Eur. Phys. J. Appl. Phys. **32**, 213–221 (2005)
DOI: 10.1051/epjap:2005083

THE EUROPEAN PHYSICAL JOURNAL APPLIED PHYSICS

Blood viscosity measurement: an integral method using Doppler ultrasonic profiles

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Received: 26 October 2004 / Received in final form: 25 May 2005 / Accepted: 24 June 2005
Published online: 30 November 2005 –

Abstract. The aim of this work is to present a new indirect and noninvasive method for the measurement of the Newtonian blood viscosity. Based on an integral form of the axial Navier-Stokes equation, this method is particularly suited for in vivo investigations using ultrasonic arterial blood velocity profiles. Its main advantage is that it is applicable to periodic as well as non periodic flows. Moreover it does not require classical filtering methods enhancing signal to noise ratio of the physiological signals. This method only requires the knowledge of the velocimetric data measured inside a spatially and temporally optimized zone of the Doppler velocity profiles. The results obtained using numerical simulation as well as in vitro or in vivo experiments prove the effectiveness of the method. It is then well adapted to the clinical environment as a systematic quasi on-line method for the measurement of the blood viscosity.

PACS. 47.15.-x Laminar flows – 47.90.+a Other topics in fluid dynamics (restricted to new topics in section 47) – 83.85.Jn Viscosity measurements

1 Introduction

Thanks to arterial blood flow models which can be used to indirectly determine clinical parameters in haemodynamics, considerable progress has been made towards developing entirely non-invasive methods of investigating vascular function [1,2] using echographic and velocimetric ultrasonic measurements as input data. Among these parameters, the blood viscosity, which is one of the main factors on which the arterial blood flow depends, is of particular importance for at least two reasons. First, it is known to be a physiopathological parameter, since the rheological behaviour of blood can be greatly affected by disease. Secondly, it is necessary to determine the blood viscosity to be able to evaluate two of the most important parameters in clinical haemodynamics, namely the arterial input impedance [3] and the wall blood shear stress [4].

Measuring the blood viscosity directly with a viscometer is a quasi non-invasive method, since it requires only a small sample of blood but at the same time, this method is based on the assumption that the viscosity is the same in both the in vitro and in vivo blood flow. However, very few methods are available so far for the indirect and non-invasive determination of the blood viscosity in situ, and only three of these can be applied in Haemodynamics. In the method developed by Tanguy et al. [5], a least square method is used to resolve a cumbersome system of four unknowns including the Newtonian viscosity. To solve the system, it is necessary to determine instantaneous velocity profiles inside a cross-section of the duct. This method proved to be unsuitable for use with ultrasonic velocimetric data both because of the temporal and spatial derivatives of the velocity profiles involved, and because of the effects of the near wall velocities, which cannot be very accurately measured with classical Doppler ultrasound systems [6]. The second method available for indirectly determining the blood viscosity is that proposed by Sieber [7]. This method involves the identification of rheological constitutive parameters using arbitrary constants and velocity profile data as inputs. Although it is compatible with non Newtonian behaviour, it is valid only in the case of steady flow, which seriously restricts the scope of the method. In addition, the accuracy of this method has not yet been tested using real or realistically simulated physiological signals. The third and last method in question was developed by the present authors [8]. This method based on a linear theory is applicable to pulsed arterial flows in medium-sized and large arteries and involves frequency analysis of the measured Doppler ultrasound velocity profiles. The frequency parameter, $\alpha = \overline{R}\left(\frac{2\pi F_c}{\nu}\right)^{1/2}$ where $\overline{R}$ is the mean radius, ν the kinematic viscosity and f the

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fundamental frequency of the cardiac cycle, is determined at each harmonic associated with the measured velocity profile, the mean radius and the fundamental frequency are both given, and the viscosity is deduced. This method is suitable for use with ultrasonic Doppler velocimetry, as it requires only three optimised velocity measurement locations. That in the centre is the one which always gives the most accurate evaluation of the velocity as a function of time. This method has been tested both in vitro and in vivo, and it has proved to be satisfactory using the second, third and fourth harmonic of the velocity profile. However, this method is valid only in the case of periodic flows, or quasi periodic one as encountered in some in vivo situations.

The aim of the present study was to develop an alternative method for indirectly determining the blood viscosity, which is applicable to both periodic and non periodic flows. In addition, this method, with which it is also necessary to determine the instantaneous velocity profiles, is particularly suitable for dealing with the problems associated with the noise which disturbs the actual physiological signals. This is an integral method, which unlike the derivative methods, improves the signal to noise ratio associated with quantities calculated from the measured velocity profiles.

2 Principle of the method

For the sake of clarity and simplicity, we present in this section the principle of the method in the case of a rectilinear rigid or elastic pipe with small radial wall displacements.

It is well known that the non Newtonian behaviour of the blood is mainly due to thixotropic effects and structural kinetics phenomenon (aggregation and disaggregation). After passing through the aortic valve, the blood can be considered as a fully disaggregated suspension. The aggregation phenomenon then requires a characteristic time of a few seconds to occurs. Due to the value of the mean flow rate, the aggregation phenomenon and, on that account, the non Newtonian behaviour, takes place in the distal arteries only.

