Fluid-structure interactions in cardiovascular mechanics 1

Biofluid and Transport

Time: 8:30 - 10:00
Date: 12th July 2018
Location: Liffey MR2

Posters for this session are on display on Thursday 12th July in Liffey A.

Chairs: Francesco Migliavacca and Claudio Capelli

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O1602 Fluid-Structure Interaction Simulation of Heart Valve Dynamics

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Abstract

Fluid–structure interaction (FSI) in biological systems often involves multiple 3D immersed bodies with complex geometry undergoing large structural displacements and inducing very complex flow phenomena. In biomedical engineering, FSI modeling is playing an increasingly important role because the coupling of fluid flow and tissue mechanics is vital to many physical phenomena. One obvious example is the heart valve dynamics in the cardiovascular system, where the pumping of blood from the heart is the result of the large deformations experienced by the valve leaflets and the myocardium, the anisotropic nonlinear elastic behavior of the cardiac tissue, and the pulsatile hemodynamic loads during the cardiac cycle. The use of FSI cardiac models for clinical evaluation as well as for the assessment of medical devices is a topic of active investigation. In-silico testing of devices using FSI models can support better device design as well as help to expedite and augment production development. Moreover, in the domain of diagnostics and therapy planning, patient-specific models of the cardiovascular system are actively pursued, with the vision of eventually supporting decision-making process, as well as to provide insight into surgical planning applications, ultimately supporting improved clinical outcomes. The objective of this presentation is to review and discuss some of the state-of-the-art research efforts on heart valve FSI, in particular in the areas of FSI modeling of prosthetic heart valves and left ventricle-aortic valve-mitral valve coupling. The FSI methodologies, such as smoothed particle hydrodynamics coupling with FEA, will be illustrated with examples.
Computational fluid-structure interaction methods and their use in the design of cardiovascular assist devices

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Abstract

Computational fluid-structure interaction methods and their use in the design of cardiovascular assist devices

Advanced simulation is an integral part of the design process in many industries, especially in the aerospace and automotive sectors. Simulations are employed routinely as part of the product design cycle and are aimed at reducing the associated costs. When it comes to medical devices, especially those designed and built to support cardiovascular function, simulation approaches are still in their infancy, and are largely restricted to academic investigations. This may be explained by the difficult nature of the underlying physiological and mechanical phenomena, the resulting modeling challenges, and the lack of general-purpose commercial software capable of addressing these problems with the required high levels of robustness and fidelity. This presentation focuses on a class of computational fluid-structure interaction (FSI) methods developed by the author and his collaborators to address the modeling challenges in cardiovascular devices. Applications to ventricular assist devices (VADs) and bioprosthetic heart valves (BHV) are shown to illustrate the power of these methods, and their potential to address the key challenges in medical-device design.
Comparison of Hemodynamic and Structural Indices of Ascending Thoracic Aortic Aneurysm as predicted by 2-way FSI, CFD Rigid Wall Simulation and Displacement-Based FEA

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Abstract

Introduction: Hemodynamic and structural simulations of ascending thoracic aortic aneurysm (ATAA) could represent a potential tool to tailor patient-specific indications by extrapolating indicators of aneurysm risk other than aortic size. This study aims to assess whether both finite-element analysis (FEA) and computational fluid-dynamic (CFD) would give similar predictions of structural and hemodynamic parameters as compared to a more complex 2-way fluid-solid interaction (FSI) analysis of ATAA

Methods: Forty two patients with ATAA as underwent aortic size evaluation by ECG-gated CT angiography were investigated. 2-way FSI analyses were carried out using an approach previously developed by the authors’ group to couple both the structural (ABAQUS) and fluid (FLUENT) solvers [1]. ATAA wall mechanics adopted a fiber-reinforced structural model based on experimental tensile testing data and multiphoton imaging quantifications of aneurysm collagen distribution [2]. To include patient-specific hemodynamics, echocardiographic transaortic jet velocity was set as the inflow velocity condition while a Windkessel model using brachial-artery pulse pressure measurements was used for each outlet. Differently, CFD simulations were performed with the aortic valve geometry at fully opened shape whereas the loading condition of FEA consisted of a displacement field that was determined by temporal tracking of aortic luminal surfaces from diastole to systole using a mathematical algorithm [3].

Results and Discussion: The stiff ATAA wall structure (diameter change over cardiac beat of 4.2±2.4%, n.42) led to unremarkable changes in the peak wall shear stress (WSS) magnitude between CFD and 2-way FSI as demonstrated by Bland-Altman plots. However, CFD fails to predict the spatial distribution of WSS when compared to those of deformable FSI modeling. At Pearson and Spearman correlation analyses, there were no statistical differences in the helical flow index and blood pressure index of FEA/CFD versus deformable 2-way FSI. Our results using FEA analysis with patient-specific displacement-based boundary conditions highlighted that ATAA does not only expand due to blood flow but also moves from systole to diastole due to cardiac beating, with difference from patient to patient. Indeed, we found that FEA using displacement-based boundary conditions revealed a dissimilar distribution of Mises stress when compared to the flow-induced stress resulting from 2-way FSI modeling. Beyond technological improvements of FSI modeling, aortic root motion has a direct influence on aortic deformation and intramural stress distribution, and therefore should be considered to obtain reliable simulations of ATAA physiology.

Acknowledgements

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References:


A new explicit fluid-structure scheme for cardiac valves simulation

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Abstract

Introduction We are interested in the interaction of elastic thin-walled bodies immersed in an incompressible viscous fluid. Cardiac valves are the main motivation of the present study. To our knowledge, the fluid-structure coupling schemes used in this context are usually implicit or semi-implicit. This yields unconditional stability but at the price of solving a coupled system at every time step. The design of explicit coupling schemes - i.e., schemes that call the two solvers only once per time-step - is of major interest, especially for three-dimensional studies. However, the major drawback of the existing approaches (e.g., [2]) is that either stability or accuracy demands severe time-step restrictions (e.g., parabolic-CFL) or correction iterations. We propose a new solution that overcomes these difficulties.

Methods The fluid is modelled with the Navier-Stokes equations in a Eulerian setting, and the valves are described by a Reissner-Mindlin shell type model in Lagrangian formulation. The coupling is enforced through Lagrange multipliers [3]. Contrary to what is usually done in Fictitious Domains (e.g., [1]), the Lagrange multipliers are approximated with Dirac masses. Our coupling scheme is an extension of the Robin-Neumann splitting proposed in [4] to the present unfitted mesh framework. It treats implicitly the coupling of the fluid with the solid inertia and explicitly the coupling with the solid elastic effects. In addition, the choice of Dirac masses for the Lagrange multipliers, combined with a consistent lumped-mass approximation in the solid, yield a very efficient implementation in the fluid solver which makes the coupling scheme explicit. The resulting scheme is provably stable in the energy norm.

Results Numerical examples, including cardiac valves, will be exposed. Comparisons in terms of accuracy and computational efficiency with respect to the implicit scheme will be discussed and will illustrate the benefits of the new explicit scheme.

