TURBULENT MICROSCALE: THE KEY PARAMETER TO EVALUATE MECHANICAL BLOOD TRAUMA IN TINY ROTARY BLOOD PUMPS.

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ABSTRACT -- The implantation of axial flow pump in the chest cavity is a promising new trend on supporting the failing heart. However, the compromise between the external device dimensions, the levels of induced shear stresses on blood cells and the energy consumption must be properly tuned to obtain a reliable *in vivo* operation. Regarding mechanical blood trauma, the major challenge in designing tiny pumps is to find an attractive compromise between the need for a finite clearance between the rotor tip and the housing and the relative importance of such gap as a source of leakage flow. Turbulent energy cascade concepts were used to give some highlights underlying the sources of hemolysis in tiny rotary blood pumps. The link between viscous dissipation at the cellular scale and the measurable turbulent quantities, as well as its potential use for design improvement of axial flow pumps were also discussed.

Keywords: Hemolysis prediction, Implantable device, Axial flow pump, Turbulent microscale, Reynolds stresses

INTRODUCTION

In the last years rotary blood pumps have been recognized as an alternative to pulsatile devices in the long term heart assistance (Nosè, 1995). One of the greatest challenges to be faced in order to produce a totally implantable rotary blood pump is the reduction of its dimension to fit in the chest cavity. Different design concepts have been employed to minimize the hemolysis which inevitably occurs when blood flows through artificial devices and recent developments revealed that it is possible to produce reliable devices (Buttler *et al.*, 1994; Jarvik, 1995).

The compromise between the external device dimensions, the levels of induced shear stresses on blood cells and the energy consumption must be tuned to obtain a reliable device operation. Device miniaturization demands for an efficient energy converter, hydrodynamically optimized impeller and low hemolytic pumping action. The major problem in reducing the pump dimensions is the inevitable increasing of the flow velocity. The smaller cross sectional area and the smaller amount of fluid volume pumped at each rotor revolution will require an increase of the rotational speed of the rotor to generate physiological flow. As consequence there will be an increasing of the velocity gradients near the rotating surfaces and the walls, exposing the sensitive blood cells to shear stress thresholds for lysis. Efforts have been made to design improvement of the existing devices as well as new designing concepts have been devised to produce new pumps. However,

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additional efforts should be made to determine the cause of some problems already hidden among successes and failures of implantable devices: Is there a limitation for reducing dimensions of rotary blood pumps? How to achieve a proper compromise between small device dimensions and induced shear stress on blood ?

The aim of the present report is to give some highlights underlying the sources of hemolysis in tiny rotary blood pumps using basic fluid dynamics concepts.

TURBULENT FLOW AND BLOOD CELL TRAUMA

The behavior of red blood cells (RBC) under different levels of shear rate has been reported in the literature in the last decades (Rand, 1964; Blackshear *et al.*, 1966; Forstom, 1969; Sutera and Mehjardi, 1975; Sallam and Hwang, 1984). All these studies agree that, at low rates of shear, red cells behave like solid particles as they are carried along the flow. On the other hand, at sufficiently high shear rates, the cells behave like liquid drops, deforming into ellipsoids while holding a constant orientation aligned to the flow, phenomenon called *streamlining* (Leverett *et al.*, 1972). In such situation the RBC membranes rotate about their own cytoplasm, a non-standard motion known as *tank-treading* (Keller and Skalak, 1982). Figure 1 shows four frames of the computer simulation conducted in our laboratory on shearing and fragmentation of a red blood cell. Frame 1 depicts a erythrocyte in its unloaded state (biconcave geometry). Frames 2 and 3 represent the action of a shear flow on the cell membrane, which cause the cell deformation into a prolate ellipsoidal shape while its long axis become oriented along the flow direction. Frame 4 shows a fragmented red blood cell resulted by the action of excessive shear stress. The equations governing the membrane deformation, the coupling with the Navier Stokes equations as well as their implementation into a computer program will be reported, at a proper time, elsewhere.

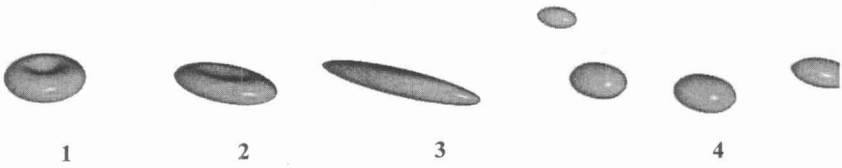

Figure 1 - Deformation and fragmentation of a red blood cell in a shear flow. Frames were taken in different phases of the membrane loading simulation.

