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MULTISCALE MODELING OF CEREBRAL BLOOD FLOW

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ABSTRACT

A multiscale model of the human cerebral vasculature has been developed which includes a three-dimensional (3D) CFD model of the circle of Willis (CoW) and fractal tree models of all regions of small cerebral vasculature, namely Anterior, Middle and Posterior Cerebral Arteries (ACA, MCA, PCA). The realistic 3D CoW model was constructed from the medical imaging data with the use of 3D Slicer segmentation software. The flow model in the fractal tree models of ACA, MCA and PCA has been developed with the effects of blood vessel structural property, arterial size-dependent blood viscosity and non-parabolic velocity profile incorporated. The coupling of the CFD model (solved with ANSYS CFX) and the fractal tree models (solved with Matlab mathematic library) has been extended from one-way to two-way method in this work. The coupled model has been used to predict the transient blood flow in cerebral arteries and study the effect of occlusion on flow distribution in the brain.

INTRODUCTION

The brain vasculature is a complex network, which is divided into six major distribution territories, namely the anterior cerebral arteries (ACA), middle cerebral arteries (MCA) and posterior cerebral arteries (PCA). The blood is supplied to these distribution territories through four major carotid and vertebral arteries, which converge in the circle of Willis (CoW). The CoW with its communicating arteries is acting as an equalizing distributor and provides some redundancy for additional blood supply reliability (see Edvinsson et al., 2002). In each of the blood distribution territory, the blood vessels branch out and becomes progressively smaller. It has been established through many studies that the most common form of branching in cardiovascular systems is bifurcation (Li, 2004) and the vascular network branching down to the capillary level is generally governed by fractal scaling principles (Mandelbrot, 1983, Zamir, 1999, and Lapi, 2008).

Due to the great difference in the sizes of cerebral blood vessels and the big number of small arteries involved in supplying blood to the brain, a multiscale approach to the modelling of the cerebral blood flow would be needed. In the past, the blood flow in the circle of Willis has been simulated using three-dimensional (3D) or simplified one-dimensional (1D) models (Hillen et al., 1986, Alastruey et al., 2007, David & Moore, 2008). Without considering the flows in smaller arteries, the previous CFD simulations of the CoW had to employ simplified boundary conditions for the six outlets. The simplified boundary conditions for CoW could be modeled by resistors in parallel (Cebral et al., 2003), porous media as terminating blocks (Moore et al., 2006), or simply described as constant pressure (Alnaes, 2007).

In this work, a hybrid modeling method based on the detailed three-dimensional CFD description of the large blood vessels, in particular, the circle of Willis, couple with a model of pulsatile blood flow in the branching networks of small arteries has been developed. The combined model is able to provide a detailed description of the transient blood flow not only in the CoW, but also in complex networks of ACA, MCA, and PCA, without a need of using approximate boundary conditions.

MODEL DESCRIPTION

Patient-specific model of the circle of Willis

The transient flow in the 3D model of the circle of Willis was described with use of the CFD method. The numerical modelling was performed using the commercial CFD package ANSYS CFX-11, which has a coupled solver and uses an unstructured mesh based on the finite element method.

A patient-specific geometry of CoW was acquired from CT scans and converted to the surface (STL) and then finite element models using Slicer-3D, Meshlab and ICEMCFD. This patient-specific geometry (Figure 1) was missing the left posterior communicating artery (PCoA) and had a penal anterior communicating artery. This anatomical configuration with missing PCoA occurs in 9% of the population (Alastruey et al., 2007).

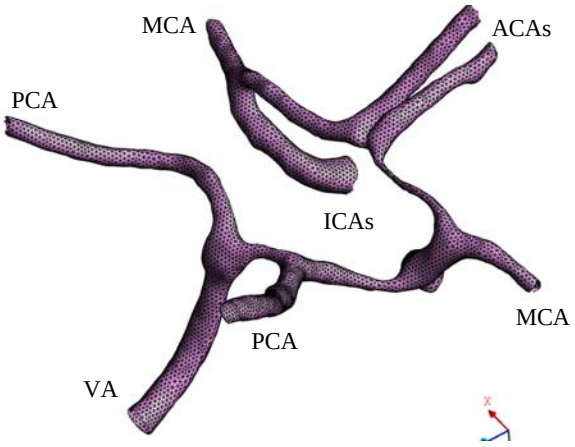

Figure 1: A patient-specific CoW geometry and computational mesh.

The mesh consisted of tetrahedral elements, and the solution was mesh-independent with the total number of elements equal to 74,250.

Cerebral vascular branching tree model

Using the fractal scaling concept, physiologically meaningful branching tree models of cerebral small vasculatures can be generated. In this work, the method of Constrained Constructive Optimization (CCO) (Schreiner et al., 1996, 2006) was employed to construct the ACA, MCA, and PCA models. Essentially, the CCO method involves geometric and structural optimisation, which minimise the total volume of the blood vessels and ensure uniform blood perfusion in the designated domain.

