◆ EXPERIMENTAL INVESTIGATION ◆

Computational Modeling of Distal Protection Filters

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Purpose: To quantify the relationship between velocity and pressure gradient in a distal protection filter (DPF) and to determine the feasibility of modeling a DPF as a permeable surface using computational fluid dynamics (CFD).

Methods: Four DPFs (Spider RX, FilterWire EZ, RX Accunet, and Emboshield) were deployed in a single tube representing the internal carotid artery (ICA) in an in vitro flow apparatus. Steady flow of a blood-like solution was circulated with a peristaltic pump and compliance chamber. The flow rate through each DPF was measured at physiological pressure gradients, and permeability was calculated using Darcy's equation. Two computational models representing the RX Accunet were created: an actual representation of the filter geometry and a circular permeable surface. The permeability of RX Accunet was assigned to the surface, and CFD simulations were conducted with both models using experimentally derived boundary conditions.

Results: Spider RX had the largest permeability while RX Accunet was the least permeable filter. CFD modeling of RX Accunet and the permeable surface resulted in excellent agreement with the experimental measurements of velocity and pressure gradient. However, the permeable surface model did not accurately reproduce local flow patterns near the DPF deployment site.

Conclusion: CFD can be used to model DPFs, yielding global flow parameters measured with bench-top experiments. CFD models of the detailed DPF geometry could be used for "virtual testing" of device designs under simulated flow conditions, which would have potential benefits in decreasing the number of design iterations leading up to in vivo testing.

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Surgical or endovascular treatment of carotid artery occlusive disease is pursued for symptomatic patients with 50% to 99% stenosis (based on diameter) in the internal carotid artery (ICA) having perioperative stroke or

death risk <6% or for asymptomatic patients with 60% to 99% stenosis having a perioperative stroke or death risk <3%.¹ The surgical treatment, carotid endarterectomy (CEA), has been shown to be effective for both symp-

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Figure 1 ♦ The 4 DPFs used in the experiments: (A) Spider RX, (B) FilterWire EZ, (C) RX Accunet, and (D) Emboshield.

tomatic^{2,3} and asymptomatic⁴ patients. The endovascular treatment, deploying a stent over the plaque lesion, evolved in the face of concerns over intraprocedural plaque embolization causing neurological events. This prompted the development of cerebral protection devices, such as distal protection filters (DPFs), to trap and remove plaque particulates from the bloodstream distal to the site of stent deployment. Although there is debate over the utility of DPFs during carotid artery stenting (CAS),⁵⁻⁷ the American College of Cardiology Foundation concedes to the use of DPFs during CAS.¹

Recently, results from CREST (Carotid Revascularization Endarterectomy vs. Stenting Trial), a comparison of CAS with cerebral protection to CEA in the treatment of carotid artery disease, were published. The data showed similar outcomes between CEA and CAS, with a stroke, myocardial infarction, and death rate of 6.8% for CEA and 7.2% for CAS.⁸ This difference was not significant ($p=0.51$). The outcome of CREST can be considered further evidence in favor of the use of DPFs for CAS.

The use of DPFs can reduce antegrade blood flow due to pores obstructed by plaque emboli, which can have an adverse effect on CAS patient outcome. This “slow-flow” phenomenon has been visualized qualitatively in vivo using angiography⁹ and quantified in vitro in a single vascular phantom setup.¹⁰ Previous studies have reported patients with flow reduction due to occluded DPFs ranging

from 7.9%¹¹ to 31%.¹² Patients experiencing “slow-flow” had significantly more strokes or deaths within 30 days of CAS than patients who maintained normal flow (9.5% versus 2.9%; $p=0.03$).⁹ Another study found an 8.0% stroke rate for flow-impaired cases compared to a 1.1% stroke rate for normal flow cases ($p=0.11$).¹³ Quantifying flow noninvasively in the treated vessel and knowing the effect of a deployed DPF on the local flow conditions (e.g., pressure drop and the ICA velocity field) would provide valuable information for the design of future generations of cerebral protection devices.

The purpose of this investigation is 2-fold: (1) quantify the relationship between the ICA pressure gradient and velocity in the presence of a DPF and (2) determine the feasibility of modeling computationally a DPF as a porous medium. To accomplish these goals, in vitro experiments and computational fluid dynamics (CFD) modeling are utilized to analyze the effect DPFs have on the ICA flow dynamics.

