Calculation of steady and unsteady shear levels in hemodialysis
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Abstract
Hemolysis has gained a growing interest in chronic hemodialysis therapies, as it is responsible for patient's discomfort and several long term complications. Shear induced hemolysis is studied in a 14G axisymmetrical catheter needle model with finite element techniques, using physiological boundary conditions. High shear spots and flow separation zones are studied to identify the sources of hemolysis. The most sensitive area lies around the needle tip.

1 Introduction
The study of shear in hemodialysis needles originates from the fact that ill-performant needles can cause large levels of hemolysis to occur in dialysis patients. Since patients on chronic hemodialysis therapy commonly suffer from a certain degree of anemia, augmented blood cell destruction has a more profound impact in these patients than it would have in healthy subjects. Hemolysis not only results in anemia. Typical symptoms are, from low to high levels of hemolysis, nausea, hypotension, abdominal pains, dyspnoe, shock and ventricular arhythmia. In the long term, also liver iron overload and splenomegaly are observed.

Hemolysis can occur at any place in the extracorporeal blood circuit, where local bio-incompatibility causes the red blood cells to be damaged (Perez [1]). Many of the hemodialysis sources can be suppressed by correctly operating the dialysis system and using adequate dialysate fluids and biocompatible devices. Even under these optimal conditions, in present-day single needle dialysis systems, hemolysis may be observed, the needle being the most prominent source (Hombrouckx [2]).

Many authors have demonstrated that shear stresses are involved in mechanical red blood cell damage and destruction [3-5], in both low shear flows by cell-surface contacts (Hochmuth [7]) and high shear flows when a threshold shear level has been threspassed. This threshold level depends non-linearly on the time during which a blood cell is exposed to shear (Leverett [6]). These two damage mechanisms can be present in dialysis catheter needles. Firstly, shear levels are expected to be high in this area, for needles have the largest flow/diameter ratio in the extracorporeal circuit. This is immediately clear from...
the wall shear stress formula for laminar steady Poiseuille flow in a circular conduit:

$$\tau_{wall} = \frac{4\mu Q}{\pi R^3} \quad [\text{Pa}]$$

with \( \tau \) the shear stress [Pa], \( Q \) the blood flow [ml/s], \( R \) the radius [cm] and \( \mu \) the dynamic viscosity of normal blood [about 0.0036 Pa\cdot s]. Secondly, catheter needle flow includes flow direction changes and conduit diameter variation, that induce secondary flows and flow separation (Leonard [8]). At these locations vortices can develop, resulting in prolonged shearing of the blood cells. They also provoke lower velocities near the wall resulting in lower wall shear rates. Thus cell-surface contacts become possible.

Even when shear levels in a dialysis needle are subhemolytic, they still can be dangerous because a single blood cell will pass many times through the needle during a hemodialysis session. These subsequent shear stress loads on the cell may result in a diminished life time.

When one is studying needle performance characteristics, two factors are important. First the resistance to flow should be as low as possible. This is measured by obtaining the pressure drop across the needle for different blood flows. Secondly it is advised to have no peak shear stresses or large regions with flow separation. These latter characteristics are difficult to determine by experiments in actual needles, because of their small sizes. Model studies are the appropriate way to study these needle flow phenomena. In the present study, flow is analyzed in a 14 Gauge catheter needle by using a numerical model for both steady and unsteady flow conditions.

2 Numerical Modelling

2.1 Solution method

In order to compute the flow characteristics in a dialysis needle, the complete Navier-Stokes equations should be solved together with the continuity equation. Since these equations generally cannot be solved analytically, a numerical method is used. Especially the finite element method is attractive for this problem, since it allows to subdivide the region of interest quite easily and complex geometries give no extra problems. Also a more dense gridding is possible in loci with high (velocity) gradients. The finite element method used, is based on the pressure velocity formulation of the discretized Navier-Stokes and continuity equations for incompressible laminar flow. They can be written as:

$$M \frac{\partial u}{\partial t} + S(u)u + N(u)u - L^T p = F$$ \quad (momentum equation)

$$Lu = 0$$ \quad (continuity equation)

with \( u \) the velocity, \( M \) the mass matrix, \( S \) the stress matrix, \( N(u)u \) the convective terms, \(-L^Tp\) is the pressure term and \( F \) is the force term. Because \( S \) and \( N \) may be non-linearly dependent on the velocity, the momentum equation is linearized by Newton’s method before the equations are solved. The solution of the pressure and velocity is decoupled by use of the penalty function method (Cuvelier [9]).

