28th Nordic Seminar on Computational Mechanics

Personalization of a Cardiac Computational Model using Clinical Measurements

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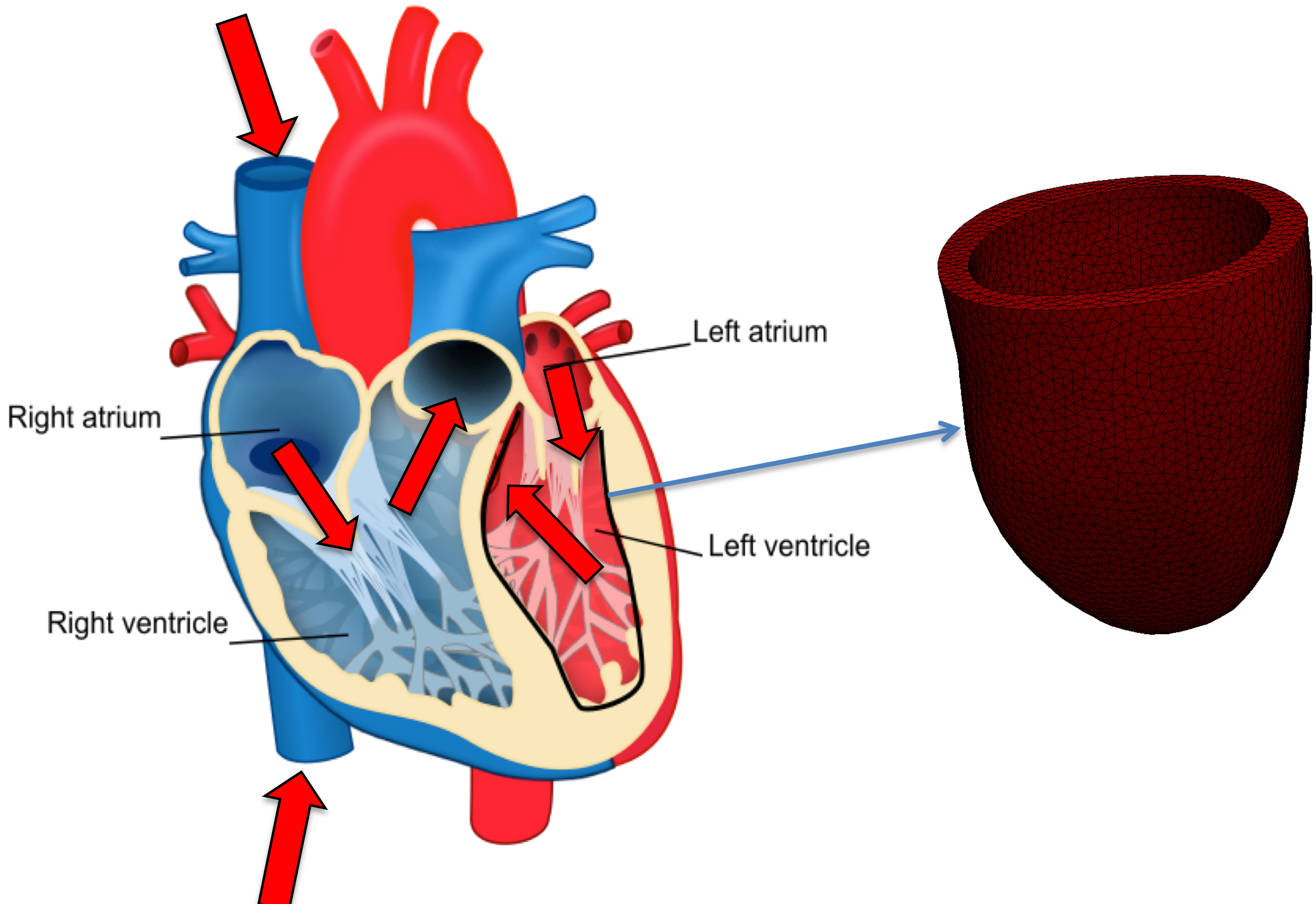
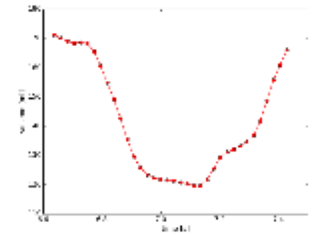
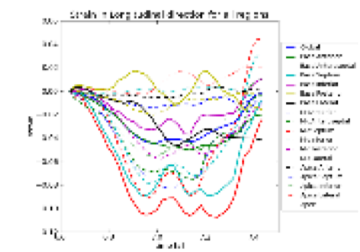
The heart is muscular organ that pumps blood through the blood vessels to the organs and the tissue in the body



LV Volume



LV Strain



LV Pressure

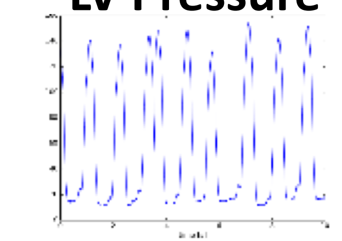
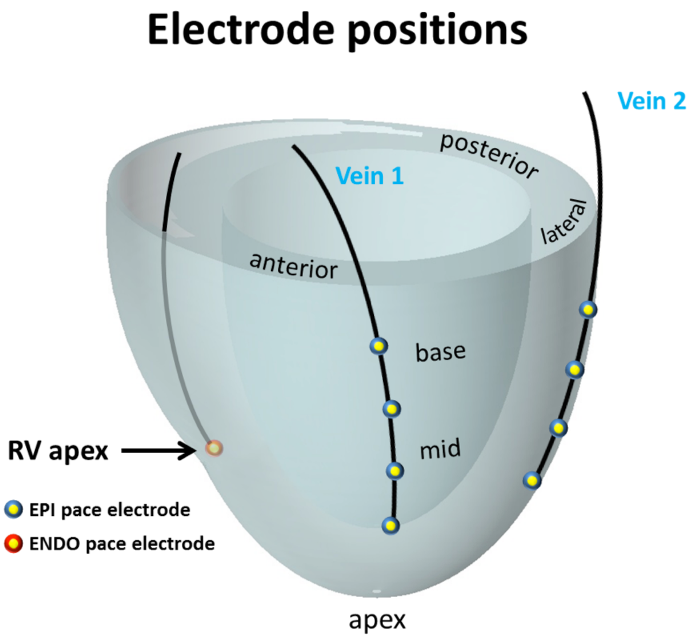
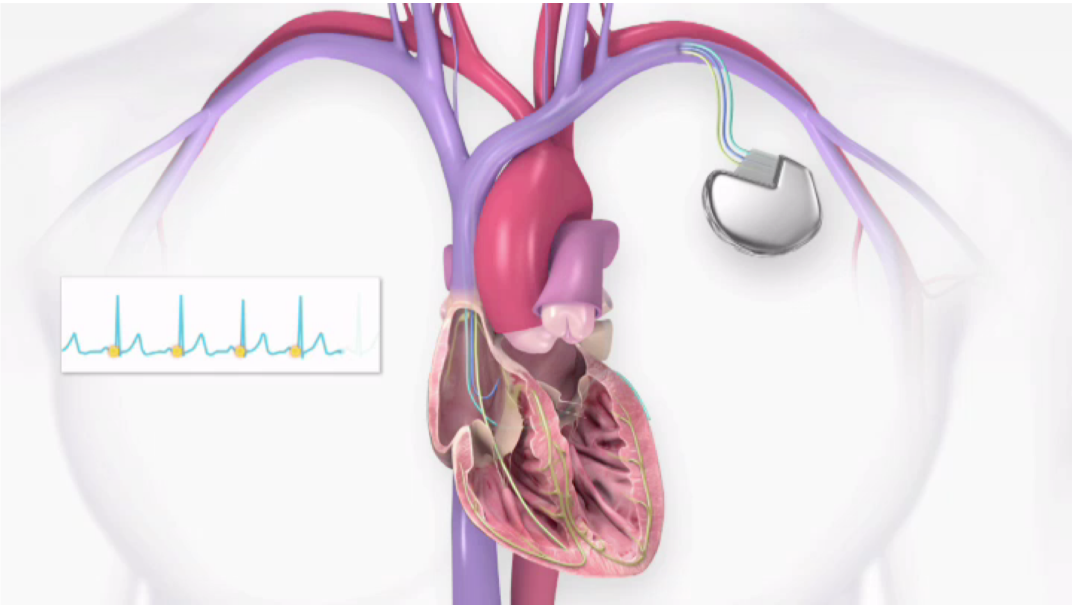


