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Computational fluid dynamics of cerebral aneurysm coiling using high-resolution and high-energy synchrotron X-ray microtomography: comparison with the homogeneous porous medium approach

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Abstract

Background—Computational modeling of intracranial aneurysms provides insights into the influence of hemodynamics on aneurysm growth, rupture, and treatment outcome. Standard modeling of coiled aneurysms simplifies the complex geometry of the coil mass into a homogeneous porous medium that fills the aneurysmal sac. We compare hemodynamics of coiled aneurysms modeled from high-resolution imaging with those from the same aneurysms modeled following the standard technique, in an effort to characterize sources of error from the simplified model.

Materials—Physical models of two unruptured aneurysms were created using three-dimensional printing. The models were treated with coil embolization using the same coils as those used in actual patient treatment and then scanned by synchrotron X-ray microtomography to obtain high-resolution imaging of the coil mass. Computational modeling of each aneurysm was performed using patient-specific boundary conditions. The coils were modeled using the simplified porous medium or by incorporating the X-ray imaged coil surface, and the differences in hemodynamic variables were assessed.

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Results—X-ray microtomographic imaging of coils and incorporation into computational models were successful for both aneurysms. Porous medium calculations of coiled aneurysm hemodynamics overestimated intraaneurysmal flow, underestimated oscillatory shear index and viscous dissipation, and over- or underpredicted wall shear stress (WSS) and WSS gradient compared with X-ray- based coiled computational fluid dynamics models.

Conclusions—Computational modeling of coiled intracranial aneurysms using the porous medium approach may inaccurately estimate key hemodynamic variables compared with models incorporating high-resolution synchrotron X-ray microtomographic imaging of complex aneurysm coil geometry.

INTRODUCTION

Intracranial aneurysm hemodynamics have been linked to aneurysm formation, growth, and rupture.¹ Aneurysm hemodynamics have been studied with computational fluid dynamics (CFD) modeling, which relies on computer reconstructions of aneurysm anatomy and blood flow. Previous reports have shown that the accuracy of CFD calculations is substantially improved by patient-specific anatomic models² and boundary conditions,^{3–5} and changes in aneurysm hemodynamics have been shown to predict treatment success or failure after endovascular coiling.⁶⁷ However, the computational modeling of aneurysm coiling often relies on the assumption that coils can be modeled as a homogeneous porous medium that fills the entire aneurysmal sac,⁸⁹ because coil configuration in aneurysms is geometrically complex and detailed information is not available from normal clinical imaging. The degree to which this technique influences CFD results is not known. Before incorporating hemodynamic calculations into clinical decision-making, an understanding of the influence of this assumption on hemodynamic metrics computed from blood flow simulations is necessary to further refine CFD modeling of coiled aneurysms.

Microtomography provides anatomic detail at a resolution (150 μm to up to 1 μm) not available with conventional CT technology, typically ranging between 2 mm and 400 μm . It has previously been applied to high-resolution visualization of aneurysm stents.¹⁰¹¹ There has been relatively little application of microtomography to the accurate modeling of intracranial aneurysm coils,¹² and no previous reports of applying microtomography from a high energy synchrotron beam to reduce the coils' beam-hardening artifact. We studied the difference between the homogeneous porous medium approach to coil modeling in CFD and CFD incorporating high-resolution synchrotron X-ray microtomographic imaging of coils deployed inside three-dimensional (3D) printed in vitro aneurysmal models.

METHODS

Patient population and endovascular measurements

Two patients with unruptured aneurysms who were undergoing endovascular aneurysm treatment were enrolled in this study, which was approved by the University's institutional review board. Three-dimensional rotational angiography was obtained immediately before and after treatment to create the patient-specific anatomic models. Blood flow velocity and pressure measurements were obtained using a dual-sensor pressure and Doppler velocity

guidewire and analysis workstation (ComboWire and ComboMap; Volcano Corp, San Diego, California, USA) every 5 ms at four peri-aneurysmal locations, as previously described.¹³ These measurements were used as patient-specific inflow boundary conditions in CFD calculations and for validation of CFD results at the other three measurement locations (outflow and two interior to the domain locations), as specified below.

