In vitro assessment of the hemodynamic effects of a partial occlusion in a vena cava filter

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Purpose: Flow fields surrounding a vena cava filter were studied with a noninvasive method of flow visualization to examine the underlying hemodynamic factors that contribute to its function and patency.

Methods: The photochromic technique was used to measure axial velocity profiles and wall shear stress distributions in a 12F titanium Greenfield filter partially occluded with a simulated volume of entrapped thrombi. These results were compared with similar measurements obtained for an unoccluded filter and with a theoretic prediction of the velocity profile and wall shear stress without a vena cava filter. Shear stress distributions were obtained along the vessel wall and for the partially occluded filter along the surface of the simulated clot.

Results: The unoccluded filter was observed to have little effect on the flow field. In the case of the partially occluded filter, the results of these measurements show that caval blood flow is preserved by the creation of an annular region of increased flow around the periphery of the clot. Within this region high shear stresses that develop as a result of the increased flow are observed along the vessel wall and along the surface of the simulated clot. No vortices or turbulence were observed with either the unoccluded or the partially occluded filter.

Conclusions: The elevated levels of shear stress may be a factor in the lysis of trapped clots observed in vivo. Although increased shear stress is reported to cause thrombogenesis in an in vivo study in canines, particularly under turbulent flow conditions, the levels of shear measured in this study around the simulated clot were well below such values. (J Vasc Surg 1997;25:663-72.)

Vena cava filters are designed to provide protection against pulmonary embolism by preventing clots from migrating to the pulmonary circulation.1,2 To be effective a filter must efficiently trap clots of varying size while preserving the patency of the inferior vena cava. Currently, a variety of filters are available, and a number of clinical trials3-6 and in vitro studies7-10 have been conducted to determine their relative effectiveness. Although there is no substitute for prospective clinical trials, little or no information has been obtained from these studies on the mechanical functioning of such devices, a factor essential to their continued improvement.

The performance of a vena cava filter is influenced by several factors. If a filter is to be effective, it must successfully capture all emboli exceeding a certain size. In addition, efficient clot lysis within the filter is essential, if caval patency is to be ensured. Clot entrapment and lysis are most affected by the flow dynamics created by the filter’s presence. Although a number of in vitro studies have examined the flow dynamics of vena cava filters, little quantitative information has been obtained on the hemodynamics of a partially occluded filter and on the relationship between flow dynamics and clot entrapment and lysis. In part, this is due to the use of dye injection, a flow visualization technique that provides only qualitative information.7,9 In addition, the complexities of the surrounding flow fields have limited the observations made with this technique to qualitative estimates of the degree of mixing as indicated by the injected dye stream.
MATERIAL AND METHODS

In this study the flow field surrounding a 12F titanium Greenfield vena cava filter was examined with a noninvasive flow visualization and measurement technique known as the photochromic dye method. Measurements were performed with the filter in two states: with an unoccluded filter, that is, with no entrapped clots, and partially occluded with a simulated clot volume. These measurements were used to assess the effects of the embolus on the surrounding flow field and to determine how the hemodynamic effects of the filter might be altered to influence its effectiveness, particularly with respect to the lysis of the entrapped thrombi.

Velocity and wall shear stress were considered to be important parameters to measure in this study. In particular, the level of shear stress at the surface of the entrapped clot volume may affect the rate of lysis. Measurement of wall shear stress distributions is especially difficult given the three-dimensional flow field and the complex geometry of an inferior vena cava filter. Conventional flow measurement techniques and even computational methods are limited in their ability to accurately predict shear stress distributions in such situations. However, the photochromic method is well suited to measuring wall shear stress under such conditions as indicated by its previous application in a number of in vitro studies.11-14

Photochromic flow visualization and measurement technique. The photochromic method requires the addition of a normally colorless dye to the test fluid, thus creating a solution that reacts when exposed to ultraviolet (UV) light and becomes opaque. A laser is used to selectively "tag" portions of the fluid. Subsequently, the motion of these elements can be tracked optically, and estimates of velocity and wall shear stress can be made.

The photochromic flow visualization and measurement system used in this study is shown in Fig. 1. It consisted of a pulsed nitrogen laser (λ = 337 nm), a convex lens (f = 380 mm), an electronic strobe, a high-resolution charge-coupled device (CCD) camera with a digital interface (Kodak MegaPlus 1.4), a macro lens and bellows system, a digital frame grabber (DIPIX P360, Ottawa, Ontario), a motorized stage and controller, and a programmable waveform pump (UHDC, London, Ontario). A computer (IBM 486 PC compatible) was used to control and synchronize the individual components of the system and to record and analyze images from the CCD camera.

