GLOBAL COMPUTATIONAL MODELING OF CARDIOVASCULAR BLOOD FLOW

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Abstract *The cardiovascular system is an internal flow network with multiple branches in which three-dimensional and pulsatile blood flow circulates. We propose a new paradigm of simulation-based biomedical engineering, which is designed as a useful means for predictive medicine. In order to achieve this paradigm a global computational modeling of cardiovascular blood flow is highlighted and some preliminary results are presented.*

1. INTRODUCTION

The cardiovascular system is an internal flow network with multiple branches in which blood flow circulates. The heart pumps blood through the arterial tree of blood vessels. The blood vessels distribute blood to different organs, supply themselves with nutrition and transport wastes. The arteries adapt to varying blood flow and pressure conditions by enlarging or shrinking to meet changing hemodynamic demands.

Blood flow in arteries is three-dimensional and unsteady because the cyclic nature of the heart pump creates pulsatile conditions in all arteries. The blood is pumped out of the heart during *systole*. The heart rests during *diastole*, and no blood is ejected. Pressure and flow have characteristic pulsatile shapes that vary in different parts of the arterial system. The tree structure of branching, tapering tubes results in that the blood flow spans a wide range (1~4000) of Reynolds number (*Re*) and generally the viscous and inertial forces work together in a moderate Reynolds number (100~1000), resulting in vortical flow in most of arteries. The existence of pulsation leads to another non-dimensional parameter commonly referred to as the Womersley parameterα (10~20 in large human arteries), which can be interpreted as the ratio of the unsteady forces to the viscous forces.

Blood flow under normal physiologic conditions is an important field of study, as is blood flow under disease conditions. The majority of deaths in developed country result from cardiovascular disease, most of which are associated with some from abnormal blood flow in arteries. Detection and quantification of the abnormal blood flow in arteries serve as the basis for surgical intervention. In the Computational Biomechanics Project started in RIKEN last year, we are aiming at providing a paradigm of simulation-based biomedical engineering for the study of arterial blood flow. We will build up an integrated computational system, which should lead to the prediction of individual hemodynamic flows in any patient, the development of diagnostic tools to quantify disease, and the design of devices to mimic or alter blood flow.

2. SIMULATION-BASED BIOMEDICAL ENGINEERING

The simulation-based biomedical engineering will be established under an integrated combination of the computer science, the mechanics and the medicine.

First, it should provide an object-oriented software environment to create a specific computer model combining data input from medical-imaging sources as CT and MRI, geometry reconstruction containing automatic grid generation, solution strategies for multi-biophysics and visualization techniques. Development of techniques for automatically creating geometry quickly from medical-imaging sources is an outstanding issue because it will largely reduce the impediment for physicians to make a computational model, which commonly costs approximately 80-90% of an

analyst's time even within the context of computer-aided engineering design. Algorithm development for solution of multi-biophysical problems in human circulatory system is also very challenging because the sophisticated multi-branching network in our body extremely enhances the difficulties for computer modeling. We should solve problems as how to treat appropriate boundary conditions at inlet and outlet as well as the initial conditions, how to deal with the interactions between blood flow and compliant blood vessel, and how to validate our simulated results by means of *in vivo* and *in vitro* measurements or some self-consistent methodology. In addition, parallel architectures will be the systems for most of these modeling of arterial blood flow.

Secondly, from the viewpoint of mechanics, we will have some important subjects. One of the features of blood flow in arteries is the biological response to hemodynamics, which is essential to maintain normal functioning of the circulatory system and may constitute adaptation or pathological disease. We know little about the non-linear mechanics associated with this biological adaptation and it will be a challenging issue. Problem of wave propagation in a tube and multi-branched network with compliant walls is also an important issue, which may provide adequate boundary conditions in modeling blood flow in the arterial tree structure. Mass transport, e.g., the blood cells transport which functions in nutrient and waste transport throughout the body as well as the particle transport between the vessel wall and blood is another hot topic currently because it may provide important insight into understanding of the generation of arterial disease. Some mechanics linking or combining the macro-hemodynamics and this micro-mechanics may be an important task in the future.

