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Approximating hemodynamics of cerebral aneurysms with steady flow simulations

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ABSTRACT

Computational fluid dynamics (CFD) simulations can be employed to gain a better understanding of hemodynamics in cerebral aneurysms and improve diagnosis and treatment. However, introduction of CFD techniques into clinical practice would require faster simulation times. The aim of this study was to evaluate the use of computationally inexpensive steady flow simulations to approximate the aneurysm's wall shear stress (WSS) field. Two experiments were conducted. Experiment 1 compared for two cases the time-averaged (TA), peak systole (PS) and end diastole (ED) WSS field between steady and pulsatile flow simulations. The flow rate waveform imposed at the inlet was varied to account for variations in heart rate, pulsatility index, and TA flow rate. Consistently across all flow rate waveforms, steady flow simulations accurately approximated the TA, but not the PS and ED, WSS field. Following up on experiment 1, experiment 2 tested the result for the TA WSS field in a larger population of 20 cases covering a wide range of aneurysm volumes and shapes. Steady flow simulations approximated the space-averaged WSS with a mean error of 4.3%. WSS fields were locally compared by calculating the absolute error per node of the surface mesh. The coefficient of variation of the root-mean-square error over these nodes was on average 7.1%. In conclusion, steady flow simulations can accurately approximate the TA WSS field of an aneurysm. The fast computation time of 6 min per simulation (on 64 processors) could help facilitate the introduction of CFD into clinical practice.

1. Introduction

Growth and rupture of cerebral aneurysms have been associated with the intra-aneurysmal hemodynamics (Hashimoto et al., 2006). Better understanding of hemodynamics could improve diagnosis and treatment. In the last decade, computational fluid dynamics (CFD) simulations have been employed to study the relationship between hemodynamics and rupture (Cebral et al., 2011b, 2011c; Xiang et al., 2011; Miura et al., 2013) and the hemodynamic effect of endovascular treatment (Kakalis et al., 2008; Cebral et al., 2011a; Larrabide et al., 2013; Morales et al., 2013). In particular, research has focused on the wall shear stress (WSS), which is a key regulator of vascular biology and pathology (Malek et al., 1999; Dolan et al., 2013).

The majority of studies use unsteady, *pulsatile* flow simulations to capture the changing flow rate and inertia effects during the cardiac cycle. The enormous amount of generated data is typically reduced by extracting and analyzing the time-averaged (TA) (Cebral et al., 2011c; Miura et al., 2013; Xiang et al., 2011), peak systole (PS) (Castro et al., 2009; Shojima et al., 2004) or end diastole (ED) flow field (Jou et al., 2008; Fukazawa et al., in press). Some researchers have argued, however, that in many cases the main flow features in the aneurysm remain relatively stable throughout the cardiac cycle (Müller et al., 2012; Mantha et al., 2009) and that computationally much less expensive *steady* flow simulations might already provide clinically relevant hemodynamic information (Cebral et al., 2011c). The shorter time required to create these simulations could aid in the introduction of CFD into clinical practice. So far, comparisons between steady and pulsatile flow simulations have mostly been qualitative and based on few cases, but before steady flow simulations can be relied upon, it is essential to quantify for a large population the accuracy with which they can approximate aspects of the pulsatile flow field in the aneurysm.

The aim of this study was to quantify the error with which steady flow simulations can approximate the aneurysm's TA, PS and ED WSS fields derived from pulsatile flow simulations. Since

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this error might depend on the flow rate waveform (FRW) imposed at the inlet of the vascular model, we systematically varied the heart rate, pulsatility index, and TA flow rate. The study's dataset included both terminal and lateral aneurysms, covering a wide range of aneurysm volumes and shapes.

2. Methods

2.1. Blood flow modeling

Aneurysm models included in this study were drawn from a large database created within the EU project @neurIST (Villa-Uriol et al., 2011). Patient-specific vascular models, represented by triangular surface meshes, were constructed by segmenting three-dimensional rotational angiography (3DRA) images using a geodesic active regions approach (Bogunović et al., 2011). These models were smoothed using a geometry-preserving smoothing algorithm (Nealen et al., 2006) to reduce small surface perturbations without displacing vertices by more than 0.02 mm. Touching vessels were removed. All inlet and outlet branches were clipped perpendicular to their axes and were, respectively, 12 and 4 diameters in length. To define the aneurysm region, the aneurysm neck was manually delineated and a neck surface was automatically created from this curve. This approach has been demonstrated to have a low interobserver variability for volume and surface area measurements of the aneurysm (Larrabide et al., 2011). All mesh editing operations were performed in @neuFuse (B3C, Bologna, Italy) (Villa-Uriol et al., 2011), a software application developed within @neurIST.

Unstructured volumetric meshes were created using the octree approach with ICEM CFD 13.0 (ANSYS, Canonsburg, PA, USA). Meshes were composed of tetrahedral elements of 0.24 mm, and three prism layers with a total height of 0.08 mm and a side length of 0.12 mm. The total number of elements ranged from 1.2 to 3.8 million, the density from 1420 to 2950 elements per mm^3, depending on the surface-area-to-volume ratio of the computational domain. This mesh setup was selected following mesh dependency tests performed on both cases of experiment 1 (see Section 2.3). These tests demonstrated the mesh independency of the WSS field for steady flow simulations under the highest inflow rate conditions considered in this study.