Discussion The new method is very promising and outperforms the coupling schemes we are aware of. Nevertheless, an efficient management of the contact still has to be devised in order to address cardiac valves in realistic physiological conditions.

Acknowledgements This work has been supported by the project MIVANA and the companies Kephalios and Epygon.

References


O1606 Fluid-Structure Interaction analysis of a total artificial heart

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Abstract

Study: Heart failure is a progressive and often fatal pathology and one of the main causes of death in the world. An alternative to heart transplantation is a totally implantable artificial heart (TAH). The CARMAT ‘embedded biomimetic heart’ is a bioprosthetic pulsatile electro-hydraulic biventricular support to replace heart in patients with and-stage heart failure. It mimics the functioning of the natural heart, minimizing the risk of thrombosis and offering patients independence [1]. The preclinical evaluation of the hemodynamic shear stress by using a robust numerical methodology is then essential to predict the hemocompatibility performance and to optimize the ventricular design. Computational Fluid-dynamics (CFD) and Fluid-Structure interaction (FSI) models can be employed to investigate the fluid dynamic behavior of blood, as well as the kinematics of the TAH components.

The aim of the study is to develop a computational model able to accurately reproduce the hemodynamics inside the left chamber of the TAH together with the displacement of the aortic and mitral valves and the membrane. The modeling strategy combines a FSI simulation based on a Non-Boundary Fitted Method with a CFD analysis that includes moving boundaries. In particular, the movement of the valves and the membrane is considered in the CFD simulation. Lastly, the evaluation of the critical hemodynamics indexed to predict the blood damage are calculated based on the CFD results.

Methods: The left chamber is composed of oil and blood compartments, with a thin membrane separating the two fluids. The flow into the oil compartment deforms the membrane, which drives the blood into the ventricle through the mitral valve and out through the aortic valve. Oil flow rate (5.4 L/min), static atrial pressure (0 mmHg) and aortic pressure (120 mmHg) were applied as boundary conditions. Subsequently, CFD analyses of the blood domain were carried out by prescribing the membrane displacement obtained from the FSI simulation with the dynamic mesh. Two different CFD simulations were carried out, with and without the valves. Results were achieved in terms of membrane displacement, valve kinematics and shear stress, pressure and velocity fields of blood.

Results: Results from FSI simulations led to realistic kinematics of the valve leaflets and membrane. The comparison between FSI and CFD velocity fields confirmed that the presence of the valves in the CFD model is essential to reproduce realistic blood dynamics. The adopted strategy of sequential simulations allowed us to obtain accurate fluid-dynamic features, hardly caught by the FSI analyses (based on a fixed grid technique), such as wall shear stresses, vortices and local vector velocity fields nearby the interface between solid structures and fluid.

Modelling of a human mitral valve within left ventricle with fluid-structure interaction

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Abstract

Computational modelling of the mitral valve mechanics, particularly within the context of the left ventricle (LV) environment, can improve our understanding of the valvular-ventricular interaction, and potentially lead to more efficient MV repair and replacement. To date, very few studies have reported valvular-heart modelling with fluid-structure interaction (FSI). In this study, we present an integrated MV-LV model based on in vivo magnetic resonance images of a healthy volunteer. The new MV-LV model incorporates important valvular features, LV contraction, nonlinear soft tissue mechanics, and FSI.

The MV-LV model is developed using hybrid immersed boundary-finite element framework implemented within the open-source IBAMR software. We first use this model to simulate cardiac function from diastole to systole for a healthy volunteer. In diastole, a populated-based LV end-diastolic pressure (EDP=8 mmHg) is applied to the left atrium side. To model myocardial active relaxation in diastole, an additional pressure ($P_{endo}$=16 mmHg) is applied directly onto the endocardial surface. In systole, myocardial active contraction is triggered simultaneously across the whole ventricle. Our results agree with in vivo measurements reasonably well. For example, in diastole, blood enters into ventricle with increased volumetric flow rates across the MV, and gradually decreases towards end-of-diastole. In systole, the increased LV pressure due to the myocardial contraction closes the MV and pumps blood through the aorta valve (AV). The maximum flow rate across the AV is 468 mL/s, the total ejection duration is around 243 ms, and the ejection fraction is 51%. These are comparable to clinical measurements.

We further investigate how $P_{endo}$ affects the MV-LV dynamics by varying its value from 8 mmHg to 16 mmHg, and the effects without $P_{endo}$ but with increased EDP from 8 mmHg to 20 mmHg. We find that with an increased $P_{endo}$, the peak flow rate across the MV during the filling phase becomes higher with more ejected volume through the aortic valve, associated with more prolonged ejection duration, shorter isovolumetric contraction duration, and higher ejection fraction. On the other hand, if we do not apply $P_{endo}$, a non-physiologically high EDP is needed for the required ejection fraction. For example, with EDP=8 mmHg, the ejection fraction is only 29%. Only when EDP=20 mmHg, the LV pump function is comparable to the case with EDP=8 mmHg and $P_{endo}$=16 mmHg.

In summary, a coupled valvular-heart model has the potential to quantify cardiac function in a more comprehensive way and can provide more pathological details for risk stratification.

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INTRODUCTION

Cardiovascular minimally invasive interventions have revolutionized the way to handle congenital [1] and acquired structural heart diseases [2]. The success of planning such interventions also relies on the capability of predicting the mechanical interactions of devices and implantation site [3]. While current imaging technique allows an accurate description of the implantation site, inferring its mechanical properties still represents a challenging task. The aim of this study was to develop a framework of modelling able to include specific mechanical characteristics as derived from imaging techniques.

METHODS

An image-based framework was set up to extrapolate mechanical properties of cardiovascular structures from Phase Contrast Magnetic Resonance Imaging (PC-MRI). This framework was tested both with in vitro and in vivo data. First, an experimental mock circulatory loop (MCL) was assembled to acquire characteristics of a 3D-printed TangoBlackPlus FLX980 (TP) [4] model of a blood vessel under pulsatile conditions. The MCL was positioned inside the scanner to acquire the PC-MRI of the phantom's middle-cross section, while sensors registered simultaneously flow and pressure data. Flow-area loop (QAL) method [5] was used to estimate phantom's material properties in terms of Young's modulus (E). The TP elastic modulus was also compared against standard uniaxial tensile tests conducted on fifteen 3D-printed TP dogbone-like specimens. The QAL was applied also on a PC-MRI dataset of a real patient case to derive in vivo mechanical characteristics. Finally, fluid-structural interaction (FSI) simulations were conducted on both simplified and patient-specific models (ANSYS Workbench 17.2) to evaluate the predictive capability of the models in terms of area deformation.

RESULTS

QAL applied to the segmentation of phantom's PC-MRI data provided a Young's modulus of 0.22±0.04MPa, while tensile tests computed E=0.50±0.02MPa. Results of FSI simulations for E=0.50MPa showed excellent agreement with PC-MRI measurement in terms of deformed area (i.e. absolute difference=1.28mm², equivalent to 0.87%). A difference was found for the QAL based case (i.e. absolute difference=32.78mm², equivalent to 18.67%). Concerning the clinical case, the E value was estimated to be 0.06MPa and the difference between FSI simulation and PC-MRI maximum area was 23.15% (i.e. 148.31mm²).

