Computers in Biology and Medicine 49 (2014) 83–94

Contents lists available at ScienceDirect

Computers in Biology and Medicine

journal homepage: www.elsevier.com/locate/cbm

Computational analysis of the importance of flow synchrony for cardiac ventricular assist devices

Matthew McCormick [a], David Nordsletten [b], Pablo Lamata [b], Nicolas P. Smith [a,b,c,*]

[a] Department of Computer Science, University of Oxford, Wolfson Building, Parks Road, OX1 3QD, UK
[b] Department of Biomedical Engineering, King's College London, The Rayne Institute, 4th Floor Lambeth Wing, St Thomas' Hospital, SE1 7EH, UK
[c] Faculty of Engineering, University of Auckland, 20 Symonds St, Auckland, New Zealand

ARTICLE INFO

Article history:
Received 18 November 2013
Accepted 28 March 2014

Keywords:
Fluid–structure
Computational model
Cardiovascular
Tissue mechanics
Computer model

ABSTRACT

This paper presents a patient customised fluid–solid mechanics model of the left ventricle (LV) supported by a left ventricular assist device (LVAD). Six simulations were conducted across a range of LVAD flow protocols (constant flow, sinusoidal in-sync and sinusoidal counter-sync with respect to the cardiac cycle) at two different LVAD flow rates selected so that the aortic valve would either open (60 mL s^{-1}) or remain shut (80 mL s^{-1}). The simulation results indicate that varying LVAD flow in-sync with the cardiac cycle improves both myocardial unloading and the residence times of blood in the left ventricle. In the simulations, increasing LVAD flow during myocardial contraction and decreasing it during diastole improved the mixing of blood in the LV cavity. Additionally, this flow protocol had the effect of partly homogenising work across the myocardium when the aortic valve did not open, reducing myocardial stress and thereby improving unloading.

1. Introduction

Heart failure is the leading cause of hospitalisation among older adults in Western society with a lifetime risk of 20% at age 40. Despite improved medical and surgical techniques, mortality after the onset of heart failure remains high, ranging from 20 to 50% [6]. Orthotropic heart transplantation is recognised as the best therapy for end-stage heart failure [26]. However, approximately 20 to 30% of potential recipients die while waiting for a donor heart [29]. Due to this shortage, left ventricular assist devices (LVADs) are often used as a bridge to transplant [1].

The role of these LVAD pumps is to reduce the mechanical load on the heart by pumping blood from the left ventricular (LV) apex directly to the aorta, with the implantation of these devices significantly reducing both LV pressure and volume [9]. Post implantation, it is standard practice for clinicians to tune LVAD flow so that the aortic valve opens occasionally to prevent it fusing shut [27]. However, the impact of valve opening on myocardial unloading and the residence times of blood within the ventricle remains unknown. Both these factors are of critical importance with respect to improving treatment outcome for patients – too much unloading can lead to myocardial atrophy, while too little results in the myocardium remaining over-stressed [14]. A further consideration is the impact of LVAD flow on blood residence times, where inadequate recirculation has the potential to increase the risk of thrombosis formation [2]. Tuning the device to optimise for these factors involves varying both LVAD flow rate and LVAD flow synchrony – i.e. whether the LVAD cannula outflow is constant or varies through the cardiac cycle. However, these parameters result in substantial variation in cardiac behaviour, ranging from determining whether the aortic valve opens at all, through to the extent to which LV volume changes through the cardiac cycle.

A central difficulty for this type of optimisation is the challenge of observing cardiac function and cardiovascular flows under LVAD support using standard medical image modalities, such as MRI and echocardiography, due to the positioning of the pump, along with its metallic components. This context motivates the application of mathematical modelling techniques as an investigative tool for studying the behaviour of the ventricle under LVAD support and analysing its efficacy as a pump. For such analyses to facilitate the optimisation of LVAD support, the interaction at the core of ventricular function needs to be addressed – i.e. the coupling between blood flow in the ventricular chamber and the myocardium. As a result, coupled fluid–solid mechanical models are required with the ability to support investigations into the impact of LVAD support on ventricular hemodynamics and myocardial mechanics.

* Corresponding author at: Faculty of Engineering, University of Auckland, 20 Symonds St, Auckland, New Zealand.
E-mail address: np.smith@auckland.ac.nz (N.P. Smith).

http://dx.doi.org/10.1016/j.compbiomed.2014.03.013
0010-4825/

Several coupled fluid–solid mechanical LV models currently exist in the literature, ranging from the pioneering work of McQueen and Peskin [17,25] through to recent models incorporating greater degrees of physical realism, in particular, in the description of myocardial behaviour [22]. These models have been used to investigate blood flow within the ventricular cavities and the efficiency of the heart as a pump from diastole [3] through to systole [11,22,32,33]. Recently, we [15,16] have extended a non-conforming finite element fluid–solid mechanics scheme [21] to facilitate the simulation of LVAD supported LVs through the full cardiac cycle. Using a fictitious domain (FD) [31] method to prescribe the LVAD cannula, the application of this approach enables the interaction between the cannula and the myocardial wall to be captured, facilitating the simulation of the full range of cardiac behaviour.

In this study we apply this framework for the first time to a patient customised geometry to present the first (to our knowledge) numerical investigation into the impact of aortic valve opening and LVAD flow synchrony on ventricular hemodynamics and myocardial mechanics. Specifically, the developed model is applied to investigate the mixing of blood within the LV chamber, as well as the efficiency of myocardial work transduction under different LVAD flow protocols.

