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Blood flow reduction of covered small side branches after flow diverter treatment: A computational fluid hemodynamic quantitative analysis ☆

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A B S T R A C T

Small side branches related brain infarction remains one of the major concerns for flow-diverter devices. However, among several factors, whether this high-profile stent would significantly block blood flow into small side branches remains unclear. The authors quantitatively evaluate blood flow reduction due to the deployment of flow-diverter devices using computational fluid dynamics approach. Thirty one patient-specific anterior inferior cerebellar artery geometries were employed. The flow-diverter device was hypothetically embedded into the basilar trunk, and to cover the anterior inferior cerebellar arteries. The blood flow reduction of each anterior inferior cerebellar artery following flow-diverter device deployment was calculated, with independent validations for both inflow and outflow conditions. Efficient diameters of the anterior inferior cerebellar arteries were calculated to evaluate any correlation with blood flow reduction after flow-diverter devices.

The blood flow reduction ratio was shown to be $3.61 \pm 1.94\%$. There was moreover no significant difference of either inflow or outflow boundary conditions during the simulation. The results were calculated approximately as a modest linear correlation between the blood flow reduction ratio and the size of anterior anterior inferior cerebellar arteries which had a mean efficient diameter of 1.12 ± 0.36 mm (range from 0.31 mm to 1.91 mm), and the R^2 was 0.361. When covered by flow-diverter devices, the mechanical blood flow reduction in anterior inferior cerebellar arteries was found to be low with a maximum value estimated to be less than 8%. Therefore, mechanical blood flow reduction is probably not the leading factor contributing to small side branches related brain infarction.

1. Introduction

Flow-diverter devices (FDDs) are designed as high-profile structures (Kulcsar et al., 2010) to conduct blood flow within parent arteries so as to exclude aneurysms from circulation (Kallmes et al., 2007). Preliminary clinical applications from different reports indicate encouraging results compared with the traditional surgery or coil embolization (Byrne et al., 2010; Lubicz et al., 2010; Nelson et al., 2011). Pooled data published recently including 1451 patients with 1654 aneurysms showed a complete aneurysm occlusion rate of 76% at 6 month (Brinjikji et al., 2013). However, early infarction rate was reported to reach up to 5% in which posterior circulation had a significantly higher odds ratio than that of anterior circulation and usually result in serious consequences (Brinjikji et al., 2013). Some authors argued that FDDs might either block the orifices of side branches or narrow them into insufficient size which could contribute to side branches related brain infarction shortly after FDDs deployment. Moreover, they stated that under certain situations, small side arteries with a mean diameter around 400 μm in basilar artery and around 320 μm in posterior cerebral artery covered by FDD would significantly lose their orifices size, in which blood flow might be considerably reduced (Kulcsar et al., 2010). Therefore, the question whether the high-profile mesh of FDDs would significantly mechanically block the blood flow passing into small side branches was raised (Brinjikji et al., 2013; Kulcsar et al., 2010). Till date, this question has not been clearly confirmed.

Computational fluid dynamics (CFD) technology is accepted as an effective method to study cerebral vascular hemodynamics since last decade (Xiang et al., 2011). A porous media technique

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has been introduced to simulate the hemodynamic changes within aneurysms due to FDDs treatment, with the results found to agree with the clinical outcomes obtained in previous studies (Zhang et al., 2013a, 2013b). In that study, the authors examined the hemodynamic forces and the jet flow changes after FDD deployment. They concluded that in order to successfully occlude the aneurysms, the flow resistant force should be larger than the dynamic force and the jet flow speed should be reduced. In this study, we aim to quantitatively evaluate changes of blood flow into small side branches after FDDs treatment.

2. Materials and methods

This research was approved by the ethics committee of local university. All data collected under the current research project was patient-data blinded. Consequently, the sex, age, and primary diseases were not specified. Sixteen patients, who were approved of as exhibiting healthy vertebra-basilar systems except one patient with a left superior cerebellar artery aneurysm, were included in this study. Computed tomography angiographies (CTAs) of these 16 patients were selected from our database to construct patient-specific geometries.

2.1. Individual specific geometries

Posterior circulation harbored a higher rate in early infarction after FDDs deployment than that of anterior circulation (Brinjikji et al., 2013). Moreover, tiny vessels as perforators are difficult to be segmented from CTA studies so far. Thus, to perform a comprehensive CFD procedure on various sizes of small arteries and to simplify the calculation, the basilar artery was defined as the target vessel of FDD deployment. Anterior inferior cerebellar arteries (AICAs), if any, served as side branches. Cross-sectional CTA images with a thickness of 0.625 mm were read into MIMICS 14.0 (Materialise Company, Leuven, Belgium) to generate individual specific 3 dimensional vascular geometries. All visible branches, if any, including AICAs, superior cerebellar arteries, and posterior cerebellar arteries arising from basilar artery were segmented.

