Computational estimation of fluid mechanical benefits from a fluid deflector at the distal end of artificial vascular grafts

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Abstract
Intimal hyperplasia at the distal anastomosis is considered to be an important determinant for arterial and arteriovenous graft failure. The connection between unhealthy hemodynamics and intimal hyperplasia motivates the use of computational fluid dynamics modeling to search for improved graft design. However, studies on the fluid mechanical impact on intimal hyperplasia at the suture line intrusion have previously been scanty. In the present work, we focus on intimal hyperplasia at the suture line and illustrate potential benefits from the introduction of a fluid deflector to shield the suture line from unhealthily high wall shear stress.

Key words: vascular bypass, artificial grafts, CFD modeling, wall shear stress, computational design evaluation

1. Introduction
In vascular bypass surgery, as well as surgery for venous access, autologous vessels are used as far as possible. However, because of occasional lack of autologous material of sufficient quality, artificial grafts are often used as a substitute. Unfortunately, artificial grafts are more prone to occlude postoperatively [2]. Anastomotic designs can be categorized to either the type of end-to-end anastomosis or the type end-to-side anastomosis (Figure 1). The latter is typically employed in vascular bypass surgery as well as in constructions of vascular access grafts, used for hemodialysis. Anastomotic intimal hyperplasia (IH) at the lower anastomosis (Figure 1) is considered to be one of the principal causes of graft failure. By attaching a vein patch or a vein cuff at the lower anastomosis, the rate of graft patency seems to be improved [3, 18, 21]. The mechanism of the improvement has been suggested to be related to factors such as fluid mechanical adjustments, compliance mismatch, and to the venous material itself [3, 6, 20, 12, 16, 22].

Artificial graft ends that mimic the geometries obtained by the surgical techniques mentioned above have been suggested to significantly improve the patency [19]. However, still there is room for further geometric...
design improvement [14, 15]. In particular, fluid mechanical effects around the suture line needs to be drawn into focus [14, 15, 24]. It is therefore desirable to include protrusions representing the suture line in the geometric model (Figures 2–4).

IH seems to be closely coupled primarily to low or high wall shear stress (WSS), but also to related measures [13, 11]. A WSS outside the range 0.5–3 Pa is considered closely related to IH, as disclosed by van Tricht et al. [23]. Arteries seem to establish a radius that, under normal conditions, results in a mean WSS in the range 1–2 Pa [8]. Therefore, IH induced by low WSS may be a way for the arteries to restore the local WSS back to the normal range [3, 10]. However, the process of IH stimulated by high WSS most likely receives positive feedback from the increased WSS that follows, which means the process does not end until the site is occluded.

The aim of the present study is to computationally evaluate the idea of redirecting a potentially high WSS away from the suture line by introducing a slight stricture—a fluid deflector (FD)—to shield the suture line from excessive fluid mechanical forces. The design change due to the FD will affect the manufacturing process of the artificial graft, that is, the FD is introduced before the surgical procedure. The result will be increased WSS just proximal to the suture line, within the artificial graft, where intimal hyperplasia is not expected to be stimulated by a high WSS [17].

2. Model

We model the blood flow using the Navier–Stokes equations for an incompressible fluid,

\[
\frac{\partial \mathbf{u}}{\partial t} - \nabla \cdot \left( \frac{\eta}{\rho} \left[ \nabla \mathbf{u} + (\nabla \mathbf{u})^T \right] \right) + (\mathbf{u} \cdot \nabla) \mathbf{u} + \frac{1}{\rho} \nabla p = 0, \\
\nabla \cdot \mathbf{u} = 0,
\]

where \( \mathbf{u}(\mathbf{x}, t) \) denotes the fluid velocity and \( p(\mathbf{x}, t) \) denotes the pressure at point \( \mathbf{x} \) and time \( t \). The parameters \( \eta \) and \( \rho \) denote the viscosity and density, respectively. The no-slip condition models the fluid–solid interaction, that is, \( \mathbf{u} = \mathbf{0} \) along the vessel and graft walls.
Blood is complicated to model and displays many non-Newtonian properties. In particular, at low shear rates, blood exhibits a high apparent viscosity while there is a reduction in the blood’s viscosity at high shear rates. [?] reviews various models describing the non-Newtonian viscosity and yield stress of blood flow. Here, we model the shear-thinning aspects of blood with the same viscosity model as used by Longest and Kleinstreuer [14],

\[ \eta(\dot{\gamma}) = \left( \sqrt{\eta_\infty} + \frac{\sqrt{\tau_0}}{\sqrt{\lambda} + \sqrt{\dot{\gamma}}} \right)^2 \]  

