TECHNICAL NOTE

VELOCITY PROFILES IN ANNULAR PERFUSION CHAMBER MEASURED BY LASER-DOPPLER VELOCIMETRY

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Abstract—The fluid flow in the annular perfusion chamber of Baumgartner developed to study platelet vessel wall interaction was examined with laser-Doppler velocimetry. A laminar and stable flow with a Reynolds number of ≤ 50 was measured at flow rates up to 3 ml s^{-1} . No turbulence was found. The wall shear rate directly determined from measured velocity profiles agreed well with theory. The experiments underlined the necessity to work with vessels of uniform thickness and a smooth surface.

INTRODUCTION

The adherence of blood platelets to the subendothelium of the damaged vessel-wall is a critical and early event in the formation of a haemostatic plug or thrombus (Stemerman, 1979). *In vitro* studies of platelet adherence to subendothelium are mostly performed in an annular perfusion chamber developed by Baumgartner (Baumgartner and Haudenschild, 1972; Baumgartner and Muggli, 1976). The flow condition, especially the vessel-wall shear rate, is a critical parameter in these studies (Turrito and Baumgartner, 1979). Until now the vessel-wall shear rate was calculated based on theoretical considerations, assuming a fully developed flow of a Newtonian fluid. The theory of Newtonian fluid flow through cylindrical annulus was described by Bird *et al.* (1960). Figure 1 shows the theoretical velocity profile for such flow.

Local velocity in the annulus is given by

$$V_z(r) = \frac{2Q}{\pi R_0^2} \left[\frac{1 - \left(\frac{r}{R_0}\right)^2 - \frac{1-k^2}{\ln k} \ln\left(\frac{r}{R_0}\right)}{(1-k^4) + \frac{(1-k^2)^2}{\ln k}} \right] \quad (1)$$

where $V_z(r)$ = local velocity (cm s^{-1}), Q = flow rate ($\text{cm}^3 \text{s}^{-1}$), R_0 = radius of the chamber (cm), R_1 = radius of the central rod (cm) and $k = R_1/R_0$.

In practice an artery segment, which is turned inside out, is mounted on the central rod and the shear rate at the vessel wall is calculated from

$$\gamma_w = \left. \frac{dV_z(r)}{dr} \right|_{r=R_1} = \frac{2Q}{\pi R_0^3} \left[\frac{\frac{2k}{1-k^2} + \frac{1}{k \ln k}}{1+k^2 + \frac{1-k^2}{\ln k}} \right] \quad (2)$$

where γ_w = vessel wall shear rate (s^{-1}) and R_1 = radius of central rod including thickness of the vessel. A characteristic parameter, the position of the maximum velocity (R^*) can be calculated from

$$\frac{dV_z(r)}{dr} = 0 \text{ yielding } R^* = R_0 \left[\frac{k^2 - 1}{2 \ln k} \right]^{1/2}. \quad (3)$$

At the position where the artery segment is placed the annulus is narrowed due to the thickness of the artery resulting in larger values for R_1 and k . As a consequence the velocity profile is changed according to the new geometry and the position of the velocity maximum is shifted. However, these theoretical considerations have not been verified by direct measurements. Our objective was to study the flow behaviour in the perfusion chamber in order to obtain answers to the following important questions: Are the measured velocity profiles in agreement with theory? Which entrance length does it take for the flow to adapt to the narrowed annulus at the leading edge of the artery segment? Can vessel wall shear rate be determined accurately from the measured velocity profiles at the region of the artery segment? Is turbulence occurring near the edge of the artery segment. To obtain answers to these questions the technique of laser-Doppler velocimetry was used which permits measurement of velocity profiles in the perfusion chamber with high local resolution and accuracy.

METHODS

The laser doppler velocimetry

The basic principles of the laser doppler velocimetry have been described by Drain (1980). Laser doppler velocimetry, in this case the 'forward scattered method', is based on the detection of light scattered by particles passing the inter-

Fig. 1. Theoretical velocity distribution in a cylindrical annulus. Coordinates r and z refer to Fig. 2 and Table 1.

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ference section of two identical, crossing laser beams. A particle passing the light and dark bands of the interference section, the so called 'fringe pattern', will scatter light with rapid variation of intensity in time. The frequency of this intensity variation is called the Doppler-frequency of the particle and this is linearly proportional to its velocity. This Doppler frequency is determined by a Laser-Doppler frequency tracker from the forward scattered light collected with a photodetector. Local velocity is determined by the relation

$$V = f_d \lambda / 2 \sin \phi \quad (4)$$

where V = local velocity, f_d = doppler frequency, λ = wavelength of light in the suspending medium and ϕ the angle between the laser beams in the suspending medium. The fringe pattern can be used as a non-invasive probe for local velocity measurement. Velocity profiles were measured by scanning the perfusion chamber with the fringe pattern. The position of the fringe pattern in the chamber was determined with a screw micrometer, which had an accuracy of about $5 \mu\text{m}$ precision. The laser Doppler velocimeter was equipped with an He-Ne laser (Spectra Physics, 'Stabilito TM, model 120'), a home build photodetector, Krohn-Hite filters (model 3200 R) and laser Doppler frequency tracker (T.N.O. Physical Technics, The Netherlands, type 1032). This system has been described in more detail elsewhere (Corver *et al.*, 1983). The width of the fringe pattern is about $100 \mu\text{m}$, while the width of the cylindrical annulus is $1.5\text{--}2.0 \text{ mm}$. This allows the measurement of flow profiles from velocities determined at 15–20 independent positions across the annulus.