2.2. Simulation methods

Grid and boundary conditions independent analysis was solved in previous work (Qian et al., 2010; Zhang et al., 2013a, 2013b). Moreover, one patient with the smallest AICA was applied to prove the numerical accuracy using different grid size settings, which confirmed that the results were acceptable with maximum grid size less than 0.1 mm producing mesh elements more than 2 million cells in total (Supplementary materials). In order to form fully developed velocity profile, the inlets were extended to the 20 times of vertebral artery diameter in upstream normal directions. Outlets domains were also extended in their downstream normal directions to 100 mesh layers (10 cm) for sufficient recovery of blood pressure. Grid was generated using a commercial software package ICEM CFD 14.1 (ANSYS, Canonsburg, PA, USA). Virtual FDDs geometries were generated with a thickness of 100 μm (Zhang et al., 2013a, 2013b) and embedded smoothly within the basilar arteries to cover bilateral AICAs so that the blood flow was forced to pass this layer before going into AICAs (Fig. 1).

The governing equations employed in the current study were the Navier–Stokes equations. To simplify the calculations, the vessel walls were assumed to be rigid and no-slip. The blood was assumed to be Newtonian fluid with a density of 1060 kg/m^3 and a dynamic viscosity of 0.0035 Pa s. The pulsatile flow was approved to be comparable to the steady flow on two patients when considering the porous layer might behave differently (Supplementary materials). On the other hand, this result was also verified by an early study (Augsburger et al., 2009). To save the solver time, the flow was treated to be steady. At the outlets zero static pressure was prescribed, while at the inlets a normal speed of 0.3 m/s which was normalized using the data measured by duplex sonography on 50 individuals was given (Seidel et al., 1999). FDD was set as porous medium with a permeability of 1E-9 m^2 and a loss coefficient of 8703/m based on in vitro measurement of a Silk stent (Augsburger et al., 2011). The equations were solved using the CFX 14.1 solver package (ANSYS) on a HP Z800 workstation (Hewlett-Packard Company, Palo Alto, CA, USA).

2.3. Flow conditions and validation

The main concern of the current study was to determine the blood flow reduction due to FDD deployment. The blood flow reduction was defined as the blood flow before stent minus the blood flow after FDD placement. Therefore, the mass flow at each AICA outlet was calculated by using CFX 14.1 post-processing package (ANSYS) before and after stent deployment. To check the independence of results from inflow and outflow conditions, two kinds of validation were carried out. The outlet pressure conditions at posterior cerebral arteries and superior cerebellar arteries were imposed to be 100 Pa and 1000 Pa respectively which is reasonable to cover the pressure gradient between proximal outlets as AICAs and distal outlets as superior cerebellar arteries and posterior cerebral arteries in vivo (Henderson et al., 2010). The inflow was imposed to be $\pm$20% of average inflow condition; 0.3 m/s. It was expected to perform simulations under a reasonable flow range in vertebral artery.

To investigate the relationship between blood flow reduction after FDDs and vessel size, mean diameters of the AICAs were calculated. Considering that the morphology of AICAs rarely remained as regular cylinders, we defined an efficient (mean) diameter of a regular circle which had an area equal to the AICA.

2.4. Statistic analysis

The blood flow reduction ratio was defined as the ratio of mass flow reduction to the mass flow of AICA before stent deployment. The correlation of diameter and blood flow reduction ratio was analyzed. The one-way analysis of variances (ANOVA) was applied to solve the impacts of boundary conditions on mass flow reduction. All the data recordings and processing are presented as mean and standard deviation. The data analysis was performed on SPSS 17.0 (SPSS Inc., Chicago, USA). Statistic significance was defined as $p < 0.05$.

3. Results

Sixteen geometries were segmented from the patients' CTA images, of which 15 patients had bilateral AICAs while the last one had only one AICA that could be segmented. As a consequence, 31 AICAs were introduced in this study.

3.1. Mass flow reduction after FDD deployment

The mass flow reduction of AICAs after FDDs deployment is summarized in Supplementary Table 1. To be brief, the blood flow reduction ratio varies depending on different geometries with a mean value of 3.61 $\pm$ 1.94% (Supplementary Table 1).

3.2. Results variations on flow boundary conditions

For the effects of outflow conditions, we found that the mass flow through AICAs before stent would be increased by 6.93 $\pm$ 3.67% and 68.55 $\pm$ 35.70% at the prescribed values namely 100 Pa and 1000 Pa respectively. In order to determine the influences which might be introduced by the outlet different pressure conditions on the blood flow reduction after FDDs deployment, one-way ANOVA test was carried out based on the mean blood flow reductions. One-Sample Kolmogorov–Smirnov test was applied in each group showing normal distribution with $p \geq 0.05$. The statistic results indicated an F value of 0.384 and a P value of 0.682 revealing no significance of difference when considered the outlet boundary conditions impacting on the flow reduction of AICAs (Supplementary Table 1).

The tests of flow changes of inflow conditions were performed to validate the influence of blood flow reduction in AICAs, the mass flow through into AICAs before stent was increased to 24.27 $\pm$ 14.63% at an increase of 20% flow velocity into inlets and decreased to 23.08 $\pm$ 10.30% at a decrease of 20% flow velocity into inlets respectively. One-way ANOVA test showed that the F value was 1.340 and the P value was 0.267 (Supplementary Table 1). The results indicated that the blood flow reduction changes in AICAs after FDDs deployment was limited to be influenced by inflow conditions.

In summary, under physiologically considerable inflow and outflow conditions, the blood flow reduction of AICA due to FDD treatment did not undergo significant change, whilst the maximum blood flow reduction ratio was found to be less than 8% (Fig. 2).