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Finite Element Modeling of Endovascular Intervention Enables Hemodynamic Prediction of Complex Treatment Strategies for Coiling and Flow Diversion

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Abstract

Endovascular interventions using coil embolization and flow diversion are becoming the mainstream treatment for intracranial aneurysms (IAs). To help assess the effect of intervention strategies on aneurysm hemodynamics and treatment outcome, we have developed a finite-element-method (FEM)-based technique for coil deployment along with our HiFiVS technique for flow diverter (FD) deployment in patient-specific IAs. We tested four clinical intervention strategies: coiling (1–8 coils), single FD, FD with adjunctive coils (1–8 coils), and overlapping FDs. By evaluating post-treatment hemodynamics using computational fluid dynamics (CFD), we compared the flow-modification performance of these strategies. Results show that a single FD provides more reduction in inflow rate than low PD coiling, but less reduction in average velocity inside the aneurysm. Adjunctive coils add no additional reduction of inflow rate beyond a single FD until coil PD exceeds 11%. This suggests that the main role of FDs is to divert inflow, while that of coils is to create stasis in the aneurysm. Overlapping FDs decreases inflow rate, average velocity, and average wall shear stress (WSS) in the aneurysm sac, but adding a third FD produces

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Conflict of Interest Statement

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minimal additional reduction. In conclusion, our FEM-based techniques for virtual coiling and flow diversion enable recapitulation of complex endovascular intervention strategies and detailed hemodynamics to identify hemodynamic factors that affect treatment outcome.

Keywords

Intracranial aneurysm; flow diverter; flow diverter with adjunctive coils; virtual stenting; virtual coiling

1. Introduction

Endovascular intervention is emerging as the predominant means for treating intracranial aneurysms (IAs). In coil embolization, coils deployed in the IA sac cause thrombotic occlusion of the aneurysm. In flow diversion, flow diverters (FDs), which are densely-braided, mesh-like stents deployed across the IA orifice, redirect blood inflow away from the IA, thereby causing its thrombotic occlusion and parent vessel reconstruction. Despite the success of endovascular interventions, poor treatment outcomes have also been reported: 30% of coiled IAs experience aneurysm recurrence (Raymond et al., 2003), and 24% of FD-treated IAs fail to completely occlude at 6 months (Brinjikji et al., 2013). Since these strategies work primarily by modifying aneurysmal flow, post-treatment hemodynamics may play a critical role in treatment outcome. Computational tools for simulating clinical interventions and hemodynamics in patient-specific IAs may help evaluate the impact of various intervention strategies and identify hemodynamic factors that affect treatment outcome.

Several coiling and FD modeling techniques have been reported, but most of them employed simplified representations of device geometries and deployment (Appanaboyina et al., 2009; Byun and Rhee, 2004; Cebal and Lohner, 2005; Dequidt et al., 2008; Kakalis et al., 2008; Larrabide et al., 2008; Morales et al., 2011c; Schirmer and Malek, 2010). It is unclear if they can capture the wide variation of deployed device configurations to enable accurate post-treatment hemodynamic analysis which could explain treatment outcome variations.

To simulate accurate FD deployment, we established a FEM-based high fidelity virtual stenting (HiFiVS) technique (Ma et al., 2012; Ma et al., 2013) and validated it *in vitro* (Ma et al., 2013). HiFiVS has allowed us to investigate the hemodynamic modifications of emerging clinical strategies for FD treatment, for the first time (Ma et al., 2014; Xiang et al., 2014).

Expanding this modeling capability to encompass most current IA intervention strategies, we introduce a new FEM-based coil deployment technique. It explicitly models pre-shaped coils and their deployment mechanics. Using this coiling method and HiFiVS, we simulated four classes of clinical intervention strategies in a patient-specific IA: coiling (1–8 coils), single FD, FD with adjunctive coils (1–8 coils), and overlapping FDs. By evaluating post-treatment hemodynamics via computational fluid dynamics (CFD), we assessed hemodynamic modifications following treatment. Results demonstrated that our FEM-based simulations recapitulate complex endovascular treatment strategies and enable post-

treatment hemodynamic predictions that are supported by several clinical and *in vitro* deployment and flow studies.