Mechanical blood trauma by shear stress may be considered the main cause of hemolysis in artificial devices since there are evidences (Leverett *et al.*, 1972) that neither cell-solid surface interaction nor cell-cell interaction appear to play an important role in the blood cells fragmentation present in shear flows. As stressed by many investigators, the effects of the viscous shear stresses and the exposure time dominate.

The effects of the flow regime (laminar or turbulent) on hemolysis tests in vitro, performed mainly in cone-and-plate and Couette viscometers, apparently did not affect the test results. Experiments of Sutera and Mehjardi (1975) have suggested that hemolysis induced by shearing in a concentric cylinders viscometer at a particular stress was not accelerated by the presence of turbulent fluctuations. Later, although the authors were not convinced by their own explanation about the results (Sutera, 1977), they have concluded that the eddy size of the turbulent flow in the gap between concentric cylinders was too large to produce damage in the red blood cells present in the fluid sample. The basic concept which should be filtered from their conclusions is that the way in which turbulence affects the RBC in a shear flow. It clearly depends on how the turbulent eddies are dissipated in the flow. The phenomenology of such dissipation is known, in the basic turbulence literature (Hinze, 1959), as Energy Cascade.

The basic idea of this theory (experimentally observed in many real situations) is that larger eddies are continuously transferring their energy to smaller eddies through inertial interaction. At the same time viscosity effects and, with them, dissipation become more and more important for the smaller eddies. When the smallest eddies, at the end of the Energy Cascade, are the size of a red blood cell, they can be dissipated through interaction with the cell and transfer their energy to the cell membrane. Such dissipation causes the membrane rupture with the consequent release of hemoglobin into the plasma. On the other hand, if the smallest eddies are greater than the erythrocyte, the cell will be convected away and its membrane will experience only the relative motion in respect to the turbulent fluid.

Kramer (1970) seems to be the first to recognize that the characterization of the red cell trauma due to turbulence may be described by three essential influence parameters, namely the size of the turbulent eddies, the turbulent shearing stresses and the time in which RBC remain in zones of high turbulence. His experiments with perforated disks showed that in the same flow regime, the hemolysis rate may vary at stenoses with the same pressure loss but different geometry.

The dimension of the smallest eddies (Ld) in a turbulent flow (also called turbulent microscale or Kolmogorov length scale) is related to the local large-scale turbulent Reynolds number by the following relation (Kolmogorov, 1941),

$$\frac{L}{L_d} \approx \left(\frac{uL}{\nu}\right)^{3/4} \tag{1}$$

where, L is the characteristic length of the flow (for example, the tube diameter in tube flows, or the gap between the rotor and the housing in axial flow pumps), u is the velocity fluctuation (considering isotropic turbulence) and ν is the kinematic viscosity [m^2/s], ratio between dynamic viscosity μ [kg/ms] and density ρ [kg/m^3], of the fluid.

Since u is directly related to the Reynolds normal stress (σ_n),

$$u = \left(\frac{\sigma_n}{\rho}\right)^{1/2} \tag{2}$$

an expression for Ld may be derived

$$L_d = \frac{kL^{1/4}}{\sigma_n^{3/8}} \tag{3}$$

where $k = \nu^{3/4}\rho^{3/8}$

Using the information arising from the Energy Cascade theory and the Kolmogorov microscale calculations, Baldwin *et al.* (1994) has estimated the smallest eddies lengths for forward and for regurgitant flow (52 μm and 6 μm, respectively) in a prosthetic mechanical heart valve (Björk-Shilley type) installed in the Penn State artificial ventricle. The calculations suggested that the regurgitant flow is likely to be more damaging to blood cells than the forward flow because the turbulent microscale in regurgitant flow is of the order of magnitude of a red blood cell (≈ 8μm). Similar strategy was employed by Jones (1995) to demonstrate that the correlation between Reynolds shear stress and blood cell damage is not direct. It is important to keep in mind that Reynolds Stresses are convective acceleration terms, and the term *shear stress* derives from their effect on the mean velocity profile, not on the local forces (White, 1991). As Jones pointed out, Reynolds stresses arise from the larger turbulent eddies and, as mentioned above in the present paper, the majority of viscous dissipation occurs at smaller length scales. Unfortunately, direct measurements of the latter quantity imposes an excessive spatial resolution for the available instrumentation. On the other hand, a methodology for non-intrusive measurements of Reynolds stresses is established both for laser Doppler anemometer (Woo and Yoganathan, 1986) and pulsed Doppler ultrasound (Hasenkam *et al.*, 1988).

