Journal of Biomechanics 48 (2015) 604–612

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Computational modelling of left-ventricular diastolic mechanics: Effect of fibre orientation and right-ventricle topology

Arnab Palit[a,*], Sunil K. Bhudia[b], Theodoros N. Arvanitis[c], Glen A. Turley[a], Mark A. Williams[a]

[a] WMG, The University of Warwick, Coventry, UK
[b] University Hospitals Coventry and Warwickshire, Coventry, UK
[c] Institute of Digital Healthcare, WMG, The University of Warwick, Coventry, UK

ARTICLE INFO

Article history:
Accepted 31 December 2014

Keywords:
Fibre structure
Finite element methods
Ventricular diastolic mechanics
Bi-ventricular geometry

ABSTRACT

Majority of heart failure patients who suffer from diastolic dysfunction retain normal systolic pump action. The dysfunction remodels the myocardial fibre structure of left-ventricle (LV), changing its regular diastolic behaviour. Existing LV diastolic models ignored the effects of right-ventricular (RV) deformation, resulting in inaccurate strain analysis of LV wall during diastole. This paper, for the first time, proposes a numerical approach to investigate the effect of fibre-angle distribution and RV deformation on LV diastolic mechanics. A finite element modelling of LV passive inflation was carried out, using structure-based orthotropic constitutive law. Rule-based fibre architecture was assigned on a bi-ventricular (BV) geometry constructed from non-invasive imaging of human heart. The effect of RV deformation on LV diastolic mechanics was investigated by comparing the results predicted by BV and single LV model constructed from the same image data. Results indicated an important influence of RV deformation which led to additional LV passive inflation and increase of average fibre and sheet stress–strain in LV wall during diastole. Sensitivity of LV passive mechanics to the changes in the fibre distribution was also examined. The study revealed that LV diastolic volume increased when fibres were aligned more towards LV longitudinal axis. Changes in fibre angle distribution significantly altered fibre stress–strain distribution of LV wall. The simulation results strongly suggest that patient-specific fibre structure and RV deformation play very important roles in LV diastolic mechanics and should be accounted for in computational modelling for improved understanding of the LV mechanics under normal and pathological conditions.

1. Introduction

Cardiovascular disease (CVD) is a major health problem worldwide and can lead to heart failure (HF) (National Health Service, 2010; American Heart Association, 2014). Epidemiological studies, by Paulus et al. (2007) and Wang and Nagueh (2009), stated that at least 50% of the HF patients have left-ventricular diastolic dysfunction with normal systolic pump function. Despite the wide variety of surgical and pharmacological treatments developed over the past, diastolic dysfunction is still a common and less well-understood medical condition. Finite element (FE) modelling, in combination with new cardiac imaging modalities and advanced simulation tools, can be used to analyse the diastolic mechanics of the left ventricle (LV). Such simulation models can provide a greater insight of the physiology of such patients and thereby predict their response to medical and surgical interventions. However, FE modelling solely on passive inflation mechanics of LV is very limited (Table 1).

The myocardium is the functional tissue of the ventricle. Recent histological studies with a higher level of magnification introduced the concept of fibre-sheet (laminar) architecture of the myocardial wall (LeGrice et al., 1995; Hunter et al., 1998). Histological studies confirmed that the myocardial fibre angle varies from $+50^\circ$ to $+70^\circ$ in the sub-epicardial to almost 0° in the mid-wall, from -50° to -70° at sub-endocardial with respect to the local circumferential direction of the LV (Holzapfel and Ogden, 2009). Even though the fibre-sheet orientation is histologically similar for human hearts, differences exist in fibre-angle between individual subjects (Buckberg et al., 2008). In addition, the fibre-sheet archtecture may also alter in diseased hearts, such as in myocardial infarction due to tissue remodelling (Buckberg et al., 2008). Therefore, sensitivity of LV diastolic mechanics to the details of fibre structure is an important issue and has not been carried out for bi-ventricular (BV) model (Table 1).