2. Methods

2.1 Aneurysm Model

As a test case, we used a patient-specific internal carotid artery (ICA) aneurysm, which in reality was treated with three overlapping Pipeline Embolization Devices (PED, Covidien, Irvine, CA) (3.25×14mm, 3.25×12mm, and 3.25×10mm) and one adjunctive coil (7mm×30cm). The aneurysm and vascular geometry were reconstructed from 3D rotational angiography using the software VMTK (www.vmtk.org).

2.2 Workflow For Coil Deployment Modeling

We implemented the FEM-based workflow in Abaqus/Explicit v6.12 (SIMULIA, Providence, RI). In this coiling workflow, we incorporated aspects adapted from our HiFiVS technique for FD. The workflow (Fig. 1) consists of 3 parts: pre-processing, coil deployment, and post-processing.

Pre-Processing—We first generated the aneurysm sac and coil geometries for the coiling simulations. The aneurysm sac of the IA test case was isolated from its parent vessel in ICEM CFD v.14 (ANSYS, Canonsburg, PA) by creating a virtual surface across the neck plane, on which we created a small hole to allow for insertion of a microcatheter. This procedure eliminated the need to simulate clinically-difficult coil deployment strategies such as “jailing”, “coil-through”, or “coil-stent” (Spiotta et al., 2012; Wells-Roth et al., 2005). The volume of the isolated aneurysm was found to be 210 mm³.

Since most clinical coils are pre-shaped, also referred to as complex, 3D, or framing, we modeled a pre-shaped coil in this study. We generated the geometric centerline of this complex coil in MATLAB (MathWorks, Natwick, MA) based on previously-derived 3D parametric equations (Babiker et al., 2013a). We assumed the coil to be a continuous cylindrical structure, which was a reasonable simplification of the tertiary 3D shape (tightly wound metallic wires which form a primary cylindrical geometry) of real endovascular coils. Considering the aneurysm volume, the coils length and primary cylindrical diameter were chosen to be 11cm and 0.3mm respectively. Based on these dimensions, coil packing density (PD), defined as the percentage of the aneurysm sac volume occupied by the coil, was 3.7%.

FEM Simulation of Coil Deployment—The coil centerline was imported into the FEM simulation and discretized as 3D Euler-Bernoulli beam elements. To minimize coil-to-coil surface penetrations during deployment, coils were meshed with a resolution of 0.45mm (1.5 × the coil diameter, as described previously (Babiker et al., 2013a)). This resulted in a coil mesh size of approximately 245 beam elements. We assigned the coil a linearly-elastic constitutive relation with isotropic material properties similar to platinum, with a density of 0.0213 g/mm³, Young’s modulus of 7.5 GPa, and Poisson’s ratio of 0.39 (Babiker et al., 2013a).

Before deployment into the IA sac, we packaged the pre-shaped coil into a virtual microcatheter. The microcatheter was modeled as a 3D rigid discrete shell extrusion, with a length of 11.1cm and diameter of 0.5mm. The coil was pulled into the microcatheter from its stress-free, pre-shaped, configuration by aligning one end of the coil with the proximal end of the microcatheter and assigning a displacement boundary condition of 111mm. The microcatheter was fully constrained, and the FEM simulation was run under a 2s dynamic-explicit time-step. Contact was modeled using the “general contact” algorithm in Abaqus/Explicit. For coil packaging, tangential frictional behavior was simplified as frictionless and normal behavior was modeled as “hard”-contact.

Next, the coil was deployed into the aneurysm sac. Perpendicular to the neck plane surface, the microcatheter and packaged coil were forwarded slightly through the hole in the virtual neck plane. The microcatheter and aneurysm were assumed to be rigid and were constrained during the simulation. A displacement boundary condition of 111mm was applied to the proximal end of the coil to push the coil into the aneurysm. To prevent significant inertial effects, a 2s time-step was defined for deployment. Friction coefficients of 0.2 and 0.4 were defined for coil-to-coil and coil-to-aneurysm contacts, respectively (Babiker et al., 2013a). The internal strain energy produced by straightening the coil during packaging caused the coil to spring back to its complex shape when unsheathed into the aneurysm sac.