2. Materials and methods

2.1. Model framework

Derived from the principles of conservation of mass and momentum, and as outlined in detail in our previous publications [16,22], we have developed a model that provides a physiological description of the myocardium and ventricular blood flow. In brief, the model was solved using a non-conforming Galerkin finite element scheme to enable varying degrees of refinement to adequately resolve the blood and myocardial spatial domains. This scheme enables high levels of physiological detail (including the complex fibre architecture [13] and biophysically based constitutive laws) to be incorporated.

To resolve the physical system, ventricular blood flow and myocardial mechanics were modelled using the arbitrary Lagrange–Eulerian form of the Navier–Stokes equations [20] and the quasi-static finite elasticity equations [23], respectively. To enforce continuity between the solid myocardial wall and the fluid ventricular chamber, velocities were equated over their common interface [21]. This constraint was applied by introducing a Lagrange multiplier to enforce equal, but opposite, tractions across the endocardial boundary. To incorporate the LVAD cannula into the model, a zero velocity boundary condition was implemented on the cannula wall using the fictitious domain method whereby a second Lagrange multiplier was applied to the FEM weakform. This method enables the cannula boundary to move through the fluid domain, resolving the numerical issues resulting from the deformation of the fluid mesh [15]. Additionally, it has been demonstrated that application of the fictitious domain terms yields adherence to the velocity constraint weakly [30,31], and the method is applied to many cardiovascular applications. Furthermore, the combination of the two Lagrange multipliers implicitly resolves the contact problem of an immersed rigid body in a deformable chamber. As a result the model system is capable of resolving the complete range of cardiac motion – including contact between the myocardium and the LVAD cannula [15].

Solving the fluid–solid mechanical model through a whole cycle requires the addition of accurate systemic constraints on the flow model. This was achieved by integrating the 3D FSI model with a 0D Windkessel representation of systemic circulation. In this work, we coupled the Shi and Korakianitis 0D Windkessel model [28] using a fixed point prescribed flow rate technique [16]. Using this technique, flow was prescribed according to the pressure gradient across the valve using Bernoulli's equation for the conservation of energy along the same streamline. Valve opening was prescribed to occur when LV lumen pressure exceeds aortic sinus pressure. To approximate opening and closing in the 3D model, the valves were defined as functions on the mitral and aortic boundaries, with the radius of the open valve assumed to be proportional to flow rate. The proportionality constant was fitted to match observed human data, $\approx$ 46 ms and $\approx$ 24 ms for the mitral [34] and aortic [24] valves respectively, see Appendix A for details.

To capture the mechanical properties of the myocardium, the finite elasticity stress tensor was defined as a combination of passive and active components. The stress further incorporated information about myocardial structure, by the introduction of an orthonormal fiber tensor, to denote the fibre, sheet and normal directions of the tissue [7,19]. In this paper, the passive constitutive law was defined using a modified form of the Costa constitutive law [4] based on the strain energy functions W and W_{iso}, where W represents the Costa constitutive contribution and an isotropic stiffness component (see [22] for details of the incorporation of this component). Additionally, to approximate the interaction between the cannula base and the myocardium, the myocardial wall was assumed to be stiffer at the junction between the LVAD cannula base and the myocardial wall. Active contraction in the tissue was generated using the Niederer contraction model [18] chosen due to the limited number of parameters enabling a more unique fit to patient data [18]. This 6 parameter model captures the length dependent rates of tension development, along with peak tension.

2.2. Patient model

This framework was applied to a patient specific LV geometry which was constructed based on 422 short axis CT image slices taken at end diastole from a 53 year old heart failure patient with an implanted LVAD, all data was acquired as part of a local ethics committee at the German Heart Centre approved protocol consistent with the principles expressed in the Declaration of Helsinki and informed consent was obtained from the patient. The spatial resolution of the image stack was 0.4 mm $\times$ 0.4 mm, in the CT image plane, and 0.6 mm in the through plane direction. Digitisation of the image data was performed by Phillips Research and the resulting binary segmentation was used to construct the geometric myocardial mesh. Fig. 1 highlights each stage of the mesh generation procedure. A cubic Lagrange myocardial mesh was constructed from an ellipsoidal template using an automated meshing tool that implements the procedure previously outlined [12]. Mean error from the fitting procedure (with respect to the normal distance between binary data and the fitted mesh) was 0.72 ± 1.05 mm. The final fitted cubic Lagrange mesh was interpolated from the warped cubic Hermite geometry. The resulting cubic hexahedral mesh consisted of 324 elements, with a through wall thickness of 3 elements. An idealised fibre geometry, $\pm 60^{\circ}$ with respect to the endo/epicardial surfaces, was defined within the myocardial geometry.

Within the ventricular cavity a linear tetrahedral fluid mesh, consisting of $\approx 3.2 \times 10^4$ elements, was constructed using the software package CUBIT,[1] with a characteristic mesh length of 3.2 mm. The linear mesh was modified to provide a curvilinear description (quadratic Crouzeix–Raviart [5] elements) of the cavity by projecting surface nodes onto the endocardial surface. Internal nodes were unchanged maintaining the linear spatial description

[1] http://cubit.sandia.gov.