DISCUSSION

Results of this preliminary study show the feasibility of a non-invasive method to improve the current patient-specific modelling of cardiovascular structures from PC-MRI (see figure). Further investigations should be conducted to reduce the differences between direct and image-based
measurements, with the final aim of refining modelling environments to predict the mechanical feasibility of cardiovascular interventions for each patient.

**ACKNOWLEDGEMENT**

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**REFERENCES**

On the chordae structure and dynamic behaviour of mitral valve

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Abstract

Introduction

Mitral valve (MV) dysfunction, including mitral valve stenosis, prolapse, and regurgitation, is one of the most common valvular heart diseases and hence has attracted significant research interest. Computational modelling of human MV function can improve our understanding of MV biomechanics, which is important for improving surgical procedures and medical therapies. However, because of the challenges of modelling the highly complex MV structure, its deformation, and its interaction with the left ventricle, only limited progress in multi-physics modelling of the MV has been made to date.

We develop a mitral valve (MV) model that includes physiologically detailed descriptions of the leaflets and the chordae tendineae. Three different chordae models — complex, “pseudo-fibre”, and simplified chordae — are compared to determine how different chordae representations affect the dynamics of the mitral valve. To quantify the highly complex system behaviour resulting from the fluid-structure interaction (FSI), an energy budget analysis of the coupled MV FSI model is performed.

Methods

Our MV model is based on physiologically-detailed MV geometry from multi-slice CT scans of a normal mitral valve at middle diastole, from a 61-year-old male patient, which incorporates detailed leaflet thickness and chordal information. Both leaflets and chordae are modelled as fibre-reinforced hyperelastic materials.

We use an immersed boundary-finite element (IB/FE) method for our dynamic MV modelling. The simulation has been carefully verified against the commercial software ABAQUS under static loading conditions. Energy budget is conducted with introduction of different energy terms appearing in the system: change of the kinetic energy, kinetic energy flux, rate of work done by the applied pressure, the rate of energy dissipation and the rate of change of elastic strain energy in the immersed structure.

Results & Discussion

Our results show that the complex and pseudo-fibre chordae models yield good MV closure during systole, but that the simplified chordae model leads to poorer leaflet coaptation and an unrealistic bulge in anterior leaflet belly. Energy budget analysis shows that the MV models with complex and pseudo-fibre chordae have similar energy distribution patterns, but the MV model with the simplified chordae consumes more energy, especially during valve closing and opening. Interesting flow patterns and vortex formulation are seen in all three cases.

In general, we show that the complex chordae and pseudo-fibre chordae have similar impact on the overall MV function, but that the simplified chordae representation is less accurate. Because a pseudo-fibre chordal structure is much easier to construct and less computationally intensive, it may be a good candidate for modelling MV dynamics or interaction between the MV and the heart in patient-specific applications.

Acknowledgements

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Waiver Programme of the University of Glasgow.
A numerical approach for the analysis of carotid atherosclerosis severity

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Abstract

INTRODUCTION

Strongly related to the pulsatile flow ejected from the left ventricle, the geometry of the arteries and the compliance of the microcirculation, the mechanical properties of the arterial wall play an important role in determining the characteristics of blood flow within the arterial system.

The presented work aims to develop a tool to help surgeons in defining the risk level of plaque growth and rupture in stenosed carotid artery (CA), in order to prescribe adequate treatment for patients. The correlations between the blood flow parameters and the stability of the plaques in the carotid bifurcation is investigated using both Computational Fluid Dynamics (CFD) and full Fluid Structure Interaction (FSI) computational models.

METHODS

The pulsatile blood pressure causes the periodic deformation of the carotid wall. Therefore, in order to accurately describe the phenomenon, a fluid-structure interaction (FSI) model has been opted for.

CFD and FSI models of the CA were implemented using both simplified and patient specific geometries. A variation of hemodynamic simulations parameters was conducted in order to verify their accuracy. Hyperelastic material was adopted for the arterial wall while the blood was assumed a generalized newton fluid.

RESULTS AND DISCUSSION

The comparison between healthy and stenosed arteries showed a diversity of flow disruptions near the stenosis area resulting in a high shear stress in the plaque zone and low shear stress in the after plaque zone (Figure 1), in agreement with recent works showing that this disruption in the flow behavior may cause endothelial dysfunction (1) (2).
Figure 1: right: WSS distribution for stenosed and healthy carotid artery at t=1.7s (systolic phase) Left: 5a: Wall shear stress at after plaque area 1b: Wall shear stress at plaque area during 3 cardiac cycles

The plaque rupture is affected by the local WSS and the local strength of the tissue. Low shear stress (LSS) zones are areas of vivid proliferation of atherosclerosis plaques. The LSS can promote inflammations and cause plaque instability (3).

The presented results highly depend on the geometry and the inflow conditions, which confirms the necessity of patient specific simulations to obtain more accurate results.

BIBLIOGRAPHY


Instability of elastic tubes conveying power law fluids

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Abstract

Introduction
At the present day, the modeling of elastic tubes conveying Newtonian fluids is used for studying the dynamics of biological fluids. Instability of elastic tubes naturally occurs in several physiological processes [1, 2], e.g. blood flow in arteries. When the pressure inside of the tube is substantially lower than external pressure, the tube loses stability that yields the onset of oscillation. It is known, however, that blood in small vessels demonstrates non-Newtonian rheology, which previously was not taken into account in stability analyses. This paper is devoted to investigation of flutter of elastic tubes conveying power law fluid based on a simple 1-D model, which similarly used in [3].

Methods
3-D equation of motion (Navier-Stokes equation for the power law fluid and the tube equation of motion) are reduced to 1-D problem by integrating over the tube cross-section under assumptions: (a) wave length is large comparing to the tube radius \( R \); (b) Wave frequency is sufficiently small so that the flow inside the tube can be considered as quasi-steady. Finally, the dimensionless system of two 1-D equations is derived:

\[
\frac{\partial Q}{\partial z} + \frac{\partial (\pi R^2)}{\partial t} = 0,
\]

\[
\frac{\partial Q}{\partial t} + \frac{\partial}{\partial z} \left( \frac{(3n + 1)Q^2}{(2n + 1)\pi R^2} \right) + \frac{16Q^n}{\pi ReR^{3n+1}} + \frac{1}{\pi R^2} \frac{\partial R}{\partial z} + \frac{\partial^3 R}{\partial z \partial t^2} - \frac{\pi R^2 N}{\partial z^3} = 0.
\]

Where \( Q \) is flow rate, \( \beta \) is the radial stiffness of the tube, \( m \) is wall surface density, \( \lambda \) is the longitudinal tube tension, and \( n \) is power law index. For Newtonian fluid \( n=1 \). Case of \( n<1 \) corresponds to pseudoplastic media; case of \( n>1 \) corresponds to dilatant media.