Post-Processing—While the FEM simulations utilized beam element discretization of the coil geometry centerline, CFD analysis required a solid representation of the coil in order to apply surface-based boundary conditions. To unambiguously define the surface of the coil, we swept the centerline of the deployed coil in the CAD software Creo Parametric v.2 (PTC, Needham, MA) with a circular profile of diameter 0.3mm to convert it to a 3D solid model. We then exported the solid coil model in the Parasolid CAD file format for subsequent CFD simulations. The 3D coil model was then merged with the original reconstructed vascular model.

2.3 FD Deployment Modeling

For FD deployment, we used the HiFiVS method described before (Ma et al., 2013). The HiFiVS workflow included all critical mechanical steps that affect the final FD configuration. The FD model in this study mimics the FDA-approved PED.

2.4 Different Clinical Treatment Strategies

Coiling—In total, 8 coils were sequentially deployed into the IA model, achieving 3.7%–29.6% PD for 1–8 coils. These simulations aimed to model the common clinical practice of using coils alone to treat IAs.

Single FD; FD with Adjunctive Coils—To simulate these two clinical strategies, we deployed a FD (3.25×14mm) using HiFiVS. We merged this result with results of 1–8 coils to simulate FD with adjunctive coils.

Overlapping FDs—As illustrated in Fig. 2, we simulated two independent FD deployments from the microcatheter, and superimposed the results to allow the FDs to

mechanically expand into one another. This avoided modeling computationally expensive FD wire-to-wire interactions which are inherent to the clinical procedure of overlapping FDs. Friction between overlapped FDs was assumed to behave the same as between wires within a single FD (Ma et al., 2013).

We simulated a total of three overlapping FDs ($3.25 \times 14\text{mm}$, $3.25 \times 12\text{mm}$, and $3.25 \times 10\text{mm}$). To evaluate apposition between overlapped FDs, we calculated 3D-volumetric dehiscence ratios, defined previously to quantify conformity between a deployed FD and vessel wall (Ma et al., 2012). Following a similar procedure, enclosed FD volumes were found by calculating their convex hulls in ICEM CFD, which were used to calculate FD-to-FD dehiscence ratios. A ratio of 0 signifies perfect conformity.

Sweeping Devices into 3D Models—The deployed FD wire strands and coil centerlines were swept with circular profiles of diameters 0.03mm and 0.3mm respectively to provide them continuous 3D surfaces.

Validation of Coil Packaging and Deployment—To validate the coiling simulation workflow results, we compared coil packaging and coil deployment simulations with *in vitro* deployments and *in vivo* deployment data (described in Supplementary Material Part 1).

2.5 Computational Fluid Dynamics Simulations

To evaluate the hemodynamic modifications of the four intervention strategies described above, we performed CFD using Star-CCM+ v.9 (CD-adapco, Melville, NY). Computational grids for each scenario were created using a unique meshing strategy (Supplementary Material Part 2). To obtain the aneurysmal flow fields, the Navier-Stokes equations were solved under the assumptions of steady state, laminar, and incompressible flow. Blood was modeled as a Newtonian fluid with a density of 1056 kg/m^3 and a dynamic viscosity of 3.5 cP. The vessel wall and devices were assumed to be rigid. All surfaces were defined as no-slip boundaries. A typical flow rate for the ICA was used (Xiang et al., 2011), giving an inlet velocity of 0.35 m/s. A traction-free boundary condition was prescribed at the outlet.

We investigated hemodynamic characteristics such as blood streamlines, intra-aneurysmal velocity magnitude and wall shear stress (WSS) distributions, inflow rate, aneurysm-averaged velocity, and aneurysm-averaged WSS.

3. Results

3.1 Device Deployment Results

Coiling—The coils sparsely covered the neck plane and lied along the periphery of the aneurysm sac at low PDs (<11%), but coil neck coverage increased and coils distributed more evenly in the aneurysm sac at higher PD (Fig. 3A).

Single FD—The deployed FD had uniform mesh density across the IA orifice (Fig. 4A).

FD with adjunctive coils—We merged the deployment results for the coil embolization scenarios with the single FD to simulate FD with adjunctive coils (Fig. 3B).

Overlapping FDs—We simulated the overlapping FDs strategy for up to 3 FDs, starting with the deployment results for the single FD (Fig. 4A). Visually, the mesh density across the aneurysm neck increased with deployment of successive FD. 3D dehiscence ratios between the 1st–2nd FDs and 2nd–3rd FDs were calculated to be 0.154 and 0.031 respectively.