The perfusion chamber

The perspex outer shell of the chamber was replaced by glass and mounted in a square glass box to avoid artifacts due to refractions of the laser beams at the interfaces. The space between the chamber and the glass box was filled with Ondina 17 and a mixture of geraniol–dieseloil (3:2) was chosen as perfusion fluid. The viscosity ($\sim 6 \text{ cp}$) of this latter mixture is slightly different from Newtonian viscosity of blood (3–5 cp) (Merril, 1969), but this has no influence on the results of these experiments. All the materials are commercially obtainable Shell-products (Shell-Rotterdam, The Netherlands) and they have the same index of refraction. Refractions therefore only occurred at the glass-air interfaces. These glass-air refractions were cancelled out by the parallel geometry of the glass box. As a model for the artery segment a piece of rubber tubing with length and thickness, similar to artery segments used in practice, was mounted on the central rod of the chamber. Measurements were performed with a stationary flow at a flow rate of 3 ml s^{-1} . The inner radius of the chamber (R_0) was 3.66 mm and the radius of the rod, without artery segment, was 1.72 mm .

RESULTS

Velocity profiles were measured along the length of the rubber tubing and at various positions in front of the leading edge and behind the trailing edge, as defined by the direction of the flow. The velocity profiles at various indicated positions are plotted in Fig. 2 and compared with theoretical profiles calculated from equation (1).

Another criterion to test whether the measured velocity profiles comply with the theoretical ones, the measured and calculated position of the maximum velocity were compared. These calculated and measured R^* values are listed in Table 1.

Vessel wall shear rates were determined from the slope of the velocity profiles at the wall surface and also presented in Table 1.

The wall shear rate calculated from equation (2) at an average value of $k = 0.70$ and $R_0 = 3.66 \text{ mm}$ was 1000 s^{-1} . So the choice of a Newtonian fluid as a model for blood is

Fig. 2. Velocity profiles in the perfusion chamber along the artery segment. z -coordinates of the position of the profiles refer to Table 1. Dotted lines: measured velocities; drawn lines: theoretical profiles (equation 1).

correct since blood behaves completely Newtonian above shear rates of about 100 s^{-1} (Merril, 1969). The position of the wall surface of the rubber tubing was found by determination of the position with fd corresponding to $V_z = 0$. The accuracy of this determination was about $10 \mu\text{m}$.

DISCUSSION

The shape of the measured velocity profiles were in good agreement with Fig. 1 and were almost identical to theoretical ones calculated from equation 1. Only at the edges of the artery segment measured profiles differ from theory. In front of the leading edge flow is disturbed. However, at the leading edge the flow is almost immediately adapted to the narrowed annulus so the entrance length under this flow condition is very small. Behind the final edge some disturbance was found and the flow adapted to the original situation after approximately 5 mm . This flow disturbance, however, did not influence the flow conditions at the vessel wall surface. The deviations of the measured position of the maximum velocity from the theoretical values were only small.

These experiments also elucidate that no turbulence occurs at the edges or elsewhere in the part of the chamber that was investigated. The Reynolds number for annular flow, accord-

Table 1. Comparison of the position of velocity maximum and wall shear rate obtained from theory and experiments

z coordinate (cm) of measured position (arbitrary scale)	Position of maximum velocity (mm)		
	R* theory	R* experimental	Wall shear rate (s ⁻¹)
14.00	2.61	2.61	—
13.80	2.61	2.64	—
13.60	2.61	2.90	—
13.50	3.02	3.04	800
13.40	3.00	3.04	700
13.20	2.98	2.98	775
13.00	3.00	2.98	750
12.50	3.09	3.07	1500
12.20	3.07	3.04	1500
12.05	3.15	3.04	2150
12.00	2.61	3.08	—
11.90	2.61	2.94	—
11.80	2.61	2.75	—
11.70	2.61	2.70	—
11.50	2.61	2.61	—

ing to Bird (1960), was ≤ 50 under the chosen experimental conditions. This is far below the critical value of about 2000, the laminar-turbulent transition area for normal pipe flow. In practice, always a piece of 3 mm is cut off the edges of the artery segment to avoid that possible turbulence in these areas influences the platelet adhesion number. From our present results it is evident that any increased platelet adherence at these ends is not caused by turbulence.

Another critical point is the uniform thickness of the vessel wall. Although the rubber tubing was visually smooth when mounted on the rod, Fig. 2 shows that this was not the case. This leads to a large variation of R_1 and k and subsequent variation in wall shear rate along the artery segment as shown in Table 1. In our laboratory we have recently started to use human umbilical artery instead of renal artery. Human umbilical arteries are thinner and have a more uniform thickness than renal arteries. The advantage of a more constant and reproducible wall shear rate is confirmed by the present results. Especially when a perfusion chamber with a smaller diameter is used to obtain high shear rates (up to 3000 s⁻¹), the use of thin vessels with uniform thickness is extremely important.

From these results we may conclude that flow in the annular perfusion chamber is laminar and behaves according to theory, provided thin vessels of uniform thickness are used.

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