LARGE-SCALE SIMULATION OF THE HUMAN CRANIAL ARTERIAL TREE: UTILITY IN HYDROCEPHALUS

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BACKGROUND

Characterization of cerebral hemodynamics is essential for understanding complex intracranial dynamics in normal and diseased conditions, diagnosing and treating patients, and designing medical devices. To this end, we have developed a procedure for generating physiologically accurate models of the cerebral hemodynamics by coupling clinical data and multiscale modeling with advances in parallel supercomputing. A typical multiscale numerical simulation of blood flow in the human cranial arterial tree, running on hundreds of thousands of processors on petaflop computers, requires no more than several CPU hours per cardiac cycle. Here we present two demonstrative examples of utilizing 3D Computational Fluid Dynamics (CFD) to investigate the cerebral hemodynamics in normal and hydrocephalic subjects.

METHODS

A. Model description

A straight-forward modeling of the intracranial hemodynamics is computationally prohibitive even on the emerging petaflop supercomputers, so a hierarchical approach based on the vessel size is proposed. We present the results of recent simulations of cerebral hemodynamics in a macrovascular network consisting of tens of arteries and bifurcations, down to a diameter of 0.5 mm. The anatomical model of the macrovascular network is patient-specific and is comprised of all the arteries that can be accurately imaged using conventional medical imaging techniques, such as high resolution Magnetic-Resonance Angiography (MRA) and Computed Tomography (CT). Smaller arteries, arterioles and the capilary bed are considered a microvascular network. The subpixel dynamics of the microvascular network are modeled by prescribing proper boundary conditions. Due to the geometric complexity of the cranial arterial system, we employ high-order spectral/hp element methods for solving the governing flow (3D Navier-Stokes) equations. Highly scalable parallel implementation allows the dramatic reduction of overall computation time of the 3D large-scale simulations. The full-scale modeling aproach of a human arterial tree is discussed in much greater detail in our recent publication [1].

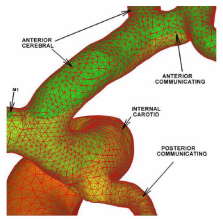

Fig. 1. Computational mesh

B. Anatomical modeling

The geometric models of the cranial arterial networks were reconstructed from high-resolution MRA and DSA CT images. Due the to high complexity of the arterial networks, which include tens of arteries, inlets and outlets, several numerical approaches to reconstruct the vessel wall geometries were implemented. The inner part of a typical arterial network is reconstructed by extracting iso-surfaces from MRA data with a predefined threshold. The smaller vessels, such as posterior temporal, are reconstructed using another technique: initially, the arterial medial axis is computed and the vessel diameter along the medial axis is approximated. Then, ring-like contours are seeded along the medial axis to form a pipe-like structure. Once all parts of the arterial tree are extracted, the data is uploaded into mesh generation software, where additional editing is performed to integrate all parts into one smooth surface, eliminate remaining small features, and form the inlet and outlet regions of the arteries. As a final step, a finite element mesh is generated.

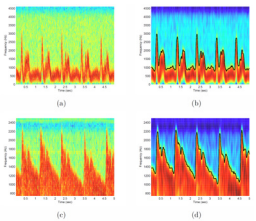

Fig. 2. Spectrogram of the audio Doppler ultrasound data: CCA (a) and MCA (b); filtered spectrogram and the corresponding maximum frequency envelope: CCA (c) and MCA (d).

C. Boundary conditions

In our models we account for the cerebral autoregulation mechanism by prescribing a scalable and efficient type of pressure boundary condition applicable to flow domains with multiple outlets [3]. This method allows us to impose accurately and in a straight-forward manner in-vivo measured flow rates at terminal outlets. Measuring the flow rate, unlike measuring the pressure, is a straightforward procedure that is performed using non-invasive techniques. Two such techniques that we employ are Transcranial Doppler Ultrasound (Fig.1) and Phase-contrast MRI (Fig. 2). The reconstruction of the velocity fields from Doppler ultrasound (CDUS) or PC MRI is described in [4]. In order to impose inflow/outflow boundary conditions in simulations of flow in an arterial network with multiple branches, the flow rate data collected at different arterial segments must be synchronized. Such synchronization can be performed by simultaneously recording Electrocardiogram (ECG) signal.