3.2 Aneurysmal Flow Patterns

For all treatment scenarios, we plotted streamlines and velocity magnitude contours to visualize the intra-aneurysmal flow patterns. Results showed an impingement jet on the distal-side of the aneurysm in the untreated scenario (Fig. 4B and 4C), which created a vortex-like flow pattern.

Coiling—The coils disrupted the impingement jet observed in the untreated aneurysm (Fig. 3C and 3E). At lower PD (<11%), this flow pattern became less structured and the velocity of the flow decreased in the aneurysm sac, but flow continued to penetrate the aneurysm volume. At higher PD (>11%) however, less flow penetrated the aneurysm volume, as revealed by the velocity magnitude contours.

Single FD—The single FD (Fig. 4B and 4C) did not disrupt the vortex-like flow pattern observed in the untreated aneurysm but apparently decreased the velocity of the inflow and of the impingement jet.

FD with adjunctive coils—Trends in blood streamlines and velocity magnitude contours for FD with adjunctive coils (Fig. 3D and 3F) mimicked the trends with the coiling scenarios, but the impingement jet was weaker than that without the FD.

Overlapping FDs—The velocity of the impingement jet decreased in magnitude with deployment of successive FDs (Fig. 4B and 4C). However, velocity magnitude contours were similar between the 2 FDs and 3 FDs scenarios.

3.3 WSS Distributions

In all treatment scenarios, the placement of devices decreased WSS magnitude (Fig. 3G, 3H, and 4D) compared to the untreated scenario.

Coiling—Pockets of elevated WSS were observed near the neck plane (Fig. 3G), but overall, coils decreased the WSS magnitude in the aneurysm.

Single FD—The single FD decreased the WSS magnitude in the aneurysm compared to the untreated case (Fig. 4D).

FD with adjunctive coils—WSS distributions (Fig. 3H) were similar to those in the coils only scenarios (Fig. 3G).

Overlapping FDs—As shown in Fig. 6D, the deployment of successive FDs continually decreased the WSS magnitude.

3.4 Comparison of Aneurysmal Flow Reduction by Different Intervention Strategies

Quantitative flow modifications for coiling and FD with adjunctive coils is shown in Fig. 5. Compared to the untreated scenario, a reduction in aneurysm-averaged velocity and WSS was observed for all coiling scenarios.

We also plotted the relative values for the single FD (compared to the untreated case) for all three flow parameters in Fig. 5. It was clear from this comparison that the majority of the inflow reduction (therefore flow diversion) was caused by the FD and not the coils. However, adding coils to the single FD provided additional reduction in all flow parameters analyzed.

In the overlapping FDs scenarios (Fig. 6), a continual decrease in all flow parameters was observed with deployment of successive FDs. However, the decrease in flow parameters was marginal after deployment of the 3rd FD compared to the 2nd FD.

4. Discussion

Endovascular interventions — notably coil embolization and flow diversion — have become the predominant mode for treating IAs. Despite their success as minimally-invasive alternatives to open skull surgery, reports of IA recurrence and incomplete occlusion at treatment follow-up emphasize the uncertainty of treatment outcomes (Brinjikji et al., 2013; Raymond et al., 2003). Currently there is no way to predict endovascular intervention outcomes. Post-treatment hemodynamics is believed to be a critical factor in treatment outcomes. However, describing it requires the ability to perform realistic virtual intervention and flow analysis in patient-specific IA models.

To this end, we have developed FEM-based deployment techniques for FD (Ma et al., 2012) and coils (in this work). These simulation tools enable detailed post-treatment hemodynamics through recapitulation of complex endovascular intervention strategies.

4.1 Comparison with Previous Device Modeling Studies

Several previous *in silico* studies introduced device deployment methods and for some, post-treatment hemodynamics were investigated after virtual coiling or flow diversion (Appanaboyina et al., 2009; Babiker et al., 2013a; Byun and Rhee, 2004; Cebal and Lohner, 2005; Dequidt et al., 2008; Kakalis et al., 2008; Larrabide et al., 2008; Morales et al., 2011c; Schirmer and Malek, 2010). Overall, our hemodynamic results show qualitative (Fig. 3 and 4) and quantitative (Fig. 5) trends that are similar to previous findings. Coils tend to reduce intra-aneurysmal velocity, WSS, and disturb aneurysmal flow patterns, but their hemodynamic effects tend to stabilize at higher PD, similar to the findings of Morales et al. (2011a). On the other hand, FDs reduce aneurysmal inflow, velocity, and WSS, though not as profoundly as coils.