Patient and aneurysm characteristics are shown in table 1. Both aneurysms were treated with stent-assisted coiling. Patient 1 was treated with a Neuroform EZ stent (Stryker Endovascular, Kalamazoo, Michigan, USA) and patient 2 with an Enterprise stent (Codman Neuro/DePuy Synthes, Raynham, Massachusetts, USA), and both were treated with Target coils (Stryker). Pressure and velocity measurements with the dual-sensor Doppler guidewire were successful in both patients. Endovascular device placement (stents and coils) was successful in both in vitro models.

Aneurysm model creation

Our method of creating patient-specific in vitro models of cerebral aneurysms has been previously described.¹⁴ Briefly, a 3D reconstruction of the vascular lumen was created by segmenting the images with Vascular Modeling Toolkit (<http://www.vmtk.org>) software. An acrylonitrile butadiene styrene model of the vessel lumen was created at 1:1 scale by a 3D printer and then cast in silicone rubber (OOMOO 25; Smooth-On, Macungie, Pennsylvania, USA). The acrylic model was then removed and this silicone mold was used to create an additional lumen model from casting wax (Freeman, Avon, Ohio, USA). The wax model was cast in a clear polyester resin (Clear-Lite; TAP Plastics, San Leandro, California, USA) and cured for 24 hours. Finally, the wax was melted away until only the final resin model of the aneurysm and parent vessel remained.

Models were then ‘treated’ by experienced neurointerventionalists. Each model was treated with the same endovascular devices (stents and coils) in the same sizes and sequence as the actual patients’ treatments. Every attempt was made to match the placement of these devices in vitro to the final angiographic conformation in vivo.

Synchrotron X-ray microtomographic imaging

The models were imaged by irradiation with a parallel and monochromatic X-ray beam (beam line ID19) at the European Synchrotron Radiation Facility (<http://www.esrf.eu>) in Grenoble, France. The X-ray beam energy was set to 222 keV in order to reduce beam-hardening effects observed with conventional CT and microtomography due to the strong absorption of the platinum content of the aneurysm coils. The model was rotated 360° while inside the beam and the X-ray radiation transmitted through the model was converted by a YAG 2000 scintillator into visible light that is imaged by a digital camera (PCO Edge; PCO, Kelheim, Germany). The resultant pixel size for patient models 1 and 2 was 25.86 µm and 3.3 µm, respectively. Pixel size, which is a function of optics and detector pixel size, was increased between patient 2 and patient 1 to allow for easier segmentation without loss of image quality. The obtained set of radiographs was then used to reconstruct the 3D coil microstructure using a filter back projection algorithm. Images were obtained by setting the camera as close as possible to the sample in absorption mode,¹⁵ which reveals the sample

microstructure provided that the sample constituents present significant differences in X-ray absorption. Coil segmentation was verified by comparing the volume of the numerical coil surfaces with the volume of the physical coils placed in each model.

Homogeneous porous medium computational modeling

The endovascular coil mass was first modeled as a homogeneous porous medium, as is standard in the literature.⁸⁹ The aneurysmal sac was numerically segregated from the rest of the computational domain using a reconstruction of the healthy parent artery¹⁶ and the entire aneurysmal sac was assumed to be uniformly packed with the homogeneous porous medium that represents the coils. To model the added flow resistance of endovascular coils, a pressure loss term with both inertial and viscous components was added to the momentum equation throughout the aneurysmal sac. Permeability and form factor, two geometric parameters, determined the contribution of viscous and inertial losses, respectively, with the overall pressure loss in the media. Coil permeability was determined using the capillary theory of Kozeny⁸ and form factor was determined using a commonly cited study in the literature¹⁷ that measured pressure loss across endovascular coils packed in a straight tube (see online supplementary material). These parameters were incorporated into the CFD simulation and the porous medium model was activated only in the aneurysm sac volume.

The entire geometry of the patient (aneurysm and parent vessel) was meshed in StarCCM+ (CD-adapco, Melville, New York, USA) using tetrahedral elements with a global grid size of 0.2 mm. Four boundary layer elements were used with a total layer thickness of 66 μm . The total number of elements for the porous medium models of patients 1 and 2 were 9.0×10^5 and 8.5×10^5 , respectively.

