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# Modeling the Effects of Myocardial Fiber Architecture and Material Properties on the Left Ventricle Mechanics during Rapid Filling Phase

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**Abstract:** The objective of this research is to study the effects of myofiber volume fraction and fiber orientation on the deformation mechanics of left ventricular (LV) during diastole. The human LV was simulated by three-dimensional finite element (3D-FE) model. The LV geometrical model was represented as a thick-walled ellipsoid truncated at two thirds of major axis. 3D Fiber network, with parallel myofiber bundles to reproduce the globally anisotropic behavior of cardiac tissue, were embedded within the LV model. The LV wall thickness in the reference unstressed state (at hypothetical zero pressure applied inside the LV internal cavity) is divided into n different concentric thin layers (n = 3, 5, 7 and 9 layers). The presence of blood (incompressible fluid) inside the LV cavity was also simulated. Four scheme models of fiber angle definition (one symmetric, two asymmetric and one complex pattern) were used throughout this study. Complex pattern of fiber-structures are derived from diffusion tensor magnetic resonance imaging (DT-MRI) data. Simulation results have shown that the proposed FE model was able to reproduce experimental ventricular volume during the rapid filling phase and that the complex fiber orientations are in good agreement with the measurements.

**Keywords:** Finite element analysis, myofiber architecture, myocardial material properties, left ventricle, rapid filling phase

## 1 Introduction

Despite the wide variety of research in medicine and bioengineering treatment strategies developed over the last half century, heart disease remains the primary threat to human life. The clinical studies have been shown that at least 50% of heart failure patients have diastolic dysfunction.

Modelling the diastolic mechanics of the human myocardium particularly the left ventricle (LV), which is the main pumping chamber, play an important role in a better understanding the performance of the heart in health and disease states [1].

The human LV is a fibrous structure that composed of helical networks of contractile muscle fibers bundles (myofibers) held together by a mesh of collagen fibers, which oriented at different angles throughout the LV wall in form of sheets that are separated by a complex structure of cleavage surfaces [2,3,4]. Fiber architecture is a key feature of the myocardium, where the fiber orientation plays a significant role in both systolic deformation and early diastolic function [5].

The myofibers could be fully described spatially by two inclination projection angles; the helix angle ($\beta$) and the transverse angle ($\eta$). Nowadays, with advances in imaging technologies, the orientation of the fiber angles can be measured via diffusion tensor magnetic resonance imaging (DT-MRI). Previous studies have shown that the helix angle ($\beta$) varies continuously from approximately +60° at the endocardium to -60° at the epicardium whereas the transverse angle ($\eta$) varies continuously from approximately +15° at the base to -15° at the apex [6].

To the author's knowledge, no finite element (FE) models have been specifically designed, to date, to study the effect of the myocardial fiber volume fraction on the deformation of the human LV during the diastole phase, where the fiber architecture plays a significant role in early diastolic function. In this study, we investigated the

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effect of changes in the distribution of fiber architecture on the behavior of the human LV during the rapid filling phase using three dimensional (3D) FE model with detailed fiber orientation data obtained in *vivo* which provides a reliable description of both muscle fiber orientation and material characteristics. The resulting model is used to simulate the mechanics of the LV during diastole phase. The results were compared with published experimental measurements of the human LV during the rapid filling phase [7].

## 2 Model description

### 2.1 Geometry and FE model

3D-FE models that simulates the human LV were designed using the ANSYS® finite element software. The geometry of the existing LV model was simplified by an ellipsoid truncated at two-thirds of the major axis, (see Figure. 1a). Figure.1b shows the initial shape of a typical FE mesh used for the present computations, while Figure. 1c shows the end-diastolic deformed shape of the FE mesh. Figure.1d shows the section view in the FE mesh in order to clarify the shape of hydrostatic fluid elements used for modeling the LV internal cavity.

Since the heart is enclosed in the ribcage and surrounded by the lungs and the diaphragm, the expansion of ventricle during diastole is influenced by these surrounding organs and tissues. This effect has been modeled here by elastic foundation 3D structural surface effect element with a stiffness $K_f$ = 0.02 kPa, to approximately mimic the boundary conditions imposed by the surrounding tissue on the external surface of the FE models [8,9].