Results
Analyzing this system, instability conditions of axisymmetric long-wave are derived in a closed form. The instability is possible only for \( n=0.611 \) (Fig.1).
Condition on the tube stiffness and power law index \( n \) are found such that local growth of axisymmetric perturbations is possible. Absolute instability can occur only for \( n<1/3 \) (Fig.1). This instability is not related to non-axisymmetric collapse of the tube always observed at flutter of collapsible tubes, and can occur at essentially positive transmural pressures. Instability analysis conducted in this study can be important for the blood transport modeling in the cardiovascular system.
**Fig. 1.** Regions of absolute (AI) and convective (CI) instability.

**References**

A study on stent-artery Compliance and wall shear stress using fluid -structure interaction simulation

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Abstract

Cardiovascular stents are used to treat coronary artery disease; however the success is limited by restenosis after the stent implantation. Presence of stent alters the haemodynamics near the walls. Low wall shear-stress promotes neointimal hyperplasia that leads to restenosis.

Computational fluid dynamics (CFD) is used to study effects of stent design parameters on blood flow and walls shear stress. However in these studies the artery wall and stent are considered rigid. Although FSI simulations require greater computational resources but can provide more insight in to other parameters such as the effect of stent material property on deformation and artery stresses during the pulsatile blood flow. The objective of this study is to investigate the effect of stent artery compliance on hemodynamic quantities using a fluid structure interaction (FSI). Simulation are carried out for rigid (no FSI) and realistic (FSI) model. In FSI model the artery wall and stent were modeled as linear elastic and blood is modeled as incompressible Newtonian fluid. To study the effect of stent material compliance mismatch, metal and polymer material properties for stent are considered. The result for rigid and FSI models show similar trends for wall shear stress distribution and justify the use of rigid wall assumption. The comparative result for two different material properties show that wall shear stresses and wall strains are higher while wall stresses are lower at the more compliant section.
References


P4525 The nonlinear dynamics of Woven Dacron Aortic Prostheses conveying pulsatile blood flow

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Abstract

Introduction

Large diameter (12-30 mm) cardiovascular vessel replacements with Dacron prostheses still represent the current accepted clinical practice. Surgeons perform vascular prosthesis implantation to exclude the compromised arterial portion from luminal pulsatile blood flow. Wide knowledge about the distinctly different mechanical properties of the Dacron implants with respect to the native arteries is available in literature [1] while the dynamic behavior of these prostheses has not yet been investigated.

Methods

We model a woven Dacron thoracic prosthesis with the nonlinear Novozhilov shell theory applied to an orthotropic circular cylindrical slightly corrugated shell. The hyperelasticity of the fabric is also considered. Material properties have been obtained experimentally on a sample of a Maquet 175428P Hemashield Platinum Woven Double Velour Vascular Graft.

Surface waves of the corrugation can be modelled introducing a sinusoidal geometric imperfection on the shell geometry for both axial and radial displacements. A pulsatile blood flow model is considered by applying physiological waveforms of velocity and pressure during the heart beating period approximated through Fourier series [2]. Coupled fluid-structure Lagrange equations of motion for a non-material volume with wave propagation in case of pulsatile flow are employed. Two boundary conditions reproducing the continuous monofilament suture and the simple interrupted suture are investigated.

Preliminary experimental results on the dynamics of the Dacron prosthesis subjected to pulsatile flow are presented. A hydraulic circuit has been built with a Harvard pulsatile blood pump, a compliant chamber, magnetic flowmeters, catheter pressure sensors. The dynamical response of the prosthesis is measured by the non-contact laser doppler sensors.

Results

Several superharmonic resonance peaks appear in the physiological frequency range by including higher harmonics in the Fourier expansion of the physiological waveforms of pressure and velocity. Large amplitude oscillations are observed in the physiological frequency range during exercise conditions.

Discussion

Since vibrations of the artificial vessel walls are activated for certain heart rates, the related high stress concentration combined with the fatigue cycles of the heart beats, could contribute to material deterioration. The growing understanding of the dynamic behavior of vascular prostheses used in clinical practice could help controlling their common long-term adverse
effects and might inspire the design of new generation prosthesis.

Acknowledgements

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References


Abstract

Introduction

Short term results of coarctation of the aorta (CoA) treatment with surgery, balloon dilation (BD) or stent implantation are excellent. Yet, long-term cardiovascular morbidity and mortality remain high. The effects of treatment modality on long-term vascular function remain uncharacterized. The goal of this study is to demonstrate the feasibility of a fluid-structure interaction (FSI) computational modeling based on patient specific data to assess vascular function in this patient population.

Methods

We created a FSI finite element numerical approach based on patient-specific anatomy and functional data, followed by the computation of hemodynamic and biomechanical biomarkers. We selected two CoA patients per treatment type. The aortic lumen and wall were reconstructed from patient-specific cardiac magnetic resonance (CMR) data. The FSI computational model used the Arbitrary Lagrangian-Eulerian formulation assuming that blood flow was governed by the Navier-Stokes equations for Newtonian and incompressible fluids and that the aortic wall was linear, elastic and isotropic in its structure and its stiffness based on patient-specific functional data. Velocity waveforms based on patient-specific flow rates derived from CMR were mapped into the aortic inlet and descending aortic outlet. At the remaining outlets, we imposed linear absorbing boundary conditions to avert spurious backflow.

Results

Peak velocities at the intervention site were 2.5 m/s for stent, 1.8 m/s for surgery and 1.6 m/s for BD. Qualitatively, flow vortices were more marked in the stent than in the surgery and BD groups. Peak wall shear stress (WSS) was between 200 and 250 dyn/cm² for all groups (Fig.1), but the area-averaged WSS varied (157 and 168 dyn/cm² for the two stent patients, 60 and 180 dyn/cm² for the two surgery patients and 90 to 115 dyn/cm² for the two BD patients). In addition, the two BD patients had the lowest area of luminal surface subjected to elevated WSS (260 and 330 mm²) compared to the two surgery (530 and 1150 mm²) and two stent patients (910 and 1050 mm²). Finally, the aortic compliance assessment showed that maximum wall displacement was 1.64 and 2.26 mm for the BD patients, 0.67 and 1.47 mm for the surgery and
1.12 and 0.6 mm for the stent patients.
Conclusions

FSI computational modeling based on patient-specific data is feasible. Our preliminary results suggest that none of the three CoA treatments restores normal aortic hemodynamics, particularly at the intervention site. This exploratory study suggests that CoA treatment with BD may be associated with better preservation of the aortic vascular function while stent placement with worse vascular outcomes. Our results require validation in large-scale studies and FSI may become a useful adjunct tool in assessing the post-treatment hemodynamics in patients with CoA.

