Measurement of blood viscosity using mass-detecting sensor

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Abstract

A newly designed mass-detecting capillary viscometer is extended to measure the viscosity of whole blood over a range of shear rates without the use of anticoagulants in a clinical setting. In the present study as proof of principle, a single measurement of liquid-mass variation with time replaces the flow rate and pressure drop measurements that are usually required for the operation of a capillary tube viscometer. Using a load cell and capillary, we measured the change of mass flowing through capillary tube with respect to the time, \( m(t) \), from which viscosity and shear rate were mathematically calculated. For water and adulterated bloods, excellent agreement was found between the results from the mass-detecting capillary viscometer and those from a commercially available rotating viscometer. Also, the mass-detecting capillary viscometer measured the viscosity of unadulterated whole blood without heparin or EDTA. This new method overcomes the drawbacks of conventional viscometers in the measurement of the whole blood viscosity. First, the mass-detecting capillary viscometer can accurately and consistently measure the unadulterated blood viscosity over a range of shear rates in less than 2 min without any anticoagulants. Second, this design provides simplicity (i.e. ease of operation, no moving parts, and disposable) and low cost. © 2002 Elsevier Science B.V. All rights reserved.

Keywords: Blood viscosity; Capillary; Mass detecting technique

Nomenclature

\( h \) \hspace{1cm} \text{height} \\
\( L_C \) \hspace{1cm} \text{capillary length} \\
\( m \) \hspace{1cm} \text{mass} \\
\( \dot{m} \) \hspace{1cm} \text{mass flow rate} \\
\( P \) \hspace{1cm} \text{pressure} \\
\( Q \) \hspace{1cm} \text{volume flow rate} \\
\( t \) \hspace{1cm} \text{time}

\text{Greek symbols}

\( \rho \) \hspace{1cm} \text{density} \\
\( \phi_C \) \hspace{1cm} \text{capillary diameter} \\
\( \phi_F \) \hspace{1cm} \text{falling tube diameter} \\
\( \eta \) \hspace{1cm} \text{non-Newtonian viscosity} \\
\( \mu \) \hspace{1cm} \text{Newtonian viscosity} \\
\( \dot{\gamma} \) \hspace{1cm} \text{shear rate} \\
\( \tau \) \hspace{1cm} \text{shear stress}

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1. Introduction

It is commonly known that blood viscosity plays a critical role in determining the work of heart that is exerted on the vascular system (Dintenfass, 1969; Kensey and Cho, 1992; Fossum et al., 1997). Intuitively, it is easy to understand that as blood becomes more viscous (or sticky) the amount of work needed to pump it through the arterial system should increase. Since whole blood viscosity is directly correlated to the work of the heart, it is of importance to measure whole blood viscosity like blood pressure. At present, blood pressure is measured routinely; however blood viscosity is not measured and its importance is not understood or appreciated.

Although there are many methods and instruments for measuring viscosity, most current technology, while useful in a research setting, is not optimal for day-to-day clinical use (Chandler and Schmer, 1986). Furthermore, most existing viscometers produce viscosity measurement one shear rate at a time. In order to measure blood viscosity that is shear-dependent, one needs to repeat the measurement over a range of shear rates by varying either rotating speed or driving pressure, which leads to a time-consuming process. At present, due to the time-consuming processes most blood viscosity measurements require anticoagulants in blood to prevent blood coagulation, which result in including the effects of anticoagulants that may increase or decrease blood viscosity depending on the types of anticoagulants (Singh and Coulter, 1973; Reinhardt et al., 1990). Hence, these methods are not suited for viscosity measurement of the unadulterated blood, which should be completed within a few minutes. Therefore, there has been a need to develop a viscometer, which can measure the viscosity of unadulterated blood. The objective of this paper is to develop a new viscometer to measure the viscosity of unadulterated whole blood without heparin and anticoagulants.

Recently, Kim et al. (2000a,b) introduced a new scanning-capillary-tube viscometer that uses a charge-coupled-device sensor array to measure the viscosity of unadulterated whole blood using CCD sensor array. This scanning capillary tube viscometer can produce viscosity data in a low shear range by extending the shear rate range as low as 1 s⁻¹ for unadulterated human blood at body temperature. However, this method requires a non-transparent test fluid due to its optic measurement, which is not applicable to transparent fluids such as serum and plasma.

Meanwhile, Shin et al. (2000) introduced a new mass-detecting capillary viscometer that uses a load cell to measure the viscosity of Newtonian and non-Newtonian fluid. This mass-detecting capillary viscometer can produce viscosity data in a low shear range by extending the shear rate range as low as 1 s⁻¹ for water within a minute.

