configuration of the hemoglobin strongly depends on the presence of oxygen. When the hemoglobin contains oxygen it is called oxyhemoglobin, otherwise it is called deoxyhemoglobin. There has been extensive experimental and theoretical research to determine the magnetic properties of red cells. It has been generally accepted that red cells are paramagnetic with a magnetic susceptibility $\chi_r \approx 2.2 \times 10^{-5}$. Therefore a strong magnetic field induces dipolar interaction, which aggregates red cells.

A typical red cell is a disk of $\sim$7.7 μm in diameter and 2.6 μm in thickness. The magnetic permeability of a red cell is μ_r with $\mu_r = \mu_0(1 + \chi_r)$ and the magnetic permeability of the plasma is μ_f. Since plasma mainly contains water, it is diamagnetic, $\mu_f \approx \mu_0(1 + \chi_w)$, where the magnetic susceptibility of water is $\sim\chi_w = -5.4 \times 10^{-6}$.

We can approximate a red cell by an oblate ellipsoid of revolution, $(x^2 + y^2)/a^2 + z^2/c^2 = 1$, with $a = 3.85$ μm and $c = 1.3$ μm. In a magnetic field, the interaction energy between the magnetic field and the red cell is given by

$$U = \frac{a^2 c(\mu_f - \mu_r)H_0^2\{2[\mu_f + (\mu_r - \mu_f)n]\sin^2\theta + [\mu_r + \mu_f + (\mu_f - \mu_r)n]\cos^2\theta\}}{6[\mu_r + \mu_f + (\mu_f - \mu_r)n][\mu_f + (\mu_r - \mu_f)n]}, \tag{6}$$

where θ is the angle between the magnetic field and the red-cell symmetry axis (z axis), which is perpendicular to the disk. The constant n is equal to 0.63214 from the formula $n = (1 + e^2)(e - \tan^{-1} e)/e^3$ with $e = \sqrt{(a/c)^2 - 1} = 2.7876$. The torque on the red cell is

$$N = -\frac{\partial U}{\partial \theta} = \frac{a^2 c(\mu_f - \mu_r)^2 H_0^2(3n - 1)\sin 2\theta}{6[\mu_r + \mu_f, + (\mu_f - \mu_r)n][\mu_f + (\mu_r - \mu_f)n]}. \tag{7}$$

Because $3n - 1 > 0$, it is clear that the stable position is $\theta = \pi/2$: In a strong magnetic field, red cells have their symmetric axis perpendicular to the field.

Let us denote the applied magnetic field as the x direction. The red cell is polarized with the induced dipole moment $\vec{m} = \beta^{(x)} H_0 \vec{e}_x$, where $\beta^{(x)} = \frac{\Omega}{4\pi[\mu_f/(\mu_r - \mu_f) + n^{(x)}]}$, $n^{(x)} = (1 - n)/2 = 0.1839$, and Ω is the volume of the red cell. When the induced dipolar interaction is stronger than the thermal motion, the red cells will align in the field direction to form short chains. From the requirement that the dipolar interaction is stronger than $k_B T$, the applied magnetic field should be on the order of 1 T or above.

The photographs in Fig. 3(a) show that the red blood cell is randomly distributed in the plasma where there is no magnetic field applied. After a field of 1.33 T was applied for 1 min, the red blood cell aggregated to form short chains [Fig. 3(b)]. If the field was very strong and the red cells were allowed to have sufficient time to aggregate, they could form thick chains. As shown in Fig. 3(c), a long and thick chain was obtained after a strong magnetic field of 1.33 T was applied for 12 min.

We note that the red cells do not stack along their symmetry axis as the polarization along this direction is the weakest. Just exactly as suggested by the calculations presented here, the aggregation is always along the disk diameter direction,

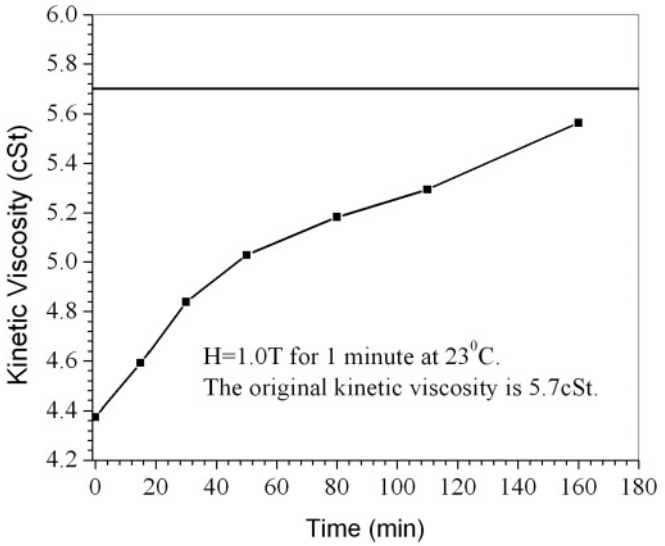

FIG. 4. The kinetic viscosity of a blood sample at 37 °C dropped from 5.7 to 4.35 cS after applying a magnetic field of 1.33 T for 1 min. The viscosity then gradually increased to return to the original rheological state.