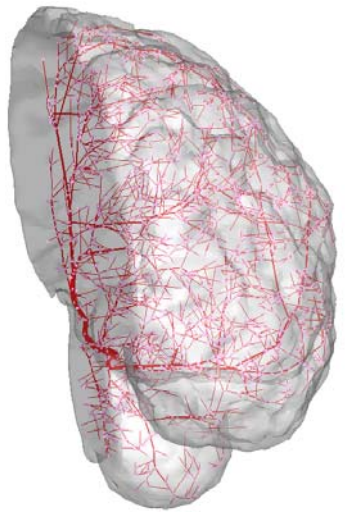

Figure 2: An example of vascular branching tree model generated for MCA.

Distributions of pressure and velocity in a vascular branching network can be obtained by solving a system of ordinary differential equations (ODEs) describing the flow conservation at bifurcation points. The effect of the blood vessel compliance and blood flow variable viscosity (caused by Fahraeus effect) can be taken into consideration. Details of the flow model for vascular networks can be found in the work by Bui et al. (2009).

The vascular branching networks of ACA, MCA, and PCA used in this work comprised 2779, 5559, and 4159 segments, respectively. It was assumed that the cerebral vasculatures of the left and right halves of the brain were symmetrical across the central plane. The network flow model was solved with use of the MATLAB ODE solvers.

Model coupling

In the past, one-way coupling of the CFD and vascular branching tree flow models was employed (Šutalo et al., 2009). In one-way coupling, pressure at the inlets to all cerebral vascular networks was assumed to be equal to the systemic pressure and the transient flows at the roots of the ACA, MCA, and PCA vascular networks were calculated and used as outflow boundary conditions in the CFD simulation of the CoW flow. In this work, a method has been developed to fully couple the CoW and the vascular network flow models. In an iterative procedure, actual pressures at the CoW outlets are used to determine the flow in each vascular network, which is then applied as boundary conditions in the CoW CFD simulation. The network flow model and the MATLAB ODE solver have been compiled into a dynamic link library (DLL), which can be called from CFX Fortran user subroutine, and the iterative procedure is conducted until the convergences of pressure and flow solutions are achieved at every simulation step.

RESULTS

Pulsatile pressure ranging 80–125 mmHg with a period of 0.7 s was specified at the CoW inlets – internal carotid (ICAs) and vertebrobasilar (VA) arteries (Figure 3). The flow in the cerebral microcirculatory system is characterised by low pulsatility Therefore, a constant capillary pressure of 25 mmHg was assigned at the terminals of the vascular branching networks. The density of the blood was assumed to be 1050 kg/m^3, and the blood flow was assumed be Newtonian with viscosity equal to 0.0036 Pa.s.

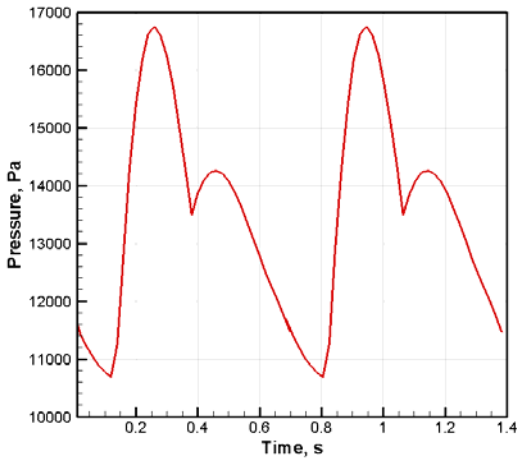

Figure 3: Pulsatile systemic pressure.

The prediction results are shown in Figures 4-7. Due to this CoW geometry specifics, the flows in the left and right branches of the brain are significantly different (Figure 5) and the geometry effect is most pronounced in the ICA flow. The difference in the flow predictions resulted from the coupling method is displayed in Figure 6. The one-way coupling approach, which was based on the assumption about uniform distribution of pressure in the CoW, produced significantly higher in- and out- flow predictions. In addition to the prediction of the flow in the normal condition, the computational model was also used to simulate the variation of the flow through the circle of Willis as a result of possible physiological and pathological changes in the network of small cerebral vasculature. Figure 7 demonstrates the variation of the blood flows in the brain when the peripheral resistance of ACAs increased due to an uniform 10% vasoconstriction in ACA networks. Vasoconstriction is the narrowing of the blood vessels, which can be caused by various physiological and/or pathological conditions.

