Impact of Side Branches on the Computation of Fractional Flow in Intracranial Arterial Stenosis Using the Computational Fluid Dynamics Method

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Background: Computational fluid dynamics (CFD) allows noninvasive fractional flow (FF) computation in intracranial arterial stenosis. Removal of small artery branches is necessary in CFD simulation. The consequent effects on FF value needs to be judged. *Methods:* An idealized vascular model was built with 70% focal luminal stenosis. A branch with one third or one half of the radius of the parent vessel was added at a distance of 5, 10, 15 and 20 mm to the lesion. With pressure and flow rate applied as inlet and outlet boundary conditions, CFD simulations were performed. Flow distribution at bifurcations followed Murray's law. By including or removing side branches, five patient-specific intracranial artery models were simulated. Transient simulation was performed on a patient-specific model, with a larger branch for validation. Branching effect was considered trivial if the FF difference between paired models (branches included or removed) was within 5%. *Results:* Compared with the control model without a branch, in all idealized models the relative differences of FF was within 2%. In five pairs of cerebral arteries (branches included/removed), FFs were 0.876 and 0.877, 0.853 and 0.858, 0.874 and 0.869, 0.865 and 0.858, 0.952 and 0.948. The relative difference in each pair was less than 1%. In transient model, the relative difference of FF was 3.5%. *Conclusion:* The impact of removing side branches with radius less than 50% of the parent vessel on FF measurement accuracy is negligible in static CFD simulations, and minor in transient CFD simulation. **Key Words:** Side branches—fractional flow—intracranial arterial stenosis—computational fluid dynamics.

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Background

Intracranial artery stenosis (ICAS) is a major cause for ischemic stroke and transient ischemic attack with its etiology still not well known.[1,2] Impaired hemodynamics related to the stenosis is considered an important mechanism for the ischemic event.[3,4] In recent years, as a novel concept, fractional flow (FF) has been put forward to substitute the measurement of lumen diameter restriction, which could not accurately evaluate the risk of stroke because of the inconsideration of complex hemodynamic effects.[5] FF is defined as the ratio of maximal blood flow in the presence of stenosis to the blood flow in the normal artery, and can be approximately calculated as the ratio of the pressures in the areas posterior and anterior to the stenosis, as in the calculation of fractional flow reserve (FFR), which has been developed as a gold standard to reflect the hemodynamic significance of vascular stenosis in the cardiovascular field.[6,7] With the advantage of delineating lesion-specific ischemia, FFR has been used to guide the selection of patients for percutaneous coronary intervention.[8] In the cerebrovascular field, a few pilot studies have also shown that FF may be a useful predictor of recurrent stroke in patients with symptomatic ICAS.[9,10] The analysis of specific influencing factors on cerebral FF, a promising parameter to reflect the hemodynamic significance of ICAS, is therefore significant.

Computational fluid dynamics (CFD), based on vascular geometry derived from clinical imaging, provides a new method for the noninvasive assessment of FF.[8,11,12] However, noninvasive FF is far from mature in gauging the severity and in treating ICAS. As a relatively new concept in the field of ICAS, FF needs further research into the details to validate its clinical implications. Thus, simplified models with acceptable accuracy are appropriate in the current stage. But the extent of simplification needs to be judged. Moreover, the modeling process also calls for simplification. To perform CFD simulation, 3D vascular geometry is required, together with the properties of blood and the vessel wall, and the inlet and outlet boundary conditions. In practical cerebrovascular simulation, due to the limited ability of current imaging techniques to accurately and fully delineate small vessels, it is inevitable to trim off some small branches. Because a side branch diverts blood flow from its parent artery with the local flow pattern influenced, the removal may cause hemodynamic changes in the parent arteries. In particular, the pressure and consequent the FF value may be affected. Hence, the reasonability of removing the side branches during CFD modeling should be judged.

Methods

In CFD modeling, the radius of branches needed to be trimmed off is usually less than half of the radius of the parent vessel. To study the branching effect, we built ideal models of identical stenosis, with the location and the radius of the branch varying among different models. By comparing the pressure and the FF derived from CFD in different models (with and without branches), the effect of branching on FF can be investigated; a relative difference of more than 5% was deemed as significant.

Idealized Models of Vascular Stenosis

The idealized 3D model of vessel with stenosis was created as a long cylinder with a radius of 1.5 mm, having a focal stenosis with a 70% area reduction (Fig 1). Sufficient elongations were sustained before and after the stenosis (25 and 50 mm, respectively) to include the regions that might be affected by turbulence. A branch with one third or one half of the radius of the parent vessel was added at distances of 5, 10, 15, and 20 mm posterior to the lesion, respectively, to generate different models. Hence, 1 reference model without a branch and 8 models with branches of different radii and the distance to the stenosis were studied using the CFD method.

CFD Modeling of Idealized Vascular Models

Computation of mesh, simulation of blood flow, and evaluation of hemodynamic characteristics of the arterial models were performed using the ANSYS 15.0 software package (ANSYS, Inc., Canonsburg, PA). A mesh was created in all the models with a maximum element size of .25 mm globally and .1 mm at the inlet and the outlet. The settings and assumptions for the simulation of the blood flow were as follows: blood was an incompressible Newtonian fluid with a constant viscosity of .004 $kg\cdot m^{-1}\cdot s^{-1}$ and a density of 1060 kg/m^3. The vessel wall was a rigid, noncompliant wall with a no-slip assumption. A total pressure of 110 mm Hg was applied at the inlet and a flow rate was applied at the outlets. The mass flow rate at the outlet was calculated by the following formula: flow rate = mean velocity × outlet area × density. The mean velocity was set as 35 cm/s, which is the typical mean flow velocity of cerebral arteries. The flow rate at the outlet of the branches was determined by Murray's law.[13] Thus, the total inlet flow rate would be added by the branch. From the reference model, control models for one third and one half of the radius groups were made. As in the reference model, there were no branches in the control models. However, the flow rates of the control models for one-third and one-half branching groups, were respectively set identical to other models within the same group (increased according the size of branches). The simulation of blood flow was fulfilled by solving the Navier–Stokes equations. The convergence criterion for the relative residual of all dependent variables was set as 10^{-4}.

Evaluation of the hemodynamic characteristics was performed on the pressure field of the models. As shown