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Finite element modeling of embolic coil deployment: Multifactor characterization of treatment effects on cerebral aneurysm hemodynamics

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ABSTRACT

Endovascular coiling is the most common treatment for cerebral aneurysms. During the treatment, a sequence of embolic coils with different stiffness, shapes, sizes, and lengths is deployed to fill the aneurysmal sac. Although coil packing density has been clinically correlated with treatment success, many studies have also reported success at low packing densities, as well as recurrence at high packing densities. Such reports indicate that other factors may influence treatment success. In this study, we used a novel finite element approach and computational fluid dynamics (CFD) to investigate the effects of packing density, coil shape, aneurysmal neck size, and parent vessel flow rate on aneurysmal hemodynamics. The study examines a testbed of 80 unique CFD simulations of post-treatment flows in idealized basilar tip aneurysm models. Simulated coil deployments were validated against *in vitro* and *in vivo* deployments. Among the investigated factors, packing density had the largest effect on intra-aneurysmal velocities. However, multifactor analysis of variance showed that coil shape can also have considerable effects, depending on packing density and neck size. Further, linear regression analysis showed an inverse relationship between mean void diameter in the aneurysm and mean intra-aneurysmal velocities, which underscores the importance of coil distribution and thus coil shape. Our study suggests that while packing density plays a key role in determining post-treatment hemodynamics, other factors such as coil shape, aneurysmal geometry, and parent vessel flow may also be very important.

1. Introduction

Cerebral aneurysms are pathological, sac-like dilations of blood vessel walls in the brain that commonly occur at major branch points in the Circle of Willis (Brisman et al., 2006). When a cerebral aneurysm ruptures, the subsequent damage is lethal in 45% of cases (Bederson et al., 2009). Accordingly, it is critically important to treat cerebral aneurysms effectively either before or very shortly after rupture. Endovascular coiling is the most common treatment both before and after rupture (Lin et al., 2011). The treatment consists of deploying a sequence of embolic coils of different shapes and sizes into the aneurysmal sac, with the intent of filling the sac and thereby reducing aneurysmal inflow. Reducing aneurysmal inflow may initiate subsequent thrombosis within the sac, leading to occlusion of the aneurysm and its eventual exclusion from circulation, which is one definition of a successful treatment outcome (Piotin et al., 2007). Other factors, including coil material, may also influence aneurysmal thrombosis (White et al., 2008b).

To facilitate effective endovascular coiling, physicians typically target a high packing density (defined as the percentage of the aneurysmal volume occupied by coils). A wide range of coils with different stiffness, lengths, shapes, and sizes can be added to the deployment sequence in order to achieve that goal among others (e.g., coil stability). The coils are constructed from a thin metal wire that is wound into a secondary helical structure as shown in Fig. 1a, which is then shaped into a tertiary structural configuration. Coils can have many different tertiary structures or "shapes". The two most common are helical and complex, which are illustrated in Fig. 1b. Helical coil structures take the form of a

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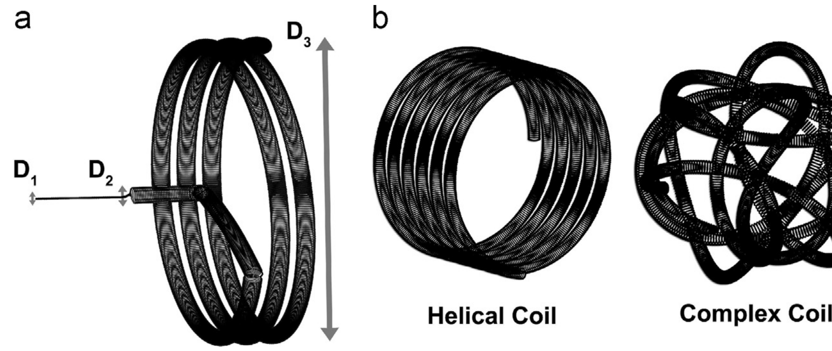

Fig. 1. Illustration of coil structural characteristics showing the metal wire diameter D_1 (ranges from 0.00175 to 0.003 *in*), the helical wind diameter D_2 (ranges from 0.010 to 0.020 *in*), and the diameter of the tertiary shape D_3 (a). Examples of helical and complex tertiary coil shapes (b).

helix, while complex structures take on a spherical shape. Coil shape and shape size (D_3 in Fig. 1a) influence aneurysmal filling and coil distribution in the sac, while the thickness of the metal wire (D_1 in Fig. 1a) and the diameter of the helical wind (D_2 in Fig. 1a) determine coil stiffness (White et al., 2008a).

Packing density is currently the only coil deployment parameter that has been clinically correlated with aneurysmal occlusion. However, many studies have reported occlusion at low packing densities and recurrence at high packing densities (Piotin et al., 2007). Such reports indicate that other factors, including coil distribution, may influence treatment success. Specific effects remain poorly understood, however, due in part to limitations of *in-vivo* flow measurement techniques in the context of coiled aneurysms.