As shown in Table 1, most of the passive simulations carried out in earlier studies were based on either animal heart or idealised geometry and mostly used 'Fung-type' transversely isotropic

* Corresponding author. Tel.: +44 24 7615 1798; fax: +44 24 7652 4307.
E-mail address: arnab.palit@warwick.ac.uk (A. Palit).

http://dx.doi.org/10.1016/j.jbiomech.2014.12.054
0021-9290/

Table 1
Previous work on passive inflation of LV with key attributes considered in the simulation.

	Single LV	Bi-ventricle	Effect of RV deformation	
			Stress–strain	EDPVR
Animal heart/Idealised geometry	Horowitz et al. (1986) Humphrey and Yin (1989) Guccione et al. (1995) Costa et al. (1996) Vetter and McCulloch (2000) Usyk et al. (2000)	Stevens et al. (2003)		
Transversely isotropic	Horowitz et al. (1986) Humphrey and Yin (1989) Guccione et al. (1995) Costa et al. (1996) Vetter and McCulloch (2000) Wang et al. (2009)			
Orthotropic	Wang et al. (2013) Usyk et al. (2000)	Stevens et al. (2003)		
Human heart	Wang et al. (2009) Wang et al. (2013)	Objectives and research of the paper		
Effect of fibre-sheet orientation on Stress–strain	Wang et al. (2013)			
Effect of fibre-sheet orientation on EDPVR				

constitutive law (Costa et al., 1996; Guccione et al., 1995; Vetter and McCulloch, 2000). In contrast, a simple shear test of pig's myocardium (Dokos et al., 2002) clearly exhibited its orthotropic nature. Diastolic modelling were carried out using 'Fung-type' law (Usyk et al., 2000) and 'pole-zero' law (Stevens et al., 2003) to incorporate material orthotropy. The material parameters in these orthotropic models were merely used as weighting factors, rather than any physical significance (Göktepe et al., 2011), and some of these parameters were highly correlated (Wang et al., 2013). Recently, Holzapfel and Ogden (2009) developed a constitutive law that considered the locally orthotropic tissue architecture and the parameters of this model were closely related to the characteristic microstructure of myocardium. With the advancement in imaging modalities over the years, patient-specific human heart geometry, created from MRI images, was used for the simulation (Wang et al., 2009, 2013). However, bi-ventricular modelling of human heart for passive inflation mechanics is very limited due to complexity of the ventricular geometry (Table 1). In addition, RV deformation seems to have a significant effect on EDPVR (End Diastolic Pressure Volume Relation) and fibre stress–strain distribution of LV wall (Sun et al., 2009). The determination of this effect could lead to more accurate stress–strain prediction in passive inflation, as well as in systole.

In our research, passive inflation of LV during diastole was carried out using 'Holzapfel–Ogden' material law with the bi-ventricular geometry developed from cardiac MRI of human heart. Following were the strategies to carry out the research objectives (Table 1). First, the passive inflation modelling using bi-ventricular geometry was carried out and validated with experimental results. Consequently, the effect of changes in underlying fibre-angle distribution on the bi-ventricular model prediction (i.e. EDPVRs and fibre stress–strain distribution) was investigated. Next, the same simulation was conducted with single LV model and compared the results with bi-ventricular model prediction to identify the effect of RV deformation on the diastolic mechanics of LV.

2. Material and methods

2.1. Ventricular mesh geometry

ECG gated, breathe hold, and steady state free precession (SSFP) cine cardiac magnetic resonance imaging (CMRI) were used to capture the images of a normal human heart in a dedicated MRI suite in 'University Hospitals Coventry and Warwickshire Trust' (UHCW), Coventry, UK. BSREC ethics approval (REGO-2012-032) was obtained to carry out the research on anonymised human data. Stack of short-axis images at early diastole were segmented in Mimics 16.0 (Materialise, Belgium) using semi-automatic procedure to construct initial ventricular geometry. The correct position of basal-atrium plane was identified using long-axis images of early diastole. The initial ventricular geometry was then cut by the basal-atrium plane to remove the extra volume generated from short-axis images. Finally the ventricular mesh geometry with linear tetrahedral elements was constructed in 3-matic 8.0 (Materialise, Belgium). A brief overview of the mesh geometry reconstruction was described in Palit et al. (2014) and is shown in Fig. 1.