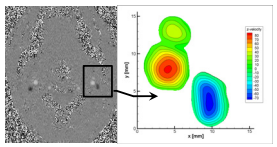

Fig. 3. PC MRI-based velocity field reconstruction: Left: Phase-contrast MR image of blood flow at mid C2 level. Bright and dark circular spots enclosed into square section indicate the location of an artery and a vein. Right: map of the normal to the plane velocity component (in cm/s) in the artery (positive values) and in the vein (negative values)

D. CFD Numerical simulations

Numerical simulations were performed on Texas Advanced Computing Center supercomputer using a CFD code NEKTAR developed at Brown University. NEKTAR implements the high-order spectral/hp element method for spatial discretization of the computational domain [2]; the method provides very high spatial resolution and is suitable for complex geometries. Spectral/hp element method provides a dual path to convergence (i.e. decay of numerical error): (a) h-convergence, with the accuracy of the solution depending on the size of elements; and (b) p-convergence with the accuracy depending on the order of polynomial approximation and on the smoothness of the approximated solution. In the case of a smooth solution, exponential rate of convergence is obtained.

RESULTS

We have performed numerical simulations of pulsatile blood flow in cranial arterial networks of a healthy and a hydrocephalic volunteer. The velocity profiles, the zones of reversed flow and wall shear stresses and pressure distributions were obtained and visualized for different time instants throughout the cardiac cycle. These results reflect the complex nature of the flow, pressure and shear stress patterns. In the healthy case (Figs. 4-6) the blood flow in all the braches of the Circle of Willis remains mostly uniform throughout the cardiac cycle, suggestive of an efficient blood circulation. In the hydrocephalic case (Figs. 7-9) the simulations reveal a steep rise in the arterial pressure and elevated pulsatility due to the abnormal anatomy, elevated intracranial pressure and the need to maintain sufficient blood perfusion to the brain.

Cranial arterial network of a healthy volunteer

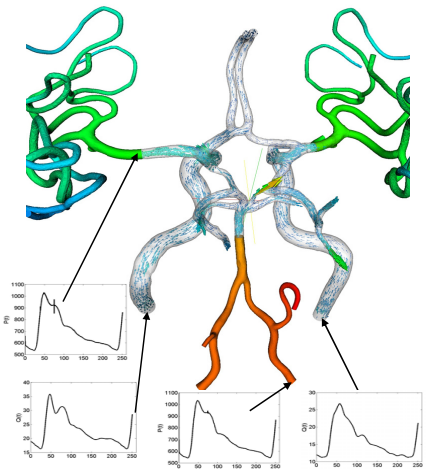

Fig. 4. Flow pattern in the Circle of Willis. Arrows depict the instantaneous flow direction; colors represent pressure distribution (red and blue colors indicates high and low pressures respectively). Flow rate and pressure waveforms in the select branches are also shown.

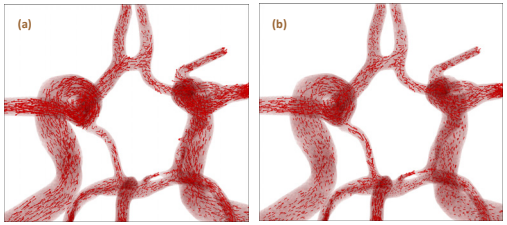

Fig. 5. Peak systolic (a) and end diastolic (b) velocities. The simulation indicates that blood supplied through the left ICA feeds both hemispheres of the brain.