Operation of the system is as follows. A UV beam produced by a single pulse of the nitrogen laser is focused by the convex lens into the flow model. The UV beam reacts with the photochromic dye to produce a dark blue trace within the fluid. The trace is illuminated by backlighting the dye trace with the electronic strobe to freeze its motion and to obtain maximum contrast. Displacement of the trace by the flow field was recorded with the CCD imaging system and was used to estimate velocity and wall shear stress. In these experiments the path of the laser beam, the CCD camera, and its field of view were fixed in space, while the flow phantom and filter were mounted on the motorized stage and could be precisely positioned with respect to the path of the laser beam. Thus by moving the stage a dye trace could be produced at any position within the flow phantom.

Fabrication of the flow phantom. The UV transparent flow phantom (Fig. 2) required to use the photochromic flow measurement system was fabricated by polymerizing methyl methacrylate monomer to form a straight tube with an internal diameter of 20 mm. A notch was cut away to allow for the insertion of a flexible ring (Sylgard) for embedding of the hooks of the Greenfield filter. This arrangement was intended to simulate the in vivo situation, where the hooks are embedded into the vessel wall to prevent migration of the filter. For the partially occluded experiments the filter was fitted with a simulated thrombus volume (Fig. 2, b). This occlusion was modeled as a cone with a hemispheric base, an approximate clot volume of 1500 mm$^3$, and a maximum cross-sectional area of 110 mm$^2$ (corresponding to an area reduction of 35%) that was contoured to fit the filter. It was machined from the same material as the UV transparent flow phantom.

The flow phantom was supported by a mount that allowed it to rotate about its longitudinal axis. This procedure was intended to permit the examination of any desired plane in anticipation of the three-dimensional asymmetric flow field surrounding a vena cava filter.

To ensure that the flow entering the filter was laminar and fully developed, the flow phantom tube was constructed with an entrance length of 160 cm. This corresponded to 80 tube diameters and ensured a parabolic velocity profile under the flow conditions used in these experiments. The desired inlet conditions were confirmed with the photochromic method, and the measured centerline velocities upstream of the filter deviated by less than 1% from the predicted values. In addition, the vicinity of the interface between the Sylgard insert and the flow model was examined for any evidence of flow disturbances, and no deviation was observed.
Flow conditions. The test fluid used in the flow loop was deodorized kerosene (Shell-Sol 715) with a trace amount (50 ppm) of photochromic dye (1',3',3'-trimethylindoline-6-nitro-benzospiropyran) dissolved into the test fluid to facilitate the formation of dye traces. The kinematic viscosity of this fluid (1.8 cS) is different from that of blood (3.5 cS) or that of the test fluids used in previous work. Nevertheless, the principle of dynamic similarity of flows was used by adjusting the flow rate to match the Reynolds number observed in vivo.

For this study the Reynolds number was defined as

\[ Re = \frac{4Q}{\pi vvD} \] (1)

where \( Q \) is the flow rate, \( D \) is the diameter of the lumen, and \( v \) is the kinematic viscosity. To match the in vivo flow conditions of approximately 2 L/min of blood, a flow rate of approximately 1 L/min of the test fluid was used. Based on the internal diameter of the flow phantom (20 mm) and the physical properties of blood and the test fluid, both of these flow conditions correspond to a Reynolds number of 600, thus ensuring their dynamic similarity.

Limitations. As is the case with in vitro simulations, some limitations were associated with our flow phantom. First, a rigid flow phantom was used, in contrast to the compliant inferior vena cava. Second, a Newtonian test fluid was used. Blood is known to behave in a non-Newtonian fashion, particularly at low flow rates. However, in a vessel of this diameter these effects will not be significant.

RESULTS

The flow field surrounding the filter was examined with the photochromic flow visualization and

![Photochromic flow visualization and measurement system.](image)
measurement system in two states, with and without the simulated clot volume. In each state measurements were made in horizontal and perpendicular planes passing through the longitudinal axis of the test section to observe the effects of any asymmetry in the flow fields.

The magnification of the CCD camera was adjusted so that the resulting field of view encompassed slightly more than one half the inner diameter of the tube or 10 mm, extending from the center of the vessel to the inner wall as shown in Fig. 3.