The current simulation-based biomedical engineering will provide a useful tool for surgeon to plan their treatment or operation decision-making of cardiovascular disease. Conventionally, this decision-making process is based largely on diagnostic imaging, experience and empirical data, which are insufficient to predict the outcome of a given treatment for an individual patient because of the multitude of therapeutic choices. The current computational biomedical engineering system will provide a new paradigm for the surgery planning, which can, in a short-and long term, make the medicine more predictive in the future.

3. GLOBAL COMPUTATIONAL MODELING

Characterized by the multi-branching arterial tree structure and the pulsation of heart pumping, blood flow is of four-dimensional nature, i.e., spatial 3D and 1D in time. In a sense, some local hemodynamics may be a result of interaction of different arteries upstream and even at different instants in the same artery. This implies that conventional developed or flat inflow conditions may be not adequate for the computational modeling of blood flows in the arterial tree. Actually, our recent study has demonstrated that the secondary inflow conditions can affect the flows downstream significantly at high and medium Reynolds numbers. In order to determine the exact blood flows in a given individual's vascular system, models should faithfully represent individual anatomical features and simultaneously mimic the realistic physiologic flow conditions. We, therefore, propose a methodology of global computational modeling of blood flow in human circulatory system.

The blood vessel system is characterized by its 3D geometry that involves shape in diameter, spatial bending and twisting; its tree structure that involves branching and bifurcation with discontinuities at joining points; and its dynamic motion mostly seen in coronary arteries that involves translation, rotation, extension and contraction. As an idealized model of the first approximation, we have recently developed a feasible and efficient computer-aided method to reconstruct 3D images of geometry involving dynamic changes as well as the discretization of computational domain, and the tree structures of cardiovascular blood vessel system.

Since computational modeling of blood flow requires solving, in the general case, the three-dimensional transient-flow equations in deforming blood vessels the arbitrary-Lagrangian-Eulerian (ALE) description of media is employed, in which the fluid and wall domains are allowed to move to follow the distensible vessels and deforming fluid subdomain. The governing equations are the three-dimensional, incompressible, unsteady Navier-Stokes equations written in strong conservation form for mass and momentum. They are discretized in a manner of

finite volume method (FVM) and are solved in a time-marching manner using the pseudo-compressibility technique by adding a pseudo time derivative of pressure to the continuity equation. The time-dependent solution is obtained by introducing an inner iteration to drive the divergence of velocity to vanish at each physical time step. The algorithm is implemented by splitting the inviscid term in a manner of MUSCL scheme of a 3rd order upwind differencing, and by differencing the viscous term in a 2nd order central scheme. The implicit approximate factorization scheme of Beam & Warm is employed to decompose the governing equations into three sweeps. Details of the algorithm can be found in Liu et al. 1998.

Boundary condition specification for multi-branched arterial networks is one of the most challenging issues in modeling blood flow. We are working on 1D wave propagation model trying to deal with the outlet boundary conditions in a systematic way. The vessel walls are treated as being rigid as a first approximation though the vessel-diameter change during the cardiac cycle is often observed to be approximately 5-10% in most of the major arteries. For the dynamic effect due to acceleration / deceleration of the moving wall, pressure divergence at the wall surface, derived from the local momentum equations, is introduced for the pressure condition. At inlet, with consideration of the influence of the secondary flow the velocity profile is defined in a generalized fashion as:

$$\boldsymbol{V} = U_{axe}(\mathrm{r})\mathbf{n} + V_\theta \boldsymbol{\tau_\theta} \quad (1)$$

where U_{axe} denotes axial speed perpendicular to the local spatial plane and V_θ is circumferential velocity.