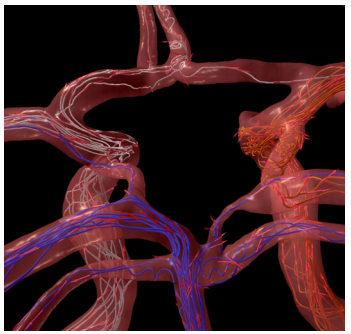

Fig. 6. Instantaneous blood flow streamline patterns. Colors vary among streamlines originating from different inlets.

We acknowledge Greg Foss (Pittsburgh Supercomputing Center) for providing this figure.

Cranial arterial network of a hydrocephalic volunteer

The lack of communicating arteries in left anterior and posterior communicating arteries in this subject results in a completely disconnected left and right networks. Here we present the results of (the larger) right arterial network only.

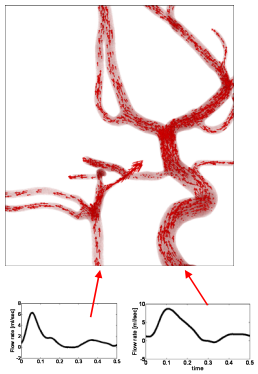

Fig. 7. Flow pattern in the Circle of Willis; arrows depict the instantaneous flow direction. Flow rate and waveforms in the internal carotid and basilar arteries are also shown; there is a noticeable phase shift between the two waveforms.

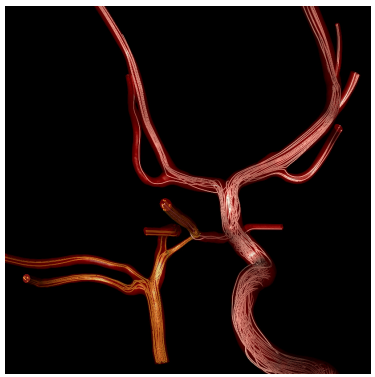

Fig. 8. Instantaneous blood flow streamline patterns. Colors vary among streamlines originating from different inlets.

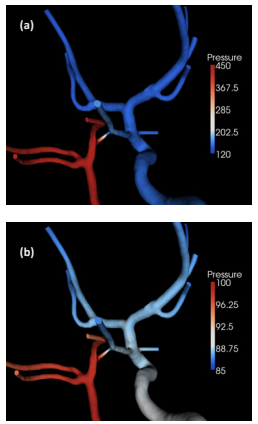

Fig. 9. Pressure distribution during peak inflow into (a) the basilar artery and (b) end diastole.

SUMMARY

Human intracranial dynamics models are powerful research tools to enhance the understanding of pathophysiological mechanisms and provide realistic parameters for comparing patients with normal controls. Realistic patient-specific models of the cerebral vasculature system should be based on accurate high resolution anatomical models and physiologically correct hemodynamic conditions. We propose a hierarchical approach that couples arterial networks of different scales and can possibly lead to an efficient simulation of the arterial tree on the new emerging petaflop systems. According to our rough estimates, a single multiscale simulation of the human brain vascular network will require about 27.7 wall-clock hours per cardiac cycle on 110,000 processors; such systems are currently available (see www.top500.org).

We have presented the results of large-scale simulations of normal and hydrocephalic human cranial arterial trees as demonstrative examples. Consistent with previous findings, our 3D simulations of human intracranial dynamics predict cerebral hypoperfusion and elevated resistivity and pulsatility indices in patients with hydrocephalus.

There is much more complexity to be added in the modeling paradigm, including blood rheology, biochemistry, blood-endothelium interactions, etc. Our goal is to account for all the physiological aspects without making prohibitive assumptions about clinically relevant hemodynamic parameters. In order to fulfill this requirement, we plan to incorporate measurable physiological data such as pressure, velocity, cerebral perfusion and blood vessel movements during the cardiac cycle. The data will be utilized in the subsequent analysis and numerical simulation steps in the form of boundary conditions as a validation tool. This will then provide the possibility of systematic studies in investigating clinical pathologies that will involve several cardiac cycles. Our models can be utilized to predict the hemodynamic effects and outcomes of clinical interventions such as shunting and endoscopic third ventriculostomy.

REFERENCES

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