Fig.1: WSS (dyn/cm²) at the intervention site for all patients, throughout the cardiac cycle.
BD1 and BD2 – Balloon dilation patients; Surg1 and Surg2 – Surgical repair patients; Stent1 and Stent2 – Stent placement patients.
Computational Fluid Dynamics with Fluid-Structure Interaction in FEBio

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Abstract

We have implemented a novel framework for computational fluid dynamics (CFD) with fluid-structure interaction (FSI) into FEBio, a free, open-source finite element software specialized for the biomechanics community.\textsuperscript{1} FSI is increasingly utilized in biomechanics for physiologically modeling complex biological processes such as cardiovascular mechanics, cerebrospinal mechanics, and joint lubrication, among many others. Traditionally, FSI methods follow either the arbitrary Lagrangian-Eulerian or the immersed boundary approach.\textsuperscript{2,3} Instead, we use mixture theory, successfully employed in FEBio for modeling reactive deformable porous media,\textsuperscript{4} while extending our recent CFD implementation.\textsuperscript{5} The FSI-domain is modeled as an isothermal, viscous, and compressible fluid-solid porous mixture. The solid component is assumed to have negligible mass, stiffness, and frictional drag, thus acting as a freely deformable mesh containing the fluid.

Nodal degrees of freedom include the solid deformation, the fluid velocity relative to the solid, and fluid dilatation. Virtual work equations for the FSI-domain enforce momentum balance for each constituent, simplified with the above assumptions. We also use the integrated form of the mass balance for the solid and fluid, and include the kinematic constraint between fluid dilatation and velocity. Surrounding solid domains are modeled using finite deformation solid mechanics. The solid and FSI-domains are strongly coupled at their interfaces. The equations are linearized, discretized, and solved using Newton’s method.
Our FSI formulation was verified by ensuring energy conservation and comparing against benchmark problems.\textsuperscript{6,7} In Figure 1, we inspect a pulse wave through a bifurcated artery upon application of a pulsatile inlet velocity profile. With only CFD, the fluid pressure was established instantly as it propagated through a rigid wall. In FSI, the pulse wave propagated finitely over time due to the flexible arterial wall, consistent with physiological observations.

This study describes the implementation and validation of a novel FSI formulation in FEBio. The fluid domain is represented as a special fluid-solid mixture. Unlike traditional methods, the FSI equations are in the Eulerian frame and solved monolithically. We use our previous approach of using dilatation as an unknown, which does not require stabilization schemes.\textsuperscript{5} This framework vastly expands the modeling capabilities of FEBio. We also plan to implement an iterative solving scheme and adaptive meshing for efficiency.

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REFERENCES


P4528 Evaluation of compliance of pre-stressed patient-specific pulmonary vasculature using blood-flow arterial wall interaction and a shape matching inverse algorithm

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Abstract

Introduction. Accurate evaluation of blood arterial-wall interaction (B-AW-I) aids diagnosis of right-heart disease (RHD) such as pulmonary hypertension, where flow reduction ($Q$) in pulmonary arteries (PAs) results in increased upstream pressure ($p$). Previous B-AW-I studies have not modeled the pre-stress in the artery due to $p$ loading and tethering, potentially overestimating the wall compliance ($c$), defined as the ratio of maximum difference in lumen area to the corresponding $p_{\text{max}} - p_{\text{min}}$ over the cardiac cycle. This study utilizes an inverse algorithm to determine the \textit{in vivo} pre-stress and obtain PA compliance.

Methods. The PA vasculature was reconstructed from angiographic MRI. Phase-contrast MRI provided the transient $Q$ profiles at the main, left- and right-branch PAs (MPA, LPA, RPA), while catheterization data provided time-varying pressure at the MPA ($p_{\text{MPA}}$) and LPA ($p_{\text{LPA}}$). The Mooney-Rivlin model was used to formulate the hyperelastic response of the arterial wall, and constants were obtained from pressure-diameter ($p$-$d$) relationships using the method described by D’Souza et al (Prog. Ped. Cardio. 2017). The LPA’s $p$-$d$ relationship was found to be different compared to MPA, which is accounted for by using different Mooney-Rivlin models. A shape matching algorithm was implemented where forward Finite Element computations were combined with backward displacements on the \textit{in vivo} artery to obtain the load-free artery. The B-AW-I was computed under patient-specific boundary conditions, and the compliances, $c_{\text{MPA}}$ and $c_{\text{LPA}}$ were calculated.

Results. The load-free and \textit{in vivo} PA with the location of PC-MRI planes is shown in Figure 1A. The load-free shape showed a 33% lower outer $d$ and a 24% lower inner $d$ compared to \textit{in vivo} geometry. Computed $p$ and $Q$ were validated with experiments (Figure 1B; $N$- numerical, $E$- experimental) where $p_{\text{MPA,N}}$ matched generally well with $p_{\text{MPA,E}}$; the mean $p_{\text{MPA,N}}$ was 7% higher than mean $p_{\text{MPA,E}}$ over the cardiac cycle. Similarly, $Q_{\text{LPA,N}}$ agreed well with $Q_{\text{LPA,E}}$ at all-time points; mean $Q_{\text{LPA,N}}$ was 11% lower than mean $Q_{\text{LPA,E}}$. The numerical and experimental $p$-$d$ relationships were compared for the MPA and LPA, which were then utilized to compute $c_{\text{MPA}}$ and $c_{\text{LPA}}$. Although higher $d_N$ are seen compared to $d_E$, the slopes of the numerical $p$-$d$ curves follow the experimental data. $c_{\text{LPA,N}}$ was 0.044 mm$^2$/Hg, which was 11.8% lower than $c_{\text{LPA,E}}$ (0.05 mm$^2$/Hg). Similarly, $c_{\text{MPA,N}}$ was 0.132 mm$^2$/Hg which was 31.7% lower than $c_{\text{MPA,E}}$ (0.194 mm$^2$/Hg). Importantly, $c_{\text{MPA}}$ was found to be 2-3 times higher than $c_{\text{LPA}}$.

Conclusion. Patient-specific numerical models of B-AW-I incorporating wall pre-stress can evaluate compliance of healthy or diseased vasculature, thus aiding the diagnosis of RHD.
Figure 1. A) Load-free vs. in vivo shape for the PA. B) Pressure (p) and flow (Q) comparisons for the MPA and LPA, respectively. C) Pressure-diameter (p-d) relationship comparison for the MPA and LPA.
Abstract

Introduction

Competent mitral valve (MV) closure depends on a well-orchestrated force balance across the anterior and posterior (PML) leaflets, the chordae tendineae, the mitral annulus, and the papillary muscles. Chordae rupture disrupts the link between the MV and the left ventricle (LV), causing mitral regurgitation (MR), the most common valvular disease. Cardiac modeling can help us better understand the functional and biomechanical environment of the LV-valve complex under healthy and diseased states. However, accurately modeling left heart (LH) dynamics has been challenging, since it is a fluid-structure interaction (FSI) phenomenon that involves large deformations experienced by the leaflets and the cardiac wall, the nonlinear elastic behavior of the valvular tissue, and the pulsatile hemodynamic load. Towards this goal, FSI LH models with detailed chordae structure are develop in this study to investigate the impact of PML chordae rupture on LV-valve dynamics.