For these reasons, we assume a Newtonian homogeneous fluid, as generally admitted in large or medium size arteries. The presence of plasma layer (zone of low red blood cells concentration, near the vessel wall) has no measurable effects in such arteries. With such hypothesis the kinematic viscosity can be assumed to be constant (time and space). Moreover, the non-stationary viscous flow inside the straight cylindrical duct is assumed to be fully developed. This assumption is not restrictive which is well satisfied in numerous arterial sites. The Navier-Stokes equations thus reduce to the axial one, where the nonlinear terms have been dropped. Using cylindrical coordinates (r, θ, z) (axisymetrical flow) and writing $w(r, z, t)$ to denote the axial velocity, the axial Navier-Stokes equation then reduces to:

$$\frac{\partial w}{\partial t} = -\frac{1}{\rho}\frac{\partial p}{\partial z} + \nu\left(\frac{\partial^2 w}{\partial r^2} + \frac{1}{r}\frac{\partial w}{\partial r}\right) = -\frac{1}{\rho}\frac{\partial p}{\partial z} + \frac{\nu}{r}\frac{\partial}{\partial r}\left(r\frac{\partial w}{\partial r}\right) \tag{1}$$

with the boundary radial conditions $w\left(\overline{R}\right) = 0$ and $\frac{\partial w(0)}{\partial r} = 0$.

Assuming the axial velocity profiles to be known, we integrate this axial equation (1) spatially to eliminate the radial derivatives. At each radial position r, we can then write:

$$\int_0^r \frac{1}{r'}\left[\int_0^{r'} r''\frac{\partial w}{\partial t}\mathrm{d}r''\right]\mathrm{d}r' = -\frac{1}{\rho}\frac{\partial p}{\partial z}\frac{r^2}{4} + \nu\left[w(r) - w(0)\right]. \tag{2}$$

Writing this equation (2) for $r = \overline{R}$, we can eliminate the pressure gradient and finally obtain the following expression for the viscosity:

$$\nu_{r,t} = \frac{\overline{R}^2\displaystyle\int_0^r \frac{1}{r'}\left[\int_0^{r'} r''\frac{\partial w}{\partial t}\mathrm{d}r''\right]\mathrm{d}r' - r^2\int_0^R \frac{1}{r'}\left[\int_0^{r'} r''\frac{\partial w}{\partial t}\mathrm{d}r''\right]\mathrm{d}r'}{\overline{R}^2 w(r) + (r^2 - \overline{R}^2)w(0)} = \frac{\mathrm{Num}(r,t)}{\mathrm{Denom}(r,t)} \tag{3}$$

or

$$\nu_{r,t}\,.\,\mathrm{Denom}(r,t) = \mathrm{Num}(r,t). \tag{3'}$$

Although $\nu_{r,t}$ is obviously a constant we have written $_{r,t}$ to express that once the instantaneous axial velocity profiles are known, the constant viscosity can be computed at each time t at any radial position r, except at the centre of the tube and at the wall. The problem now consists of determining the viscosity, which can be optimised, since its value can be computed at any time for a set of chosen values of r. A few comments about this optimisation, relating to the physics of the arterial blood flow, will help to understand how the solution was obtained.

In the case of purely sinusoidal flows, the shape of velocity profiles varies as a function of the frequency parameter α: these profiles are parabolic for small α, flat or plug-like for large values of α, and exhibit complex shapes for intermediary α. In the case of a complex time dependent flow, such as the physiological pulsatile flow we are dealing with here, the velocity profiles are the sum of the profiles associated with each harmonic, including the mean one, which is parabolic. Their shape will also depend on the amplitude of the mean profile relative to the pulsatile one. In all these profiles, viscous effects are predominant in the oscillatory boundary layer, the thickness of which is given by $\delta = \frac{\overline{R}}{\alpha}$. As above mentioned α is defined for purely sinusoidal flows by $\alpha = \overline{R}\left(\frac{2\pi F_c}{\nu}\right)^{1/2}$, and in the general case of non periodic flows, we define $\alpha_g = \overline{R}\left(\frac{1}{\nu T}\right)^{1/2}$ where T is a characteristic time, for instance the characteristic systolic

$$\nu_r = \frac{\overline{R}^2 \int_0^r \frac{1}{r'} \left[\int_0^{r'} r''(w(t_2) - w(t_1)) \mathrm{d}r'' \right] \mathrm{d}r' - r^2 \int_0^R \frac{1}{r'} \left[\int_0^{r'} r''(w(t_2) - w(t_1)) \mathrm{d}r'' \right] \mathrm{d}r'}{\int_{t_1}^{t_2} \left[\overline{R}^2 w(r) + (r^2 - \overline{R}^2) w(0) \right] \mathrm{d}t}$$

time. Therefore, the accuracy of the indirectly determined blood viscosity will be α-dependent. For this reason we propose the following two step method of evaluating the viscosity, using expression (3).

First, we carry out a time integration of (3) over an interval $[t_1, t_2]$, which is chosen to ensure that the signal to noise ratio will be sufficiently high. In practice, this time interval will be deduced from the centre-line velocity data; for example, t_1 could be the first time at which the velocity is greater than half its maximum value (during the systolic acceleration phase) and t_2 the first time at which it drops again below this half-maximum value. After this initial step, the expression of the viscosity can be expressed as:

$$\bar{\nu}_r \int_{t_1}^{t_2} \mathrm{Denom}(r,t)\mathrm{d}t = \int_{t_1}^{t_2} \mathrm{Num}(r,t)\mathrm{d}t$$

$$\Rightarrow \bar{\nu}_r = \frac{\int_{t_1}^{t_2} \mathrm{Num}(r,t)\mathrm{d}t}{\int_{t_1}^{t_2} \mathrm{Denom}(r,t)\mathrm{d}t} \quad (4)$$

or:

See equation above

which get rid of the time derivatives of the velocity.