3D Fiber network, with parallel muscle-fibers (myofibers) bundles to reproduce the globally anisotropic behavior of cardiac tissue, were embedded within continuum 3D solid element. To study the effect of total myofiber volume fraction ($\phi_{tot}$) on the LV mechanics during diastole, FE computations were conducted with four values of $\phi_{tot}$ (0.04, 0.06, 0.08 and 0.10). The value $\phi_{tot}$ equals the total volume of myofibers divided by the volume of LV wall.

### 2.2 LV myofiber architecture

***Symmetric orientation:*** The helix inclination angle ($\beta$) varies linearly manner from +60° at the endocardium to the circumferential direction ($\beta$ = 0°) in the mid-wall of the LV to -60° at the epicardium. Similarly, the transverse angle ($\eta$) varies linearly from -15° at the apex to the circumferential direction ($\eta$ = 0°) in the equatorial region of the LV to +15° at the base.

***Asymmetric orientation:*** For the first model (-30°: +60°), the helix angle ($\beta$) varies in a linear manner through wall thickness from -30° at the epicardium to +60° at the endocardium. In the second model (-60°: +30°) the helix angle ($\beta$) varies also in a linear manner through wall thickness from -60° at the epicardium to +30° at the endocardium. The transverse angle ($\eta$) varies linearly from -15° at the apex to the circumferential direction ($\eta$ = 0°) in the equatorial region of the LV to +15° at the base.

***Complex orientation:*** Based on DT-MRI measurements carried by Rohmer [10] the myofiber angle evaluated by averaging the values of the helix angle ($\beta$) across the wall in eight different regions. The corresponding variations of the fiber helix angles ($\beta$) through the eight regions are as follows; septum-basal region (-60° : +40°), anterior-basal region (-40° : +60°), lateral-basal region (-20° : +50°), posterior-basal region (-20° : +60°), septum-apical region (-50° : +40°), anterior-apical region (-20° : +60°), lateral-apical region ( -20° : +50°), and posterior-apical region (-20° : +60°) vary smoothly across the LV wall thickness from a negative angle at epicardium to a positive angle at endocardium respectively. The transverse angle ($\eta$) equal zero through all the LV wall.

### 2.3 Modelling procedure, boundary and initial conditions

In all FE computations, the initial state of the myocardium was assumed to be stress-free; accordingly, the LV pressure was zero at the beginning. The boundary conditions for the intracavital pressure-time curve was formulated based on the previously published measurements performed by Nonogi [7] for the pressure-volume and volume-time behavior for the LV during the rapid filling phase of a healthy human heart.

To prevent rigid body motion of the model, degrees of freedom for all nodes at the LV base were suppressed in the long axis direction; allowing the ventricular cavities at the basal level to expand but preventing an axial displacement. To eliminate the remaining degrees of freedom, and to avoid possible excessive deformations of elements of the apex, node at the base center (pressure node) was fixed laterally.

The sensitivity of the FE computation to mesh size was investigated by using four different mesh sizes. These models consisted of: 8,064 total number of elements (three layers/elements through wall thickness), 11,520 total number of elements (five layers/elements through wall thickness), 16,128 total number of elements (seven layers/elements through wall thickness), and 25,920 total number of elements (nine layers/elements through wall thickness).