Since these methods employ simplifying assumptions in their device and deployment modeling, it is unclear if they can model the subtleties of deployed device configurations and capture the hemodynamic modifications that could account for uncertainties of treatment outcomes. Several coiling methods use artificial forces or mathematical space-filling algorithms to deploy coils (Babiker et al., 2013a; Cebal and Lohner, 2005; Dequidt et al., 2008; Morales et al., 2011c). Thus, they may produce unrealistic deployed coil configurations. Simpler coiling methods model coils as porous media or simple shapes in pursuit of computational efficiency (Byun and Rhee, 2004; Kakalis et al., 2008; Schirmer and Malek, 2010). However, the assumption of uniform coil distribution in the aneurysm sac is invalid except at high PD (Morales et al., 2013). Furthermore, a retrospective study of 255 coiled IAs found a non-significant difference between aneurysm recurrence and PD, with a mean PD of 27% for both recurrent and stable aneurysms (Piotin et al., 2007). This suggests that even if overall post-treatment hemodynamic modifications of coils tend to stabilize at higher PD, this effect may not be enough to characterize success or failure of treatment.

Similarly, the majority of FD deployment techniques use fast deployment via artificial force expansion, which may produce unrealistic deployed FDs (Appanaboyina et al., 2009; Larrabide et al., 2008). However, capturing subtle variations in deployed FD configurations may be important when considering post-treatment hemodynamics. For example, poor treatment outcomes can be caused by localized transition zones in the FD's porosity, which allow persistent aneurysmal inflow (Darsaut et al., 2013). In addition, FD conformity analysis in a previous (Ma et al., 2012) and current study demonstrate that deployed FDs do not completely conform to the vessel wall nor to each other in cases of overlapped FDs. This effect may be important to capture, since poor apposition of single and multiple deployed FDs could increase the risk of treatment complications such as endoleak, in-stent thrombosis, and a compromised vessel lumen diameter (Lylyk et al., 2009).

These factors highlight the importance of accurate device modeling, since post-treatment hemodynamics, and thus treatment outcome, may be dependent on localized aspects of deployed coil and FD configurations. Our FEM-based coiling method aims to provide mechanical insights to coil deployment, which could enable us to better understand the aneurysm healing process following intervention. In addition, our HiFiVS technique for virtual FD deployment incorporates the full mechanics that affect the final configuration of deployed FD. Taken together, our FEM-based virtual intervention tools address the need for accurate device modeling.

4.2 New Insight to the Hemodynamic Effects of Endovascular Treatment Strategies

Several clinically significant findings have emerged from our hemodynamic analysis of four classes of intervention: coiling, FD with adjunctive coils, single FD, and overlapping FDs. For the “classic” interventions using coils and FD only, we have found that a single FD accomplishes more inflow rate reduction than coiling (PD up to 30%) as shown in Fig. 5A. This interesting result is supported by *in vitro* observations (Babiker et al., 2013g). However, coils at any PD (up to 30%) result in more reduction in aneurysm-averaged flow velocity than a single FD (Fig. 5B).

The results of our *in silico* investigation of FD with adjunctive coils supports clinical observations of this emerging clinical strategy. Due to reports of incomplete occlusion and delayed rupture after FD placement (Brinjikji et al., 2013), it has been recommended that a small number of coils be added in the IA sac in conjunction with FD (Byrne and Szikora, 2012). A recent clinical report (Lin et al., 2015) found a higher occlusion rate and a lower risk of retreatment in patients treated with FD with adjunctive coils compared to patients treated with FD alone. Clearly, there is a keen need to understand how these treatment strategies differ in terms of hemodynamic modifications. We have found that the addition of coils at any PD (up to 30%) to a single FD reduces average intra-aneurysmal velocity and WSS beyond that achieved by the single FD alone. Furthermore, the addition of coils produced no further inflow rate reduction until the coil PD exceeded 11%. At low coil PD (<11%), adjunctive coils therefor may provide a scaffold for accelerating thrombus formation but have limited hemodynamic effect.