Secondly, we look for an optimum region $[r_1, r_2]$ where it will be possible to compute with a suitably high signal to noise ratio a mean viscosity given by:

$$\hat{\nu} = \frac{\sum_{r_1}^{r_2} \bar{\nu}_{ri}}{N} \quad (5)$$

where ν_{ri} is the viscosity computed by equation (4) with $r = ri$ and N is the number of radial positions ri between r_1 and r_2.

In practice, this optimum region has to be determined depending on the shape of the measured velocity profiles and, therefore, depending on the value of α or α_g, as we will see in the following section.

3 Numerical simulation

In order to assess the validity of the method, we carried out a numerical simulation using computed velocity profiles obtained with the linear theory of propagation applied to the periodic flow of a Newtonian fluid inside a finite cylindrical viscoelastic duct with small wall deformations. This linear model proved to be valid for most of the medium-sized and large arteries, as long as no complex geometry are involved [9]. The main advantage of this model, in the context of our simulation study, was the existence of analytical solutions which could be used to express the velocity profiles as a function of the centre-line velocity and the α parameter. Assuming $w_{\mathrm{cl}}(t)$ to be the known instantaneous centre-line velocity written in the Fourier space, we then look for solutions to the velocity profiles $w(y,t)$ expressed in the same space using a reduced radial coordinate $y = \frac{r}{R_i}$ where R_i is the instantaneous radius. Therefore, if we write:

$$w_{\mathrm{cl}}(t) = A_0(0) + \sum_1^N \{A_n(0)\cos(\omega_n t) + B_n(0)\sin(\omega_n t)\} \quad (6)$$

$$w(y,t) = A_0(y) + \sum_1^N \{A_n(y)\cos(\omega_n t) + B_n(y)\sin(\omega_n t)\} \quad (7)$$

then the analytical solutions to the velocity profiles $w(y,t)$ [10] can be obtained after solving the linearized Navier-Stokes equations and the equation of continuity at each of rank i harmonic, which yields:

$$A_0(y) = A_0(0)\left(1 - y^2\right) \quad (8a)$$

$$A_n(y) - iB_n(y) = (A_n(0) - iB_n(0)) \times \left\{ \frac{J_0(\beta_n y) - J_0(\beta_n)}{1 - J_0(\beta_n)} \right\}. \quad (8b)$$

In these equations, i is the imaginary part, J_0 is the zero order Bessel function of the first kind, and we have:

$$\beta_n = \alpha n i^{3/2}, \quad \omega_n = 2\pi n F_c. \quad (9)$$

In our simulation study, we first have to consider purely oscillatory flows to study separately the effects of α, and we then will give the results obtained with simulated physiological complex flows with a non-zero mean value. From relations (8a and 8b), one can easily generate a physiological flow by choosing a given centre-line velocity which will be the input of the model. Values of α and λ (index of pulsatility defined as $\lambda = \frac{\max(w_{\mathrm{cl}}(t))}{\mathrm{mean}(w_{\mathrm{cl}}(t))}$) will characterise the simulated site of the arterial system.

On the other hand, however, the method used to indirectly determine the viscosity is an analytical one (Eq. (3)); we can expect it to give an exact value of ν when

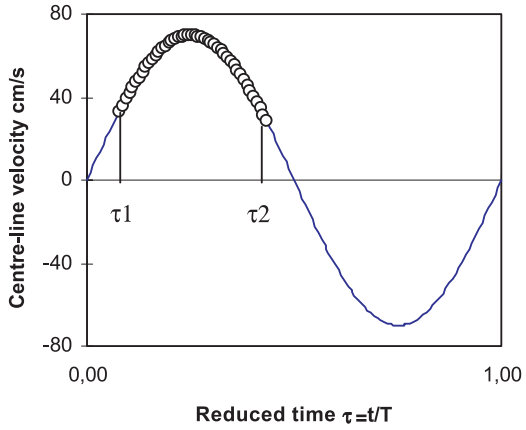

Fig. 1. Exact centre-line velocity of the simulated sinusoidal flow and time interval (∘) used in the integration process.

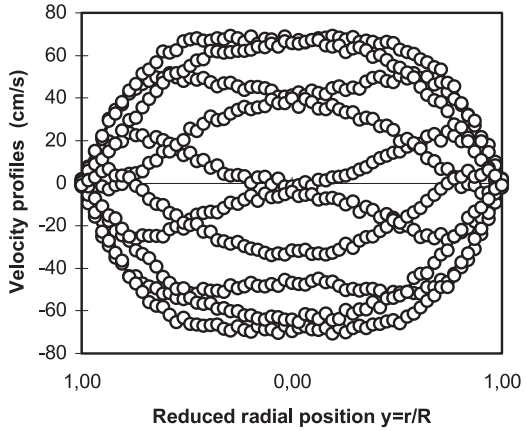

Fig. 2. Computed purely sinusoidal velocity profiles: exact (solid), with randomly added noise (∘). $\alpha = 5$.

dealing with exact computed velocity profiles. In order to simulate the actual problem, it is therefore appropriate to add a Gaussian noise to the velocity profiles. We will then be able to determine the efficiency of the method based on a realistic signal to noise ratio.