For steady flow the first term with the mass matrix vanishes and the system of equations is solved iteratively until convergence to the final solution. In the case of unsteady flow, a modified \( \theta \) method is used, similar to the implicit finite difference time scheme. Firstly, the Navier-Stokes equations are solved for an intermediate time step \( n+\theta \cdot 0 \leq \theta \leq 1 \):
Secondly, the velocity at time step \( n + 1 \) is found from extrapolation:

\[
\frac{M u^{n+\theta} - u^n}{\Delta t} + S(u^n)u^{n+\theta} + N(u^n)u^{n+\theta} - L^T p^{n+\theta} = F(u^{n+\theta})
\]

\[
Lu^{n+\theta} = 0
\]

For reasons of damping the numerical pressure oscillations, the pressure and other derived quantities like shear are always calculated at timestep \( n + \theta \). If \( \theta \) equals \( \frac{1}{2} \), a Crank-Nicolson scheme is found that has a second order accuracy. As the initial solution for the time dependent calculation, a steady state solution at the initial timestep is used.

### 2.2 Calculation Grid

To employ the finite element technique, a calculation grid is necessary. Calculation costs were reduced by choosing a two-dimensional grid. To obtain in this manner valuable three-dimensional information, it is used in an axisymmetrical coordinate system. The elements used to build the grid are 6 point triangles with velocity nodes at the vertices and at the midpoints in between for quadratic interpolation functions. The pressure and stresses are linearly interpolated between nodes at each element vertex. Figure 1 shows a mesh detail near the tip of the needle with enlarged vertical scale. The symmetry axis lies at the top, while the bottom line represents the vessel wall of the patient's fistel. In between, one half of the needle is gridded, concentrically placed in the fistel. Blood flow in the fistel is from left to right. The side wall of the needle seems to be broken. At this location, the needle wall can be either opened or closed, depending on the supplied boundary conditions. When opened, it is used to study the influence of a lateral needle hole. Because of the axisymmetric coordinate system, it is not a real hole but merely a band, splitting the needle in two pieces. Yet it provides valuable information without the expense of constructing a complete three-dimensional model.

The dimensions of the grid are chosen to represent a 14G catheter needle in a fistula.
with an internal diameter of 4 mm. The total grid consists of 2285 nodes connected by 1062 elements. The fistel is modelled as being a rigid tube. The real wall compliance is not taken into account.

2.3 Boundary Conditions
At the needle and the fistel walls the velocities are made zero. The gap in the needle wall has zero velocities either at the horizontal edges (fig 1: lateral hole is closed) or at the vertical walls of the gap (lateral hole is open). The only varying boundary conditions are for respectively the fistel entrance (bottom left), the fistel exit (right) and the needle exit (top left). At these boundaries either the pressure or the velocities are prescribed. In case of the pressure condition, fully developed flow is assumed.

For unsteady flow calculations, in vivo measured pressure curves are used. These curves are entered in the program by suitable Fourier series. Figure 2 and 3 show the periodic behaviour of respectively the fistel entrance pressure and the pressure at the needle exit. The pressure at the fistel exit either is made constant or it has the same time dependent behaviour as the entrance pressure, but with a lower mean value.

3 Numerical Results

3.1 Verification of the model
Table 1 *Shear rates, predicted and computed*

<table>
<thead>
<tr>
<th>Inflow</th>
<th>Shear (1/s)</th>
<th>Re</th>
<th>Outflow</th>
<th>Shear (1/s)</th>
<th>Re</th>
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<tbody>
<tr>
<td></td>
<td>predicted</td>
<td>computed</td>
<td>Δ%</td>
<td>predicted</td>
<td>computed</td>
</tr>
<tr>
<td>flow ml/min</td>
<td></td>
<td></td>
<td></td>
<td>flow ml/min</td>
<td></td>
</tr>
<tr>
<td>no lateral hole</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>47.4</td>
<td>1155</td>
<td>1125</td>
<td>2.6</td>
<td>142</td>
<td>573</td>
</tr>
<tr>
<td>71.4</td>
<td>1740</td>
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<tr>
<td>111</td>
<td>2700</td>
<td>2476</td>
<td>8.3</td>
<td>331</td>
<td>2656</td>
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<tr>
<td>lateral hole</td>
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<td>1564</td>
<td>1526</td>
<td>2.4</td>
<td>178</td>
<td>682</td>
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<tr>
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<td>2299</td>
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<td>3594</td>
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<td>5847</td>
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</table>
In order to verify the validity of the model, the results of steady flow analysis are compared with the theoretical prediction. To this extent the calculated wall shear rates at the needle exit under fully developed flow conditions are compared with the values predicted by the Poiseuille flow theory. The results for different blood flows are shown in table I. Blood is considered a Newtonian fluid under the given test conditions. The two values differ by less than 3 percent for all cases except for inflow without a lateral hole, where the differences are higher for Reynolds numbers greater than 200. This is probably because in this case the fully developed flow condition at the boundary does not reflect the real situation. In fact a large vortex is seen at the inner wall of the needle.