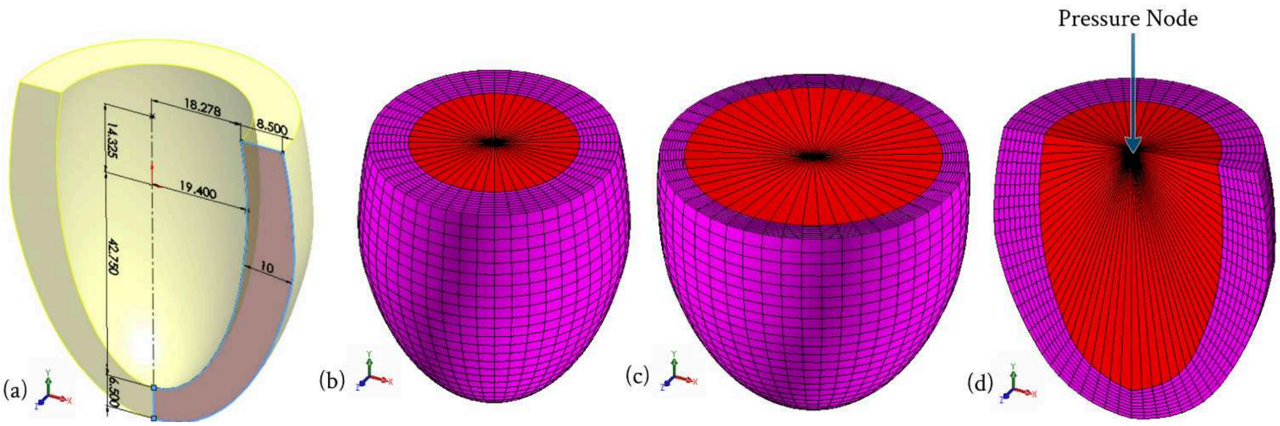


**Fig. 1:** FE model of the human LV. (a) Geometric parameters of the thick-walled ellipsoid truncated at two thirds of major axis. (b) Geometry and mesh in the undeformed state. The LV internal cavity volume is 50 ml. (c) Geometry and mesh in the deformed state. The LV internal cavity is 120 ml at the end of diastole. (d) Section view shows the shape of elements modelled in the LV internal cavity

## *2.4 Constitutive law and material properties*

***Linear elastic with large deformation:*** Material properties of the LV wall was described as a tissue-equivalent continuum (matrix) with linear isotropic properties; the Young's modulus = 20 kPa, the effective compressibility of the myocardium was simulated with a value of 0.4 for the Poisson's ratio and density = 1000 kg/m$^3$ [9]. The myofibers were characterized by a linear-elastic isotropic law with average Young's modulus = 50 kPa [11].

***Hyperelastic:*** The myocardium tissue was represented as a transversely isotropic (hyperelastic) material with relatively soft properties. Our model parameters of the hyperelastic (Ogden) models that fit the tissue mechanical behaviors are as follows; $\mu_1$ = 0.22 MPa, $\mu_2$ = 0.11 MPa, $\alpha_1$ = 11.77 and $\alpha_2$ = 14.34 [12]. To check the validity of our model two different constitutive models were employed; the first one is the three terms Ogden model following Bettendorff-Bakman [8] and represented by following parameters: $\mu_1$ = -0.03 kPa, $\mu_2$ 0.0014 kPa, $\mu_3$ = -0.05 kPa, $\alpha_1$= -45, $\alpha_2$= 38.07 and $\alpha_3$ = -14. The second is the two term Ogden model following Ghaemi [13,14] and represented by $\mu_1$ = 9.99 kPa, $\mu_2$ = 6.36 kPa, $\alpha_1$ = 2.4 and $\alpha_2$ = 2.4.

## *2.5 Result and discussion*

In order to verify the mesh density used in the above FE models, case studies with four different mesh sizes with 3, 5, 7 and 9 layers (one element per layer) through wall thickness were proposed. The maximum differences of LV internal volumes between 7-layers through wall thickness mesh and 9-layers through wall thickness mesh was less than 2% (≈1.8%). Further mesh refinement increases the computation time and may show a negligible difference of results. Therefore, the mesh density of 16128 elements (7-layers through wall thickness mesh) would be adequate from the viewpoint of efficiency, computational time costs, mesh distortion and accuracy.

Parametric study of effects of total myofiber volume fractions $\phi_{tot}$ with four different values of 0.04, 0.06, 0.08, and 0.10 was evaluated. Figure.2 shows a comparison of the obtained FE results for the LV internal volumes versus time using the adopted values of myofiber volume fractions. The FE results obtained from the LV models yields a realistic filling of the LV cavity, except for the first third of diastole, where the calculated internal volume of the LV is slightly higher than the experimental measurements especially in the case of using myofiber volume fraction $\phi_{tot}$ = 0.04.