In the mass-detecting capillary viscometer, rather than the pressure drop and flow rate measurements that are usually required in capillary-tube viscometry, a single measurement of liquid-mass variation with time provides the ability to measure fluid viscosity continuously over a range of shear rates. Throughout the development of this technique an emphasis has been placed on the simplicity of design that would result in easy and quick operation, which may be useful for day-to-day clinical use. In order to demonstrate the validity of this new mass-detecting capillary viscometer, the viscosity data were compared with data obtained
from a rotating viscometer. Also, the accuracy of the new instrument was demonstrated by measuring the viscosity of water and comparing the results with its reference value.

2. Materials and method

2.1. Instrumentation

Fig. 1 shows a schematic diagram of the mass-detecting capillary viscometer, which consists of falling tube, glass capillary tube, adapter, valve, receptacle, load cell, and computer data acquisition system. The inside diameter of the falling tube was \( \phi_f = 6 \text{ mm} \). The inside diameter and length of the capillary tube were \( \phi_C = 0.82 \) and \( L_C = 90 \text{ mm} \), respectively. The diameter and length of the capillary tube were chosen to ensure that the friction loss in the capillary tube was significantly greater than the loss in other parts of the system (Shin et al., 2000; Kim et al., 2000b).

The capillary diameter was carefully chosen to minimize the Fahraeus–Lindquist effect (Fahraeus and Lindquist, 1931). The Fahraeus–Lindqvist effect is a phenomenon that in tube flow, red blood cells (RBC) tend to migrate toward the center of the tube. The plasma-rich zone next to the tube wall, although very thin, has an important effect on blood viscosity measurement, which results in viscosity decrease with decreasing tube diameter. The wall effect was found to be neglected when the tube diameter is bigger than approximately 800 \( \mu \text{m} \) (Haynes, 1960). In addition, the length of the capillary tube was selected to ensure that the end effects would be negligible (Kim et al., 2000a). Also, the inner diameter and the length of the falling tube were carefully selected to finish one test within 2 min, a condition that is desirable in measuring the viscosity of unadulterated blood.

The essential feature in the mass-detecting capillary viscometer is the use of a precision mass balance to measure the fluid collected in the receptacle, \( m(t) \), every 0.04 s with a resolution of 0.01 g. The instantaneous fluid weights were recorded in a computer data file through an analog-to-digital data acquisition system with respect to time.

2.2. Operative procedures

Typical tests are conducted as follows: the system is turned on and connected to a computer. The data acquisition system on the computer is enabled. At that point, the experimental test run was initiated with a vein puncture on the patient using a 20-gauge stainless steel needle. The blood from the vein was first directed to the falling tube so that the initial height of the fluid in the falling tube reaches a preset position. Once this condition is achieved the needle is removed from the vein. At time \( t = 0 \), the valve was turned to allow the blood to flow through the capillary and to be collected in the receptacle as driven by the gravity head. When the fluid level in the falling tube approached datum, the test fluid stopped flowing.

Typically, it took approximately a minute for a fluid level in the falling tube to reach an asymptote for blood in the present study. In fact, the time to complete a run should vary depending on types of liquid and dimension of capillary tube. It is worth noting that the initial height was chosen to produce the maximum shear rate of approximately 400 s\(^{-1}\). If it is required to expand to higher shear rate, an initial height can easily be increased. Fig. 2 shows the fluid mass variations as function of time with water, a Newtonian fluid.

For the purpose of calibration, first, the present study used the mass detecting capillary viscometer to measure the viscosity of distilled water, which had a standard viscosity of 1.1 cP at 19 °C. Second, the viscosity of the adulterated blood was also measured and compared with using a rotating viscometer (Phipps model UDS-200) at 37 °C. Blood samples were obtained from two donors. These donors were healthy and not taking any medication. With the donor in a seated position and a cuff applied to the upper arm, blood was drawn a commercial EDTA tube (BD medical Systems).