3.1 Simulated sinusoidal flows

In the first part of this section, and for sake of clarity, it is proposed to study the influence of α on the efficiency of the method by generating purely sinusoidal flows in the physiological range α ($\alpha \in [2, 20]$), using either the exact solution or a noisy one. For this purpose, we introduced a parameter S_r giving the value of the signal to noise ratio relative to the maximum value of the computed velocities. For each α, we calculated the exact velocity profiles and generated five velocity profiles with randomly added noise, to which the method described above was applied to extract the Newtonian viscosity. As an example, Figure 1 shows the centre-line velocity used to generate a simulated sinusoidal flow for $\alpha = 5$, as well as the time interval used for the time integration of the formula (4). Figure 2 gives the exact velocity profiles calculated from the centre-line velocity in Figure 1, along with the corresponding noisy ones. In this example, as with all the results presented in this study, the level of the randomly added noise was taken to be 3% of the maximum value of the centre-line velocity, as occurs in practice with most Doppler ultrasound systems. Throughout this simulation study, the input viscosity used in the linear model was equal to 4×10^{-6} m^2 s^{-1}, which is the value of the Newtonian viscosity of blood.

The method was applied to both the exact and noisy velocity profiles. In the first part of this section, for every α, we determined the reduced ($y = r/R$) radial positions y_1 and y_2 graphically, after finding the optimum region for calculating the viscosity, using formula (5). This resolution procedure is illustrated in Figure 3, again in the case of $\alpha = 5$.

The various curves in Figure 3 give ν_r as expressed by formula (4): the line consisting of ▲ symbols gives the resolution obtained using the exact velocity profiles, while the

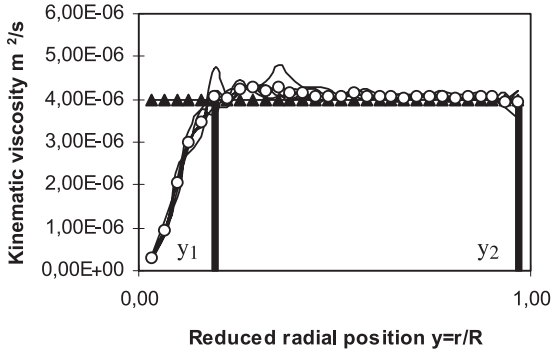

Fig. 3. Indirect determination of the viscosity: (▲) values obtained with exact velocity profiles, — values obtained when adding random noise, (∘) mean value. [y_1, y_2] is the optimum spatial region of resolution for $\alpha = 5$.

five solid lines give the resolution obtained using the noisy velocity profiles. Each of these solid lines is associated with a different random noise, but as mentioned above, the percentage of noise is always the same. The line consisting of ∘ symbols gives the mean curve calculated from the solid lines. As shown in this figure, the resolution obtained using the exact profiles yields the viscosity value introduced into the model (4×10^{-6} m^2 s^{-1}). On the other hand, to obtain the viscosity from the noisy profiles, it was necessary, as was to be expected, to determine an optimum spatial region [y_1, y_2]. In this particular case, taking the mean curve (that consisting of ∘ symbols), we obtain $y_1 = 0.19$ and $y_2 = 0.97$ mm. Lastly, applying formula (5) to this [y_1, y_2] interval, gives the viscosity $\hat{\nu} = 4.09 \times 10^6$ m^2 s^{-1}, which is in fairly good agreement with the input viscosity used in the linear model.

In addition to the results mentioned here in the particular case where $\alpha = 5$, results were also obtained for other values representative of the α physiological α range [2, 20]: these results are summarised in Table 1 for the whole α range studied [2, 12]. In line with the physiological data, we have changed the value of α by changing the mean radius value. With $\alpha \geq 12$ (Fig. 4), the results obtained were practically unchanged and the optimum regions were

Table 1. Optimal spatial region (r_1; r_2 or y_1, y_2) for the indirect determination of the viscosity in function of α and value of the mean viscosity determined inside this region. Case of simulated sinusoidal flows with superimposed noise.

α	2.	3.	4.	5.	6.	7.	8.	9.	10.	11.	12.
$\overline{R}$ mm	1.6	2.39	3.19	3.99	4.79	5.59	6.38	7.18	7.98	8.78	9.57
r_1 mm	0.88	0.85	0.72	0.77	0.77	2.34	3.29	4.17	5.41	6.23	7.1
$y_1 = r_1/R$	0.55	0.35	0.22	0.19	0.16	0.42	0.51	0.58	0.68	0.71	0.74
r_2 mm	1.34	2.24	3.09	3.86	4.63	5.40	6.17	6.95	7.72	8.5	9.26
$y_2 = r_2/R$	0.83	0.93	0.97	0.98	0.97	0.97	0.97	0.97	0.97	0.97	0.97
$\hat{\nu} \times 10^6$ m^2 s^{-1}	3.72	3.95	4.13	4.10	4.18	4.18	4.20	4.26	4.22	4.21	4.40