From Figure. 2, it can be seen that FE simulation results of myofiber volume fraction $\phi_{tot}$= 0.06 is in good agreement with the experimental results. This agrees with the results obtained from R- squared analysis (coefficient of determination method). The FE data result for $\phi_{tot}$ = 0.06 gives a good fitting with the measured points with R-squared value of 73 %. While, the calculated R-squared values for $\phi_{tot}$ = 0.04, 0.08 and 0.1 are 43 %, 70 % and 57 % respectively.

The FEA results for the LV internal volumes for different transverse angle ($\eta$) show in Figure.3. The changes in the transverse angle ($\eta$) hardly affected the volume-time relation of the LV.

To investigate the effect of helix angle ($\beta$) on the volume-time response during the rapid filling phase, LV simulation models using different sets of values for $\beta$ were performed. Four different sets of the helix inclination angle $\beta$ were used. In the first set the helix angle ($\beta$) varies from +60° at the endocardium through a variable angle in the mid-wall layers to -60° at the epicardium, and the other three sets with constant values of $\beta$ = -60°, 0°, and +60° all over the LV wall. The FE

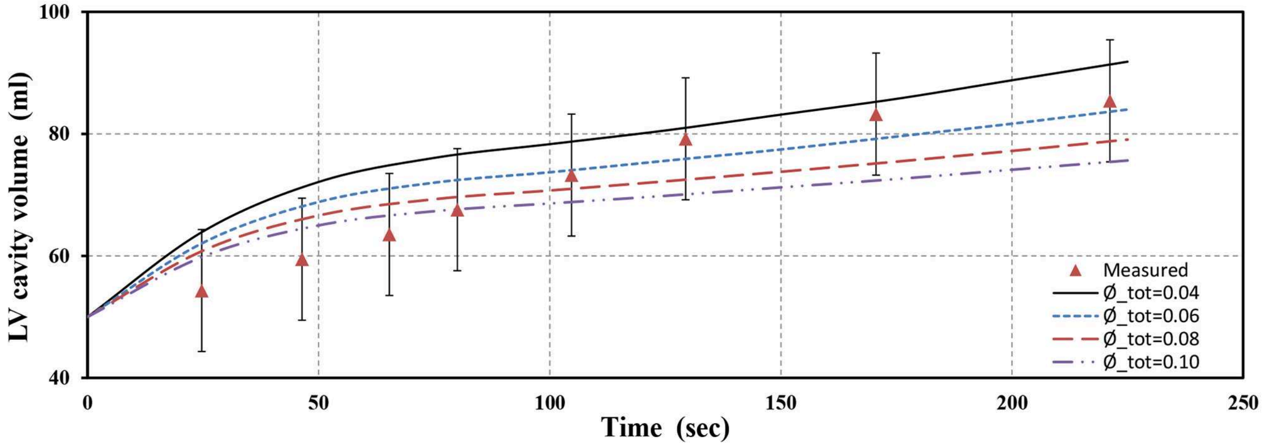


**Fig. 2:** Comparison between the LV volumes obtained from FE simulation for using different total myofiber volume fractions $\phi_{tot}$ , and the data points (triangles) obtained from published measurements by Nonogi [7] (7-layers mesh, $\eta$= -15° : +15°, $\beta$ = -60° : +60°).