It is not clear yet which turbulent quantities may be associated to the blood cell damage in shear flows, but turbulent microscales emerge naturally as the link between viscous dissipation at the cellular scale and the measurable turbulent quantities. Equations 1,2 and 3 provide the link between the large-scale quantities of the flow (dimension of the turbulence generator, L, a measure of the energy which is dissipated in the Energy Cascade) and the dimension of the smallest eddies. Figure 2 is useful to visualize the relative influence of L and σ_n on the calculation of the Kolmogorov length scale (Equation 3). The range of large-scale dimensions and Reynolds normal stresses in the horizontal and vertical axes, respectively, were taken in the literature. Reports on hemolysis thresholds (Blackshear *et al.*, 1966; Forstom, 1969) as well as information on Reynolds stresses inside artificial devices (Baldwin *et al.*,1994; Jin and Clark, 1993; Pinotti and Paone, 1996) were used in the calculations and were plotted together to give a general picture of the two extremes, from the intentionally induced blood cells damage to the hopefully avoided hemolysis.

The curve which bounds hemolytic and non-hemolytic conditions was obtained by making $L_d = 10$ μm and varying L to calculate the corresponding value of σ_n. Since 10 μm is the approximate the dimension of a stretched red blood cell, such curve gives information whether the dimension of the eddies present in the flow has the potential to cause damage to the RBC. The points displayed in the Figure 2 may lie in two different groups, namely hemolytic and non-hemolytic group. Table 1 displays the smallest eddies dimensions (L_d) obtained for each labelled point plotted in Figure 2. One should note that points with the calculated L_d less than 10 μm appear

in the hemolytic region of Figure 2, on the other hand, points with L_d greater than 10 μm appear under the boundary line, in the non-hemolytic region, as one should expect.

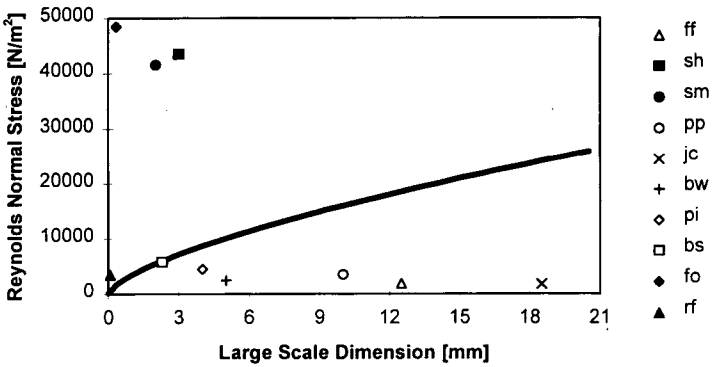

Figure 2 - Large scales quantities (Reynolds Normal Stress and scale of turbulent generator) and their relationship with hemolytic and non-hemolytic flow conditions. Table 1 gives the smallest eddies dimension for the labeled points.

Table 1 - Kolmogorov length scales of the smallest eddies obtained by Equation 3 for the labeled points in Figure 2.

Point Label	Reference	L_d [μm]
fo	Forstom (1969)	1.6
sh	Sallam and Hwang (1984)	4.6
sm	Sutera and Mehrjardi (1975)	4.8
rf	Baldwin *et al.* (1994)	6
bs	Blackshear *et al.* (1966)	11
pi	Pinotti (1996)	16
pp	Pinotti and Paone (1996)	22
bw	Bludszuweit (1995)	23
ff	Baldwin *et al.* (1994)	52
jc	Jin and Clark (1993)	78

Based on the above considerations it is possible to use the information on eddies dimension to map artificial devices, such heart valves and blood pumps, regarding potential blood trauma. It should be noticed that only Reynolds normal stresses were employed in Equation 3 to obtain the Kolmogorov length scales. This is in accordance with Jones' work which has demonstrated that turbulent shear stresses are not in direct conection to red blood cells trauma. Membrane stretching, as shown in Figure 1, is a consequence of the viscous shear exerted on RBC.

SOURCES OF EXCESSIVE TURBULENT STRESSES IN ROTARY BLOOD PUMPS

The space between the housing and the rotor tip is clearly the region with the highest potential of blood cell damage occurring in rotary blood pumps. The relative motion of the blade tip and the stationary wall (housing) induces high levels of shear stresses. Proper especification of the clearance between the blade tip and the housing involves a delicate compromise. In one hand, the pump designer should search for a maximum efficiency of the assembly impeller-housing, which requires minimization of flow leakages. On the other hand, the relative motion of the blade and the housing requires a finite clearance between the blade tip and the end-wall to prevent rubbing. Therefore the major challenge in designing tiny pumps is to find an attractive compromise between the need for a finite clearance (which is limited by constructive difficulties) and the relative importance of such gap as a source of leakage flow.