(2)

in which \( \dot{\gamma} = \left[ 2 \text{trace}(\mathbf{r}^2) \right]^{1/2} \), where

\[ \mathbf{r} = \frac{1}{2} \left[ \nabla \mathbf{u} + (\nabla \mathbf{u})^T \right]. \]

The constants

\[ \eta_\infty = 3.4 \times 10^{-3} \text{ Pas}, \tau_0 = 8.677 \times 10^{-3} \text{ Pa}, \text{ and } \lambda = 10.08 \times 10^{-3} \text{ s}^{-1} \]

are provided by fitting the constants in equation (2) to experimental data obtained from a viscometer in vitro (courtesy of Associate professor Bo Sandhagen, Department of Medical Sciences, Uppsala University, UAS, Sweden). This data consists of 30 shear rates in the range 0.0175–128.5 s\(^{-1}\) and corresponding viscosities. The relative Euclidian distance between the viscosities given by the model and the measured viscosities is around 6 %. The Quemada type model (2) can be viewed as an extension of the Casson model with an additional shear rate modifier \( \lambda \), which increases the accuracy of the shear-rate correlation [?, p. 253].

For the computations, we use the software Comsol Multiphysics to numerically solve the Navier–Stokes equations (1) in a bounded domain employing the finite element method with the Taylor–Hood element pair. At the inflow boundary of the truncated graft, we impose a parabolic inflow profile for the steady computations. In the case of end-to-side anastomoses, the proximal end of the recipient vessel is assumed to be closed and thus given a nonslip condition. At the distal boundary of the recipient vessel, we use the outflow condition of vanishing fluid forces, that is, the boundary condition

\[ \sigma n = 0, \]  

(3)

where \( \sigma = -p\mathbf{I} + 2\eta(\dot{\gamma})\mathbf{r} \) and \( n \) is the local wall normal.

As a measure for shear stress (SS), we use in the current study the quantity

\[ s(x) = \eta(\dot{\gamma}(x))\dot{\gamma}(x), \]  

(4)

which is a quantity defined locally at any point \( x \) in the domain. The WSS vector is by definition \( \mathbf{t} = (\mathbf{I} - n \otimes n)\sigma n \). Appendix A shows that \( s(x) = |\mathbf{t}| \) if \( x \) is a point on a solid wall with WSS \( \mathbf{t} \). Thus, the quantity \( s \) approximates the WSS magnitude in the vicinity of the wall.
3. Results

3.1. Qualitative behavior

The computations were performed for various anastomotic models, in both 2D and 3D, with pulsatile or constant blood flow. Here, we only discuss results from the steady calculations on grafts with a diameter of 6 mm, using a peak systolic flow of 0.5 m/s, resulting in Reynolds numbers in the range 600–800. The major conclusion is that the introduction of a FD significantly reduces WSS at the suture margin. In our first example we illustrate the idea of the FD using an end-to-end anastomosis in 2D. We represent the suture margin with two humps; we choose to shield one of these humps with a FD while leaving the other hump unshielded. The setup is from the start symmetric. By adding a FD to the upper wall and not the bottom wall, we break the symmetry and acquire a qualitative image (Figure 2) of the flow changes induced by the FD. The suture margin (s) protrudes 0.3 mm into the lumen on both the upper and lower wall and the FD (g) at the upper wall is located about 1 mm from the suture margin. The FD significantly lowers the WSS; the WSS is about three times larger on the lower (unshielded) suture line than on the upper suture line (shielded by the FD). In this case, the distal part of the FD should protrude about 0.5 mm to sufficiently lower the WSS at the suture line. In the region between the suture line and the FD, IH will likely occur, due to the presence of a recirculation zone with low WSS (< 0.5 Pa). However, as plaque is formed, the WSS will increase, initiating a self-ending process. With the idea that the flow should not be disturbed more than necessarily, the size of the FD should ideally be adapted to the surgical technique.