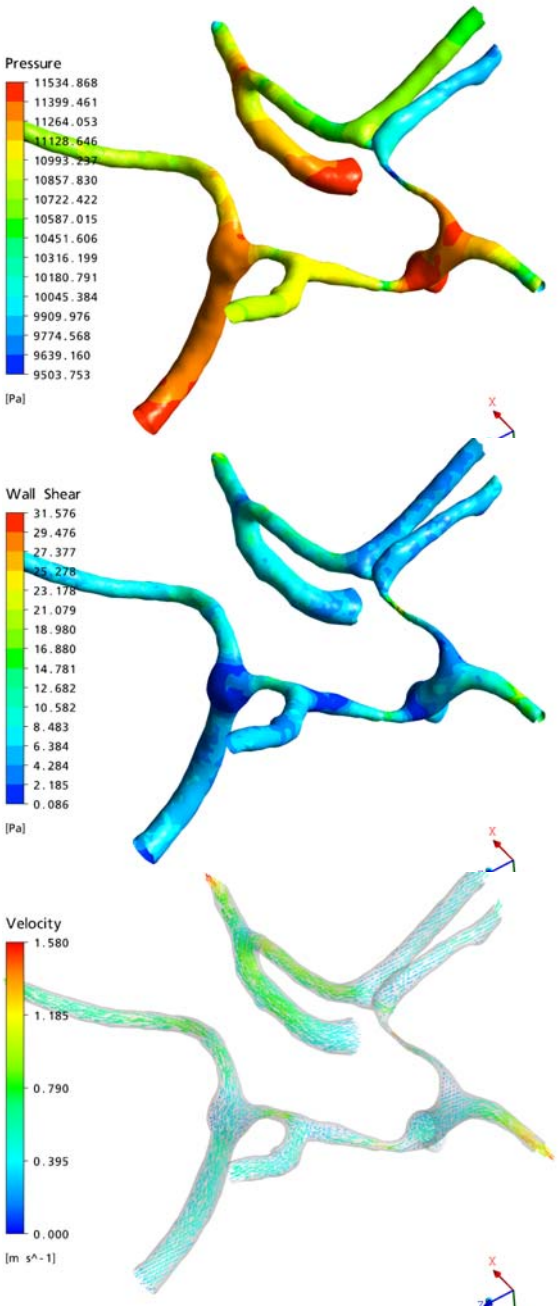

Figure 4: Instantaneous distributions of pressure, wall shear and velocity in the CoW at 0.7 s.

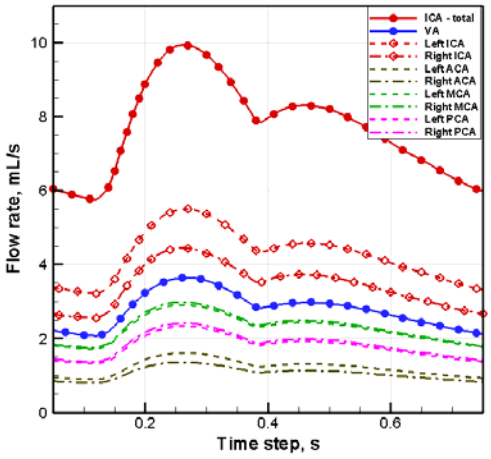

Figure 5: Flows in carotid (ICA), basilar (VA) arteries, ACAs, MCAs, and PCAs.

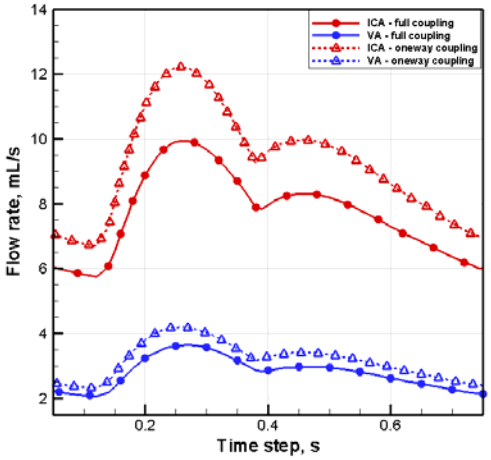

Figure 6: Comparison of the flows in ICA and VA predicted by one-way and full couplings.

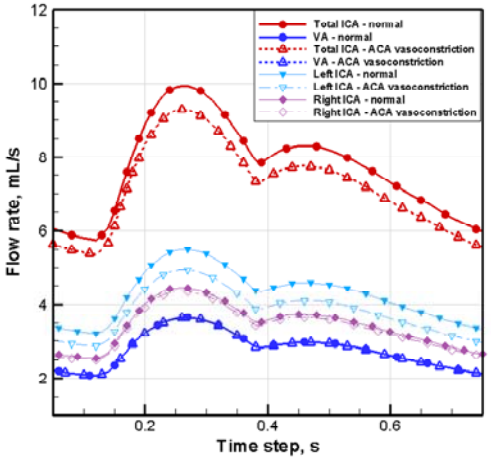

Figure 7: Flow variations due to vasoconstriction in ACAs.

CONCLUSION

A fully coupled CFD and network models of cerebral blood flow has been developed in this work. The model was applied to predict the transient flow and pressure distributions in the brain vasculature comprising a patient-specific circle of Willis geometry and fractal models of peripheral vascular networks. The models were shown to be able to efficiently provide detailed descriptions of the flow and pressure distributions at different levels of blood vessel sizes and simulate the variations of the blood flow in the major cerebral arteries when the peripheral vasculatures are subjected to various physiological and pathological conditions.

In order to improve the predictive capability of the models, the mechanisms of active regulation of blood flow need to be defined and implemented in the future model development. Further modelling work may also be conducted to comprehensively study the effect of the CoW anatomical variations on the cerebral blood flow.

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