4. SIMULATION OF CARDIOVASCULAR BLOOD FLOW

As an example of global computational modeling, we chose a prototypic cardiovascular arterial tree of the heart and proximal aorta arch involving branching arteries that supply arms and brain with blood. This computational modeling is important not only because it covers most of the important issues that will be encountered in modeling of blood flow but also the vessels (let ventricle and arteries) contained in the arterial tree exhibit flow characteristics seen in most of the arterial tree and are of great importance as they often become diseased.

4.1 Preliminary results

As a first approximation, we built an idealized model with geometry and grids (61x31x21) of the aortic arch as illustrated in Fig. 1 was reconstructed based on MRI images and by using a computer-aided method which was designed to be capable to parametrically reconstruct the blood vessel system and simultaneously to generate the computational grid systems. Radii of the aorta was defined using an empirical formula $R=R_0e^{-As}$, where R_0=1.5*cm*, A=0.035 and s denotes the curve length of the center axe of the aortic arch model.

Conceptual model of pulsatile flow in a rigid-walled tube

Velocity profile of pulsatile flow in a rigid-walled tube was computed and compared to the "Womersley solution" as illustrated in Fig. 2 at a Reynolds number of 135, a Strouhal number of 0.148, corresponding to a Womersley number of approximately 5.6. The volumetric flow rate of the incoming flow was given simply as a sinusoidal function in a form of $V(t)=1+\sin(2\pi\, t/T)$. Comparison of axial velocity distribution in radial direction is performed at four instants of a half cycle, when t/T=T/8, 3T/8, 5T/8 and 7T/8, where T is the period. We see that, even with coarse grid system (11x21x15), excellent agreement is observed at all time during a complete cycle of the pulsatile flow.

Steady helical flow in an aortic arch model

Steady flow in the aortic arch model was carried out at moderate Reynolds numbers ranging over 1000-2000 under the swirl inflow conditions as described in the previous section. Obviously, a helical flow is detected. With the swirl influence at inlet the secondary flow at top of the aortic arch, i.e. the velocity vectors as shown in Fig.3*b* show significant discrepancy from those (Fig. 3*a*) without considering the swirl. The Dean vortices now largely deform reducing to a strong single helical apparently.

Unsteady helical flow in an aortic arch model

Furthermore, unsteady flow in the model was conducted at the same Reynolds numbers under a

physiological pulsatile waveform with correspondence to a physiological Womersley number of 15. Again we see a similar helical flow from the secondary flow, the velocity vectors at the top of the aortic arch as illustrated in Fig. 4*a-c* at three moments during a pulsatile cycle but apparently it shows significant time-dependence. Note that the highest axial velocities at early systole (Fig. 4*a*) begin along the inner short path, then tend to migrate outward (Fig. 4*b*) with the secondary helical flows developed, and right after the systole the helical flows still persist. Qualitatively, our results show very similar feature of the helical flows compared with the MRI-based observation by Kilner et al. 1993.

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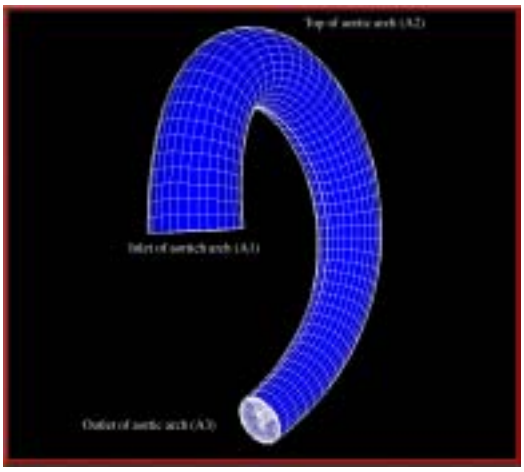