METHODS

Experimental

An experiment was devised to measure the permeability of 4 DPFs (Fig. 1): Spider RX (ev3 Inc., Plymouth, MN, USA), FilterWire EZ (Boston Scientific, Natick, MA, USA), RX Accunet (Abbott Vascular, Redwood City, CA, USA), and Emboshield (Abbott Vascular;

Figure 2 ♦ Schematic of in vitro flow apparatus.

NB: the next-generation Emboshield NAV⁶, which has an updated design, was not evaluated in this investigation). Each DPF was deployed in an in vitro flow apparatus (Fig. 2). A peristaltic pump and compliance chamber output a steady flow rate of blood-like solution (36% glycerin, 64% water),¹⁴ a Newtonian fluid with a similar average viscosity to blood ($\mu=3.5$ cP). Although blood is a non-Newtonian fluid, it can be approximated as a Newtonian fluid when the shear rate exceeds 1000 s^{-1} . The shear rate through the pores was calculated for a Newtonian fluid flowing through a pipe using Equation (1):

$$\dot{\gamma} = \frac{8v}{d}, \quad (1)$$

where v =velocity and d =diameter. The experimental shear rate was $2.82 \times 10^4 \text{ s}^{-1}$.

The carotid flow model was a Tecoflex tube (Noveon Thermedics Polymer Products, Lowell, MA, USA) with inner diameter (ID) of 5.5 mm, representing the ICA distal to the stenosis. Electromagnetic flow meters (Seametrics Inc., Kent, WA, USA) and pressure manometers (Ashcroft Inc., Stratford, CT, USA) were located proximal and distal to the carotid flow model. Flow rate data were acquired from the flow meters by a data acquisition card (National Instruments Corporation, Austin, TX, USA) controlled by a custom LabVIEW program (National Instruments Corporation, Austin, TX, USA). Pressure data were acquired by visual inspection of the pressure manometers after the flow stabilized.

A range of physiologically realistic target pressure gradients were based upon the pressure gradients measured at the minimum and maximum flow rates and 4 other flow rates (Fig. 3). The flow rate waveform measured by Doppler ultrasound in a human ICA¹⁵ is similar to the waveforms described in other studies.^{16,17} The flow rate output by the pump was adjusted manually to achieve the desired pressure gradient utilizing a trial-and-error process. The flow rate at the desired pressure gradient was measured pre- (Table 1) and post-deployment of each DPF. For comparison, an average ICA flow rate integrated over one cardiac cycle for a normal ICA is $\sim 205 \text{ mL/min}$.¹⁵

The permeability of each DPF was calculated based on the assumption that each filter

Figure 3 ♦ Flow rate waveform in the human ICA.¹⁵

TABLE 1.

Target Pressure Gradients and Flow Rates in the ICA Before Distal Protection Filter Deployment (at the initial condition)

Pressure Gradient, mmHg	Flow Rate, mL/min
7.2	138
9.5	183
11.8	239
14.1	277
16.3	322
18.6	366

allows flow to travel through its pores according to Darcy's equation for porous media. For a single tube with a porous medium:

$$K = \mu \cdot L \cdot \frac{v}{\Delta P}, \quad (2)$$

where K is the permeability of the device, μ is the dynamic viscosity of the fluid, L is the thickness of the porous membrane, and $v/\Delta P$ is the slope of the velocity versus pressure gradient curve.

The permeability of each DPF was calculated using Equation (2). The viscosity was measured using a glass capillary viscometer (CANNON Instrument Company, State College, PA, USA) before each experiment; the average value of viscosity used in the experiments was 3.5 cP. The RX Accunet membrane thickness (estimated at 50 microns) was measured from the height of the z-stack used in taking an image of a single pore using a confocal microscope.^{18,19} The relationship between the measured flow rate and pressure gradient appears to be quadratic (Fig. 4). Thus, the derivative of the quadratic equation fit to these data and evaluated at the average flow rate through each DPF was used to calculate the term $v/\Delta P$ for Equation (2).

