THE EFFECT OF BLOOD'S PSEUDOPLASTIC RHEOLOGY ON MEASURED TURBULENT STRESS

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Abstract: Blood is often modelled *in vitro* by Newtonian fluids. Local rates of fluid deformation due to turbulence, however, may be influenced by the non-Newtonian character of the blood. The aim of this study was to quantify differences in calculated turbulent stresses measured in the near-field of two fully turbulent jets with the same Reynolds number and geometry. The first jet was created with a porcine blood mixture, while the second was created with a glycerine mixture identical in apparent viscosity to the blood mixture at a shear rate of 614 s^{-1} and 25°C. Pulsed Doppler ultrasound was used to estimate turbulent stresses from 1 to 9 orifice diameters downstream. Considerably larger turbulent stresses were measured in the blood mixture jet than in the glycerine jet 1 to 5 diameters from the orifice. Local maxima in turbulent stress were observed in the blood jet 4 diameters from the orifice, but not in the glycerine jet. This work suggests that turbulent stresses measured in blood analogues in the near field of a prosthetic heart valves are different, but within a factor of two of the same stresses measured in blood.

Introduction

Devices placed within the vascular system, such as prosthetic heart valves, can alter normal vascular flow. These alterations sometimes result in strong turbulence, which is not usually present within the vasculature. If strong enough, turbulence creates stresses that can damage or destroy the cells, causing hemolysis or activation of the coagulation cascade. For this reason, devices designed to be placed within the vasculature are routinely tested for the magnitude of turbulent stresses that they create.

In some laboratories, such tests are performed *in vivo*, using animals or humans as models. In others, they are typically performed *in vitro* (in glass or plastic models of the local vasculature) using Newtonian fluids that simulate the viscosity of blood at high deformation rates. Blood, however, acts as a pseudoplastic, demonstrating shear thinning behaviour with a possible yield stress. At deformation rates above 100 s^{-1}, blood is reported to exhibit a nearly linear relationship between the shear stress applied to it and its rate of deformation [1,2]. The rate of bulk deformation blood experiences while moving over prosthetic devices usually occurs at high deformation rates. For this reason, and because non-Newtonian fluids with exactly the same response as blood to an applied stress are difficult to create, blood is modelled by Newtonian fluids with a viscosity similar to that exhibited by blood at high deformation rates.

Local rates of deformations due to turbulence, however, may be influenced by the non-Newtonian character of the blood. Turbulent shear stresses estimated from *in vivo* ultrasound measurements near prosthetic heart valves in our laboratories [3,4,5] are consistently less than turbulent stresses from *in vitro* laser Doppler velocimetry measurements performed in other laboratories. A study of smooth pipe flow using magnetic resonance imaging has shown that blood transitions to turbulence at higher Reynolds numbers than a commonly used blood analogue, a glycerine/water mixture [6]. A theoretical and numerical study has presented evidence that blood flow differs substantially from Newtonian fluid flow at low Reynolds numbers in pipes, but exhibits nearly identical pressure drops and wall shear stresses to Newtonian flow in fully developed turbulent flow past a model of a ball-and-cage prosthesis [2]. This is contradicted by reports that the non-Newtonian nature of blood results in a reduction in the viscous drag coefficient in turbulent pipe flow [7], and that pseudoplastic fluid rheological behavior causes turbulent fluctuations in confined axisymmetric jets to decay much more rapidly than Newtonian fluids under similar flow conditions [8,9]. Mann et al [10] performed the only work that has experimentally investigated the relationship between the non-Newtonian behaviour of blood and turbulence. However, only turbulent intensities were reported from these studies, and it is difficult to discern the effects of non-Newtonian behaviour on turbulent stresses in this work.

This study was performed using a geometry and measurement area identical to that of the turbulent jet experiment of Sallam and Hwang [11], the reasons being that Sallam and Hwang found what appeared to be global maxima in turbulent shear stress within their measurement area, and that the geometry of the flow in these experiments was two dimensional. The aim of this study was to test whether calculated turbulent stresses measured in the near-field of a well-defined fully turbulent flow are markedly different if measured in blood than those measured in a fluid that simulates the viscosity of blood at high shear rates. The hypothesis of this study was that the non-Newtonian

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nature of blood has a dampening effect on measured turbulent velocity fluctuations.