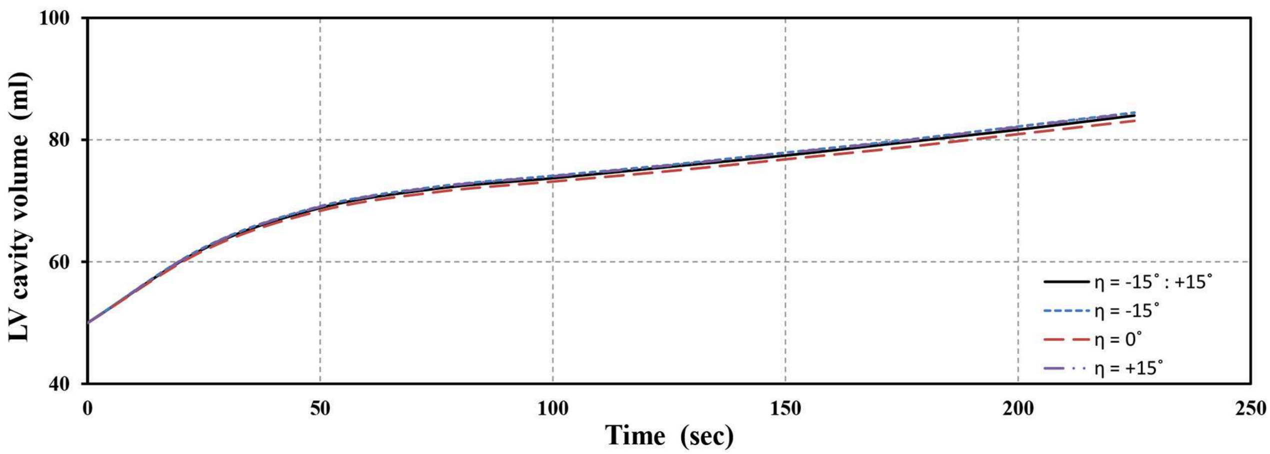


**Fig. 3:** Comparison of LV internal volumes obtained from FE simulation using both constant and variable distributions of transverse angle ($\eta$) (7-layers mesh, $\phi_{tot}$ = 0.06, $\beta$ = -60°: +60°)

results for the LV internal volumes versus time are shown in Figure.4. The obtained FE simulation results for all values of $\beta$ yields a realistic filling of the LV cavity, except for the first third of diastole for $\beta$ = -60° and +60°, where the calculated internal volume of the LV is higher in comparison with the experimental measurements. Also, it can be notice that, there is a slightly difference between the obtained FE results for myofiber distribution $\beta$ = -60° and +60°.

Figure.5 shows the comparison of obtained FE results in case of using the two asymmetric sets (-30°: +60° and -60°: +30°) with results obtained by using symmetric set (-60°: +60°). It can be seen that almost similar FE results were obtained for both symmetric and asymmetric sets. The calculated values using both systems are also in good agreement with measurement results.

Figure.6 shows a comparison of the obtained FEA results for the LV internal volumes versus time using complex myofiber orientations and myofiber without inclination. It can be seen that the oblique orientation (spiral shape) of myofibers plays an important role during the rapid filling phase, i.e. the fiber orientation has a great influence on the mechanics of the LV. Also, the calculated FE results of LV volumes for longitudinal and radial myofiber orientational distribution are over estimated. However, the obtained FE results for both the complex myofiber and circumferential orientations are in good agreement with the actual measurements.

Figure. 7 shows comparison of the calculated LV internal volumes using different material models. It can be seen that an early start of LV filling occurs when using constitutive material parameters obtained by Ghaemi [13, 14] in comparison with using constitutive material