Computational

Two computational domains representative of the RX Accunet filter were created: a filter geometry accurately modeling the different device components and a circular surface geometry representing a porous medium. The geometry of the device was generated

Figure 4 ♦ Flow rate as a function of pressure gradient through 4 DPFs tested in vitro and the average flow rate where permeability was calculated.

with GAMBIT (version 2.4.6, ANSYS, Inc., Canonsburg, PA, USA). Dimensions of the RX Accunet 6.0 DPF were obtained directly from the manufacturer, but were also verified by using digital calipers and images captured by a digital CCD camera mounted on an upright microscope.¹⁸ There was good visual agreement between an in-house photograph of the device used for the experiments and its computational representation (Fig. 5). In the second computational domain, the DPF implant was replaced by a circular surface perpendicular to the flow. Each model was embedded with complete wall apposition in a single tube representative of the ICA flow model used in the in vitro experiments.

The two computational domains were also meshed with GAMBIT using the native scripting language to control mesh size and density near the filter geometry (Fig. 6). Size functions were used to cluster more cells near the vessel wall and through the pores of the filter. Several meshes of increasing cell density were created to evaluate the independence of the simulation results with respect to mesh size (Table 2). The average inlet velocity in the z-direction was used as the parameter to determine mesh independence.

Steady, laminar flow was simulated in both models using the Fluent CFD solver (version 12; ANSYS). The pressure solver and momentum discretization algorithm were both

Figure 5 ♦ Comparison of (A) RX Accunet geometry and (B) computational domain.

second order. Pressure-velocity coupling was achieved using the SIMPLE algorithm. The convergence tolerance was set to 10^{-6} for continuity and the 3 velocity components. Blood was modeled as a Newtonian fluid with density 1.09 g/cm^3 and viscosity 3.5 cP to match the blood analog solution used in the experiments. Fluid flow simulations conducted for both geometries were run until convergence on a SGI Altix 4700 shared-memory NUMA system. The computational times and memory used for each simulation are summarized in Table 2.

The inlet and outlet boundary conditions were based on the experimental measures of pressure gradient. Due to the shorter vessel length of the computational domain compared to the bench-top phantom, the pressure gradient measured experimentally was scaled

based on the computational domain length. Thus, the scaled pressure gradient ($\Delta P = 596 \text{ Pa}$ or 4.47 mmHg) was applied to the RX Accunet model geometry using a constant, uniform inlet pressure profile and a zero traction outlet boundary condition. After simulation convergence, the computationally derived flow rate was compared to the experimentally measured flow rate. A trial-and-error process was used to redefine the inlet pressure to achieve the desired flow rate in the DPF model geometry. The resulting inlet and outlet pressure boundary conditions were then used in the circular permeable surface geometry simulations. The permeability of RX Accunet calculated from the experimental results was assigned as a porous medium boundary condition to the circular surface. Likewise, the computationally derived flow rate was compared to the experimentally measured flow rate to determine agreement. The permeability of the porous surface was then adjusted in a trial-and-error process to achieve the desired flow rate.

RESULTS

Experimental

The permeability of each DPF (Table 3) was calculated using Equation (2). Spider RX had the largest permeability and RX Accunet was the least permeable filter. For reference, the permeability of the DPFs is in the same order of magnitude of impervious materials, e.g., unweathered clay.²⁰

Computational

The average inlet z-velocity and magnitude of the shear stress vector projected onto the vessel wall were recorded for each mesh generated for the RX Accunet and porous surface geometries. A mesh sensitivity analysis was conducted by calculating the percentage change in this velocity and the average wall shear stress (WSS) relative to the densest mesh. The 8.6×10^6 cell mesh of the RX Accunet geometry had a 5.2% relative change in velocity and a 7.8% relative change in average WSS with respect to the 15.1×10^6

Figure 6 ♦ Wireframe representation of (A) entire filter geometry in cylinder and (B) filter cross section mesh.

cell mesh. Therefore, the 8.6×10^6 cell mesh was used for further analysis. The flow rate for this model was 257 mL/min (−1.9% difference from experimental data). The average inlet z-velocity and average WSS computed for the 9.3×10^6 cell circular permeable

surface mesh changed 3.8% and 9.0%, respectively, relative to the 15.1×10^6 cell mesh; thus, the 9.3×10^6 cell mesh was considered appropriate for this study. The flow rate for this model was 268 mL/min (+2.3% difference from experimental data).