Methods

A FSI framework that combines smoothed particle hydrodynamics and nonlinear finite element formulation is used. A previously developed subject-specific LH model [1] is adapted to simulate 7 chordae rupture conditions, representing a pathological process in which minimal chordae rupture precedes a more extensive rupture. The models comprise all major components of the LH (aortic root, aortic valve, LV, MV and left atrium-LA) (Fig 1a), as well as incorporate imaged-based cardiac wall motion, anisotropic hyperelastic material models, and human material properties. FSI simulations during the full cardiac cycle are performed, and the effects on MR severity, LH hemodynamics, cardiac efficiency, MV force balance, and leaflets deformation state are investigated.

Results and discussion

Based on the regurgitant volume (RV), 3 models were classified as having mild MR, 2 moderate MR, and 2 severe MR (Fig 1b). Different regurgitant jets were identified from the velocity streamlines during systole. While some non-eccentric jets were directed towards the central LA region, other eccentric jets appeared to be attracted towards the LA wall (Fig 1c). Cardiac efficiency was evaluated by the ratio of external stroke work, with a value of 75.64% for the control model. When MR was present, there was a progressive reduction in the pump efficiency due to the ineffective work of the RV (71.82% - 22.86%). Results also showed that basal/strut chordae were the major load-bearing chordae. An increased number of ruptured chordae resulted in reduced basal/strut tension, but increased marginal/intermediate load. Chordae rupture in a specific scallop did not necessarily involve an increase in the stress of the entire prolapsed leaflet. This work represents a further step towards patient-specific cardiac modeling and has the potential to improve our understanding of the biomechanical mechanisms of MR.
Fig 1. a) LH FSI model, b) flow rate across the AV and MV, and c) representative velocity streamlines at peak systole showing regurgitant jet structures.

Abstract

Introduction

Patients with a severely stenosed carotid artery are at risk of stroke. Carotid endarterectomy is performed to prevent possible plaque rupture. However, only one out of six patients benefits from this risky intervention. To improve upon clinical decision making, mechanical characterization of the stenosed carotid has been proposed, including the assessment of wall (shear) stress and material properties. However, patient-specific models require besides an accurate geometry high resolution functional data such as wall motion and flow.

In this study, a method was developed to perform patient-specific, 3-D, Fluid-Structure Interaction modelling of the carotid using ultrasound. Rather than using a 3-D ultrasound probe, the required high spatial and temporal resolution of 2-D ultrasound are used in combination with a magnetic probe tracking device.

Methods

With a novel automatic segmentation algorithm, we were able to create a simulation of the bifurcated carotid artery with minimal user interaction. Feature asymmetry (FA) of the original B-mode ultrasound is used as input for a longitudinally propagating active contour model (snake) to detect the boundaries of the lumen in 3-D. After segmentation, the frames during systole and outliers are removed. The segmented contours are transformed into a 3-D geometry using the probe tracking coordinates. The contours are converted in a surface mesh, which is smoothed using curvature flow smoothing. The patient-specific wall thickness was based on an IMT measurement, adding 30% for the adventitial layer which is not visible in ultrasound. A hexahedral mesh is created by extruding the surface mesh in the radial direction.

The method proposed is tested in vivo on 20 healthy volunteers. Four measurements per patient were performed, a slow (20s) and a fast (5s) sweep on both the left and right carotid artery. The automated segmentation were compared with manual segmentations by three sonographers. Finally, FSI was simulated to estimate the wall stress in the carotid artery and flow through the lumen, using the measured geometry, center velocity profile (Doppler) and brachial arterial pressure as input, while using a generic material model (Neo-Hookean).

Results/Discussion

The segmentation algorithm performed well with an average Similarity Index of 0.89 and an average Hausdorff distance of 1.0 mm. The estimated 99 percentile wall stresses and velocity profiles were within the range found in literature for young adults. The next step is to obtain patient specific material properties by matching the model displacements with the measured displacements with ultrasound, by iteratively updating the material properties.
Fig. 1
(A) Automatically segmented contours at different positions in the carotid artery.
(B) Lumen of bifurcated carotid artery in relation to the acquired B-mode images making a sweep.
(C) Patient-specific wall stress analysis using the finite element method.
Modelling of a continuous tornado-like blood flow in the channel of left atrium, left ventricle, and aorta

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Abstract

INTRODUCTION

A new approach has been proposed to the study of blood circulation in the heart and main arteries, based on the particular exact solution of nonstationary hydrodynamic equations for the class of self-organizing tornado-like viscous flows which was published in 1986. The study of these flows revealed a stable violation of the Reynolds analogy in the favor of heat transfer, due to a significant decrease in resistance to the streamlined surface.

The properties of tornado-like flows are similar to that of the blood flow. They are self-organizing swirling flows, characterized by a low energy losses and high mass transfer efficiency because of additional gradients occurred due to the inertia of swirling jet rotation and the special structure of boundary layer which is free from the shear stresses. The mechanism of swirling blood flow formation is determined by the instantaneous geometric characteristics of flowing channel of the heart and main arteries, changing simultaneously with the flow evolution throughout the cardiac cycle.

METHODS

The analysis was carried out basing on experimental measurements of the dynamic configuration of flow channel of the heart and the aorta using morphometry of postmortem casts, cine-angiography, aortic elastometry, MRI velocimetry, and MSCT. The exact solution of nonstationary hydrodynamic equations for a singular tornado-like jet in an attached moving cylindrical coordinate system was used as a hydrodynamic model.

RESULTS

The model of continuous tornado-like blood flow in a pulsating mode and moving boundaries was developed from the origin of the jet in the left atrium cavity up to the distal part of the Aorta. The model assumes that the generation of swirling blood flow occurs in the left atrium due to specific geometric configuration of the cavity, and directions of the flows coming from the pulmonary veins and the left atrium auricle. The flow in the left ventricular cavity is determined by dynamic configuration of the trabecular profile over the inner surface of left ventricular wall that supports evolution of the flow corresponding to the dynamic directions of tornado-like jet streamlines throughout the cardiac cycle. The aorta lumen geometry ensures the preservation of conditions for self-organization and maintenance of tornado-like jet a.o. elasticity change along the Aorta’s length.

DISCUSSION

Analysis and modeling of blood flow using the phenomenon of tornado-like self-organization allow the universal consistent concept formulation linking the flow structure and the dynamic geometry of flow channel in a wide range of states. It has been shown that in the circulation arterial segment, the necessary conditions continuously present ensuring the generation and maintenance of the dominant tornado-like jet evolution throughout the cardiac cycle.

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Validation of a direct forcing immersed boundary method solver with an application towards artificial valves

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Abstract

Introduction

Heart Valves ensure unidirectional flow through the circulatory system. Their complex geometries are often subject to calcification and impaired movement, necessitating the need for replacements. Prosthetic valves are one such replacement. While prosthetic valves are less likely to fail due to thrombosis than their mechanical counterparts, they still necessitate accurate 3D simulations to understand their short-comings. The ability to accurately predict flow patterns inside pulsatile devices necessitates fluid structure interaction (FSI) modeling, as the devices’ flexible structures constantly interact with blood. Previously, our group used the classical immersed thick boundary method (ITBM) approach [1], where the membrane is transported by the fluid. Here, we validate and explore a direct forcing (DF) approach, where membrane movement is calculated on the leaflets themselves.