For the overlapping FDs scenarios, a continual reduction in the analyzed hemodynamic parameters was observed with deployment of successive FDs (Fig. 6), suggesting that overlapping FDs aid in acceleration of aneurysm occlusion. However, this benefit diminished beyond the placement of the 2nd FD, as marginal changes in these flow parameters were observed between the deployment of the 2nd and 3rd FD.

A concern with FD has been occlusion of vital perforators, since placement of a second FD may compromise the ostia of these vessels (Dumont et al., 2013). Our results suggest that addition of coils with PD>11% to a single FD may provide enhanced aneurysm occlusion. This may be especially important in circumstances where placing multiple FD may compromise adjacent perforator vessels.

5. Conclusion

We have developed a new FEM-based coil deployment method. Combining it with the HiFiVS technique for FD deployment, we are able to recapitulate a range of current clinical intervention strategies including coiling, single FD, FD with adjunctive coils, and overlapping FDs. Our modeling techniques have allowed us to investigate the hemodynamic effect of these intervention strategies in patient-specific IAs.

6. Limitations

Our study has several limitations. First, both the aneurysm wall and the devices were assumed rigid. Second, only steady-state CFD simulations were conducted. Third, caution is required in interpreting the hemodynamic results since only one patient-specific IA was examined.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Fig. 1. Workflow for coil deployment, including three main components: pre-processing, FEM simulation, and post-processing. Device deployment results enable hemodynamic analysis.

Fig. 2.

Workflow for deploying overlapping flow diverters. After **(1)** running HiFiVS to implant the 1st FD, we **(2)** isolated the parent vessel and scaled it down by a distance of 1~2 FD wire diameters to fit entirely inside the 1st deployed FD. Then, we ran a second HiFiVS simulation to **(3)** deploy the 2nd FD in the scaled-down vessel lumen. Finally, we **(4)** superimposed the two intermediate deployed FD simulation results, **(5)** removed the scaled-down vessel and ran one or more FEM simulations to allow the 2nd FD to expand into the 1st FD.

Fig. 3.

FEM modeling and CFD results for coiling (coils only) and flow diversion with adjunctive coils (coils + FD). 1C–8C represents 1–8 coils with packing density from 3.7–29.6% (each coil adds 3.7%) **A and B.** Deployment results. Each successively deployed coil is depicted with a unique color for clarity and to help visualize changes in their configurations with deployment of successive coils. **C and D.** Blood streamlines. Each vessel inlet was seeded with 100 streamlines for consistency. The flow was from right to left. **E and F.** Velocity magnitude contours. The white gaps represent the intersection of the aneurysm mid-plane

with the coils. **G and H.** Wall shear stress contours. They reveal zones of low WSS at the apex of the aneurysm sac as the number of coils deployed increases in both intervention strategies.

Fig. 4. FEM modeling and CFD results for a single flow diverter and overlapping flow diverters. **A.** FEM modeling results for 1, 2 and 3 flow diverters. Successively deployed flow diverters are colored by blue, green, and magenta. **B.** Blood streamline results for untreated aneurysm and treatment by 1, 2 and 3 FDs. Each vessel inlet was seeded with 100 streamlines for consistency. The flow was from right to left. **C.** Velocity magnitude contours on the aneurysm mid-plane **D.** Wall shear stress distributions.

Fig. 5.

Hemodynamic quantities for coils only, FD only, and coils + FD, relative to the untreated case (with a hemodynamic baseline of 100%). **A.** Inflow rate. **B.** Aneurysm-averaged velocity. **C.** Aneurysm-averaged wall shear stress. Aneurysm-averaged values are volumetrically-averaged in the aneurysm sac. Each coiling scenario (1C–8C) corresponds to a packing density ranging from 3.7%–29.6%.

Fig. 6.

Hemodynamic quantities (inflow rate, aneurysm-averaged velocity, and aneurysm-averaged wall shear stress) for overlapping flow diverters, relative to the untreated case. A general decrease of inflow rate, aneurysm-averaged velocity, and aneurysm-averaged wall shear stress was observed with successively deployed flow diverters, but the decrease is minimal from 2 FDs to 3 FDs.