Figure 3 shows results from an end-to-side anastomosis in 2D. Here, the proximal part of the receiving vessel is closed. The suture line protrudes 0.4 mm into the lumen. The FD in the right picture protrudes 0.6 mm into the lumen and its distal edge is located about 1 mm from the suture line. Here, the FD reduces the maximum WSS at the suture margin from 8.7 Pa to 3.5 Pa. Figure 4 shows that a FD can have a shielding effect on indurations at the suture line also in end-to-side anastomoses in 3D. The graft and the
receiving vessel are modeled as cylinders with diameter 6 mm connected at an angle of \( \arctan 0.5 \) rad (about 26.5 degrees). Here a semi-sphere with radius 0.4 mm centered at the vessel wall represents a scaring on the suture line. The setup in the right image of Figure 4 includes a FD protruding 0.6 mm into the lumen. The qualitative effects are in this case the same as for the 2D cases, that is, the FD reduces the WSS at the suture margin whereas the WSS at the FD is higher than on a typical graft wall.

3.2. Shear stress trade-off studies

To explicitly study how the shape of the FD affects the trade-off between WSS on the suture line and on a FD, we perform an exhaustive search throughout a limited parameter space. Here, we use a symmetric version of the setup in Figure 2 that is, a FD is located also at the bottom wall. The shape of the FD is given by a cubic spline defined by seven control points. The control points at the ends of the FD are fixed, and the five inner control points are evenly spaced in the horizontal direction. Four distances, equispaced
between 0.0 and 0.5 mm, comprise the admissible intrusions at any of the inner control points. In total, there are thus $4^5 = 1024$ admissible designs. The diagram in Figure 5 shows the maximum WSS at the FD versus the maximum WSS at the suture line. More precisely, with “maximum WSS” we mean the maximum value of $s$, as defined in expression (4), in any of the triangles facing the boundary portion in question. The three designs on the right side of Figure 5 correspond to the red squares along the Pareto frontier (that is, the points furthest down and to the left) in the diagram to the left.

4. Discussion and Outlook

Anastomotic intimal hyperplasia (IH) at the lower anastomosis is considered to be a principal cause of artificial vascular graft failure. Hence, efforts have been made to reduce IH. However, it appears that the hemodynamic impact on the suture line has previously been underestimated. Even if a very skilled surgeon would succeed to make an absolutely smooth junction between the graft and the recipient vessel, a protrusion will nevertheless occur at this site after some time (as for instance shown by Trubel et al. [22]). Once there is a rough surface, the WSS will increase. As previously discussed, there are theoretical arguments, presented by for example van Tricht et al. [23], supporting that an increased WSS speeds up the process of intimal hyperplasia, causing a vicious circle that may be broken by introducing a fluid deflector (FD).

The present study only addresses conditions at the FD and the suture line. Other conditions, such as those at the floor [7], are not addressed. Design changes to improve these conditions will likely necessitate monitoring of quantities, different from WSS, that in the literature has been correlated to IH formation, such as oscillations in the wall shear vector, wall-normal pressure gradients, indicating local in- or outflow, or velocity-field-driven particle-transport properties at the wall [13].
The main focus of the present study is to estimate qualitative fluid mechanical effects of adding a FD at the distal end of an artificial graft to shield the suture line in steady flow on idealized geometries ignoring any fluid structure interaction. A next step could thus be to see whether anything fruitful comes out of delving in detail into the quantitative effects of fluid structure interactions in real life geometries under conditions of physiological unsteady flow.