Methods

DF methods were validated within the massively parallel in-house YALES2BIO finite volume Navier-Stokes solver [2]. The membrane force $F_m$, resulting from membrane interaction with the fluid was calculated as described in [3,4]:

$$F_m = \frac{U^d(X_k) - U^*(X_k)}{\Delta t}$$ (1)

Where $U^d$ represents the desired velocity for the solid membrane (obtained by solving the structural mechanics problem for the valve leaflets) and $U^*$ the predicted velocity for the time step, $X_k$, the solid Lagrangian membrane markers. The force was then regularized, through the use of a discrete Dirac function, on the surrounding fluid nodes and added to the Navier-Stokes equation before system (2) was solved.

$$\partial_t u + (u \cdot \nabla)u + \nabla p - \nu \nabla^2 u = f$$ (2)

Following validation through a number of 2D tests that include freely falling and rotating cylinders, the code was extended to 3D. The YALES2BIO solver was then coupled to LMGC90, finite element structural mechanics solver, in order to account for leaflet properties as seen in [1].

Results

DF proved to be a robust method for simulating FSI. Special attention was paid to regularization of the membrane force on the unstructured fluid grid in order to ensure proper constraint on the fluid. The fluid source term ($f$) was calculated from the solid force ($F_m$) with the use of a dynamically evolving regularization factor, the calculation of which is evaluated and discussed.

Discussion
The advantage of the DF method is the smooth, non-feedback induced, distribution of force performed on the Lagrangian solid markers. Valve leakage is inherent in classical IBM methods due to slight membrane permeability. While ITBM does compare well with experimental results [5], perceived numerical valve leakage and associated underestimation of the diastolic transvalvular pressure gradient prompted the need to test an alternative method. DF IBM offers another promising way of simulating artificial valves.

References

**Abstract**

**Introduction**

The surgical repair of interrupted aortic arch was used to restore its normal shape. Various graft materials in a classical half-moon shape were adopted for the surgery [1]. We proposed a crescent shape for better matching with the normal aortic arch shapes. The goal of this work is to characterize the influences of graft materials and shapes on the hemodynamics and wall mechanics of the restored aortic arch and thus the clinical outcomes.

**Methods**

A three-dimensional aortic arch model was reconstructed from the magnetic resonance images (MRI). Fluid-structure interaction (FSI) analysis were developed to systematically characterize the hemodynamics and solid wall mechanics for a variety of graft choices, including shape design and materials. Two graft shapes, i.e., the classical half-moon shape and a proposed crescent shape, were considered. The stiffness of three common used graft materials, i.e., polytetrafluoroethylene (PTFE) synthetic graft, cormatrix extracellular matrix (ECM), and pulmonary homograft, were 1000 MPa, 30 MPa, 3 MPa, and were studied by FSI analysis. The stiffness of native aortic tissue was also adopted as 0.84MPa as a control. Von-Mises stress distributions and pathological hemodynamic parameters, low time-averaged wall shear stress (TAWSS<0.2 Pa), high oscillatory shear index (OSI>0.2) and relative residence time (RRT>10 Pa/1) [2] on the aortic wall were quantified. The clinical implications were discussed.

**Results**

It was observed that the peak Von-Mises stress in crescent shape graft was 76% less than that in the half-moon shape graft (Fig 1a). low WSS, high OSI, and, RRT in crescent shape graft were significantly reduced by 87.2%, 81.4%, 89.2% respectively compared to the ones in half-moon shape pulmonary graft (Fig 1b). The peak Von-Mises stresses in PTFE, cormatrix, pulmonary, and native tissue are 1.80 MPa, 0.64 MPa, 0.20 MPa, and 0.09 MPa respectively (Fig 1c). It is clear that larger material mismatch led to higher stress concentration. However, the pathological hemodynamic parameters was not sensitive to graft materials as depicted in Fig 1d.
Fig. 1: Von-mises stress distribution and percentage area of low WSS, high OSI and RRT in half-moon and crescent shape (a, b) and in PTFE, Cormatrix, Pulmonary, Native tissue (c, d)

Discussion

The graft materials tend to induce stress concentrations on the wall, which could be aggregated by the large graft material mismatch. The blood flow dynamics were sensitive to the graft shape instead of materials. Pulmonary graft with the proposed crescent-shape might provide less alternations in the hemodynamics and wall mechanics of aortic arch. This work might shed lights on the optimum graft decisions towards better clinical outcomes.

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References

Effects of Atrial Fibrillation on Haemodynamics in the Left Atrium using 4D PC-MRI Data and Computational Fluid Dynamics

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Abstract

Atrial Fibrillation (AF) is a type of supraventricular arrhythmias and one of the most common types of cardiac arrhythmia. In 2014, it was estimated that in the UK alone around 1.4 million people suffer from AF [1], a trend which is increasing as a result of an aging population. Despite all developments in prognosis, diagnosis and treatment of AF, there are still many uncertainties regarding the underlying phenomena that cause AF and its long term effects on haemodynamics in the heart chambers. AF in Left Atrium (LA) lowers its functionality that leads to lack of perfusion and blood stasis especially in the Left Atrial Appendage (LAA), which might enhance the possibility of stroke because of thromboembolism. Since different flow characteristics attribute to various pathologies, recognition of haemodynamics of LA due to the AF would help clinicians adopt the most efficient therapeutic strategy. Over the last decade, 4D Phase Contrast Magnetic Resonance Imaging (PC-MRI) technique has continuously been improved and now can provide information about anatomy and flow characteristics in the cardiovascular system. Despite its capability, the PC-MRI data is usually accompanied with some artifacts and limited spatial and temporal resolutions. Simultaneous use of PC-MRI and Computational Fluid Dynamics (CFD) provides a robust framework to meticulously investigate the haemodynamics of heart chambers with high temporal and spatial resolutions.

In this study we numerically investigate the haemodynamics of LA for an AF patient-specific model obtained through reconstruction of the magnitude images of 4D PC-MRI data. The LA comprises four pulmonary veins as the inlets, mitral valve as the outlet and the LAA. Associated phase images are employed to extract the main flow features in LA and its boundaries in the pulmonary veins and mitral valve. The blood flow is considered incompressible, laminar and Newtonian, while a fiber-reinforced material model is adopted for the strain energy function of atrial wall. Also, the LA geometry in late diastole is considered as the initial state for the simulation. The current study invokes fully coupled fluid-structure interaction method. Therefore, to obtain the haemodynamics data in AF, Navier-Stokes equations in Arbitrary Eulerian Lagrangian (ALE) formulation were used along with elastodynamic equation to capture wall movements. This work represents the first study of its type to consider the compliance of LA in a patient suffering from AF. Furthermore, numerical results are validated against the 4D PC-MRI data by comparing different haemodynamic parameters and wall strain in the LA.