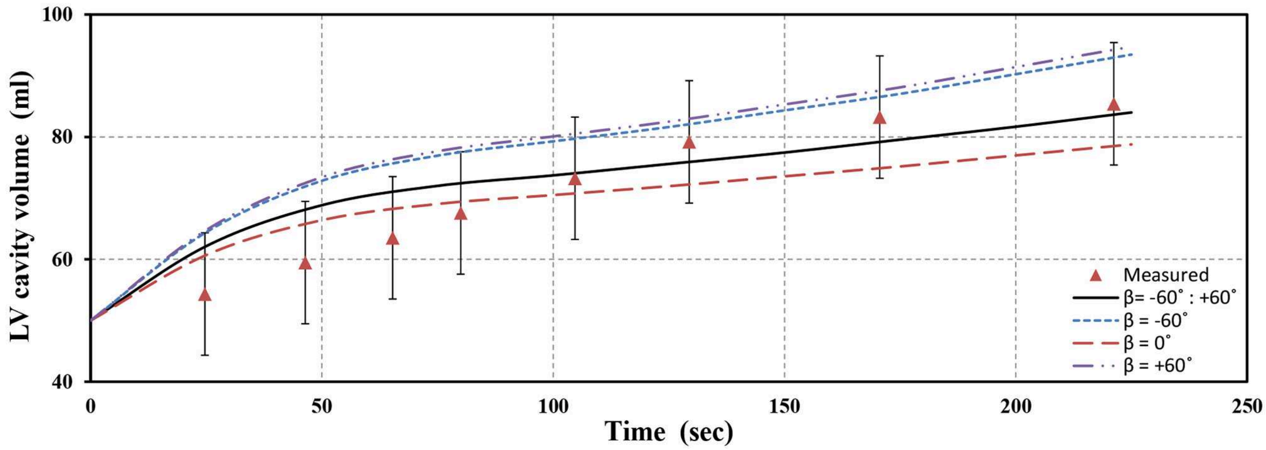


**Fig. 4:** Comparison of LV internal volumes obtained from FE simulation using both constant and variable distributions of helix angle ($\beta$) (7-layers mesh, $\phi_{tot}$= 0.06, $\eta$= -15°:+15°

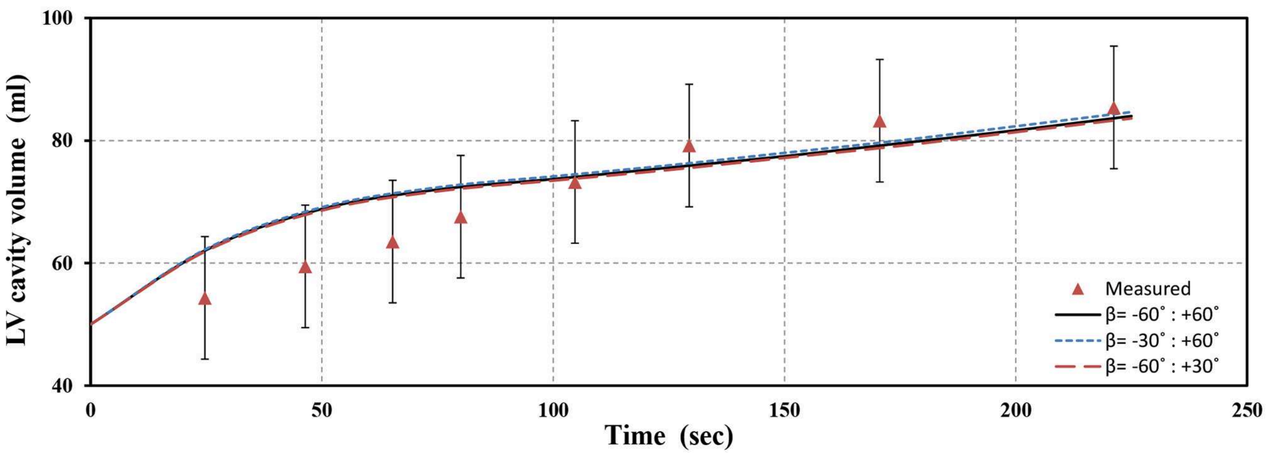


**Fig. 5:** Comparison of LV internal volumes obtained from FE simulation using both symmetric and asymmetric distributions of helix angle ($\beta$) (7-layers mesh, $\phi_{tot}$ = 0.06, $\eta$= -15°: +15°)

parameters obtained by Bettendroff-Bakman [8,9]. It can be concluded that the using material parameters obtained from our model Hassan [12] gives a good agreement with the measurements.

## 3 Conclusions

In the present study the, the human LV wall was modeled, for simplicity, as a thick-walled ellipsoid truncated at two thirds of major axis with spatial myofiber angle distribution. The ellipsoidal geometry was chosen to model the human LV, as it reflects closely to the real anatomical shape, and yet quite simple. The parametric study was designed to evaluate the effect of myofiber volume fraction and myofibers transmural orientation on the mechanical behavior of LV which undergoes inflation. The LV internal cavity was modeled by using hydrostatic fluid elements suitable to calculate the variation of blood volume and pressure during the cardiac cycle. Based on the results and discussion presented in the preceding sections, the following conclusions can be drawn:

1. The oblique orientation of myofibers plays an important role in early diastolic function.
2. The transverse angle ($\beta$) has little effect on the human LV function during the diastole.
3. The myofiber volume fraction and fiber orientation have a great influence on the mechanics of the LV during the rapid filling phase.
4. Our results show that the model results are sensitive to changes in helix angle ($\beta$) than to transverse angle ($\eta$) changes.