Image source: http://figures.boundless.com

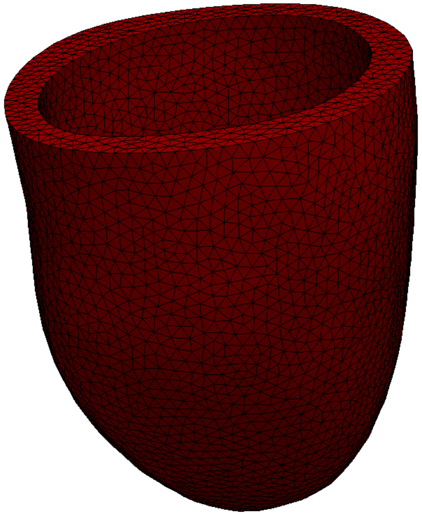
Clinicians want to improve response rates of cardiac resynchronization therapy (CRT)





Source:CCI, American Heart Asscociation

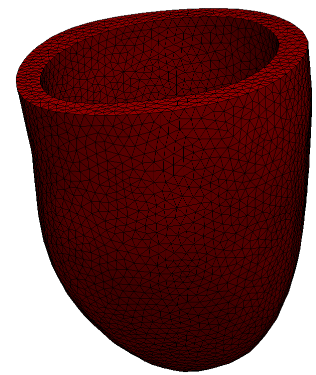
Mechanical features that can not be meaured in the clinic can be computed using a cardiac computational model



Myocardial stress can not be measured

Outline

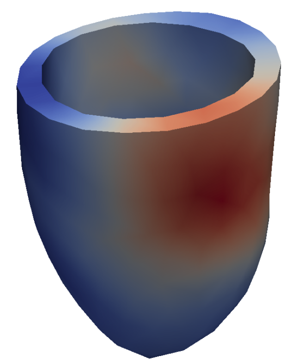
1. Building a Cardiac Computational Model



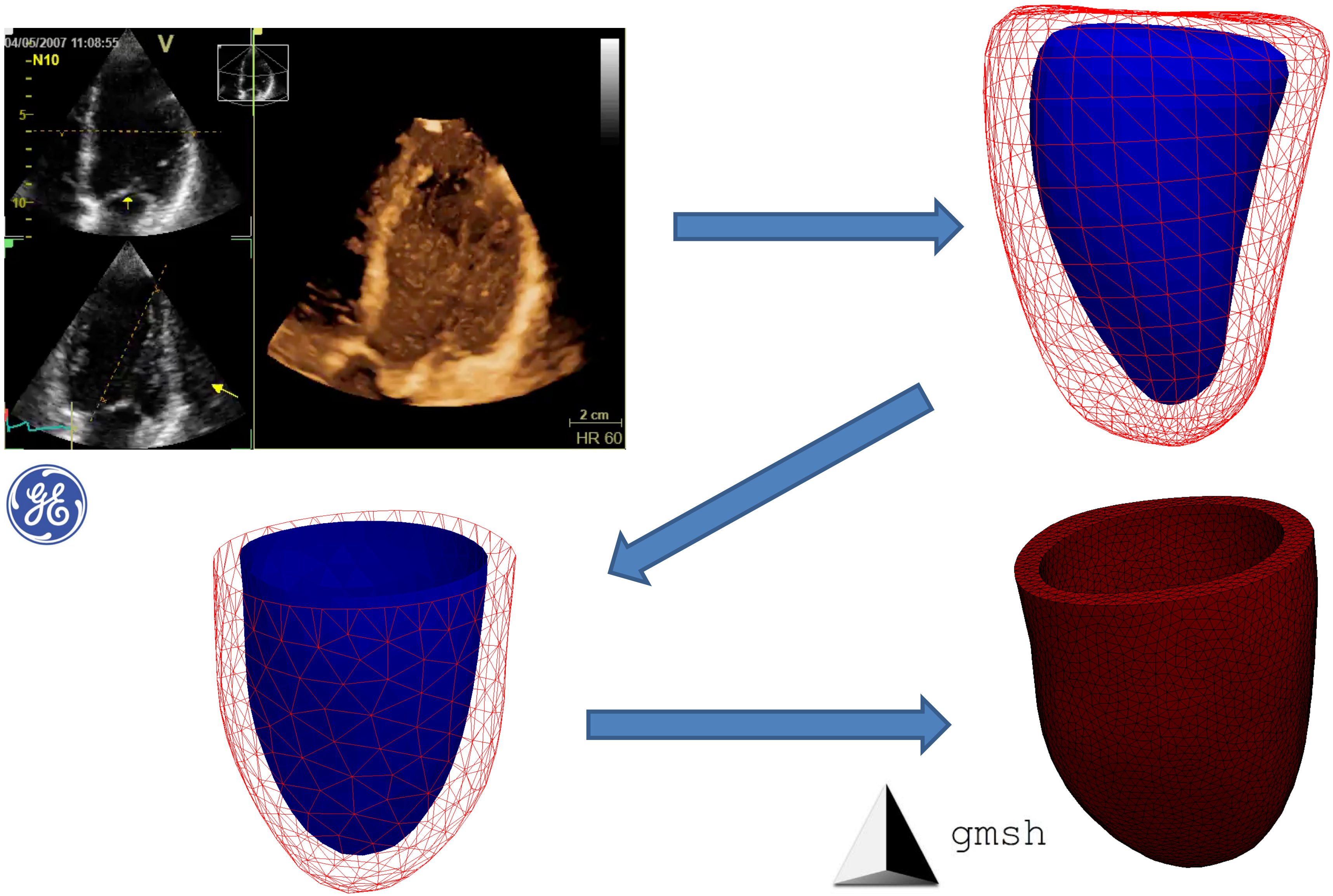
2. Constraining the model to clinical data

$$I_{\alpha}^{i} = \alpha I_{\mathrm{vol}}^{i} + (1-\alpha) I_{\mathrm{strain}}^{i}$$

