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Simulating left ventricular fluid–solid mechanics through the cardiac cycle under LVAD support

M. McCormick [a,1], D.A. Nordsletten [b,1], D. Kay [a], N.P. Smith [a,b,*]

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

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ABSTRACT

In this study we have integrated novel modifications of the standard Newton–Raphson/line search algorithm and optimisation of the interpolation scheme at the fluid–solid boundary to enable the simulation of fluid–solid interaction within the cardiac left ventricle under the support of a left ventricular assist device (LVAD). The line search modification combined with Jacobian reuse produced close to an order of magnitude improvement in computational time across both test and whole heart simulations. Optimisation of element interpolation schemes on the fluid–solid boundary highlights the impact this choice can have on problem stability and demonstrates that, in contrast to linear fluid elements, higher order interpolation produces improved error reduction per degree of freedom. Incorporating these modifications enabled a full heart cycle under LVAD support to be modelled. Results from these simulations show that there is slower clearance of blood entering the chamber during early compared to late diastole under conditions of constant LVAD flow.

1. Introduction

The application of multi-physics computational techniques to understand the integrated function of physiological systems is now rapidly emerging as a significant research focus for both the computational and life science communities. Within this field, the heart is one of the most advanced examples – with anatomically accurate models currently able to successfully quantify detailed physiology across a range of contexts. Central to these advances have been the development and application of ventricular models seeking to capture the interaction at the core of cardiac behaviour – i.e. the interaction between blood flow within the heart chambers and the passive and active mechanics of the myocardium. Driven by the needs arising within the health care communities, cardiac models have seen continued development and progression in both the models themselves as well as the numerical methods which drive them [1]. One such example, and the focus of this study, is the simulation of the interaction between fluid flow, myocardial mechanics and left ventricular assist devices (LVADs) in the cardiac left ventricle (LV).

LVADs are pumps that reduce the mechanical load on the heart by pumping blood from the left ventricular (LV) apex directly to the aorta. The implantation of these devices significantly reduces both LV pressure and volume [2]. At present patient selection for LVAD implantation is based on relatively simple, qualitatively based, clinical metrics combined with some

* Corresponding author at: Department of Biomedical Engineering, King's College London, The Rayne Institute, 4th Floor Lambeth Wing, St. Thomas' Hospital, London SE1 7EH, UK.
E-mail address: nicolas.smith@kcl.ac.uk (N.P. Smith).
[1] These authors wish to be identified as first authors on this work.

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standard hemodynamic variables [3]. Additionally, once implanted there is little customisation to the patient or tuning to the cardiac cycle. It is in this context that the ability to optimise LVAD use, via patient specific customisation, to benefit cardiac function has the potential to provide substantial clinical gains [4]. However, implanted LVADs are challenging to observe using standard medical image modalities such as MRI and echocardiography, due to positioning as well as the metallic components of the pump. This provides a niche for fluid–solid structure modelling techniques to be used as an investigative tool for studying the behaviour of the ventricle under LVAD support and analysing its efficacy as a pump.

Historically the fluid and solid mechanical models which underpin this modelling approach in the heart have been pursued independently. Advanced solid mechanical models have been developed which incorporate complex anatomical descriptions of both ventricles [5–9], and capture the nonlinear tissue behaviour in normal and diseased hearts [10,11]. Concurrently, fluid mechanical models were developed to characterise the complex three-dimensional flows through the heart chambers [12–15] using prescribed wall motion based on medical images. The need for integration of these phenomena to understand their coupled function [16,17] has placed a renewed focus on coupling fluid–solid mechanics in the ventricular chambers which, in turn, has enabled novel investigations of efficiency of the heart as a pump from diastole [18] through to systole [19–23].