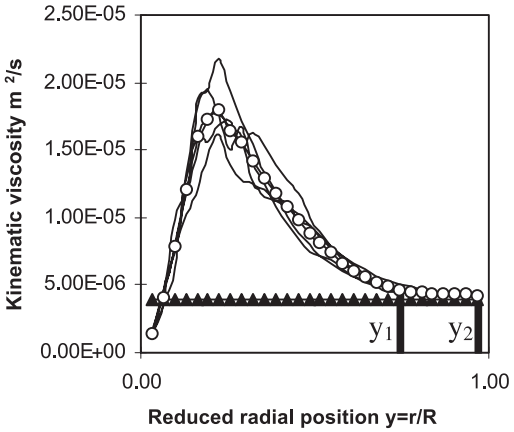

Fig. 4. Indirect determination of the viscosity: (▲) values obtained with exact velocity profiles, — values obtained when adding random noise, (○) mean value. [y_1, y_2] is the optimum spatial region of resolution for $\alpha = 12$.

confined to a near-wall area which can not be used in the model, since it was located almost entirely inside the viscous boundary layer.

On the whole, as shown by Table 1, three main points emerged from all the results obtained. First, when exactly computed velocity profiles are used in the model, the solution was accurate throughout the whole physiological tested range. This confirms that the method proposed is suitable for indirectly determining the Newtonian viscosity, at least at the sites on the flexible duct where the linear propagative theory can be applied.

Secondly, when the noise is taken into account, the spatial range of validity of the method is correlated with the value of α, and therefore, with the shape of the velocity profiles. The method proved satisfactory at intermediate α values, ranging between 3 and 9. At small α (<3) and large ones (>9), the efficiency of the method deteriorated, in terms of the viscosity obtained and the optimum region of resolution. This point reflects some simple hydrodynamic constraints: at low values of α, the velocity profile is quasi parabolic and only the slight change in shape observed can give any information about the viscosity. In the same way, at high values of α, the velocity profile tends to adopt a plug-like shape, and only slight differences in shape (which are masked by the noise) can give the required viscosity information.

Thirdly, for in vivo applications, medium size arteries will be used. In order to obtain a practical method which can be used in vivo, we have to fix a common optimum region for the whole range of intermediary α values. The data in Table 1 suggested that r_1 should be taken to be half the radius ($y_1 = 0.5$), while r_2 can be taken to be the point nearest to the wall ($y_2 = 1$). However, Doppler ultrasonic near-wall velocity measurements are known to be difficult to determine because of the convolution effects, and it is unrealistic to take these near-wall measurements into consideration, unless one implements a sophisticated deconvolution process [11]. We therefore decided to set the radial position $y_2 = 0.9$, assuming that the size of the convolution zone would be less than 10% of the arterial radius.

In the next paragraph, the method developed for indirectly determining the Newtonian viscosity will therefore be applied to some simulated physiological velocity profiles, in the intermediate α range [3,9] systematically using the optimum spatial region defined above ($y_1 = 0.5$, $y_2 = 0.9$).

3.2 Simulated physiological flows

In the second part of this section, we simulated physiological flows with a non-zero mean velocity profile by entering a pulse-like centre-line velocity including about ten harmonics and a non-zero mean value equal to 10% of the maximal value into the model. Figure 5 shows the change in the centre-line velocity resulting from the added noise. At the low diastolic values, the signal to noise ratio is very weak, and this corresponds to an extreme case relative to the real measured Doppler profiles. Figure 6 shows the corresponding velocity profiles in the case where $\alpha = 5$ (α computed by using the fundamental frequency of the signal), as in the previous paragraph.

Using the previously defined optimum region compatible with an automatic resolution procedure, Figure 7 gives the viscosity resolution applied to the velocity profiles of Figure 6, corresponding to $\alpha = 5$ and Table 2 summarises the results obtained. These can be subjected to the same analysis as in the previous section. The results show that the most relevant parameter is always the fundamental α. This, of course, is due to the fact that, as in the case of the real physiological signals, the main components of the simulated velocity profiles are the fundamental

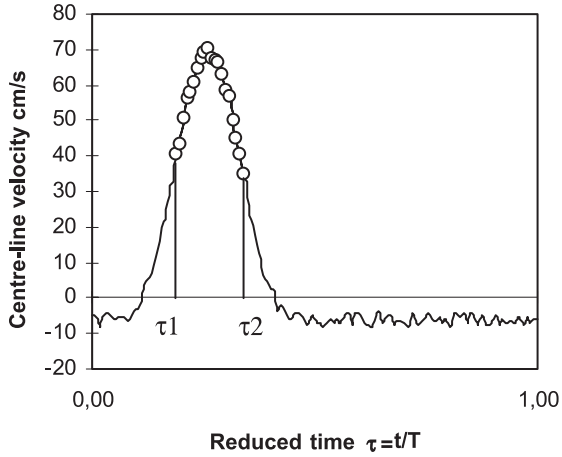

Fig. 5. Centre-line velocity for the simulated physiological flow with randomly added noise. $[\tau_1; \tau_2]$ is the reduced time interval (○) used in the integration process.