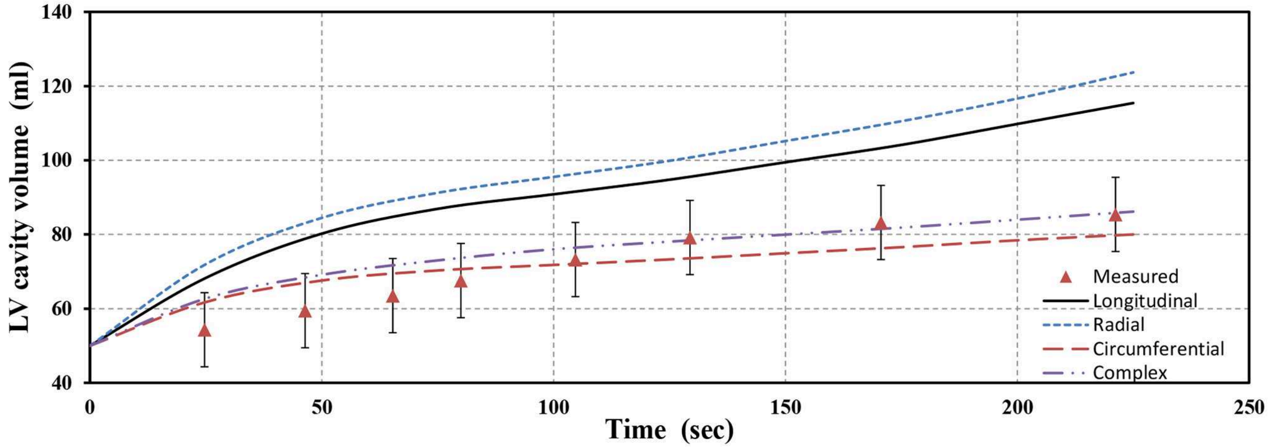


**Fig. 6:** Comparison of LV internal volumes obtained from FE simulation using different myofiber orientations (7-layers mesh, $\phi_{tot}$ = 0.06, $\eta$= zero)

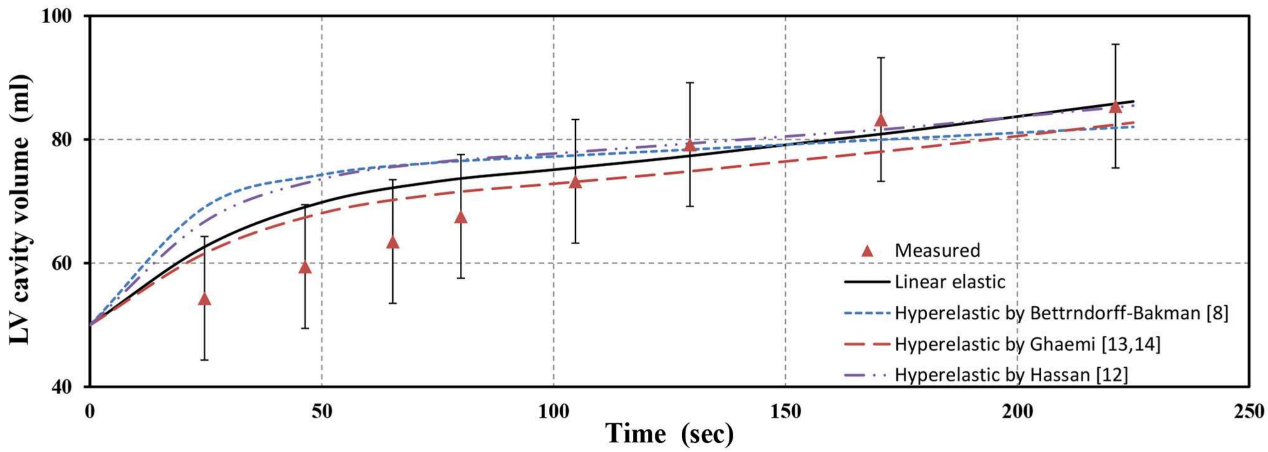


**Fig. 7:** Comparison of LV internal volumes obtained from FE simulation using different material models (7-layers mesh, $\phi_{tot}$ = 0.06, Complex myofiber orientation)


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