X-ray microtomography-based computational modeling

The 3D reconstructed surface of the endovascular coil microstructure obtained by synchrotron X-ray microtomography was manually placed inside the segmented lumen surfaces used in the creation of the in vitro models. The coil surface location was manipulated using MeshMixer software (<http://www.meshmixer.com>) such that the orientation of the coils inside the aneurysmal sac of the computational model matched that of the actual coils inside the scanned in vitro models. The merged surfaces were then converted to volumes and tetrahedral meshes were generated for each patient. A global mesh size of 0.2 mm was used in the parent vessel. The size of the elements used to discretize the surface of the coils was 20 μm , and these elements grew to 40 μm inside the remainder of the aneurysm volume.¹⁸¹⁹ Four boundary layer elements were used on both the lumen and the coil surfaces with the total layer being 66 μm and 20 μm thick, respectively. The final meshes for patients 1 and 2 consisted of 9.23×10^6 and 9.64×10^6 total elements, respectively.

CFD simulations

Finite volume fluid simulations were performed using ANSYS FLUENT (Release 14.1; ANSYS, Canonsburg, Pennsylvania, USA) and incorporated patient-specific boundary conditions obtained using the dual-sensor Doppler guidewire as previously described.¹³ One homogeneous porous medium model and one synchrotron X-ray microtomography model were created for each coiled aneurysm. Blood flow was assumed to be incompressible and

Newtonian with a density of 1050 kg/m^3 and a viscosity of 3.5 cP. At the inlet boundaries the time-dependent Womersley velocity profile was based on the in vivo patient-specific blood flow velocity measurements from the dual-sensor Doppler guidewire obtained in the proximal vasculature during the endovascular procedure. The systolic and time-averaged inlet flow rates for patient 1 were 191.93 mL/min and 133 mL/min, respectively. For patient 2, velocity inlet conditions were imposed at both the right and left vertebral arteries. The combined systolic and time-averaged flow rates in the basilar artery were 104 mL/min and 64 mL/min, respectively. Each patient contained one outlet at which a zero-pressure condition was specified. CFD simulations were run using the pressure implicit with splitting of operator pressure-correction method and second order pressure and time-stepping with a time-step of 5×10^{-4} s. Each simulation was run for three cardiac cycles. The first two cycles were discarded to ensure that the simulations were independent of initial conditions.

Values of wall shear stress (WSS), WSS gradient (WSSG), and oscillatory shear index (OSI) were averaged over the entire area of the aneurysm sac. Blood flow into the aneurysm (Q_{Aneurysm}) was calculated and viscous dissipation (ϵ), a measure of energy loss,²⁰ was integrated over the aneurysm volume (see online supplementary material). All hemodynamic parameters were calculated at both peak systole and averaged over one cardiac cycle, with the exception of OSI which was calculated only over the averaged cardiac cycle, according to its definition. Time averaging was done over 40 temporal instances throughout the cardiac cycle.

Comparisons of hemodynamic parameters between the two modeling modalities (porous medium vs X-ray microtomography) were performed. Additional comparisons of the percentage difference in hemodynamics when normalized to pretreatment values for each individual patient were also performed to provide context for the influence of different modeling modalities on the effect of aneurysm treatment.

RESULTS

X-ray microtomography-based models of the coils inside the aneurysmal sac were successfully generated and are shown in figure 1, along with each patient's 3D rotational angiography images used to create the lumen models.

Hemodynamic comparisons between CFD of models using X-ray microtomography coil reconstructions and the standard homogeneous porous medium to represent the coils are shown in figures 2 and 3, table 2, and the online supplementary table. There were substantial qualitative and quantitative differences in most hemodynamic variables between the two approaches for both patients. Visual inspection of streamlines (figure 2) shows more complex peri- and intra-aneurysmal flow in the X-ray microtomography models than in the porous medium models in both patients. The homogeneous porous medium approximation led to overestimation of the flow rate into the aneurysmal sac, Q_{Aneurysm} , while underestimating OSI and ϵ (figure 3).

In patient 1 the hemodynamics with a porous medium approximation in the coiled sac underestimated WSS and overestimated WSSG while, in patient 2, it overestimated WSS

and underestimated WSSG. These differences were seen in both the peak systolic and time-averaged conditions (table 2). Similarly divergent results were seen when hemodynamic changes were normalized to pretreatment values (see online supplementary table), where porous media models substantially exaggerated the treatment effect of coil embolization on WSSG and OSI.