A variety of computational techniques underpin these frameworks, including immersed boundary [16], finite volumes [18,21] and finite elements [23]. Recently Nordsletten et al. [22], developed a Lagrange multiplier coupling approach which allowed non-conformity within a finite element scheme, enabling areas of high spatial gradients, such as ventricular blood flow, to be modelled with a higher degree of refinement. Extending this work for studying LVAD devices, McCormick et al. [24] applied fictitious domain methods [25–28] to capture the interactions between the LVAD cannula and the ventricular myocardium. However, the introduction of this new modelling element significantly increased computation times, required careful customisation of boundary conditions and introduced stability issues in specific regions of interface between the fluid and solid meshes. These computational issues limited the ability to effectively simulate a full cardiac cycle involving active contraction. Further, the lack of coupling of the LV model with the systemic circulatory network severely limited the models ability to provide useful mechanical results.

To address these challenges, this paper presents a fluid–solid left ventricular model coupled to a 0D Windkessel model [29] of the circulatory network under LVAD support. Novel numerical contributions are outlined that facilitate reductions in simulation time by modifying the standard Newton–Raphson/line search algorithm [30]. Additionally, to reduce error in the non-conforming fluid–solid coupled finite element system, an investigation into the choice of fluid element type is performed using a number of test problems. These results detail the numerical requirements for extending the existing work to enabling simulations of the full cardiac cycle. The developed LV model is then used to simulate ventricular function through the cardiac cycle, converging on a repeating pressure volume loop. Finally, and more generally, through the integration of these new modelling elements we demonstrate the potential of computational approaches for both capturing and optimising device interactions with the cardiovascular system.

2. Methods

This section outlines the numerical scheme developed for simulating left ventricular function under LVAD support. Section 2.1 describes the model problem and governing equations with reference to appropriate weak forms. Details of the numerical implementation for simulating the supported heart through the cardiac cycle are subsequently provided in Section 2.2, including a discussion of the adapted global Newton scheme with matrix re-use in Section 2.3.

2.1. The LVAD model strong form

Simulating the mechanical effects experienced in the supported LV over the cardiac cycle (i.e. for a cycle $I = [0, T]$) requires the simulation of three primary components: (1) ventricular blood flow, (2) myocardial tissue mechanics and (3) systemic circulation under VAD support (see Fig. 1). These elements were modelled by combining an anatomically-based three-dimensional left ventricular model with a 0D Windkessel representation of the systemic circulation (and VAD device).

To model the blood flow within the ventricular chamber the arbitrary Lagrangian–Eulerian (ALE) Navier–Stokes equations [31] (with density and viscosity parameters μ and ρ) were used. Fluid velocity, $\mathbf{v}$, and fluid pressure, p_f, are both sought on a reference domain $\Lambda_f \subset \mathbb{R}^d$ (with boundary Υ_f) as schematically shown in Fig. 1. In the ALE method, this reference domain is related to a physical domain, $\Omega_f \subset \mathbb{R}^d \times I$ (with boundary Γ), through a bijective mapping (cf. [32,33]) which is used to map all state variables onto Ω_f. This mapping is approximated using a Laplace problem (cf. [34]). The boundary, $\Upsilon = \Upsilon_c \cup \Upsilon_f^D$ is partitioned into components linked to the solid mechanical model, Υ_c, and those coupled to the systemic model via fluid Dirichlet conditions, $\Upsilon_f^D = \Upsilon_{mv} \cup \Upsilon_{av} \cup \Upsilon_{vad}$ where mv, av, and vad denote the mitral/aortic valve boundaries and the VAD device outflow (again see Fig. 1). Due to the close proximity of the LVAD cannula and the tissue wall, the device within the three-dimensional fluid domain was modelled using the fictitious domain (FD) method [26–28]. Here an artificial boundary, Γ_{Fd}, was inserted and the flow enforced to be zero (note, this is posed with respect to the physical domain, as motion of the LVAD device is not prescribed by the ALE mapping).

The solid mechanical model, denotes the quasi-static finite-elasticity formulation [7,35] in the Lagrangian frame. Solid displacement, $\mathbf{u}$, and solid pressure, p_s, are both set on the reference domain $\Lambda_s \subset \mathbb{R}^d$ (with boundary Υ_s) which is also linked