Materials and Methods

Porcine blood from six 60 kg Danish Yorkshire pigs was collected in 10 1.3 L vacuum flasks partially filled with neutral citrate anticoagulant (Royal Veterinary School Pharmacy, DK). Blood transfer from the pigs to the vacuum flask was accomplished by injecting a large guage needle attached to the flask via silicone tubing into the ascending aorta. The resulting blood/anticoagulant mixture was thoroughly mixed, filtered to remove any potential large blood clots, and refrigerated until its bulk temperature was 25 ^{o}C. The final density of the blood/anticoagulant mixture was 1.067 g/cm^3. The viscous response of the collected blood was studied at a temperature of 25.0 ^{o}C and shear rates between 96 and 614 s^{-1} using a cone and plate viscometer (LVDV II+, Brookfield Engineering Laboratories, MA, USA). Figure 1 shows the apparent viscosity of blood as a function of shear rate over this range. All blood used in these experiments was collected from pigs used as experimental models in the training of medical students, and this blood was collected just prior to euthanasia. All procedures were conducted according to approval from the Danish Inspectorate of Animal Experimentation.

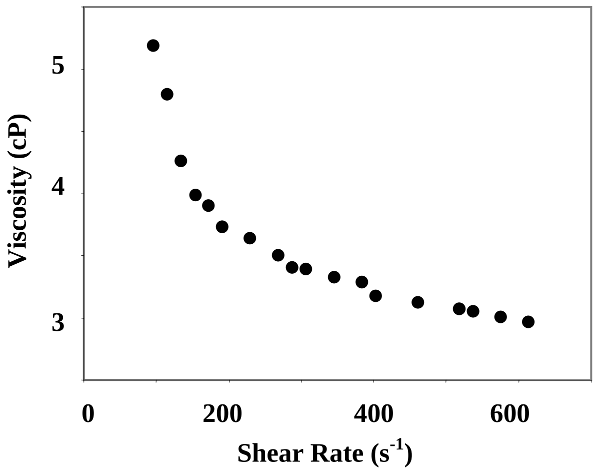

Figure 1: Viscosity of blood-anticoagulant mixture.

After viscosity measurement, the blood was loaded into a flow circuit that created a steady, fully turbulent jet with a Reynolds number of approximately 5700. The jet was created by a steady 109 mm Hg pressure difference across a stainless steel nozzle with an orifice diameter of 3 mm (Figure 2). The flow model was identical to that used by Sallam and Hwang [11], with the exception that the model top was open. Fluid exited the model through the open top.

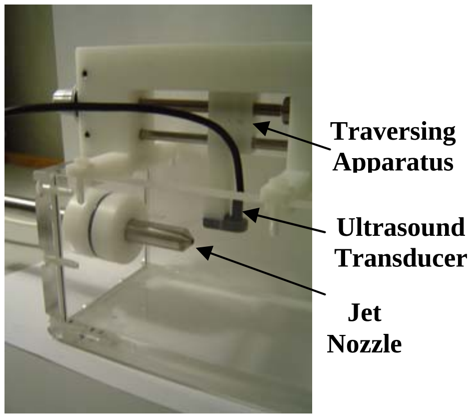

Figure 2: Flow model used in experiments.

Velocity measurements perpendicular to the jet axis were obtained using pulsed Doppler ultrasound at 90 locations along a plane intersecting the nozzle axis. A 10 MHz Doppler ultrasound transducer was operated in the pulsed mode to obtain approximately 10 000 mean velocity measurements at each spatial location. This transducer was connected to a VingMed amplifier (Vingmed, Horten, Norway, Model ALFRED) operated in the pulsed mode. The amplifier quantified velocities from the Doppler signal by a zero-crossing algorithm. Previous tests on the PDU system [12] have quantified the –3 dB cutoff frequency of the ultrasound system at 200 Hz. The sample volume for this apparatus was a cylinder approximately 1.5 mm in diameter and 1.0 mm in length. The amplifier was interfaced with a PCM data recorder (TEAC, Tokyo, Japan, Model RD 180T) for data collection. A manual traversing device was used to move the ultrasound transducer in set spatial increments with an accuracy of $\pm$ 0.1 mm. After data collection, the apparatus was visually inspected to ensure there was no blood clot deposition around the orifice, and subsequently cleaned.