Previous *in vitro* studies have quantified hemodynamics in coiled aneurysms, but have been subject to considerable limitations. For example, *in vitro* studies have not investigated the role of coil distribution on post-treatment hemodynamics and have only considered a small, usually statistically insignificant, number of cases because of the costs associated with embolic coil experiments (Babiker et al., 2010; Sorteberg et al., 2004). Conversely, *in silico* studies have simulated many coiled cases but have relied on unrealistic assumptions to simplify complex coil geometries. For example, coils have been modeled as a homogenous porous media, as a single sphere inside the aneurysmal sac, as perfectly helical tubes, and as cylinders with random trajectories inside the sac (Mitsos et al., 2007; Byun and Rhee, 2004; Schirmer et al., 2010; Morales et al., 2011).

We present an *in silico* study that uses a novel finite element (FE) approach and computational fluid dynamics (CFD) to simulate post-treatment hemodynamics in idealized models of basilar tip aneurysms. The FE approach considers the physical properties of coils and applies structural dynamics to simulate coil deployment. The approach is used to evaluate the influence of coil shape, aneurysmal neck size, packing density, and parent vessel flow on post-treatment hemodynamics in a testbed of 80 unique cases. This study represents the first time, to the authors' knowledge, that deployment mechanics and the physical properties of coils have been considered in studying the post-treatment hemodynamics of coiled aneurysms.

2. Methods

2.1. Model construction

Two idealized computational models of basilar tip aneurysms were designed in Solidworks (SolidWorks, Concord, MA): a narrow-neck (nneck) and wide-neck (wneck) model with dome-to-neck-width ratios of 1.5 and 1.1, respectively. The dimensions of both models are presented in Fig. 2. The dome-to-neck-width ratios were chosen to represent different types of aneurysm cases commonly encountered in the clinic: a narrow-neck case that would be considered for treatment with coils alone and a wide-neck case that requires either a stent or balloon during treatment (Brinjikji et al., 2009).

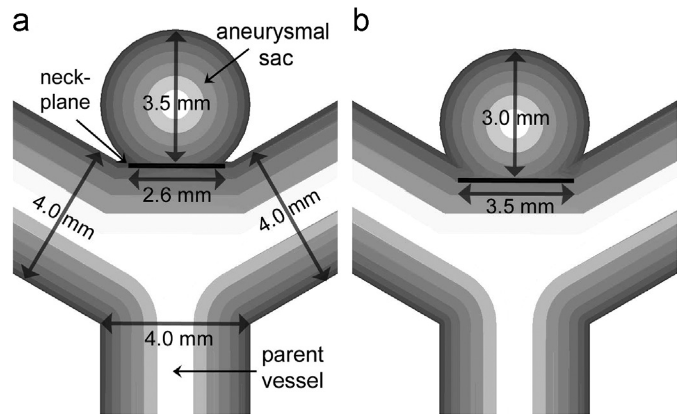

Fig. 2. Nneck (a) and wneck (b) model dimensions. Detail on model construction can be found in Babiker et al. (2010).

2.2. Modelling virtual coil deployment

Embolic coils were modeled using three-dimensional (3D) beam theory, which is similar to the approach used by Dequidt et al. (2009). Each coil was represented by a set of serially linked 3D Timoshenko beam elements in Abaqus (Simulia, Providence, RI). The beam elements were assumed to be elastic with 92% platinum and 8% tungsten material composition (White et al., 2008a), which resulted in an elastic modulus of 7.5 GPa and a density of 21.3 g/cm^3. A Poisson ratio of 0.39 was also prescribed after approximating the coils as solid beams composed primarily of platinum. This simplification followed the assumption that the coil stock wire was tightly and perfectly wound and that individual helical winds could not be stretched. Further, the cross-sectional plane of the beam element was assumed to remain plane and undistorted during deployment.

Two coils were modeled: A long framing coil with an 8 cm length and a 0.31 mm diameter (D_2), and a shorter coil with a 2 cm length and a 0.27 mm diameter. The coils were virtually placed in a 0.4 mm diameter rigid microcatheter and a rigid ellipsoidal balloon was used to constrain coils within the aneurysmal sac during deployment, which is similar to the balloon-assisted technique used *in vivo*. The balloon was modeled according to the shape of the physical balloon used in previous *in vitro* experiments conducted by our group (Babiker et al., 2010), which facilitated comparisons between computational and experimental results. More detail on the balloon is provided in the Appendix.

The coils were discretized with a mesh resolution of $1.5 \times D_2$; finer mesh resolutions resulted in considerable overclosure between adjacent coil loops. Parametric equations were applied through a subroutine to specify coil shape and loop size; concentrated load forces were exerted on the beam element nodes in each Cartesian direction, as shown in Fig. 3. The parametric equations were derived by estimating the force required to displace a beam element by $D_3/2$ and scaling/applying that force to the set of beam elements using a non-uniform distribution that modeled coil shape. Two non-uniform distributions were modeled: a complex and a helical distribution. The complex distribution was modeled as a 3D curve with multiple helical loops rotated around a sphere at different angles, which is similar to the physical geometries of complex coils as described in Hung et al. (2005). The helical distribution was simply modeled as helical loops perpendicular