References

Development of a novel autologous atrioventricular valve "biovalve" (In vitro evaluation of hydrodynamic performance by using left heart simulator)

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Abstract

Introduction and Objective: We have been developing autologous heart valves so called biovalve fabricated by in body tissue architecture technology (IBTA) for heart valve replacement such as aortic valve and pulmonary valve [1]. IBTA is based on in vivo tissue engineering utilizing the foreign body reaction and it can be easily fabricated various complex shaped autologous tissue made of collagen. We have already developed some kinds of semilunar valves such as aortic and pulmonary valve and implanted into various animals. But atrioventricular valve such as mitral and tricuspid valve have not been developed because atrioventricular valves, which consist of two leaflets with chorda tendineae and the leaflet completely and swiftly close coupled with ventricle walls, is very complex. So this study developed a mitral biobalve and a left heart simulator, and investigated the influence of valve shape on valve performance.

Experimental setup and methods: Experiment was conducted by using newly developed left heart simulator. This simulator consisted of a left heart model with flexible ventricular suck and a human main circulation model based on Windkessel. Dimensions of mitral biobalves were as follows; length of posterior leaflet was 11-17mm (anterior leaflet was anatomically set 7mm longer than the posterior), length of chordae tendineae was 22-43mm, papillary muscle position from the valve root is 30-45mm respectively. Tested valves were installed into an orifice plate located between atrium model and ventricle model, of which hole shape was same design as Carpentier Edwards Physio-Ring. And chorda tendineae were connected to papillary muscles located on a ventricle suck. The simulator was operated under the following condition; heart rate was 60bpm, stroke volume was 85mL/cycle and systolic/diastolic blood pressure in ventricle and atrium was 80-120mmHg and 2-12mmHg respectively. The valve opening-closing behavior, flow rate waveform through mitral/aortic valve and blood pressure were measured by using a high speed camera, ultra-sonic flow probes and pressure transducers.
Results: Under all experimental conditions, retrograde flow was virtually occurred during valve closing i.e. end diastole and the regurgitant volume was increasing by leaflet becoming longer. When a non-dimensional valvular size parameter, which was defined as the ratio of the sum of the length of leaflet and the length of chorda tendinea to the papillary muscle position, was bigger, the regurgitant fraction was remarkably increased. And time for valve closing was longer with the non-dimensional valvular size parameter increasing caused by hydrodynamical drag on the leaflet. EOA was linearly decreasing with smaller valve leaflet because shape of mitral biovalve had tapered leaflet. These results indicate that our developing mitral biovalve is able to ensure the compatibility with valve hydrodynamical and regenerative performance.

Modeling the effect of aortic stent-graft properties on blood pressure: Comparison between modified windkessel and fully-coupled fluid-structure interface (FSI) models

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Abstract

Introduction. Despite being a major risk factor for cardiovascular diseases, hypertension is treated, with limited success, through medication and recommended lifestyle changes. In a healthy arterial system, the large, elastic aorta acts as a capacitor, expanding and contracting to facilitate forward blood flow throughout the cardiac cycle. As people age, the aortic wall undergoes changes that inhibit this function and contribute to development of hypertension. In addition to natural, age-related stiffening, when stent-grafts are placed in the aorta to treat conditions such as aneurysm or dissection, the stent-graft’s rigidity contributes to increased blood pressure in many patients. Our goal is to investigate whether these changes could be counteracted via placement of a capacitive endovascular device.

Methods. A modified Windkessel circuit model was developed to simulate the arterial system with a stent in the descending thoracic aorta. The baseline case consisted of a typical blood flowrate and parameters reflecting average aorta properties for a patient over age seventy. Resistance and capacitance of the stent were varied and resulting pressure was recorded.

The circuit model was expanded into a FSI model, where the aorta was a neo-Hookean material and the stent linear elastic. Impedances from the heart outlet and peripheral vessels were modeled as resistances at the inlet and outlet of the aorta, respectively. Flowrate was imposed at the inlet, and outlet pressure was zero. Capacitance of the stent was varied by changing the elastic modulus of the material. Blood pressure for each capacitance was extracted from time-dependent studies.

Results. The circuit model predicted a 20 mmHg rise in blood pressure for a stent that provides no capacitance. A maximum benefit of 22 mmHg pressure decrease from baseline was observed with a capacitive stent. Using the latest guidelines for hypertension diagnosis, the baseline system exhibited Stage 2 hypertension, but a certain range of stent properties lowered the pressure to Stage 1. Baseline parameters in the FSI model yielded pressure comparable to the circuit model. The peak blood pressure decreased with increased stent capacitance in both models. FSI showed less overall change in pressure over the capacitance range tested.

This work shows the potential for a capacitive stent-graft to therapeutically reduce hypertension. Though both models produce similar trends, FSI better incorporates inertia of the blood and provides information about stress in the aortic wall.

Figure 1: (a) Results from varying stent properties in lumped parameter model, categorized by level of hypertension; (b) Comparison of baseline pressure in lumped parameter model and FSI model; (c) Effect of stent capacitance on systolic blood pressure in both models.
Abstract

The cardiovascular system is a complex interplay of a number of different phenomena that occur across a range of temporal and spatial scales. The interaction between blood and vascular structures is of particular interest given its known role in the pathogenesis of many cardiovascular diseases. A significant number of FSI methods have been developed at larger scales incorporating continuum methods. These methods have grown increasingly sophisticated, incorporating more complex material models and high resolution patient specific geometries via various medical imaging modalities. However, due to the nature of the methods, including effects such as local defects in the structure is difficult.

Discrete methods are more readily able to include these effects but have traditionally been restricted to small spatial scales due to high computational expense. However, coarse graining methods and the ever increasing accessibility of high performance computing (HPC) has loosened these restrictions to some extent.

The Vector-based discrete element method, also known as the V-model, is a recently proposed variant of the discrete element method whereby a given solid is discretised into a number of particles connected via bonds [1]. In previous studies it has been coupled to a lattice Boltzmann fluid solver via immersed boundary on CPU architecture and used to investigate cardiovascular applications [2]. However, its local and explicit nature, along with linear governing equations, means it is well suited to implementation via GPU allowing numerical analysis to take place on bedside rather than at a dedicated HPC facility, reducing financial cost and improving accessibility of the numerical analysis to clinicians.

Here an FSI method, coupling a lattice Boltzmann fluid solver and V-model structural solver via immersed boundary and implemented on GPU architecture, is validated and compared against its previous implementation on CPU. Further, complex structural phenomena such as aggregation and remodelling are presented using the V-model. Finally, the FSI method is applied to flow through an idealised, two dimensional deep leg venous valve to demonstrate the capabilities of the method in cardiovascular applications. Results indicate the potential for this solver to be integrated within clinical diagnosis tool chains, along with patient-specific imaging methods, to allow clinicians to make more informed decisions regarding patient treatment.

References