A mixture of glycerin (Bie and Bernsten, DK), water, and corn starch was then constructed to match the viscosity of the blood used in these experiments at high shear rates. Glycerin and water were mixed to give a solution with a viscosity of 2.7 cP at 25.0 ^{o}C. Corn starch, used to provide a signal for the ultrasound transducer used in these experiments, was then added until the kinematic viscosity of the mixture matched that of the collected blood within experimental error at a temperature of 25 ^{o}C and a shear rate of 614 s^{-1}. The final mixture had a density of 1.071 g/cm^3 and an apparent viscosity of 2.94 cP at an applied shear rate of 614 s^{-1}. It should be noted that this mixture was not Newtonian; a measurement of its apparent viscosity at an applied shear rate of 96 s^{-1} and temperature of 25.0 ^{o}C was 3.74 cP. This mixture was loaded into the flow circuit, and the experiment was repeated.

Power spectra of the velocity signals were examined for the presence of low frequency fluctuations indicative of coherent structures, of which none were observed. Turbulent normal stresses were calculated from mean velocity measurements using an ensemble averaging technique:

$$\sigma_{y,PDU} = \frac{\rho \sum_{i=1}^{N} (u_i - u_o)^2}{N} \qquad (1)$$

where ρ represents the density of the medium u_o the arithmetic mean of velocity measurements, u_i an individual velocity measurement, and N the total number of measurements. These stresses were corrected for errors caused by spectral noise and Doppler ambiguity by a previously defined function valid for this experimental setup [13]:

$$\sigma_y = \left(\frac{\sigma_{y,PDU} - 2}{1.3} \right) \qquad (2)$$

Calculated values of turbulent stress were plotted against measurement location to create two dimensional spatial maps of turbulent stress magnitudes. Such plots were thought to be physically relevant for comparing the fluctuating velocity fields of the two jets since the densities of the two fluids were approximately the same.

Results

Figures 3 (a) and (b) show color plots of turbulent shear stress magnitudes as a function of position for experiments with the blood and glycerine mixtures, respectively. Each intersection of the grid lines in this figure represents a spatial location at which measurements were obtained. The highest turbulent stress observed in the jet with blood was 22 N/m^2, while the highest magnitude observed in glycerine was 19 N/m^2. In general, higher turbulent stresses were measured in the blood mixture than in the glycerine mixture. However, the average difference between turbulent stresses measured at the same spatial location was only 1.2 N/m^2.

Figure 3 (a) shows that local maxima in turbulent stress were observed between 9 and 12 mm downstream of the jet orifice. As this was not expected to occur in a turbulent flow, the power spectra of the velocity signals at these locations were examined again for the presence of large low frequency fluctuations, but none were observed. The largest differences between turbulent stresses in the two experiments were found at these local maxima. The differences between turbulent stresses measured at these locations were 9 N/m^2; this magnitude was approximately 50% of the calculated turbulent stress in the blood experiment.

Table 1 shows averages of calculated turbulent stress over cross sections of the jets at various axial distances from the orifice. Averages are not reported for axial distances of 24 and 27 mm from the orifice because at these locations portions of the jets were outside the measurement areas. Table 1 shows that more turbulence is present close to the jet nozzle in the blood mixture jet than in the glycerine mixture jet. At axial distances of 18 and 21 mm (6 and 7 diameters from the orifice), measured turbulence is nearly the same in the two jets.