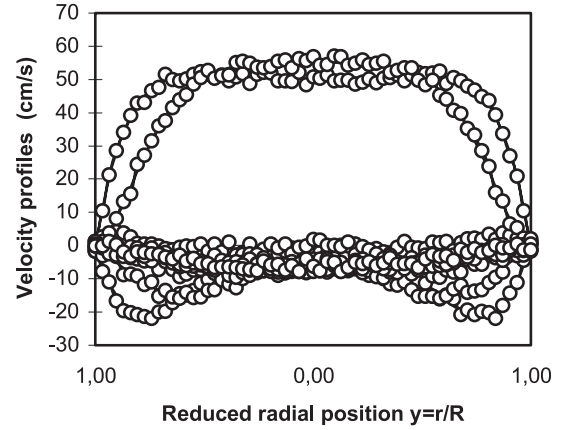

Fig. 6. Computed physiological like velocity profiles: exact (solid), with randomly added noise (○). $\alpha = 5$.

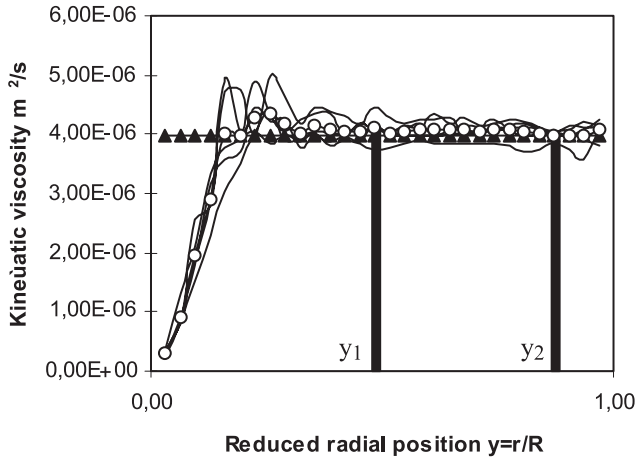

Fig. 7. Indirect determination of the viscosity in the previously fixed optimal spatial region $[y_1, y_2]$ of resolution for simulated physiological flow with a normal physiological value of the mean flow. $\alpha = 5$. (▲) values obtained with exact velocity profiles, —— values obtained when using five different simulated physiological flows with added random noise, (○) mean value.

Table 2. Indirect determination of the viscosity in the previously fixed optimal spatial region in function of α and value of the mean viscosity determined inside this region.

α	3.	4.	5.	6.	7.	8.	9.
$\overline{R}$ mm	2.39	3.19	3.99	4.79	5.59	6.38	7.18
$\hat{\nu} \times 10^6$	4.08	4.05	4.04	4.11	4.09	4.29	4.24

harmonic and the next two components. On the whole, in the given range of intermediate α values and in the given spatial region of resolution, the indirect determination of the viscosity proved to be satisfactory. In the lower α sub-range [3,7], the relative error was always found to be less than 3%, whereas at the upper limits of α, values as high as 10% were recorded.

4 Experimental results

While giving excellent results using numerically simulated flows, ($\Delta\nu/\nu < 3\%$ for $\alpha < 8$) the method described above still required to be validated by applying it to real flows. For this purpose, we carried out an in vitro study using a hydrodynamic set-up as well as applying the method to in vivo Doppler velocity profiles.

4.1 Application in vitro

The experimental circulatory system we used to generate non stationary viscous flows consisted first of a pump chamber with an electronically driven piston, and a latex tube with an interior diameter of 16 mm. This main circuit was closed by a damping system consisting of one resistance and one compliance, resulting in a completely steady flow through a quasi-rigid tube, returning to the pump chamber. The fluid tested was an emulsion of sunflower oil in water including a sufficiently large quantity of polystyrene marbles in order to ensure good echogenicity for the ultrasounds. It was confirmed using a low shear viscometer (Contraves Low Shear 40) that this fluid, like blood, showed Newtonian behaviour at a shear rate $\dot{\gamma}$ greater than 100 s^{-1}. At the Newtonian plateau, a viscosity of 10×10^{-6} $m^2 s^{-1}$ was recorded at the ambient temperature. The flows generated were all sinusoidal with an added mean flow. Instantaneous velocity profiles were measured using a 8 MHz multigate ultrasonic Doppler velocimeter (Dop 1000, Signal Processing, Lausanne, Switzerland). With this apparatus, it is possible to instantaneously measure the whole velocity profile in a section of the tube by simultaneously activating a sufficiently large number of channels (between 56 and 114), with a spatial resolution of 0.3 mm.

The same analysis can be repeated in the whole set of experiments, and in order to show the reproducibility of the method, we will give the results obtained in a series of four independent experiments carried out at the same frequency of 1 Hz and leading to a frequency parameter α equal to 6.34. Because of the presence of an