3. Verification and preliminary results



The geometry is extracted from 4D echocardiography

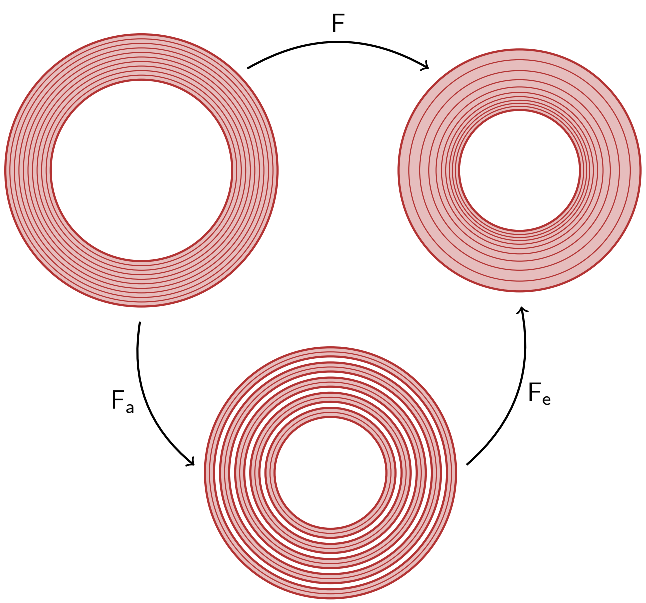


The mechanical model is based on the active strain approach

$$\mathbf{F} = \mathbf{F}_e \mathbf{F}_a$$

$$\text{Elastic deformation: } \mathbf{F}_e$$

$$\text{Active deformation: } \mathbf{F}_a$$

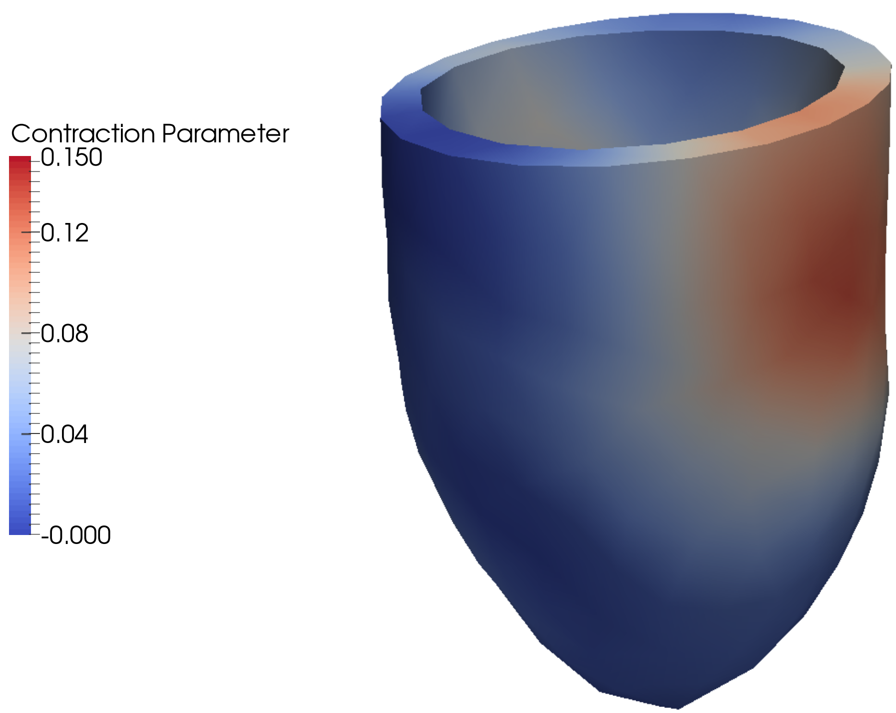


Pezzuto et. al

The active contraction is modeled by introducing a parameter which represents the relative shortening in the fiber direction

$$\mathbf{F}_a = (1-\gamma)\mathbf{f}_0 \otimes \mathbf{f}_0 + \frac{1}{\sqrt{1-\gamma}}(\mathbf{I} - \mathbf{f}_0 \otimes \mathbf{f}_0)$$

$\mathbf{f}_0$: fiber direction



Gjerald et al 2015

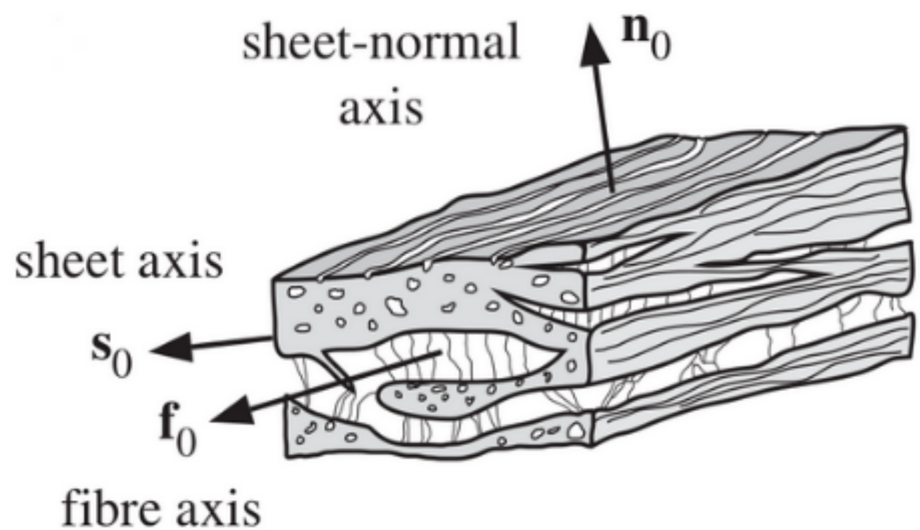
The myocardium is modelled as a hyperelastic, transversely isotropic material