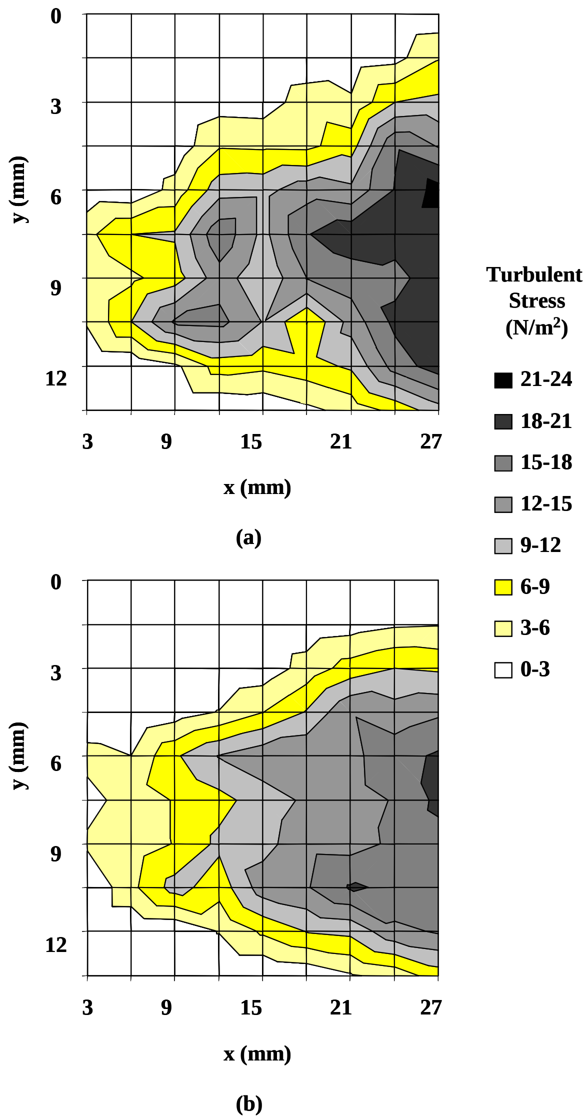

Figure 3: Turbulent stresses measured (a) in the blood jet and (b) in the glycerine jet.

Table 1: Radial averages of calculated turbulent stresses at several downstream locations.

x (mm)	$\sigma_{y,glycerine}$ (N/m^2)	$\sigma_{y,blood}$ (N/m^2)
6	3.2	7.8
9	4.3	9.1
12	7.8	11.8
15	7.2	9.3
18	10.2	10.0
21	11.2	10.9

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Discussion

Previous reports have stated that the apparent viscosity of the blood remains relatively constant above shear rates of 100 s^{-1} [1,2]. While this may hold true for whole human blood at 37°C, it did not for the anticoagulated porcine blood at 25°C used in these experiments. The apparent viscosity of the blood used in these experiments became relatively constant after a shear rate of 400 s^{-1}. The report of a constant viscosity for blood above shear rates of 100 s^{-1} may therefore not be applicable in many *in vitro* experiments, which always use anticoagulants and often use animal blood at low temperatures.

Earlier work has suggested that the non-Newtonian nature of blood either dampens [6,7,8,9] or has no effect [2] on turbulence. Although higher turbulent stresses were found in blood than in the glycerine mixture in this experiment, this does not dispute earlier findings. There are two possible reasons for this. The first is that in this study, measurements were obtained in the developing portion of the turbulent jet; from 1 to 9 orifice diameters downstream. Previous studies of pseudoplastic flow in free turbulent jets did not report this near-field turbulence [8,9]. The closest measurements they obtained were 10 orifice diameters downstream, where they reported that turbulence measurements in Newtonian and pseudoplastic fluids were nearly equivalent. In these studies, turbulence dampening started to occur 20 orifice diameters downstream of the jet [8,9]. In this study, turbulent stresses 6 and 7 orifice diameters downstream were approximately the same for the glycerine and blood mixture jets.

