Development of an Immersed Boundary Method for Pulsatile Flow Predictions in Cerebral Aneurysms

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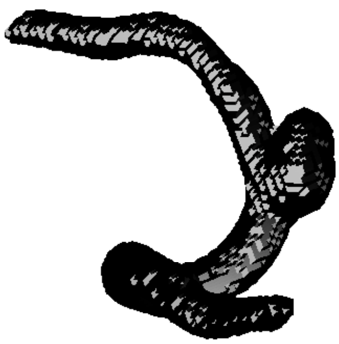
Introduction

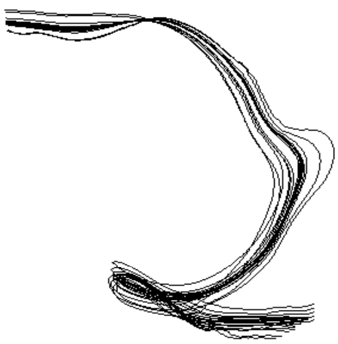
We present an immersed boundary (IB) method for the simulation of pulsatile blood flow in realistic cerebral aneurysms. The medical need is to obtain patient-specific predictions of the flow and forces inside aneurysms. This constitutes a challenging computational multiscale modeling problem, aimed at understanding and identifying the likelihood of long-time rupture from an analysis of short-time pulsatile flow. We focus on effects due to changes in the flow regime, expressed by the Reynolds number (*Re*) and study pulsatile flow at *Re* = *100*, *250* and *400*. This range corresponds to physiological cerebral flows. We analyze time-dependent flow patterns and show the shear stress response to pulsatile forcing.

Materials and Methods

The numerical model for the simulation of blood flow through vessels in the human brain is based on the incompressible Navier-Stokes equations in 3D. An IB method using volume penalization is applied to capture flow in the aneurysm geometry [Mittal and Iaccarino, 2005]. In our IB approach, complex geometries are represented on a 3D Cartesian grid by a 'masking function', which takes the value '*0*' inside fluid parts of the domain and '*1*' inside solid parts of the domain. This technique allows a 'direct transformation' of voxel-based medical imagery into a masking function, without additional steps as smoothing, local grid refinement, body-fitted grid etc. In realistic cases some reconstruction process involving segmentation and cutting away small side branches would be needed, as we are interested in the main vessels leading to, and containing aneurysms.

The example aneurysm geometry, which we are working with, was reconstructed from medical images obtained using rotational CTA [Gambaruto et al., 2011]. We convert the CTA aneurysm surface into the masking function (Fig.1a), which gives a fast approximation of the 3D-geometry. The flow inside the vessels is simulated with the use of a skew-symmetric finite-volume discretization and explicit time-stepping [Verstappen and Veldman, 2003].


(a)


(b)

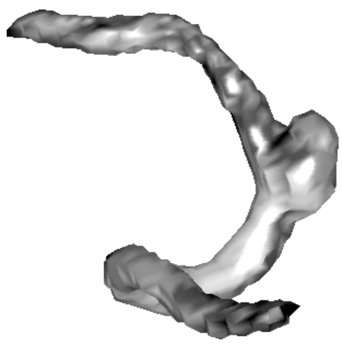

(c)

Figure 1: Three-dimensional geometry of a realistic aneurysm reconstructed from CTA data: masking function (a), velocity streamlines (b) and shear stress distribution (c).

Results and Discussion

Our IB method was validated for laminar Poiseuille flow in a pipe at *Re* = *250*. For a model aneurysm and several vessel geometries we also simulated pulsatile flow at Reynolds numbers in the range *Re* = *100 – 500* to validate the method. We proposed numerical bounding solutions to analyze the sensitivity of the numerical solution to uncertainty in the masking function [Mikhal and Geurts, 2011].

Some of the velocity streamlines of a steady flow computed at *Re = 250* are presented in Fig.1b, showing the extent by which the flow penetrates the aneurysm. For the same flow conditions we illustrate the normalized shear stress over the aneurysm geometry (Fig.1c). We observe high stresses (dark areas) near the rims of the geometry, e.g., near the aneurysm neck, while low shear stresses (light areas) are clearly visible in the aneurysm cavity.

Switching to pulsatile flow in the same aneurysm geometry we include realistic pulsatile flow forcing, for which the mass-flow rate is a pulsatile wave at the cardiac frequency. This pulse signal was measured inside the middle cerebral artery by using Transcranial Doppler sonography. The normalized signal during one cardiac cycle is shown in the inset of Fig.2.

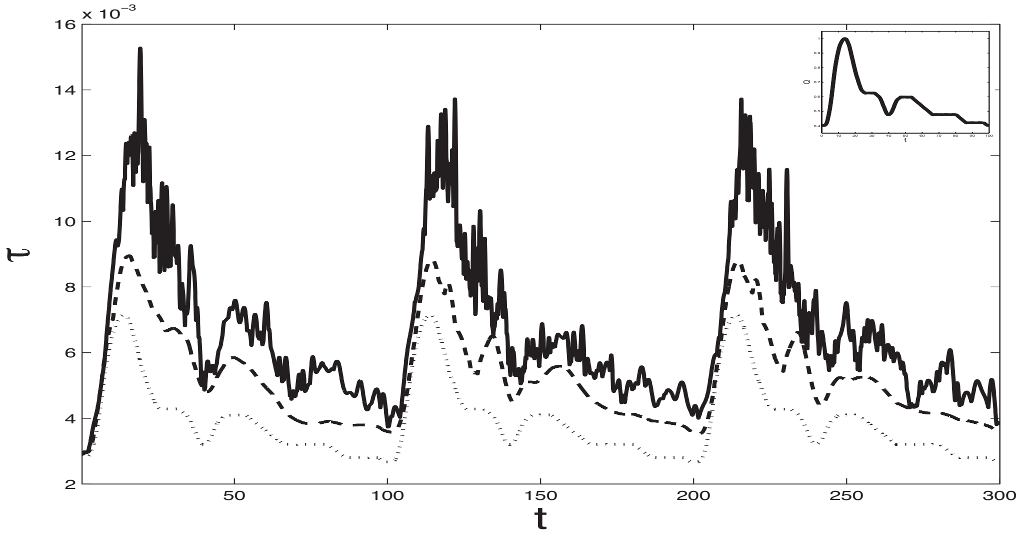


Figure 2: Pulsatile response of the maximum shear stress τ at different flow regimes: *Re = 400* (solid), *Re = 250* (dash) and *Re = 100* (dot). The inset shows one period of the mass flow rate *Q* versus time *t*.

In Figure 2 we present the shear stress response for different flow regimes. At *Re = 400* the Navier-Stokes unsteadiness dominates, while for *Re = 250* and *Re = 100* the imposed pulsatile pattern is still clearly present. The average stress levels were found to decrease with decreasing Reynolds number.

Conclusions

Time-dependent flow was computed and analyzed for different flow regimes, corresponding to physiological cerebral flows. The evolution of the shear stress was shown in response to the imposed pulsatile forcing. A transition to complex time-dependency was observed with increased aneurysm size, which might be linked to possible reasons of the aneurysm rupture.

Acknowledgements

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