2.3. Data analysis

The fluid mass data from the receptacle were analyzed in the following way to determine the viscosity of a non-Newtonian fluid. The mathematical model of the flow analysis began with the equation of the conservation of energy between the fluid level at the falling tube and datum. In deriving the viscosity relation in the capillary viscometers, the important assumptions are: (1) fully developed, isothermal, laminar flow; (2) no
velocity in the radial and peripheral directions; and (3) no slip at the walls (Macosko, 1994). With the above assumptions, one may write the governing equations as follows (Kim et al., 2000a):

$$\{P_i + \frac{1}{2} \rho V_i^2 + \rho g h_i(t)\}$$

$$- \{P_{\text{datum}} + \frac{1}{2} \rho V_{\text{datum}}^2 + \rho g h_{\text{datum}}\}$$

$$= \Delta P_e(t) + \Delta P_c(t) + \rho g \Delta h_{aw}.$$  \hspace{1cm} (1)

where $P$ is the static pressure, $\rho$ the density of the test fluid, $V$ the fluid velocity, $g$ the acceleration due to gravity, $h$ the fluid level, $\Delta P_c$ the pressure drop across capillary tube, $\Delta P_e$ the pressure drop occurring at entrance and exit of the capillary tube, and the subscripts $i$ and datum are at the falling tube and at datum, respectively. The third term in the right hand side in Eq. (1) represents the residue height at the falling tube at $t = \infty$ due to surface tension effect.

With the present experimental set-up, the pressure drop ($\Delta P_e$) caused by secondary flow patterns or eddies in the entrance and exit of the capillary tube may appear to be significant in a high shear zone. One of the accurate methods for determining $\Delta P_e$ is to make a Bagley plot with at least two shot capillaries of the same diameter (Macosko, 1994). It turned out that the contribution from the second term on the right hand side in Eq. (1) is less than 0.5%; hence this term can be neglected for all practical purpose (Shin et al., 2000).

Since $P_i = P_{\text{datum}}$ (static ambient pressure), $|V_i| = |V_{\text{datum}}|$, and $h_i(t) = h_{\text{datum}}$, the Eq. (1) can be simplified as:

$$\Delta P_e(t) = \rho g [h_i(t) - h_{aw}] = \rho g [h_i - h_{aw} - \Delta h_i(t)].$$  \hspace{1cm} (2)

where $h_i$ is the initial fluid level at $t = 0$, $h_{aw}$ is the final fluid level at $t = \infty$, and $\Delta h_i(t)$ is the fluid level difference between $h_i$ and $h(t)$. In addition, the Eq. (2) can be expressed as function of fluid mass collected in the receptacle as follows.

$$\Delta P_e(t) = \frac{4g}{\pi \phi_f^2} [m_{aw} - m_i - m(t)].$$  \hspace{1cm} (3)

where $\phi_f$ is falling tube diameter, $m_i$ the fluid mass at $t = 0$, and $m_{aw}$ the fluid mass at $t = \infty$.

It is of note that the volume flow rate is proportional to the rate of change of the mass of the fluid collected on the load cell. Hence, the corresponding flow rate in the capillary tube can be expressed as:

$$Q(t) = \frac{1}{\rho} \frac{dm}{dt}.$$  \hspace{1cm} (4)

The shear rate dependent viscosity for a non-Newtonian fluid flowing in the capillary tube is obtained from experimental data with some mathematical treatment, and the necessary equations can be found in any standard handbook (Bird et al., 1987). The shear rate at the capillary tube wall is obtained from the classical Weissenberg–Rabinowitsch equation (Macosko, 1994),

$$\dot{\gamma}_w(t) = \frac{1}{4} \frac{\dot{\gamma}_w}{\gamma_w} \left[3 + \frac{d \ln Q}{d \ln \tau_w}\right].$$  \hspace{1cm} (5)

where $\dot{\gamma}_w$ is the apparent or Newtonian shear rate at the wall.

$$\dot{\gamma}_w(t) = \frac{32\dot{m}(t)}{\rho \pi \phi_f^4}.$$  \hspace{1cm} (6)

The shear stress at the wall is given by,

$$\tau_w(t) = \frac{\Delta P_c \phi_c}{4L_c} = \frac{g \phi_c}{\pi L_c \phi_f^2} [m_{aw} - m_i - m(t)],$$  \hspace{1cm} (7)

where $\phi_c$ and $\phi_f$ are the diameter of capillary and falling tube, respectively, $L_c$ the length of the capillary tube, $\dot{m}$ the mass flow rate, and the subscript $i$ and $\infty$ represent initial and final state conditions, respectively.