$$\mathbf{F}_e = \mathbf{F}\mathbf{F}_a^{-1}$$

$$\mathbf{C}_e = \mathbf{F}_e^T \mathbf{F}_e$$

$$I_{4,\mathbf{f}_0} = \mathbf{f}_0 \cdot (\mathbf{C}_e \mathbf{f}_0)$$

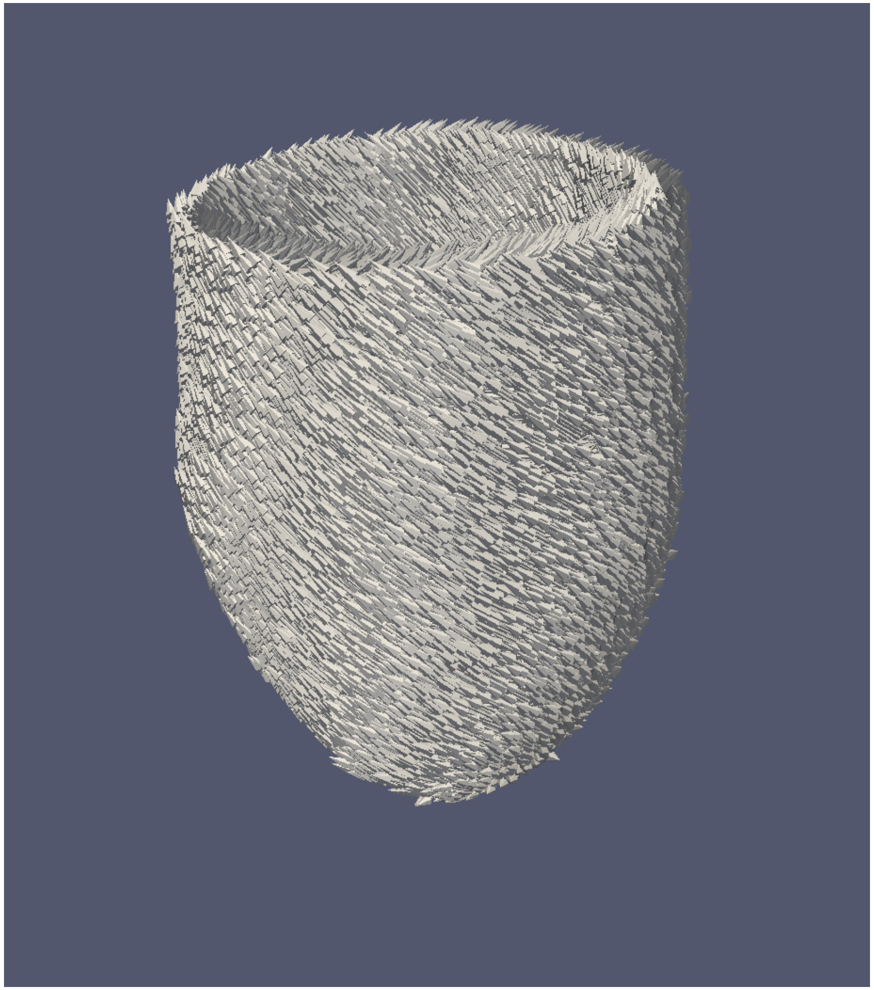
$$I_1 = \mathrm{tr}(\mathbf{C}_e)$$

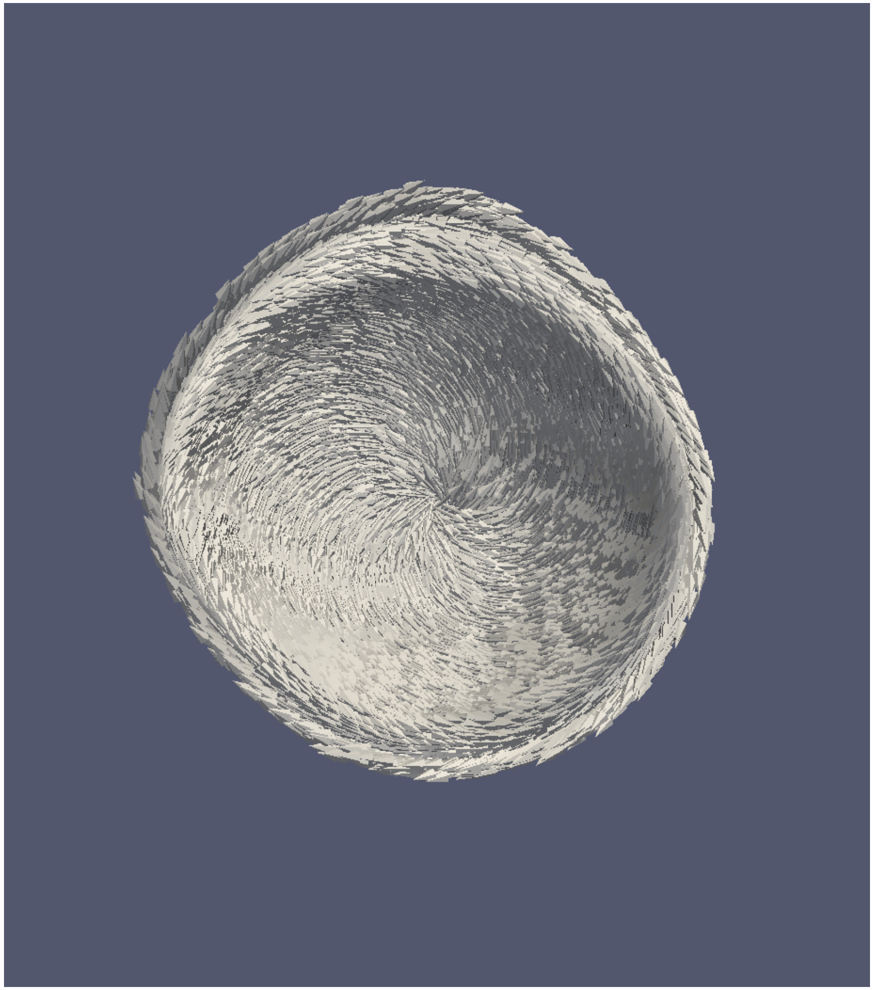


$$\mathcal{W}(\mathbf{C}_e) = \underbrace{\frac{a}{2b}\left(e^{b(I_1-3)} - 1\right)}_{\textbf{Isotropic term}} + \underbrace{\frac{a_f}{2b_f}\left(e^{b_f(I_{4,\mathbf{f}_0}-1)_+^2} - 1\right)}_{\textbf{Transversely isotropic term (fibers)}}$$

Holzapfel and Ogden 2009

Myocardial fiber orientation is assigned by using a rule-based algorithm





Bayer et al 2012

The mechanical model

Active Strain Decomposition:

$$\mathbf{F} = \mathbf{F}_e \mathbf{F}_a$$

Passive Mechanics:

$$\mathcal{W}(\mathbf{C}_e) = \frac{a}{2b}\left(e^{b(I_1-3)} - 1\right) + \frac{a_f}{2b_f}\left(e^{b_f(I_{4,\mathbf{f}_0}-1)_+^2} - 1\right)$$

$$\mathbf{C}_e = \mathbf{F}_e^T \mathbf{F}_e \qquad I_{4,\mathbf{f}_0} = \mathbf{f}_0 \cdot (\mathbf{C}_e \mathbf{f}_0) \qquad I_1 = \mathrm{tr}(\mathbf{C}_e)$$

Active Mechanics:

$$\mathbf{F}_a = (1-\gamma)\mathbf{f}_0 \otimes \mathbf{f}_0 + \frac{1}{\sqrt{1-\gamma}}(\mathbf{I} - \mathbf{f}_0 \otimes \mathbf{f}_0)$$

The mechanical model (continue)

Boundary Conditions:

Endocardial pressure from clinical data :

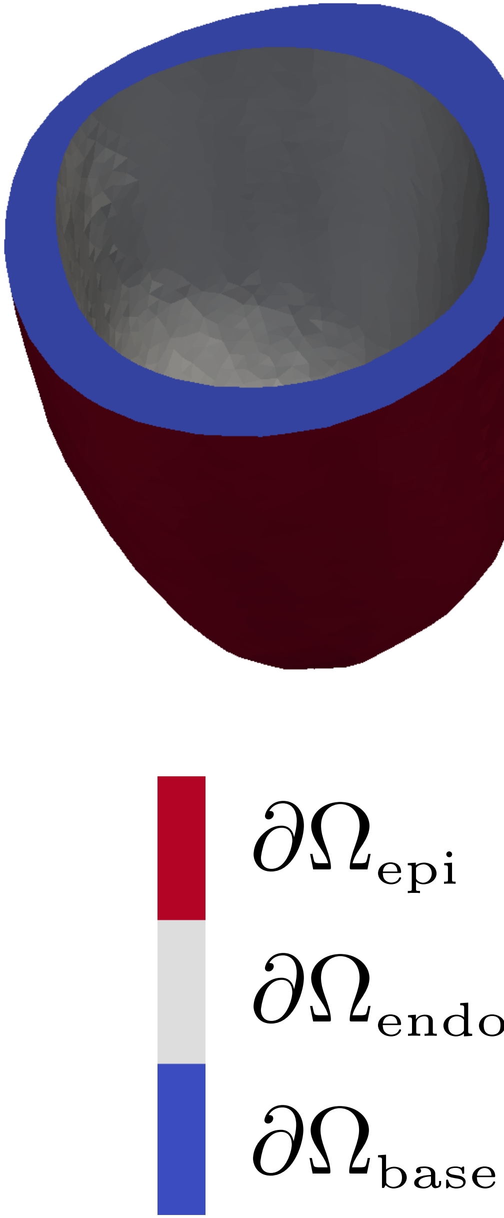
$$\mathbf{T} = J p_{\mathrm{LV}} \mathbf{F}^{-T} \mathbf{N}, \ \mathbf{x} \in \partial\Omega_{\mathrm{endo}}$$

Fix base in vertical direction. Allow some displacement in the base plane:

$$\mathbf{u} = (u_1, u_2, u_3)$$

$$u_1 = 0, \ \mathbf{x} \in \partial\Omega_{\mathrm{base}}$$

$$\mathbf{T} = -k\mathbf{u}, \ \mathbf{x} \in \partial\Omega_{\mathrm{base}}$$

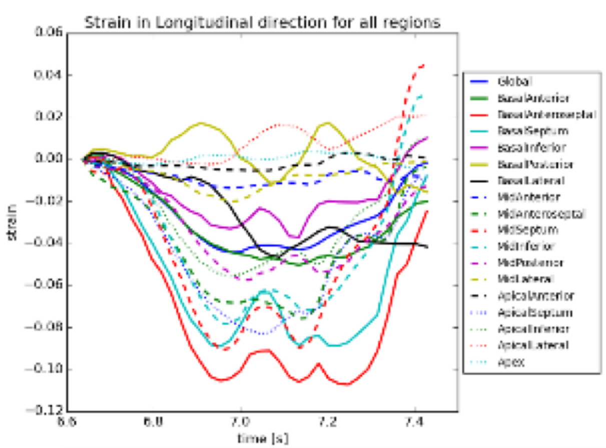


The total elastic energy functional is made specific to a patient by calibrating the 5 model parameters to clinical data

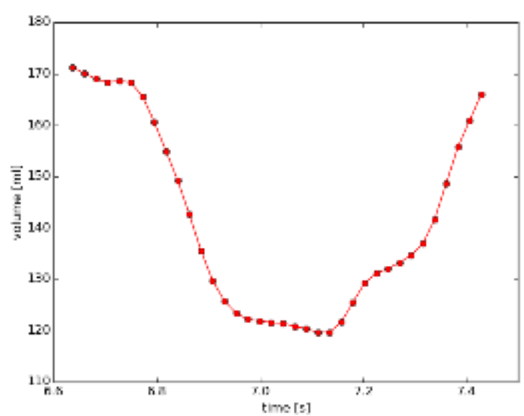
$$\Pi = \int_{\Omega} \mathcal{W}(\mathbf{C}_e) + p(J-1)dV + \text{boundary conditions}$$

$$R(\mathbf{u}, p) = \begin{pmatrix} D_{\delta \mathbf{u}}\Pi \\ D_{\delta p}\Pi \end{pmatrix} = \mathbf{0} \qquad \Pi = \Pi(\gamma, a, b, a_f, b_f)$$