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TABLE 2
Mesh Sizes (Number of Cells), Computational Time, and Gigabytes of Memory Used for the RX Accunet and Permeable Surface Models

RX Accunet Model			Permeable Surface Model		
Mesh size, $\times 10^6$	CPU Time, h	Memory, GB	Mesh size, $\times 10^6$	CPU Time, h	Memory, GB
1.2	0.89	8	0.67	1.33	4
1.5	1.80	8	1.0	3.46	4
2.1	1.80	8	1.5	4.79	4
3.6	5.01	8	2.8	10.4	4
7.4	11.7	12	5.2	25.3	4
8.6	16.9	8	7.0	13.1	12
11.1	19.1	16	9.3	33.3	8
15.1	34.1	12	15.1	32.0	16

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CPU: central processing unit.

TABLE 3
Permeability of 4 Distal Protection Filters

	Permeability, m^2
Spider RX	2.14×10^{-14}
FilterWire EZ	8.10×10^{-15}
RX Accunet	3.94×10^{-15}
Emboshield	1.26×10^{-14}

The inlet average z-velocity and pressure gradient calculated for the 2 computational domains were compared with the experimental measurements (Table 4). There was reasonable agreement between the experimental ($Q=262$ mL/min, $\Delta P=4.47$ mmHg) and the computational results for both DPF models (RX Accunet: $Q=257$ mL/min, $\Delta P=4.12$ mmHg; circular surface: $Q=268$ mL/min, $\Delta P=4.16$ mmHg).

The velocity field was evaluated near each DPF model by means of velocity vectors at cross sections perpendicular to the flow, proximal and distal to the filter and surface geometries (Figs. 7 and 8). The average velocity showed reasonable agreement between the 2 computational domains, exhibiting a parabolic shape proximal and distal to the filter and surface models (cross sections 1 and 8). However, there was a discrepancy evident in the local flow patterns within the DPF (cross sections 2-7); the velocity profile tended to flatten through the permeable surface while exhibiting disorganized flow patterns through the filter as the fluid exited the pores of RX Accunet. The in-plane velocities (V_x and V_y) were also evaluated at these 8 cross sections for both models, revealing no significant secondary flow patterns.

Regarding the magnitude of the WSS vector (Fig. 9) for both computational domains, the RX Accunet geometry yielded greater average and maximum WSS (1.56 Pa and 52.3 Pa, respectively) than the permeable surface geometry (1.29 Pa and 17.9 Pa, respectively). There were regions of nearly zero WSS where the flaps of the DPF apposed the vessel wall. WSS varied more along the length of the Accunet basket (the porous region of the filter) compared to the same

TABLE 4
Comparison Between Experimental and Computational Results

Model	Flow Rate, mL/min	Pressure Gradient, mmHg
Experimental	262	4.47
Computational (DPF)	257 (-1.9%)	4.12 (-7.8%)
Computational (permeable surface)	268 (+2.3%)	4.16 (-6.9%)

DPF: distal protection filter.

location in the permeable surface domain. This difference was due to a larger landing zone for RX Accunet device versus the infinitely thin permeable surface.

DISCUSSION

Although all distal protection filters do not have the most desirable design features of cerebral protection devices, previous investigators have found that filters with larger pore sizes and a wire mesh construction (such as the Spider RX) obstructed flow the least and performed the best in vitro under both steady and pulsatile flow conditions.^{10,21-23} In this investigation, we measured the permeability of 4 commercially available DPFs with the objective of providing a quantitative parameter that could be used as a design feature representative of DPF performance in vitro. To our knowledge, this parameter has not been measured previously; instead, most prior investigations reported the effect that DPFs have on pressure gradient,^{10,21,22} flow rate,^{21,22} and vascular impedance.²³ In previous work, we reported the porosity and pore density of 5 approved DPFs.^{18,22}

Hendriks et al.¹⁰ measured the pressure gradient and degree of flow reduction through 4 DPFs (Spider, FilterWire EZ, RX Accunet, and Angioguard) in a single 5-mm internal diameter tube setup. An input pressure of 70 mmHg was used to circulate a blood-like solution, resulting in an outflow of ~200 mL/min. In Hendriks' study, RX Accunet yielded a pressure gradient of 3.90 mmHg versus 4.47 mmHg measured in our study;

Figure 7 ♦ Velocity vectors color coded with the longitudinal velocity component (V_z) in the (A) RX Accunet model; (B) velocity contours. (C) Schematic illustrating location of the cross sections where the velocity field was evaluated. Units of velocity are m/s.

our computational model predicted a pressure gradient of 4.12 mmHg for a flow rate of ~260 mL/min. It should be noted that the distance between pressure transducers was not reported in the Hendriks' study.¹⁰