As pointed out by Yamane *et al.* (1995) axial flow pumps should rotate 40% faster than centrifugal pumps to generate the same pressure head. However, only in impeller centrifugal pumps the major portion of the flow passes through the space between the housing and the rotor tip exposing almost all the flowing blood passing through their internal passages to critical shear stresses. Since it is very difficult to achieve scale reduction of the pump without increasing the shear stresses induced in the blood flow, impeller centrifugal pumps are not suitable for miniaturization. On the other hand, axial flow pumps are suitable for miniaturization, once proper hydrodynamic design is achieved. In such cases, the blades should be designed to impart energy to the blood inducing maximum flow near the axis of rotation. Consequently, pump's efficiency will decrease, but this may be the price for reducing hemolysis in tiny axial pumps. The secondary flows, which are generated in the blade tips, contribute to increase the viscous shear stress level in the main flow and must be minimized. Guide vanes and multi-stage assembly have been studied to minimize the viscous shear stresses and RBC colision with the moving parts imposed by the pumping action of axial pumps (Sieβ *et al.*, 1995; Wakisaka *et al.*, 1995).

DISCUSSION AND CONCLUSIONS

The study of the main causes of mechanical blood trauma and its precise prediction in artificial organs are complex subjects which have received special attention in the last decades. Effects of viscous shear on the cell membrane are also presented in the literature. However, as it is pointed out by Jones, a relationship between viscous shear stresses and measurable turbulent quantities is needed because most of the quantitative research on cellular damage has used cone-plate or Couette viscometers, and the results have been correlated with laminar shear stresses. An examination of the length scales of turbulence seems to provide information on the mechanisms of turbulent-induced cell damage. The analysis reported in the present work was able to classify as

hemolytic and non-hemolytic different situations described in the literature. The link between Reynolds normal stresses and the Kolmogorov length scales of the smallest eddies present in the flow has overcome the limitation imposed by the difficulty to record high-frequency fluctuations at the dissipation scales.

Designing procedures of blood pumps should consider the approach of turbulent scales analysis in order to check potential regions of hemolysis. This is particularly important for tiny axial flow pumps which induce high velocity gradients in the flowing blood. Special experimental techniques should be improved to detect Reynolds stresses with high space resolution, using pulsed Doppler ultrasound (PDU) or a laser Doppler anemometer (LDA). The presence of moving parts inclosed in complex external geometries makes optical instruments the most adequate device to record the turbulent stresses, once the pump is made of transparent material or proper optical windows were provided. LDA has proved to be an useful tool to perform such measurements and refractive index matching technique with a blood analog fluid to access the inner parts of the device has been described in the literature for centrifugal blood pump (Pinotti and Paone, 1996).

Additional work is required to understand the exact interaction between the smallest scale eddies and the RBC, as well as the relationship between the measurable flow occurences and the percentage of hemoglobin release.

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MICRO-ESCALA TURBULENTA: O PARÂMETRO CHAVE PARA AVALIAR O TRAUMA MECÂNICO AO SANGUE EM PEQUENAS BOMBAS ROTATIVAS DE SANGUE

Marcos Pinotti[2]

RESUMO – A implantação de bomba axial na cavidade torácica é uma promisora tendência no suporte à falência cardíaca. Entretanto, o compromisso entre as dimensões externas do dispositivo, os níveis de tensão de cisalhamento induzidos nas células do sangue e o consumo de energia devem ser adequadamente balanceados para se obter uma operação segura *in vivo*. No que diz respeito ao trauma mecânico ao sangue, o maior desafio no projeto de bombas de pequena dimensão é encontrar um compromisso razoável entre a necessidade de um espaçamento finito entre o topo da hélice e a carcaça e a relativa importância deste espaçamento como fonte de escoamento secundário. Conceitos relacionados à cascata de energia turbulenta foram utilizados para destacar as fontes de hemólise em bombas rotativas de sangue de pequena dimensão. A conexão entre dissipação viscosa a nível celular e a medida de tensões turbulentas, assim como seu uso potencial para aperfeiçoamento do projeto da bomba axial, foram discutidos neste trabalho.

Palavras-Chave: Previsão da Hemólise, Dispositivo Implantável, Bomba Axial, Micro-escala turbulenta, Tensões de Reynolds

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