A total of four separate scans along the longitudinal axis were made with the filter in each state starting from the base to a position lying several tube diameters downstream of the tip. In between scans the flow phantom was rotated about its longitudinal axis by 90 degrees. Consequently the first and third scans lie in one plane, whereas the second and fourth scans lie in a second, perpendicular plane. The orientation and numbering scheme used to identify these scans is shown in Fig. 4.

Individually, each scan was composed of many photochromic profiles recorded at 1 mm intervals along the tube axis. At each position a photochromic trace was first formed with a flash delay of 0 msec to obtain the initial position and angle of the trace and its intersection with the wall. Then a second trace was formed with a flash delay of 20 msec. The information obtained from the first displacement profile made it unnecessary to precisely align the beam of the laser perpendicular to the wall and allowed the velocity profile to be determined accurately from the
relative displacement between the two profiles. In addition, the positions of the intersections with the wall obtained from the first trace facilitated calculation of wall shear stress.

**Displacement and velocity profiles.** The displacement profiles of the Greenfield filter in the unoccluded and partially occluded states are shown in Fig. 5. The profiles recorded with a flash delay of 0
Because the phantom was rotated by 180 degrees between the two scans measured in either the horizontal or vertical plane, and the laser beam was not precisely perpendicular to the longitudinal axis of the flow phantom, the profiles do not coincide at the center of the flow where the scans overlapped. As expected, this misalignment is most apparent at the center of the flow field. However, it does not affect the accuracy of the velocity or wall shear stress measurements.

Within the interior of the unoccluded filter some noise is evident in the measured profiles. This noise does not represent chaotic or turbulent flow but...
simply that the algorithm used by the automated imaging system failed to detect the photochromic trace. These failures were due primarily to the filter “legs” preventing the penetration of the trace. In other instances poor trace contrast or artifacts in the digital image prevented a trace profile from being measured properly.

In Fig. 5, C the measured axial velocity profiles are overlaid with the theoretic parabolic velocity profiles that would be obtained in the absence of the occlusion and filter to demonstrate the combined effect of the filter and the simulated occlusion. This theoretic profile is given by

\[ u(r) = \frac{32Q R^2 - r^2}{\pi D^4} \]

where \( u \) is the axial velocity, \( r \) is the radial distance to the center of the tube, \( Q \) is the flow rate, and \( D \) is the diameter of the vessel. This profile is based on the analytic solution to the Navier Stokes equation for flow in a straight tube, that is, Poiseuille flow, and is calculated assuming the same flow rate as used with the filter. It is useful to note that corresponding experimental and theoretic profiles always coincide at the intersection with the wall because of the assumption of the no-slip condition, that is, the fluid velocity is zero at the wall.

Because the viscosities of the test fluid and blood are different, the velocities measured in vitro must be scaled to obtain the predicted in vivo values. Thus the in vitro and in vivo velocities are related by

\[ \left( \frac{u}{v} \right)_{\text{in vivo}} = \left( \frac{u}{v} \right)_{\text{in vitro}} \]

where \( u \) is the axial velocity and \( v \) is the kinematic viscosity. Thus the in vitro velocities must be scaled by a factor of 1.9 to obtain the corresponding in vivo values. For comparison the maximum velocity (10.8 cm/sec) of the theoretic profile occurs at the centerline. In vivo, this would correspond to a velocity of approximately 20.5 cm/sec.

As expected, the introduction of the occlusion leads to a much greater impact on the flow field than with the filter alone. At the base of the filter the flow remains almost unaffected as indicated by the agreement between the theoretic and experimental profiles, with the exception of small regions in scans 1 and 3. In these regions the filter “legs” have intersected the measurement plane near to the wall and are partially obstructing the flow.

As the flow in the filter nears the occlusion, the axial velocities in the central region are reduced substantially. Simultaneously, the velocities around the periphery of the filter increased substantially as the fluid was redirected by the occlusion.

**Wall shear stress distributions.** Wall shear stresses were calculated from the velocity profiles by fitting a cubic polynomial to several points nearest to the wall or surface and computing the velocity gradient at the wall.\(^{15-18}\) All shear stress values measured along the vessel wall were positive, and no regions of separation or reversed flow were observed. For purposes of comparison the theoretic wall shear stress for flow in the absence of the filter is also shown. This value is given by

\[ \tau_{\text{ws}} = \frac{32\mu Q}{\pi D^3} \]

where \( Q \) is the flow rate, \( D \) is the diameter of the vessel, and \( \mu \) is the dynamic viscosity. It is representative of the flow upstream of the filter and is based on the solution to the Navier Stokes equation for steady flow in a straight tube, that is, Poiseuille flow.