DISCUSSION

Previous studies have demonstrated significant risk of aneurysm recanalization or recurrence based on immediate elevation of post-coil WSS at the aneurysm neck.⁶⁷ Other work has shown that OSI is elevated at the rupture point of untreated aneurysms, as it represents the degree of change in shear force over the cardiac cycle.²¹ Small changes in modeling techniques and boundary conditions can have a substantial effect on such hemodynamic calculations.⁴ Thus, understanding the limitations of existing approaches, as well as attempting to improve CFD accuracy, has a potential clinical impact. The improvement in CFD software and hardware, as well as an expanding body of literature relating CFD calculations to clinical outcomes,^{22–24} suggests that CFD information may soon be incorporated in clinical decision-making. It is incumbent on physicians treating these aneurysms that both the advantages and limitations of such studies are understood before applying their results to patient care.

We have shown that high-resolution synchrotron X-ray microtomography of patient-specific in vitro models of coiled aneurysms can be incorporated into CFD modeling. By comparing these high-fidelity geometric reconstructions of the coil configuration inside the aneurysmal sac with coils approximated using the standard homogeneous porous medium approach, we have identified possible sources of improvement on the treatment of coils, beyond the current homogeneous porous medium that fills the entire aneurysmal sac. This could have an effect on the accuracy of hemodynamic calculations of coiled aneurysms. By using patient-specific aneurysm geometry and physiological boundary conditions, we have attempted to create the most realistic CFD model possible for each condition in an effort to isolate the effects of the different modeling strategies on hemodynamic calculations.

Varying techniques have been employed to approximate the effects of intra-aneurysmal coils on aneurysm hemodynamics in CFD modeling. Early reports assumed that coils act as a solid barrier and velocity and flow rate drop to zero at the vessel–coil interface,⁶⁷ which is computationally economical but does not account for interstitial flow between coil wires. Others have employed either finite-element modeling²⁵ or virtual coil placement,²⁶ both of which attempt to recreate a more precise coil mass configuration inside the sac. However, these methods are both computationally intensive and rely on idealized coil geometry, which does not take into account either the complexity of the actual coil shape or its interaction with the aneurysm geometry or existing coils.

The homogeneous porous medium approach is a widely employed method of estimating the effect of aneurysmal coils because it is computationally efficient and grid-independent.^{8927–29} This technique is performed by determining the overall porosity of a coil mass based on the summative effects of diameter, length, and volume of each and every

coil. By imposing that the full volume of the aneurysm sac is fully filled with the total coil mass, the porosity, permeability, and pressure loss factor of the coil mass can be estimated and imposed on the spatial discretization via porous medium-like terms in the momentum equations.

The homogeneous porous medium approach substantially overestimated intra-aneurysmal flow while underestimating OSI and viscous dissipation in the current study. Calculations of WSS and WSSG were unreliably overestimated or underestimated in the two patients using this approach, and the effects of treatment on WSSG and OSI were exaggerated. Previous in vitro work has shown that higher packing density and greater coil uniformity reduces coil mass permeability.²⁵³⁰ The discrepancy seen in the current study may result from an overestimation of porous medium permeability related to the simplified approximation of the coil mass being homogeneously distributed throughout the entire aneurysmal sac volume, with the coils presenting a fixed cross-sectional configuration perpendicular to the flow (for purposes of calculating the pressure loss coefficients).³¹ Thus, the coil mass in this method does not impact the direction of flow entering the aneurysm. In an actual coil mass, as imaged by X-ray microtomography, the complexity of the spatial distribution and orientation of the coil may induce areas of complex heterogeneous velocity fields with higher shear and lower flow. This complexity is not captured by the homogeneous porous medium method but is responsible for the increased OSI and ϵ seen in the X-ray microtomography-based simulations; both variables would be expected to increase with higher shear and greater temporal and spatial flow variation within the aneurysm sac. While refinement of the porous medium method could reduce such discrepancies, a consistently accurate representation of the complexities of coiled intra-aneurysmal blood flow may not be possible using such a method.

One previous study of aneurysm coils using commercially available microtomography was not able to resolve the complex 3D shape of the coil mass due to this beam-hardening artifact,¹² and another microtomography study of endovascular cerebral aneurysm treatment in the literature has focused mainly on the effect of aneurysm stents.¹⁰¹¹³² By employing synchrotron X-ray microtomography, beam-hardening artifacts were eliminated in our 3D model of the aneurysm coil mass, permitting us to reconstruct the complex geometry of the coils into a high-resolution complete surface and thus directly incorporate them into CFD models. This type of coil imaging, surface reconstruction, and modeling provides a 'ground truth' for coil hemodynamic effects against which to compare the effects of other coil CFD models, and in particular the standard homogeneous porous medium.