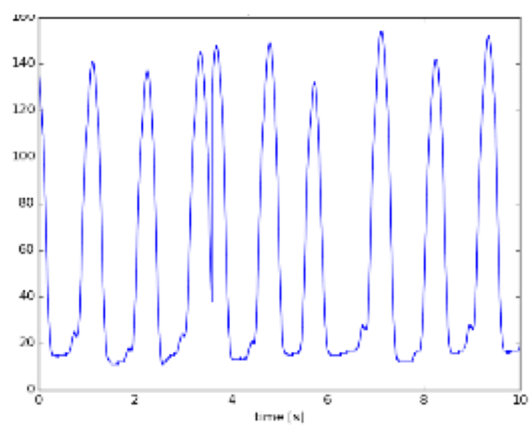
LV Regional strain



LV Volume



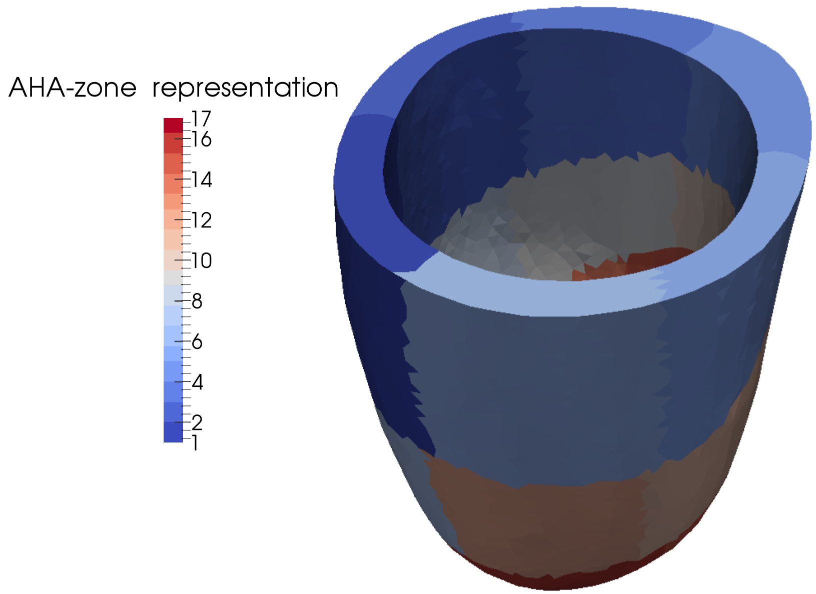
LV Pressure

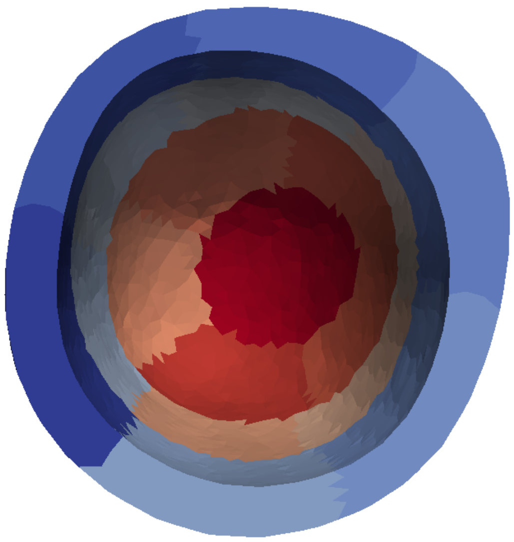


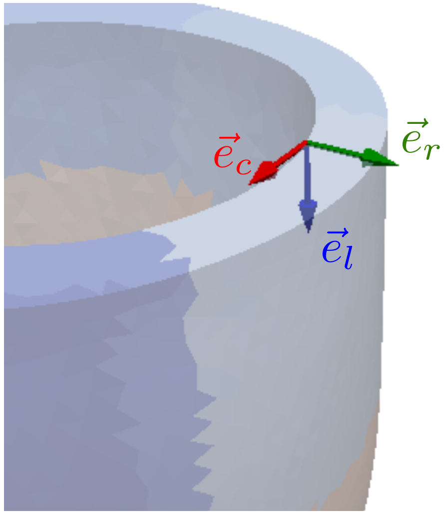
Minimize misfit

Boundary condition

The average regional strain is computed in three directions in the left ventricle(LV)







$$\Omega_{\mathrm{LV}} = \bigcup_{j=1}^{17} \Omega^j$$

$$\tilde{\varepsilon}_{k,j} = \frac{1}{|\Omega^j|} \int_{\Omega^j} e_k^T \nabla \mathbf{u} \cdot e_k \; dx$$

We minimize the difference between measured and simulated strain and volume

Volume misfit functional:

$$I^i_{\mathrm{vol}} = \frac{|V_i - \tilde{V}_i|}{V_i}, \ \tilde{V}_i = \frac{1}{3}\int_{\partial\Omega_{\mathrm{endo}}} (\mathbf{I} + \mathbf{u})\cdot J\mathbf{F}^{-T}\mathbf{N} \ dS$$

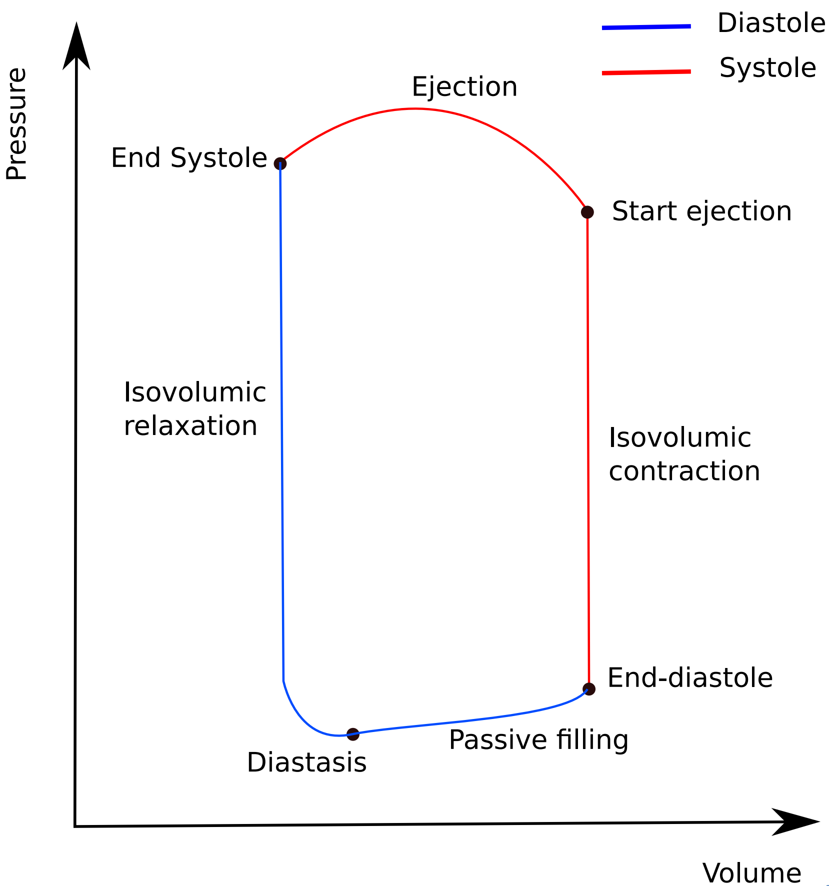
Strain misfit functional:

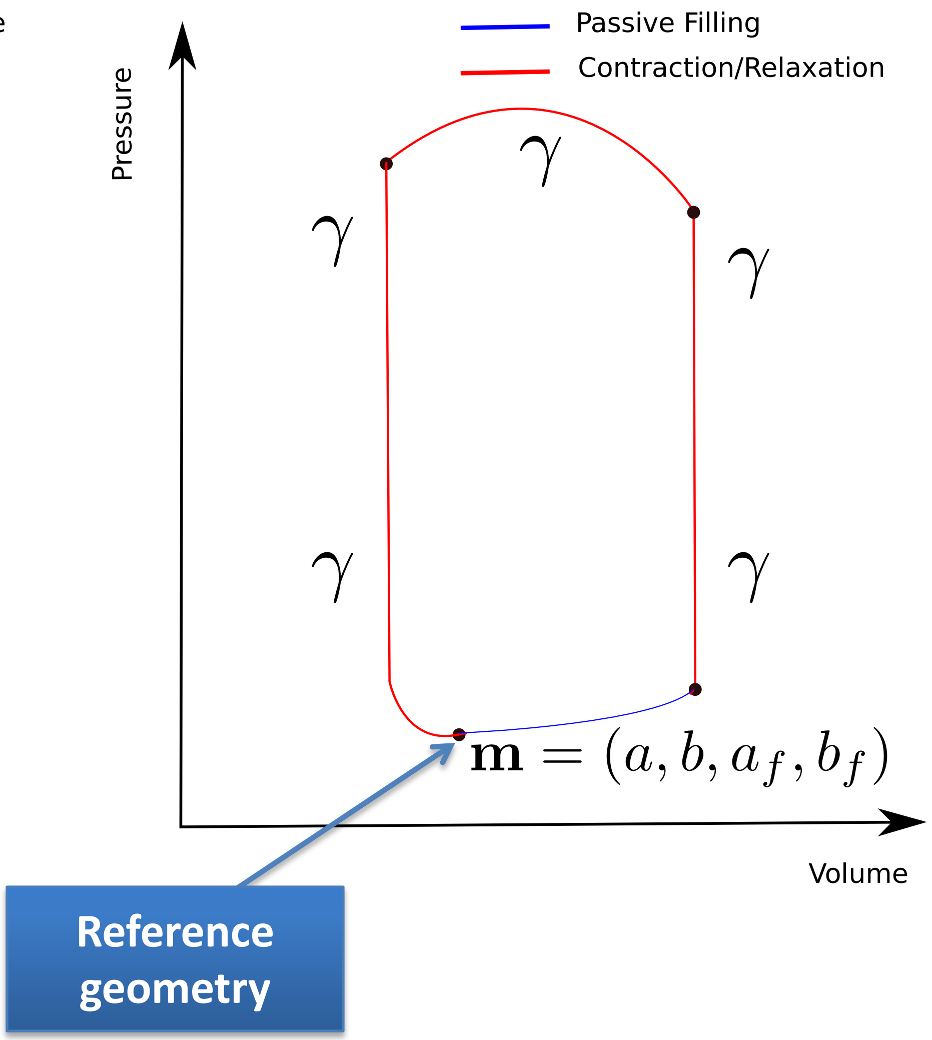
$$I^i_{\mathrm{strain}} = \sum_{j=1}^{17} \left\| \mathbf{W}(\boldsymbol{\varepsilon}^i_j - \tilde{\boldsymbol{\varepsilon}}^i_j) \right\|_2, \ \ \boldsymbol{\varepsilon}^i_j = \begin{bmatrix} \varepsilon^i_{c,j} \\ \varepsilon^i_{r,j} \\ \varepsilon^i_{l,j} \end{bmatrix}, \ \mathbf{W} = \mathrm{diag}(\omega_{c,j}, \omega_{r,j}, \omega_{l,j})$$

Total misfit functional (weighted):

$$I^i_\alpha = \alpha I^i_{\mathrm{vol}} + (1-\alpha) I^i_{\mathrm{strain}}$$

Different part of the cardiac cycle is used to calibrate the different parameters





Parameters are estimated by solving a PDE-constrained optimization problem

Passive material parameters:

$$\begin{aligned} &\underset{\mathbf{m}}{\text{minimize}} && \sum_{i=0}^{N_{\mathrm{ED}}} I_{\alpha}^{i} \\ &\text{subject to} && R(\mathbf{u}, p) = 0. \end{aligned}$$

$N_{\mathrm{ED}} =$ Number of points from beginning of passive filling to End-Diastole

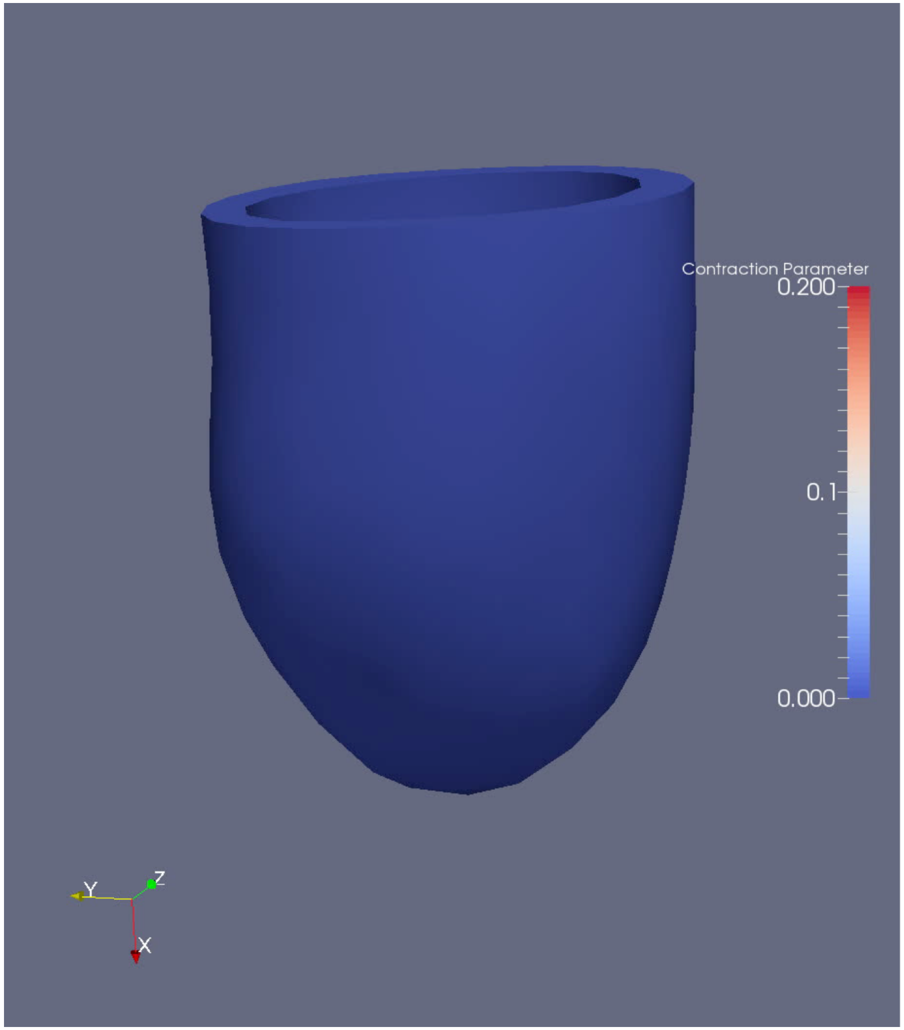
Active contraction parameter:

$$\begin{aligned} &\underset{\gamma(\mathbf{x}, i)}{\text{minimize}} && I_{\alpha}^{i} + \lambda \|\nabla \gamma\|_{L^2(\Omega)}^2 \\ &\text{subject to} && R(\mathbf{u}, p) = 0, \\ & && \gamma(\mathbf{x}, i) \in [0, 1), \ \mathbf{x} \in \Omega, i = N_{\mathrm{ED}} + 1, \cdots, N. \end{aligned}$$