In three spatial dimensions and for unsteady flows, it will be impossible to employ the strategy of Section 3.2 exhaustive parameter search, even within a modestly-sized design space. An attractive alternative is to employ sensitivity analysis using derivatives computed by the adjoint-equation approach; see for example Giles and Pierce [9] for a description of the technique. The sensitivity analysis could potentially be used to perform numerical shape optimization for the design of artificial grafts. Agoshkov et al. [1] report on a preliminary attempt in this direction, optimizing the design of a patch region in order to minimize the (squared and integrated) vorticity in unsteady 2D Stokes flow. The assumption of Stokes flow is made to simplify the mathematical and numerical issues, but is not physically relevant for graft anastomoses; in reality, the Reynolds number ranges from about 200 (in arterial bypasses) up to about 1500 (in arteriovenous grafts) as reported by Haruguchi and Teraoka [12].

In summary, our computational results suggest that a fluid deflector may be useful in cases where artificial grafts are employed (end-to-end as well as end-to-side anastomoses) and, moreover, hopefully prove beneficial in increasing the patency of already optimized graft ends. The results should be followed up by further geometrical improvements and in vivo experiments.

A. Appendix

The WSS vector is by definition \( t = (I - n \otimes n) \sigma n \), that is, the projection in the tangential plane of the fluid force vector \( \sigma n \). Here, \( \sigma = -pI + 2\eta(\dot{\gamma}) \tau \) is the stress tensor, \( \tau = \frac{1}{2} \left[ \nabla u + (\nabla u)^T \right] \) the rate of deformation tensor, and \( n \) the local wall normal. In this article, we evaluate \( |t| \) indirectly by observing

\[
s(x) = \eta(\dot{\gamma}(x))\dot{\gamma}(x),
\]

where \( \dot{\gamma} = [2 \text{trace}(\tau^2)]^{1/2} \).

Here we will demonstrate that \( s = |t| \) at solid walls. Thus, the value of \( s \) in the vicinity of the wall yields a simple and “geometry-free” estimation of the wall-shear-stress magnitude \( |t| \).

For any point \( x_w \) on the wall, we can construct a local Cartesian coordinate system with basis vectors \( e_i, i = 1, \ldots, n \), such that the wall normal is aligned with \( e_n \), and \( e_1, \ldots, e_{n-1} \) span the tangent plane at \( x_w \). Because of the no-slip condition, the partial derivatives of the velocity vector at the wall vanish in the tangential directions, that is, \( \partial u / \partial x_j = 0 \) for \( j = 1, \ldots, n - 1 \) at \( x = x_w \). Hence, the rate of deformation
tensor $\tau$ at $x_w$ reduces to

\[
\tau = \frac{1}{2} \begin{bmatrix}
0 & \cdots & 0 & \frac{\partial u_1}{\partial x_n} \\
\vdots & \ddots & \vdots & \vdots \\
0 & \cdots & 0 & \frac{\partial u_{n-1}}{\partial x_n} \\
\frac{\partial u_1}{\partial x_n} & \cdots & \frac{\partial u_{n-1}}{\partial x_n} & 0
\end{bmatrix},
\]

and we conclude that, at point $x_w$,

\[
n \cdot \tau n = e_n \cdot \tau e_n = 0.
\]

Moreover, at a solid wall, it holds that

\[
t = (I - n \otimes n)\sigma n = (I - n \otimes n)(-pI + 2\eta(\dot{\gamma})\tau)n
\]

\[
= -p(I - n \otimes n)n + 2\eta(\dot{\gamma})(I - n \otimes n)\tau n
\]

\[
= 2\eta(\dot{\gamma})(I - n \otimes n)\tau n
\]

where the last equality follows from orthogonality relation (7). Definition (5) together with expressions (6) and (8) then yield

\[
s(x) = \eta(\dot{\gamma}(x))\dot{\gamma}(x) = \eta(\dot{\gamma}(x))[2 \text{trace}(\tau^2)]^{1/2}
\]

\[
= \eta(\dot{\gamma}(x)) \left( \sum_{j=1}^{n-1} \left( \frac{\partial u_j}{\partial x_n} \right)^2 \right)^{1/2}
\]

\[
= \eta(\dot{\gamma}(x))[2|\tau n|] = |t|,
\]

which is what we wanted to show.

References