2.2. Rule-based fibre orientation generation

Laplace-Dirichlet-Region growing-FEM (LDRF) based algorithm was used to assign myocardial fibre-sheet architecture on ventricular mesh geometry. The algorithm was developed based on the method proposed by Wong and Kuhl (2012), with the amendment in identifying the different surface domains automatically. Fig. 2a illustrates the three orthonormal axes related to the characteristic microstructure of fibre-sheet orientation($\mathbf{f}$, $\mathbf{s}$, $\mathbf{n}$), the helix angle (α) and the sheet angle (β). Three inputs were required for LDRF: (1) ventricular mesh geometry, (2) Point cloud of the plane parallel to the short-axis image plane just below the basal-atrium intersection, and (3) fibre angle definition from the histological data. Three functions were then used to generate fibre map on the mesh geometry. First function automatically separated the BV geometry into 4 surface domains: LV endocardium, RV free wall endocardium, RV septum and epicardium. In the second function, local coordinate ($\mathbf{e_n}$,$\mathbf{e_c}$, $\mathbf{e_z}$) was defined on the nodes of those surfaces and subsequently calculate $\mathbf{f}$ and $\mathbf{s}$. Third function carried out the Laplace interpolation with Dirichlet boundary to assign $\mathbf{f}$, $\mathbf{s}$ and $\mathbf{n}$ for all the nodes and subsequently for each element in the mesh geometry (Fig. 2b and c). Details of the LDRF implementation were discussed in Wong and Kuhl (2012) and Palit et al. (2014).

2.3. Structure-based constitutive law for passive myocardium

The strain energy function developed by Holzapfel and Ogden (2009) was extended by Göktepe et al. (2011) using decoupled volumetric–isochoric formulation of finite elasticity. The deformation gradient($\mathbf{F}$)is multiplicatively decomposed into a volumetric part $\mathbf{F_{vol}}$ and an isochoric part $\overline{\mathbf{F}}$ as,

$$\mathbf{F} = \overline{\mathbf{F}}\,\mathbf{F_{vol}} \quad \textit{with} \quad \mathbf{F_{vol}} = J^{1/3}\mathbf{I} \quad \textit{and} \quad \overline{\mathbf{F}} = J^{-1/3}\,\mathbf{F} \tag{1}$$

such that $J = \det(\mathbf{F_{vol}})$ and $\det(\overline{\mathbf{F}}) = 1$. The right Cauchy-Green tensor is defined as$\mathbf{C} = J^{2/3}\,\overline{\mathbf{C}}$, where $\overline{\mathbf{C}} = \overline{\mathbf{F}}^{\mathbf{T}}\,\overline{\mathbf{F}}$denoted the modified tensor quantities. The myocardium tissue is an orthotropic material with fibre, sheet and sheet-normal directions denoted by $\mathbf{f_0}$, $\mathbf{s_0}$, $\mathbf{n_0}$ respectively in the Lagrangian framework. The strain energy function Ψ per unit reference volume is additively decomposed into volumetric $\Psi_{vol}(J)$ and isochoric part $\overline{\Psi}$ parts,

$$\Psi = \Psi_{vol}(J) + \overline{\Psi}(\bar{I}_1, \bar{I}_{4f}, \bar{I}_{4s}, \bar{I}_{8fs}) \tag{2}$$

where $\Psi_{vol}(J)$and $\overline{\Psi}$ are given function of J and the isochoric invariants $\bar{I}_1$, $\bar{I}_{4f}$, $\bar{I}_{4s}$, $\bar{I}_{8fs}$ respectively where,

$$\bar{I}_1 = Tr(\overline{\mathbf{C}}), \quad \bar{I}_{4f} = \mathbf{f_0} \cdot (\overline{\mathbf{C}}\,\mathbf{f_0}), \quad \bar{I}_{4s} = \mathbf{s_0} \cdot (\overline{\mathbf{C}}\,\mathbf{s_0}), \quad \bar{I}_{8fs} = \mathbf{f_0} \cdot (\overline{\mathbf{C}}\,\mathbf{s_0}) \tag{3}$$