i.e., perpendicular to the symmetry axis. Therefore, such aggregation does not affect red cells' normal function—delivering oxygen and removing waste.

III. RESULTS AND DISCUSSION

These theoretical predictions were verified in a series of experiments. In the first experiment, a capillary viscometer was used. We selected this capillary viscometer because the diameter of the measuring tube was ∼1 mm, similar to the size of some blood capillaries. It also made the Reynolds number in our experiment very small, <1.0, allowing for a laminar flow. The temperature in our experiment was maintained at 37 °C. The blood sample originally had a kinetic viscosity of 5.7 cS. After the blood was exposed to a magnetic field of 1.33T parallel to the flow direction for 1 min, the viscosity dropped to 4.37 cS, reduced by 23.3%.

The viscosity then gradually went up to return to the original value as the aggregated short red-cell chains were breaking. After 160 min, it reached 5.56 cS (Fig. 4). Once all aggregations were broken, the blood returned to its original state and the viscosity returned to the original value. If we apply the magnetic field again, we repeat this viscosity reduction process.

In our second experiment, we used a Brookfield viscometer LVDV-II+ and an ultralow viscosity (UL) adapter for the experiment. During the experiment, the sample remained inside the UL adapter and was maintained at 37 °C. This blood sample had a viscosity ∼7.0 cP, higher than the normal value. After application of a magnetic field of 1.33 T for ∼1 min, the viscosity was down to 4.75 cP, reduced by 33%. We noticed that the initial measurement does not have a stable viscosity. This could be related to the fact that the flow was circular, but we could not produce a strong circular magnetic field parallel to the flow direction. Therefore, the aggregated chains were initially not parallel to the flow direction and the measured viscosity was thus not stable. Only after the aggregated chains aligned along the flow direction, the viscosity became stable and slowly went up to recover the original value. After 200 min, the viscosity was 5.4 cP, still considerably lower than the original value (Fig. 5). This clearly indicates that our viscosity reduction method is more effective for blood with higher viscosity.

We note that the reduced viscosity 4.75–5.4 cP is within the normal value of blood viscosity. Our method can effectively bring high blood viscosity down to the normal value if we select a suitably strong magnetic field.

Blood vessels have various diameters. For example, the central blood-carrying canal, i.e., the lumen, varies from 25 mm for the aorta to 5 mm for veins, 4 mm for arteries, and 8 μm for capillaries. From our experimental results in the capillary viscometer, we believe that this viscosity reduction can be beneficial for blood flow in all kinds of blood vessels.

By selecting a suitable magnetic field strength and pulse duration, we will be able to control the size of the aggregated red-cell chains, hence to control the blood's viscosity. While this viscosity reduction only lasts for a couple of hours, the process is repeatable. Once the viscosity returns to its original value or when it goes too high, reapplying the magnetic field brings the viscosity down again. In this way, this method of magnetorheology provides an effective way to control the blood viscosity within a selected range.

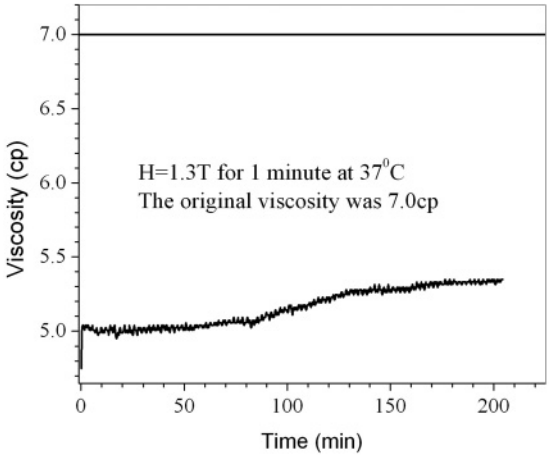

FIG. 5. The viscosity of a blood sample at 37 °C dropped from 7.0 to 4.75 cP after applying a magnetic field of 1.33 T for 1 min. The viscosity then gradually increased to return to the original rheological state.

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