The other possible explanation is that the glycerine mixture used in these experiments became non-Newtonian when corn starch was added. The corn starch particles used in this study were stiffer and considerably larger (an average of 15 μm, determined by examination with a microscope) than blood cells. A particle of corn starch can therefore be expected to dissipate more flow energy than a blood cell, as the rate of viscous dissipation of turbulent energy on a particle can be approximated by:

$$\varepsilon \approx \eta V \overline{s_{ij} s_{ij}} \qquad (3)$$

where η is the volumetric viscous modulus of the particle (larger for stiffer particles), V is the particle volume, and s_{ij} is the fluctuating rate of strain of the flow. From Equation 3, a stiffer, larger particle would dissipate more turbulent energy than a more flexible, smaller particle experiencing the same fluctuating rate of strain. Perhaps more importantly, corn starch particles act on larger scales of turbulence than blood cells due to their large size. Of course, there were likely many more blood cells in the blood mixture than corn starch particles in the glycerine mixture. It is possible, however, that the corn starch caused the slight dampening in turbulent stress observed in the glycerine mixture. Seed particle dynamics may play a role in studies of turbulent flow.

In these early stages, large differences between turbulent stresses in the two jets were found, amounting to around 50% of the calculated turbulent stress in the blood experiment. It appears, therefore, that turbulence measurements obtained in a blood analogue 1 to 5 length scales downstream can only estimate turbulent stresses occurring in blood to within a factor of 2. The reason for the presence of larger early stage turbulent stresses in the blood mixture jet relative to the glycerine mixture jet could be linked to the promotion of turbulent mixing by the red cells. During this stage, erythrocytes perhaps have not had the chance do adapt their shape and align themselves with respect to the flow conditions. Since their membranes are so elastic, red cells can promote this mixing without dissipating turbulent kinetic energy.

These findings do not explain the discrephancies between the turbulence measurements that our laboratory has obtained using pulsed Doppler ultrasound *in vivo* [3,4,5] and those obtained using laser Doppler velocimetry *in vitro*. In our leakage flow study [5], however, measurements were obtained approximately 30 diameters from jet orifices. The previous finding that pseudoplastic flow causes considerable dampening of turbulent stresses starting between 10 and 20 orifice diameters downstream may explain the difference between *in vivo* and *in vitro* results in leakage flow [8,9]. The remaining two studies [3,4] measured forward flow turbulence over a cross-sectional area approximately 1 valve diameter downstream. In these studies, the measurements were made over only 17 measurement locations. Laser Doppler velocimetry measurements have shown that turbulent stresses close to a prosthetic heart valve during forward flow are highly localized to the velocity gradients along the edges of the jets [14]. Because a sparse number of measurement locations were used in the *in vivo* forward flow studies, turbulence within these velocity gradients may not have been measured.

Of course, the differences between *in vitro* and *in vivo* results of turbulence near prosthetic heart valves could be due to differences between the measurement and analysis techniques of pulsed Doppler ultrasound and laser Doppler velocimetry. Pulsed Doppler ultrasound creates a larger sample volume than laser Doppler velocimetry. Turbulent stresses determined from pulsed Doppler ultrasound are calculated from the time average of the mean or maximum velocity of particles passing through the sample volume at a particular instant, whereas turbulent stresses determined from laser Doppler velocimetry measurements are calculated from the time average of single particles passing through the sample volume. It should be mentioned, however, that the ability of the pulsed Doppler ultrasound system used in these studies to determine turbulent stresses has been validated in forward flow of prosthetic heart valves using hot wire anemometry, a system that has a much smaller sample volume [12]. In addition, a previous study using only pulsed Doppler ultrasound has reported differences in turbulent intensities measured in blood and glycerine mixtures near prosthetic heart valves in an artificial heart device [10].

Conclusions

The hypothesis of this work, which was that the non-Newtonian nature of blood has a dampening effect on measured turbulent velocity fluctuations, was shown false in the developing regions of the turbulent jet studied in this experiment. This hypothesis could be true in flow regions far from the jet orifice, however, which were not covered in this study. This is suggested by previous studies of turbulence far from the orifice in pseudoplastic jets [8,9].

Acknowledgements

The author wishes to thank Salvatore Rotella and Alan Monrad for their work in creating the flow model used in these studies, and Benny Sørensen for his advice in the use of anticoagulants. The author is also very appreciative of Morten Smerup and Tanja Thomsen for their suggestions and work in collecting porcine blood, a very messy job.

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