Fig. 1 A aortic arch model and computational grids

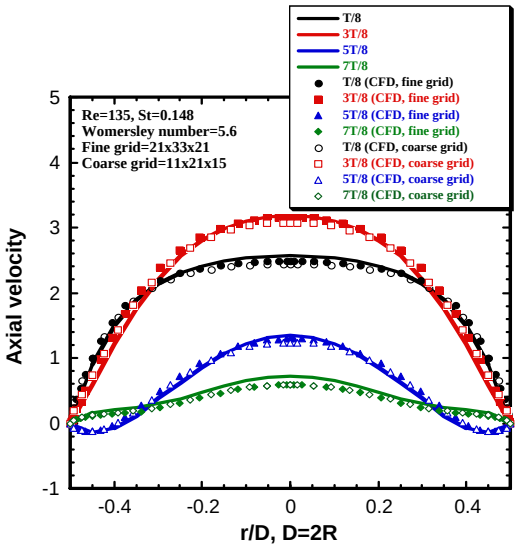
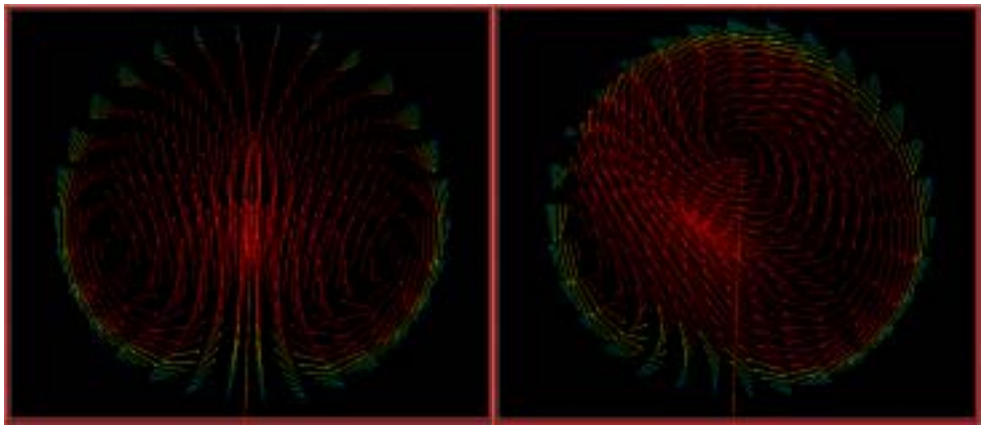
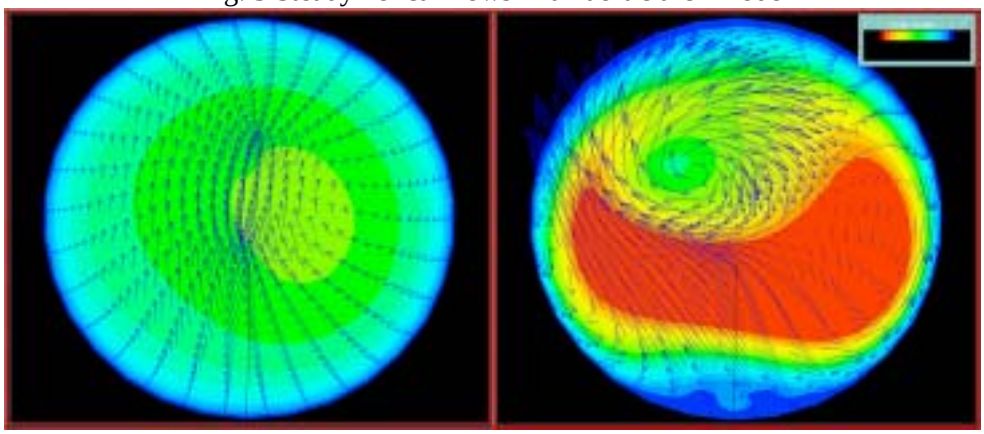


Fig. 2 Axial velocity at four instants of a half cycle in a straight tube



a without swirl at inlet *b* with swirl at inlet

Fig. 3 Steady helical flows in an aortic arch model



a early systole *b* mid-end systole

c right after systole
Fig. 4 Unsteady helical flows in an aortic arch model