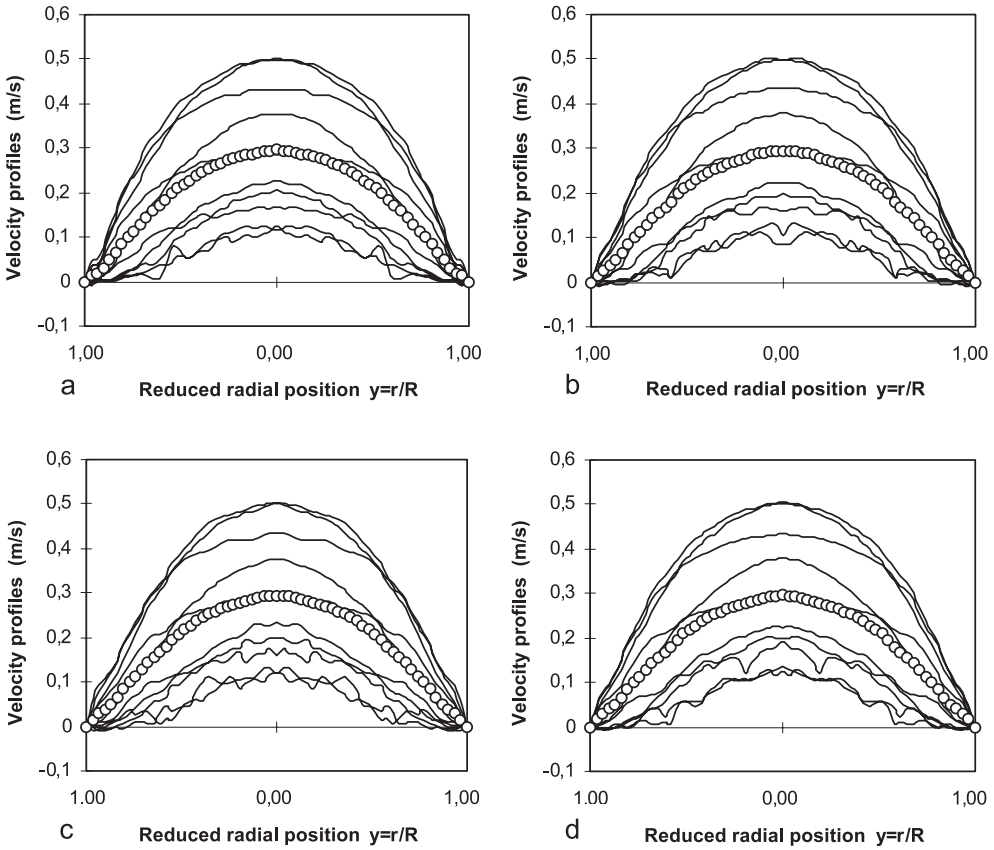

Fig. 8. Instantaneous Doppler velocity profiles (solid) with the deduced mean velocity profile (○) obtained with four different experiments, a, b, c, d.

image profile due to echoes reflected by the distal wall, we used only those measurements obtained on the first half of the tube section. Figures 8a–d give, for example, the instantaneous velocity profiles associated with the four experiments: these were measured at various equally distributed instants during the periodic cycle and have been symmetrized for the sake of the presentation based on the assumption that an axisymmetrical flow was present. The accompanying Figures 9a–d show the results obtained upon applying the above method to the indirect measurement of the fluid viscosity. This method therefore proved to give very satisfactory results, since in comparison with the directly measured viscosity of 10×10^{-6} m^2 s^{-1}, the viscosity obtained was very similar, and in any case, the relative error was never more than 5%.

4.2 Application in vivo

In order to test the feasibility of the present method under clinical conditions, we used a set of previously measured instantaneous velocity profiles. These profiles were measured on human brachial arteries at the Broussais Hospital, Paris, France, using an 8 MHz single channel ultrasonic Doppler velocimeter (Echovar Doppler pulsé, Alvar Electronic, Montreuil, France). Figure 10a shows the velocity profiles measured at various equally distributed instants during the cardiac cycle plus the mean velocity profile. Here again, as shown in Figure 10b, the method proposed gave very satisfactory results: in comparison with the directly measured blood viscosity of 4×10^{-6} m^2 s^{-1} (for $\dot{\gamma} > 100$ s^{-1}), we obtained an indirectly measured viscosity equal to 4.1×10^{-6} m^2 s^{-1}. In this experiment, the frequency parameter α was equal to 3.32, as the mean cardiac frequency was 1.22 Hz and the mean radius, 2.40 mm.

5 Discussion

The method proposed above for indirectly and noninvasively measuring the Newtonian blood viscosity is highly compatible with the hospital environment: it is an in situ method with which it is possible to obtain a systematic, quasi real time measurement of the blood viscosity, based on Doppler ultrasonic velocity profiles. This objective was reached thanks to the presented simulation study, in which we defined an automatic means of determining an optimum spatial measurement zone for computing the viscosity. This simulation study also enabled us to validate the principle of the method and to confirm that it is suitable for the problems encountered in vivo, as the noise was realistically simulated. In this respect, the presented results were obtained without requiring the methods available for