Thus, the viscosity corresponding to the wall shear rate is calculated in the form of a generalized Newtonian viscosity:

$$\eta = \frac{\tau_w}{\dot{\gamma}_w} = \frac{\rho g \phi_f^4}{8L_c \phi_c^2} \frac{[m_{aw} - m_i - m(t)]}{\dot{m}(t)} \left(3 + \frac{d \ln Q}{d \ln \tau_w}\right)^{-1},$$  \hspace{1cm} (8)

If there is enough data near the point of interest, it is possible to evaluate the derivative $d \ln Q/d \ln \tau_w = 1/n'$ where $n'$ is simply the exponent of the power law constitutive equation. The typical number of data points in the mass-detecting capillary viscometer is about 5000 over a range of shear rates. Even though the power-law exponent is used in the above equations, this does not limit the capability of the present measurement for power-law fluids. The rigorous approach can still be taken for obtaining a viscosity versus shear rate relationship for any fluid (Macosko, 1994). Thus, Eq. (8) can be described in terms of the mass measured in the mass-detecting capillary viscometer as follows:

$$\eta = \frac{\rho g \phi_f^4}{8L_c \phi_c^2} \frac{[m_{aw} - m_i - m(t)]}{\dot{m}(t)} \left(n' \frac{3n' + 1}{n'}\right).$$  \hspace{1cm} (9)

The viscosity versus shear rate information can be obtained from Eqs. (5)–(9) by measuring the mass of the collected fluid with respect to the time from which the pressure drop and flow rate can be calculated. The values of $\phi_f$ and $L_c$ must be obtained by calibration. Since Eq. (9) is non-linear, the procedure to calculate the shear rate and the corresponding viscosity is not straightforward. One of the approaches to obtain the viscosity from the general equations presented above is to adopt a finite difference technique for differentiation of Eq. (9).

3. Results and discussion

Figs. 2 and 3 provided test results obtained with distilled water at room temperature (approximately 19 °C) with the mass-detecting capillary viscometer. Fig. 2 shows the mass variation of the collected fluid as
a function of time: \( m(t) \). As time passed, the collected fluid mass gradually increased and reached \( m_\infty \) asymptotically. It is worthy to note that the rate of the collected mass decreased with time. This was associated with the pressure differential decreasing caused by the column of fluid falling toward the datum, subsequently resulting in decreasing volume flow rate with respect to time. The viscosity of distilled water was calculated from \( m(t) \) using Eq. (9).

Fig. 3 shows viscosity of the distilled water at room temperature as obtained with the mass-detecting capillary viscometer. Based on our viscosity measurement method, the viscosity of the distilled water was found to be between 1.07 and 1.09 cP at 19 \(^\circ\)C, which was within 0.5\% difference in the whole range of shear rates from the viscosity data in the literature (Lide, 1994) of 1.07 cP at the same temperature.

Fig. 4 shows the viscosity results for the adulterated human blood at 37 \(^\circ\)C. For comparison, the test fluid viscosity was also measured by the rotating type viscometer (Physica-UDS 200). The test results from the mass-detecting capillary viscometer were in very close agreement (less than 2\%) with those from rotating viscometer in a range of shear rate (3–300 \( \text{s}^{-1} \)).

Figs. 5 and 6 shows test results obtained with unadulterated human blood without introducing anticoagulants or EDTA with MDCV. The present study tested the blood viscosity with limited blood donors as a proof of principle. Fig. 5 shows the mass variation of the collected fluid as function of time, \( m(t) \). The trends of the fluid mass variations were very similar to those for water. It is important to note that the test should be completed within 3–4 min. If not, blood may begin to clot. In our measurement, one test run took less than 2 min. Fig. 6 shows the viscosity results for the unadulterated human blood for two different male donors at 37 \(^\circ\)C. The hematocrits of the blood were \( H = 47 \) and 49, respectively. The
viscosity for the case with \( H = 49 \) was consistently greater than that for the case with \( H = 47 \). The difference between the two viscosity data was very small at high shear rates greater than 100 s\(^{-1}\), whereas the difference was significant (i.e. greater than 140\%) at a low shear rate range. In other words, the viscosity measurement at low shear rate easily can tell the difference between the two blood viscosity data even having similar values of hematocrits. Thus, this fact implies that it is important to measure the blood viscosity at low shear rate as low as 1 s\(^{-1}\).

4. Conclusion

This study introduced a new method of measuring blood viscosity without anticoagulant over a range of shear rates continuously from high to low shear rate (as low as 1 s\(^{-1}\)). The feasibility and accuracy of the new viscosity measurement technique have been demonstrated for distilled water and adulterated human blood by comparing the results against an established viscosity measurement technique with a rotating viscometer. Among the advantages of this new viscometer are simplicity (i.e. ease of operation and no moving parts), quick measurement, and the ability to make accurate measurements over a relatively broad shear rate range without using anticoagulants.

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