CFD simulations were created with the commercial finite volume solver CFX 13.0 (ANSYS), using a second order advection scheme, a second order backward Euler transient scheme for unsteady simulations, and CFX' automatic time scale control for steady-state simulations. Solutions converged until the normalized residual of the WSS everywhere in the computational domain was $<10^{-5}$. Blood was modeled as an incompressible Newtonian fluid with density $\rho = 1060$ kg/m^3 and viscosity $\mu = 0.004$ Pa s. Vessel walls were assumed rigid with a no-slip boundary condition. Flow rate values (steady) or waveforms (pulsatile) were imposed at the inlet and zero-pressure boundary conditions at all outlets. For pulsatile flow simulations, the cardiac cycle was discretized in 200 uniformly distributed time steps and, to reduce the effect of initial transients, the second of two simulated cardiac cycles was analyzed. Tests performed on both cases of experiment 1 demonstrated WSS differences of $<1\%$ with respect to simulations with either 400 uniformly distributed time steps per cardiac cycle or from which the third cycle was analyzed.

Table 1
Flow rate waveform descriptors.

Descriptor	Physiological variation		
	Lower	Baseline	Upper
Heart rate [bpm][a]	52	68	84
Pulsatility index [–][a]	0.58	0.92	1.26
TA flow rate [ml/s][b]	$0.73Q_A$	Q_A	$1.27Q_A$

[a] baseline = mean, lower = mean − 2 SD, and upper = mean + 2 SD where the mean and standard deviation (SD) are taken from Ford et al. (2005) and Hoi et al. (2010).
[b] $Q_A = 48.21A^{1.84}$ where A is the inlet area in cm^2 (Cebral et al., 2008).

2.2. Flow rate waveform transformation

FRWs describe how flow rate $Q(t)$ at the inlet of the vascular models changes over time t during the cardiac cycle. In this study, we used a physiological FRW $Q^0(t)$ that was derived from phase-contrast magnetic resonance images of the internal carotid artery (ICA) of a healthy volunteer (Cebral et al., 2003). This FRW was then linearly transformed to obtain a FRW with specified heart rate (HR), TA flow rate (Q_{TA}), and pulsatility index (PI) given by $PI = (Q_{PS} - Q_{ED})/Q_{TA}$. These three variables will be referred to as FRW descriptors. The transformation is given by

$$Q(t) = aQ^0(ct) + b \quad (1)$$

where

$$a = \frac{Q_{TA}}{Q^0_{TA}} \frac{PI}{PI^0}, \quad b = Q_{TA}\left(1 - \frac{PI}{PI^0}\right), \quad c = \frac{HR}{HR^0}$$

Physiological values for the FRW descriptors (see Table 1) were derived from the literature. For each descriptor, we defined a baseline, an upper and a lower value. Values for HR and PI were obtained using the mean and standard deviation (SD) of these descriptors reported by Ford et al. (2005) and Hoi et al. (2010): baseline = mean, lower = mean − 2SD, and upper = mean + 2SD. Values for Q_{TA} were obtained using the relationship $Q_A = 48.21\,A^{1.84}$ where Q_A is in ml/s and A is the inlet's cross-sectional area in cm^2 (Cebral et al., 2008). This relationship was determined by fitting a power-law function through measurements of Q and A of the ICAs and vertebral arteries of 11 normal subjects. Values for Q_{TA} were given by: baseline = Q_A, lower = $0.73Q_A$, and upper = $1.27Q_A$. The 27% variation is the average relative error between prediction and measurement, derived from Cebral et al. (2008).

2.3. Experiments

Two experiments were conducted in this study. Fig. 1 schematically represents the workflows of both experiments.

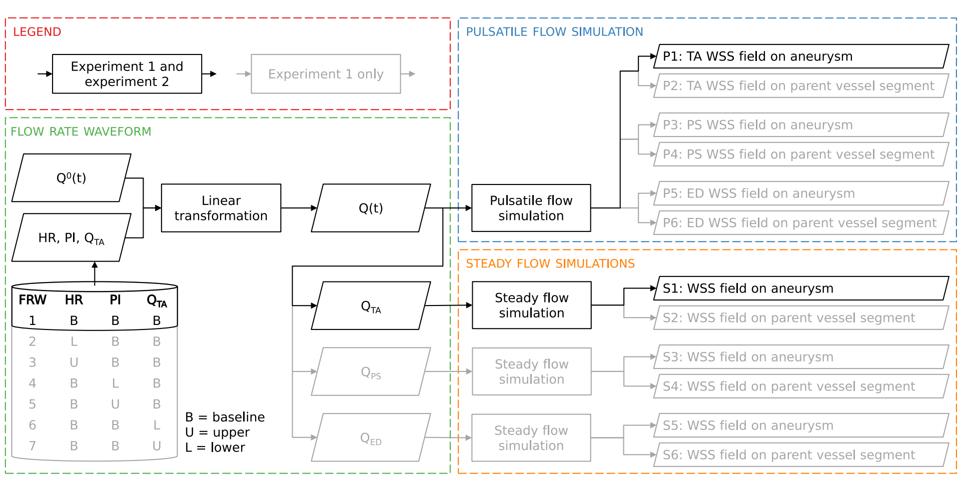

Fig. 1. Workflows of experiments 1 and 2. Sets of FRW descriptors HR, PI, and Q_{TA} were created by keeping two descriptors at baseline and varying the third from lower to baseline to upper value (see Table 1). For each set of FRW descriptors, $Q^0(t)$ was linearly transformed to obtain $Q(t)$ from which then Q_{PS} and Q_{ED} were derived. One pulsatile flow simulation was created with $Q(t)$ imposed at the inlet of the vascular model and three corresponding steady flow simulations were created with flow rates Q_{TA}, Q_{PS} and Q_{ED} imposed at the inlet. In the data analysis, WSS field S1 was compared to P1, S2 to P2, etc.