3.2 Steady State Calculations
To assess the flow resistance of this 14G needle, the pressure difference between entrance and exit of the needle (model length is 40 mm) is plotted against the flow (figure 4). The dotted curve is the reference theoretical Poiseuille pressure loss over this needle for the same diameter and flow rate. As can be seen the outflow from the needle with lateral hole closely matches this Poiseuille curve. All other flow conditions have higher pressure drops due to a deviation from the Poiseuille flow conditions. The worst case is the inflow condition, where already very high losses can occur for flows of only 100 ml/min.

![Figure 4: Pressure gradients over a 14G needle](image)

In figure 5 the corresponding maximum values of shear rate are plotted. This maximum in most cases is found near the needle tip just inside the needle: at the wall in case of outflow and between the bulk fluid flow and the vortex near the wall at inflow De Wachter[9]. Only when a lateral hole is present with the inflow condition, the maximum shear is found at the edge of the hole. The Poiseuille reference is calculated for a tube of 1.3 mm diameter, which is the diameter at the needle tip. Both in- and outflow of a needle with hole perform well, but the values obtained with the regular inflow condition are very high. In this case the fluid flow has to bend for 180 degrees over the needle tip. The values are somewhat better for a rounded tip than for the straight tip as in figure 1. This emphasises that needle edge finishing can play a non-neglectable role in the onset of hemolysis.

The region where very high shear is seen, is about 1 mm long, from the needle tip into the needle lumen. The local velocities are 1 m/s in order of magnitude. This means that a
cell will be exposed to the high shear levels during about 1ms. Leverett [6] tabulated the threshold shear rate causing immediate hemolysis for this exposure period to be about 140000s⁻¹. If we may extrapolate the calculated values, this threshold shear rate is already attained with flows as low as 200 ml/min!

![Figure 5 Maximum shear rate in a 14G needle](image)

3.3 Unsteady Calculations
Viewing the results from the steady calculations, it is important to study in more detail the worst case: needle inflow without a lateral hole. First we look at the variation of maximum shear in time and the locations where it is found. Secondly, we try to identify low shear zones where cell-surface contacts are possible.

![Figure 6 Maximum shear plotted vs. Time (left) and Flow through the needle (right)](image)
3.3.1 Variation of shear in time. In this simulation the pressures as shown in figures 2 and 3 are applied during 0.8 seconds. This is the period before the suction pressure in the needle rises to the fistel pressure and when the flow temporarily drops significantly (at about 2.4 seconds, figure 3). At the fistel exit the same pressure as at the entrance is supplied, but with a mean value that is only one third. The mean flow through the needle for these conditions is about 91 ml/min. In figure 6 the maximum shear shows a periodic behaviour similar to the fistel pressure (figure 2). This may signify that also the fistel pressure variation can be a factor that influences hemolysis. The straight curve at the right shows the relation between flow through the needle and the maximum shear encountered. It appears that they are linearly related to each other (least square fit correlation, \( r > 0.99 \)).

3.3.2 Low shear zones. To study these, the pressure boundary curves were shifted over 2.4 seconds to study the low flows through the needle. Fluid flow is still aimed inward. When the pressure difference is minimal, the flow through the needle becomes neglectable and a slow vortex appears at the outside of the needle apical hole. When the suction rises again, the flow increases and the initial vortex shrinks and finally a new one appears at the inside wall of the needle tip. With increasing pressure difference, the vortex becomes longer, and its lateral size increases. In figure 7 the boundary of the vortex is shown at different times from the beginning of the flow increase. Also at the fistel wall, where the blood flow bends into the needle, a vortex is formed (bottom right). The actual flow through the needle is shown in figure 8.
Low shear values as explained in Hochmuth [7] are found at the distal ends of these vortices (stagnation points). Here it is possible that a blood cell, which adheres during a low shear period, later is deformed, detached and damaged when the shear rises again by the repositioning of the stagnation point when the vortex length changes.

Conclusion

As explained in this article, a regular 14G catheter needle has the possibility to expose blood cells to high shear levels that cause blood cell damage and even immediate destruction. Also vortices are present that can trap blood and expose it during prolonged periods to this high shear. At the stagnation points cell-surface contacts are possible. When using a peripheral needle, one should select it carefully for the desired flow through it. The authors believe that there is a need for improvement of the present needle flow characteristics.

References