However, because of the difference in the viscosities of the test fluid and blood, the in vitro and in vivo wall shear stresses are related by

\[ \left( \frac{\tau_{\text{ws}}}{\rho \mu^2} \right)_{\text{in vivo}} = \left( \frac{\tau_{\text{ws}}}{\rho \mu^2} \right)_{\text{in vitro}} \]

Thus the predicted in vivo wall shear stress (in vivo values are shown in brackets in Fig. 6 and Fig. 7) was obtained from the values measured in vitro by multiplying by a factor of 5.

The shear stress distributions along the vessel wall are shown in Fig. 6. At the base of the filter reduced wall shear stress values can be seen, particularly in the horizontal (Fig. 6, A). As described previously these reductions are due to the filter “legs” obstructing the flow near the wall as the measurement plane is intersected.

Past the filter base the shear stress increased dramatically, reaching a maximum value of 2.3 dyn/cm\(^2\) (11.5 dyn/cm\(^2\) in vivo) or more than 7 times the nominal upstream value of 0.3 dyn/cm\(^2\) (1.5 dyn/cm\(^2\) in vivo). Further downstream the wall shear stress returned gradually to the upstream value.

In Fig. 7 the shear stress distribution along the surface of the occlusion is shown. The distribution is somewhat complex because of the intricacy of the combined geometry of the filter “legs” and the occlusion, but the trend is quite clear. The wall shear at
the base of the occlusion was 2 to 3 times the upstream value. Along the surface of the occlusion the wall shear stress increased rapidly to the widest point of the simulated occlusion. Here at this axial position where the fluid moves most rapidly in the annular flow, the wall shear stress reached its maximum value of approximately 10 times the normal level. Further along the occlusion the wall shear stress fell to less than 0.3 dyn/cm² (1.5 dyn/cm² in vivo) in the vertical plane.

**DISCUSSION**

The relative extents of the flow disturbances caused by the filter in the two states are illustrated in Fig. 8. Areas in red indicate regions where the fluid velocity is less than the normal velocity in the absence of both filter and clot. Although the inclusion of the clot results in a larger disturbance, the flow field becomes more uniform and symmetric. The region of annular flow around the periphery of the clot is accompanied by increased fluid velocities and ele-
vated levels of shear stress along the vessel wall and the surface of the clot.

An in vivo study of a similar Greenfield filter inserted in the inferior vena cava of dogs show gradual lysis of trapped clots as a result of continued caval flow. Clinical trials have also provided evidence of this process. The elevated levels of wall shear stress measured along the surface of the simulated clot may be a primary factor in the process of clot lysis observed in vivo. There is substantial evidence of a link between hemodynamic forces and thrombogenesis. In particular, the high levels of shear stress generated by turbulence have been shown to be a major factor in thrombus formation. Nevertheless these levels are well above the shear stresses measured in this study. At the levels observed in our study the shearing forces may be sufficient to cause a mechanical lysis of the trapped clot, reducing its volume and thus ensuring continued flow through the inferior vena cava. The shear stress distribution suggests that the lysing action is most potent at the widest point of the clot, where it is exposed to the highest levels of shear stress. In addition, the elevated velocities and shear stress along the vessel wall serve to prevent thrombogenesis caused by the presence of the foreign material of the filter. In fact, the elevated shear stress levels may play the primary roles in preserving the patency of the filter and the vena cava.
A previous study described the formation of a turbulent bypass channel with the introduction of a clot into a different vena cava filter design (VenaTech LGM).8 No such effect was observed with the Greenfield filter in this study. Although not conclusive, the reason for this discrepancy is not due to differences in filter design or clot volume but to different flow conditions. A flow rate of 2 L/min of water was used in the simulations with the VenaTech LGM filter. This corresponds to a Reynolds number of approximately 2100, far in excess of the normal in vivo conditions of the vena cava (Re = 600) and the level at which steady laminar flow becomes turbulent in a straight tube.19 Thus the development of turbulence downstream of a vena cava filter is unlikely under normal in vivo flow conditions except in the event of a severe reduction in cross-sectional area, that is, greater than 60%,20 caused by occlusion of the filter with one or more thrombi.

REFERENCES