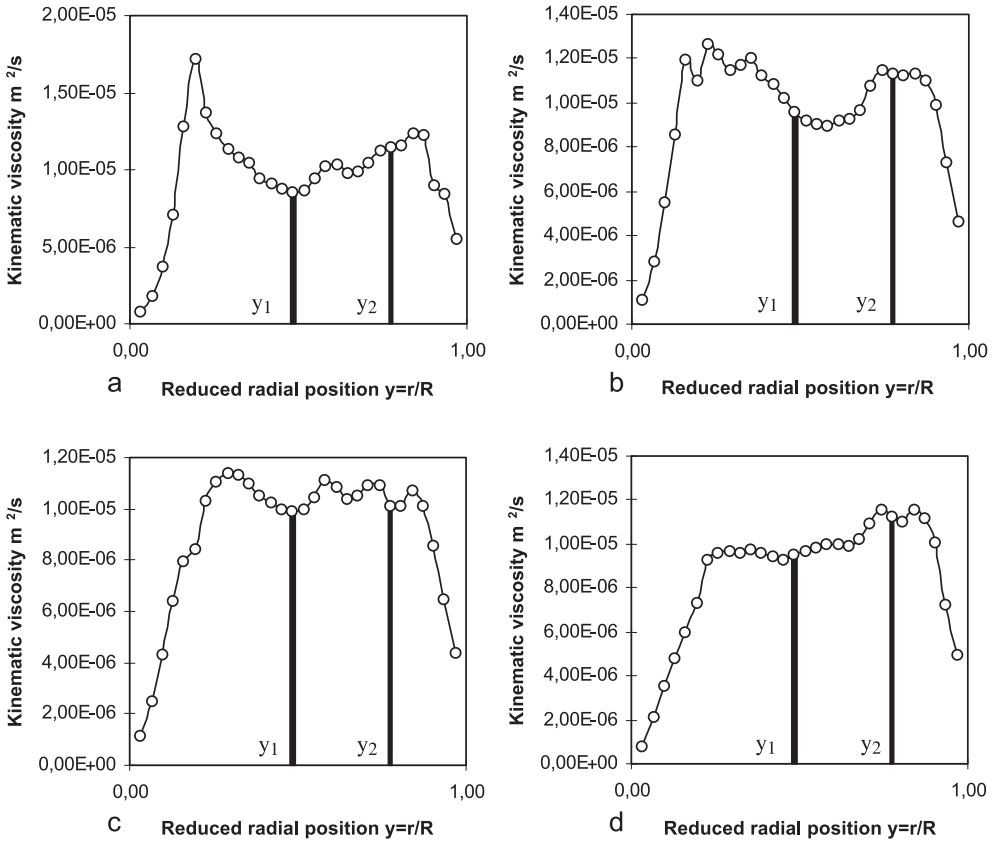

Fig. 9. Indirect measurement of the viscosity in vitro obtained in the previously fixed optimal spatial region $[y_1, y_2]$ with the four experiments a, b, c, d. The computed viscosity is for the experiment a: $\nu = 10.4 \times 10^{-6}$ m^2 s^{-1}; for b: $\nu = 10.0 \times 10^{-6}$ m^2 s^{-1}, for c: $\nu = 10.5 \times 10^{-6}$ m^2 s^{-1}, and for d: $\nu = 10.4 \times 10^{-6}$ m^2 s^{-1}.

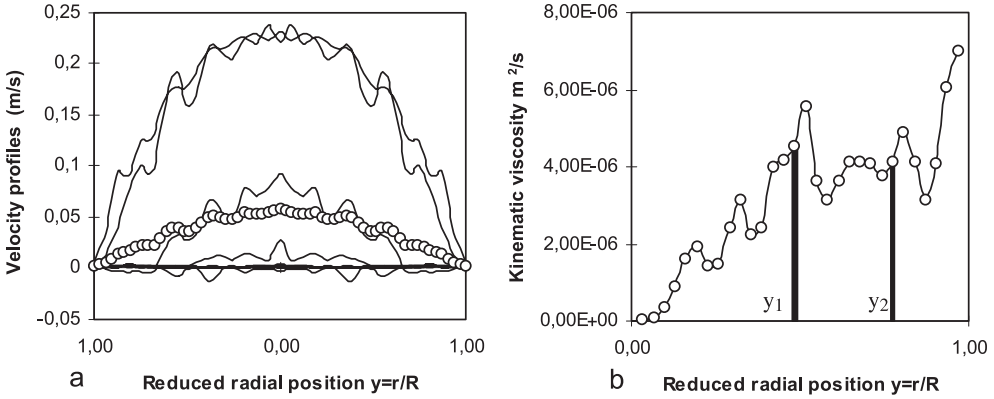

Fig. 10. (a) Instantaneous Doppler velocity profiles (solid) measured in human brachial artery with the corresponding mean velocity profile (○). (b) Associated indirect measurement of the blood viscosity computed in the fixed optimal spatial region $[y_1, y_2]$: $\nu = 4.1 \times 10^{-6}$ m^2 s^{-1}.

enhancing the signal to noise ratio, and this further confirms the efficiency of this integral method.

On the other hand, the hypothesis that we were dealing with a rectilinear duct and fully developed flow is not very restrictive, at numerous sites in the arterial system. In the same way, Newtonian like behaviour does in fact occur in the large arteries, since blood is thixotropic and that high shear rates occur in the heart, which leads to a high rate of blood disaggregation. In large arteries the aggregation dynamics has no time enough to occur and to lead to a modified viscosity (the required characteristic times being a few seconds).

Lastly, the main limitation of the method is related to the range of the frequency parameter α. In the case of physiological flows, this means that the use of the method is also limited to a specific range of arterial diameters. Since this includes the main distal arteries, these arteries will constitute the preferred field of in vivo application, as

illustrated by the example of the human brachial artery tested above.

6 Conclusion

This method, which is applicable to non stationary Newtonian flows, is based on the fact that these flows generate zones where the inertial and viscous effects are of the same order of magnitude. As a matter of fact, these zones are also present in the case of non Newtonian fluids which quickly reach their equilibrium value (non Newtonian and non thixotropic fluids), and a similar approach is now being developed to deal with fluids of this kind, using instantaneous velocity profiles to indirectly determine the parameters characterising their non Newtonian rheological behaviour.

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