On-line Appendix

CFD Simulation

Reconstruction of an aneurysm models was based on 3D DSA images of patients with an ICA aneurysm, and these images were obtained from an Axiom Artis dBA (Siemens). The 3D DSA images at the isotropic resolution between 0.1 and 0.14 mm were segmented with Volview (Kitware, Clifton Park, New York). Each aneurysm model started from the ICA and included the ICA bifurcation. A long section of the ICA proximal to the aneurysm ensured that the intra-aneurysmal flow would not be affected by the inflow boundary condition.

A commercial CFD package FLUENT (ANSYS) was used for flow analysis. An ICA waveform based on the first patient, measured by phase-contrast MRA, was used at the inlet for all cases. The mean flow rate at the ICA was 4 mL/s and peak flow rate was 5.4 mL/s. The flow rate ratio between the anterior and middle cerebral arteries was based on their sizes so that the mean wall shear stresses at these branches were the same. Each period consisted of 100 time-steps and each case was repeated for 3 periods using an implicit second-order approach. At each time-step, iteration continued until residues were reduced by 4 orders of magnitude. Grid-independence tests were conducted for each case to ensure that the difference in the intra-aneurysmal flow was less than 3%.

The aneurysm neck was first defined for each model and the amount of intra-aneurysmal flow rate at the neck was computed at each time-step. Because there was no net flow at the neck (no accumulation of blood within the aneurysm), only the flow into the aneurysm was used. The intra-aneurysmal flow rate was used to determine the inflow ratio (f), the ratio of intra-aneurysmal flow, and the flow rate at the parent vessel. The inflow ratio was averaged over a cardiac cycle to obtain a time-averaged value, which was later compared against the inflow ratio estimated by the high-frame-rate cerebral angiography.

Imaging Processing

On the logarithmically subtracted images, the signal is proportional to kcz, where k is a constant, c the concentration, z the penetration depth. The total signal from a given volume (after digital subtraction logarithmically) will be $S = \iint_{2D} kcz\, dx\, dy$, which can be written as $S = \iint_{2D} kcz\, dx\, dy = \iiint_{3D} kc\, dx\, dy\, dz = k\bar{c}V$, where V is the volume of the region of interest on 3D and x-y plane is the 2D projection plane. Thus, the average signal is proportional to the mean concentration ($\bar{c}$) after normalization by the imaging volume. This normalization process is independent of the shape of region of interest as long as the region of interest is small enough that the concentration is uniform. This normalization is especially important when the segments of the parent vessel proximal and distal to the aneurysm are pointing in 2 different directions.

One may choose to normalize the signal by the penetration depth at every pixel, but this process may require an accurate coregistration of 2D and 3D DSA images and determination of penetration depth at every pixel. Because we are interested in the mean concentration at the region of interest, our approach greatly simplifies the process involved. Our technique, therefore, has used the penetration depth of x-ray indirectly. For cases of small residual flow into a treated giant aneurysm, we then use the entire vector space for the null space of aneurysm signal. This is based on the fact that it will be very difficult to measure the signal at an aneurysm or the neck once the coils are deployed into the aneurysm, so a comparison of signals proximal and distal to the aneurysm is a better approach than a direct approach to measure aneurysm signal.