This study has several limitations. First, the small number of aneurysms modeled precludes us from precisely quantifying the effects of the homogeneous porous medium approach on CFD models of coiled aneurysms. However, our results do suggest that such effects could be quantified by refining and expanding our comparative approach with a larger dataset. Second, neither synchrotron X-ray microtomography nor standard microtomography are widely available for routine CFD study of in vitro coiled aneurysm models. However, the advantages of microtomography as the highest resolution imaging of complex coil geometry can be used as a comparison to evaluate the degree of inaccuracy introduced by other approaches to coil modeling, and to compensate for them with possible corrective factors or

refinement of the technique. This study lays the groundwork for developing more accurate modeling techniques to inform clinical decision-making of aneurysm treatment using CFD simulations.

CONCLUSION

The homogeneous porous medium approach to modeling aneurysm coils in CFD simulations may be a source of significant inaccuracies in hemodynamic computations. High-resolution synchrotron X-ray microtomography of complex aneurysm coil geometry can be successfully incorporated into patient-specific CFD models and may improve the accuracy of hemodynamic results.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1. High-resolution synchrotron X-ray microtomography of aneurysm coils in vitro (right panel) for patients 1 (A) and 2 (B), inserted into three-dimensional rotational angiography models of patient-specific vessel and aneurysm anatomy (left panel). Individual surface characteristics of each coil, as well as the complex geometry of the coil mass, can be appreciated without beam-hardening artifact.

Figure 2. Systolic streamline analysis demonstrates the differences in intraaneurysmal and perianeurysmal flow using synchrotron X-ray microtomography (left) and porous medium (right) coil modeling in patients 1 (A) and 2 (B).

Figure 3.

Comparison of hemodynamic effects of synchrotron X-ray microtomography (left columns) and porous medium (right images) computational fluid dynamics models of coiled aneurysms in patients 1 (left panel) and 2 (right panel). Wall shear stress (A) and wall shear stress gradient (B) show opposing differences between the two modeling approaches, while oscillatory shear index (C) and viscous dissipation (D) are consistently underestimated and intra-aneurysmal flow (E) is consistently overestimated in the porous medium models compared with microtomography models. All parameters are shown in the peak systolic condition with the exception of oscillatory shear index, which is averaged over one cardiac cycle. Intra-aneurysmal flow is measured as it crosses a two-dimensional plane defined as the aneurysm neck, while viscous dissipation is shown as a two-dimensional cross-section oriented through the maximal diameter of the aneurysm.

Table 1

Patient and aneurysm characteristics

Patient	Aneurysm location	Aneurysm volume (mm ³)	No of coils	Total coil length (cm)	Packing density (%)
1	Para-ophthalmic	26	3	11	20.3
2	Mid-basilar	86	3	35	18.2

Comparison of hemodynamic effects of porous media and synchrotron X-ray microtomography computational fluid dynamics models of coiled aneurysms

Table 2

Hemodynamic variable	Patient	Porous media			Microtomography			Difference (%)	
		Cycle-averaged	Peak systole		Cycle-averaged	Peak systole		Cycle-averaged	Peak systole
WSS (Pa)	1	0.312	0.623		0.426	0.8021		-31.01	-25.14
	2	0.081	0.132		0.048	0.091		51.45	37.24
WSSG (Pa/m)	1	7280	12 700		4250	8460		52.45	40.02
	2	186.25	300.52		442.60	807.53		-77.64	-91.51
OSI	1	0.0078	NA		0.0172	NA		-39.74	NA
	2	0.0153	N/A		0.0717	N/A		-129.27	N/A
Q _{Aneurysm} (mL/min)	1	10.25	18.95		7.68	15.69		28.68	18.81
	2	8.08	12.51		3.08	5.06		89.56	84.75
e (mW)	1	8.01×10 ⁻⁴	2.44×10 ⁻³		1.87×10 ⁻³	5.07×10 ⁻³		-79.82	-70.02
	2	1.27×10 ⁻⁴	3.08×10 ⁻⁴		1.30×10 ⁻⁴	3.55×10 ⁻⁴		-2.46	-13.45

OSI, oscillatory shear index; Q_{Aneurysm}, intra-aneurysmal flow; WSS, wall shear stress; WSSG, WSS gradient; e, viscous dissipation.