The order of increasing permeability (RX Accunet, FilterWire EZ, Emboshield, Spider RX) appears to be closely associated with the decreasing vascular resistance (RX Accunet, Emboshield, FilterWire EZ, and Spider RX)²¹

Figure 8 ♦ Velocity vectors color coded with the longitudinal velocity component (V_z) in the (A) permeable surface model; (B) velocity contours. (C) Schematic illustrating location of the cross sections where the velocity field was evaluated. Units of velocity are m/s.

and impedance (RX Accunet, FilterWire EZ, and Spider RX).²³ Roffi et al.¹³ found flow obstruction occurred most frequently in Angioguard XP than in FilterWire EZ or Spider RX. Although Angioguard XP was not included in our investigation, our previous findings

suggest the same trend of flow resistance as measured in vitro.^{18,21}

Previous studies have used CFD or finite element analysis to model intravascular devices.²⁴⁻²⁷ Computational modeling can aid the engineering design process when used as

Figure 9 ♦ Wall shear stress in the (A) RX Accunet model and the (B) permeable surface model. Units of wall shear stress are Pascal.

a tool for noninvasively evaluating device performance and optimizing design parameters without the need to follow a trial-and-error protocol with physical prototypes. For example, vena cava filters have been modeled to determine the efficacy of capturing blood clots²⁷ and for clot dissolution.²⁶ Cha et al.²⁴ modeled various packing densities of cerebral aneurysm coils as porous media. Recently, Conti et al.²⁵ used finite element analysis to simulate DPF deployment from its delivery sheath.

The modeling study pursued in the present work revealed good agreement (a discrepancy of up to 2.3%) between the computationally estimated average velocity and that measured in the bench-top experiments. However, the local flow patterns near the DPF were not adequately represented by modeling the filter as a circular permeable surface. In addition, the 2 computational domains had different WSS patterns at the site of DPF deployment. Most previous studies involving WSS and the carotid bifurcation attempt to correlate plaque burden with regions of low WSS. Recently, Teng et al.²⁸ measured the critical flow shear stress (CFSS) in ruptured and unruptured human carotid plaques using coupled finite element analysis and CFD. The measurements in the current study indicate that the average WSS of both computational domains is lower than the average CFSS for both ruptured and unruptured carotid plaques (9.294 Pa and 5.270 Pa, respectively). The decrease in WSS due to DPF deployment, as predicted by the RX Accunet computational model, should not

contribute to plaque formation in the ICA since filters are temporary implants deployed at an average of 20 minutes during CAS.²⁹

The in vitro setup is well-suited to study the effect of filter permeability on the bulk flow conditions in the ICA. Future generations of DPFs can be evaluated by using device permeability in a parametric evaluation of filter designs. Such a computational modeling study could be improved by using a paraboloid velocity profile at the inlet, pulsatile flow conditions, and a diseased patient-specific carotid bifurcation. To validate the results of the simulations, local flow patterns near the DPF can be compared to particle image velocimetry experiments, while wall shear stress can be validated using hot film anemometry. In addition, a circular surface perpendicular to the flow with the same pore density as RX Accunet could be modeled instead of using a permeable surface. Moreover, the permeability and pressure-jump coefficient for all approved DPFs should be modeled and compared to experimental results.

Conclusion

Following the aforementioned objectives of this investigation, the pressure gradient and flow rate in an ICA model in the presence of a DPF were quantified using a bench-top flow apparatus. These variables were adequately reproduced, within a reasonable margin of error, in a computational model of the vascular phantom with a deployed RX Accunet device. Moreover, it is feasible to generate

a computational model of the DPF as a porous medium analog; however, such a model does not yield the complex local flow patterns near the device, as compared to the actual DPF geometry. Nevertheless, the use of an experimentally estimated filter permeability led to a simplified methodology for computationally modeling DPFs by reproducing the ICA pressure gradient and flow rate induced by the device in the bench-top apparatus. Using CFD techniques to model DPFs can be useful to evaluate the effect filters have on local flow conditions in the ICA for a given parametric analysis of device permeability. The outcome of such analysis can provide valuable information on desirable filter design features for minimizing distal embolization and reducing flow disturbances in the distal ICA. The present investigation provides insight in the potential use of combined bench-top experiments and computational modeling for developing the next generation of DPFs.

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