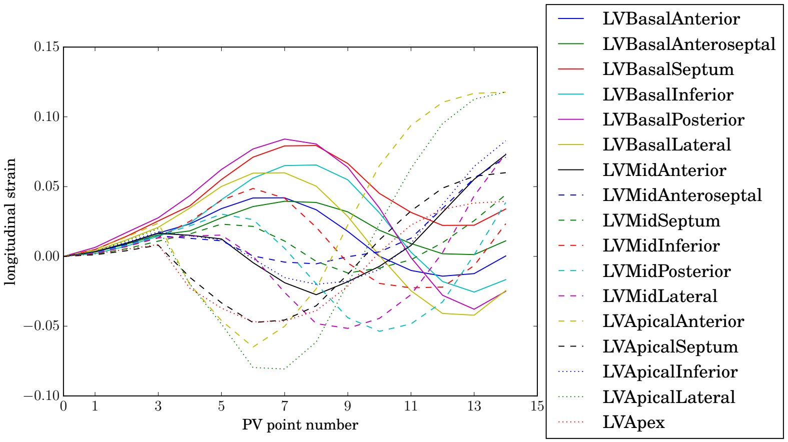
dolfin-adjoint

FEniCS project

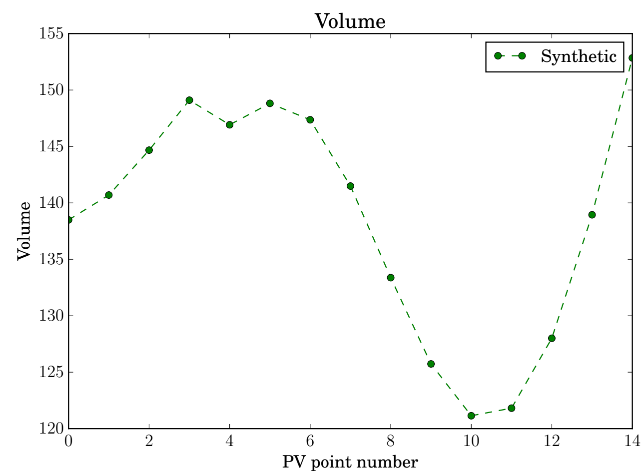
We test the model with a prescribed sequence of contraction parameters



Strain curves:



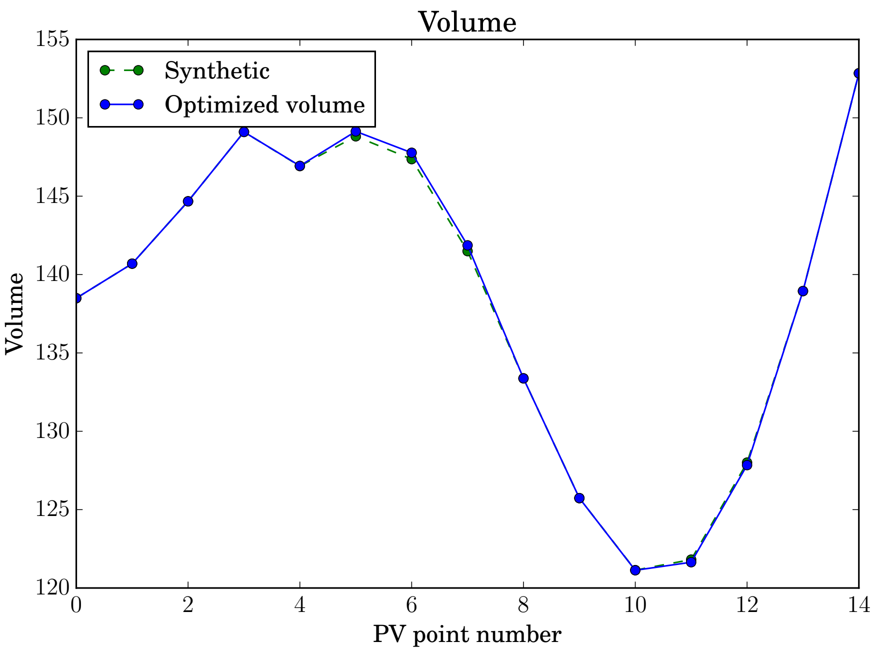
Volume:



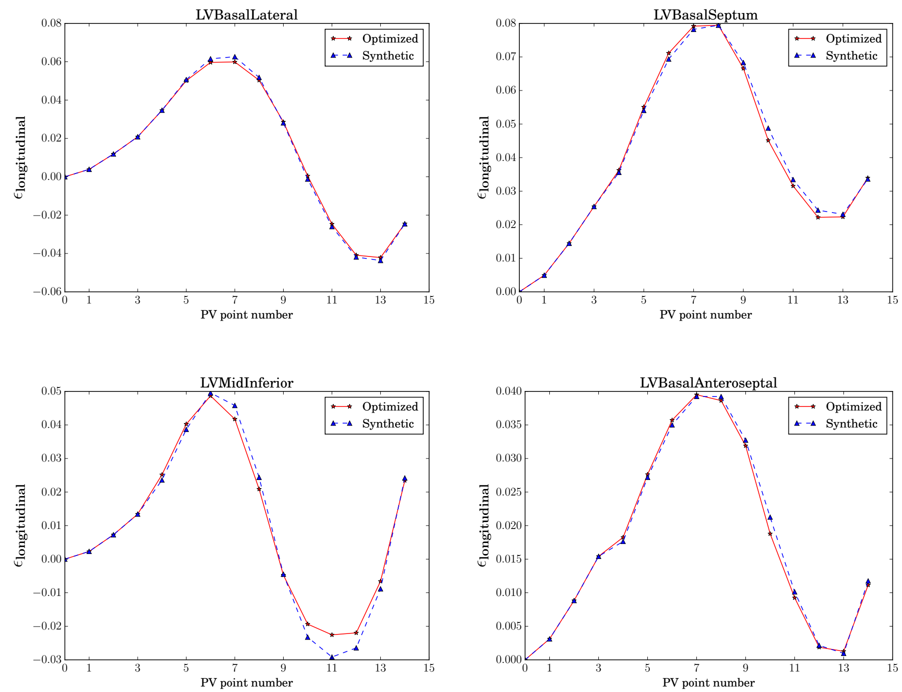
We are able to fit the volume and strain very well using synthetic data

$$\alpha = 0.5$$
$$\lambda = 10.0$$

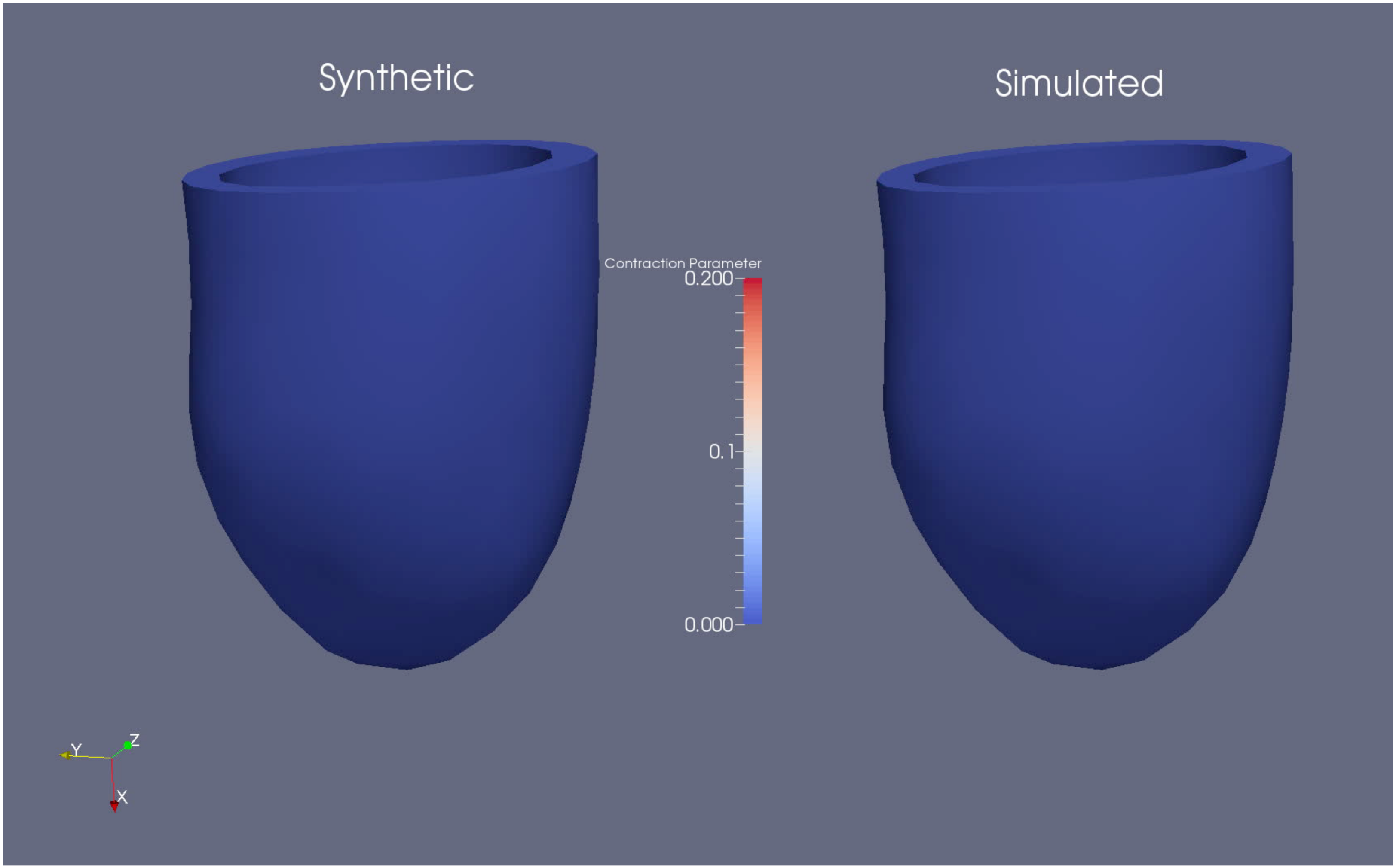
Volume match:



Strain match:

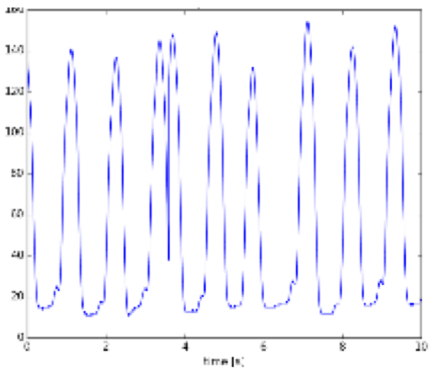


We test the model with a prescribed sequence of contraction parameters

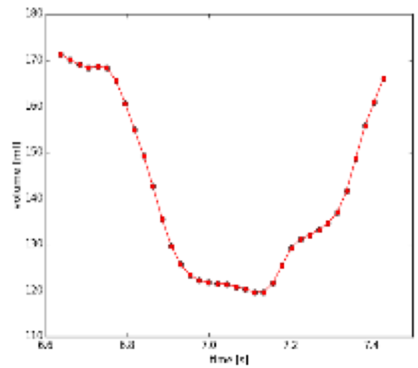


We test the model on real patient data

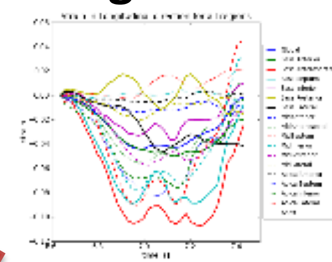
LV Pressure:



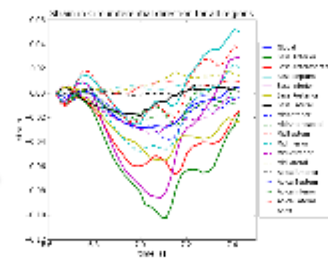
LV Volume:



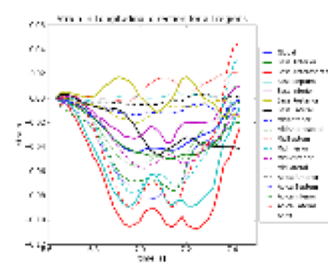
Longitudinal



Circumferential



Radial



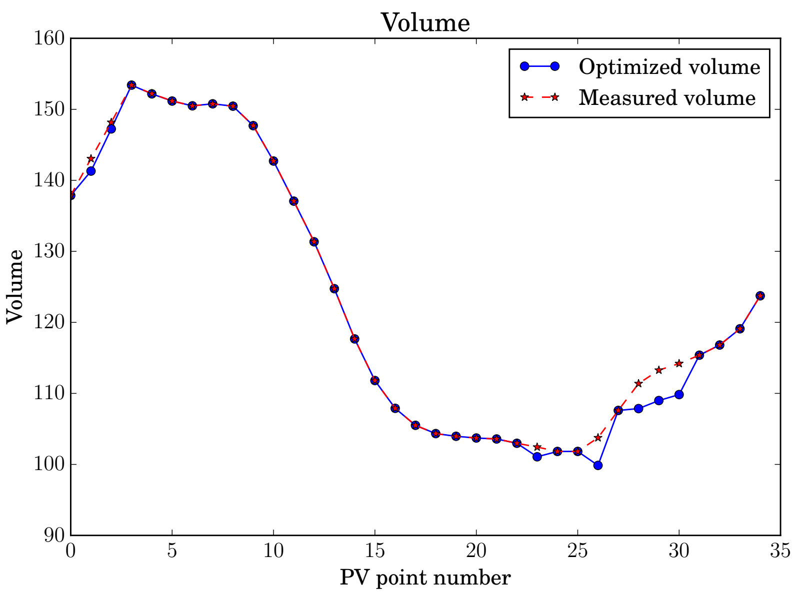
Model

Objective function

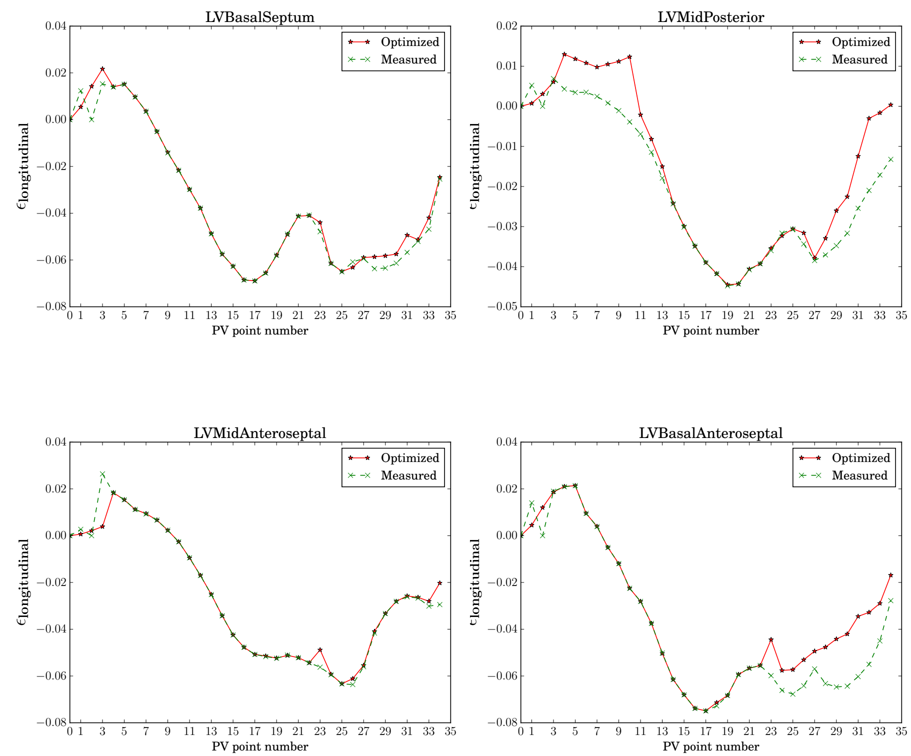
We are able to fit well the volume and strain in the regions with highest weight

$$\alpha = 0.5$$
$$\lambda = 50.0$$

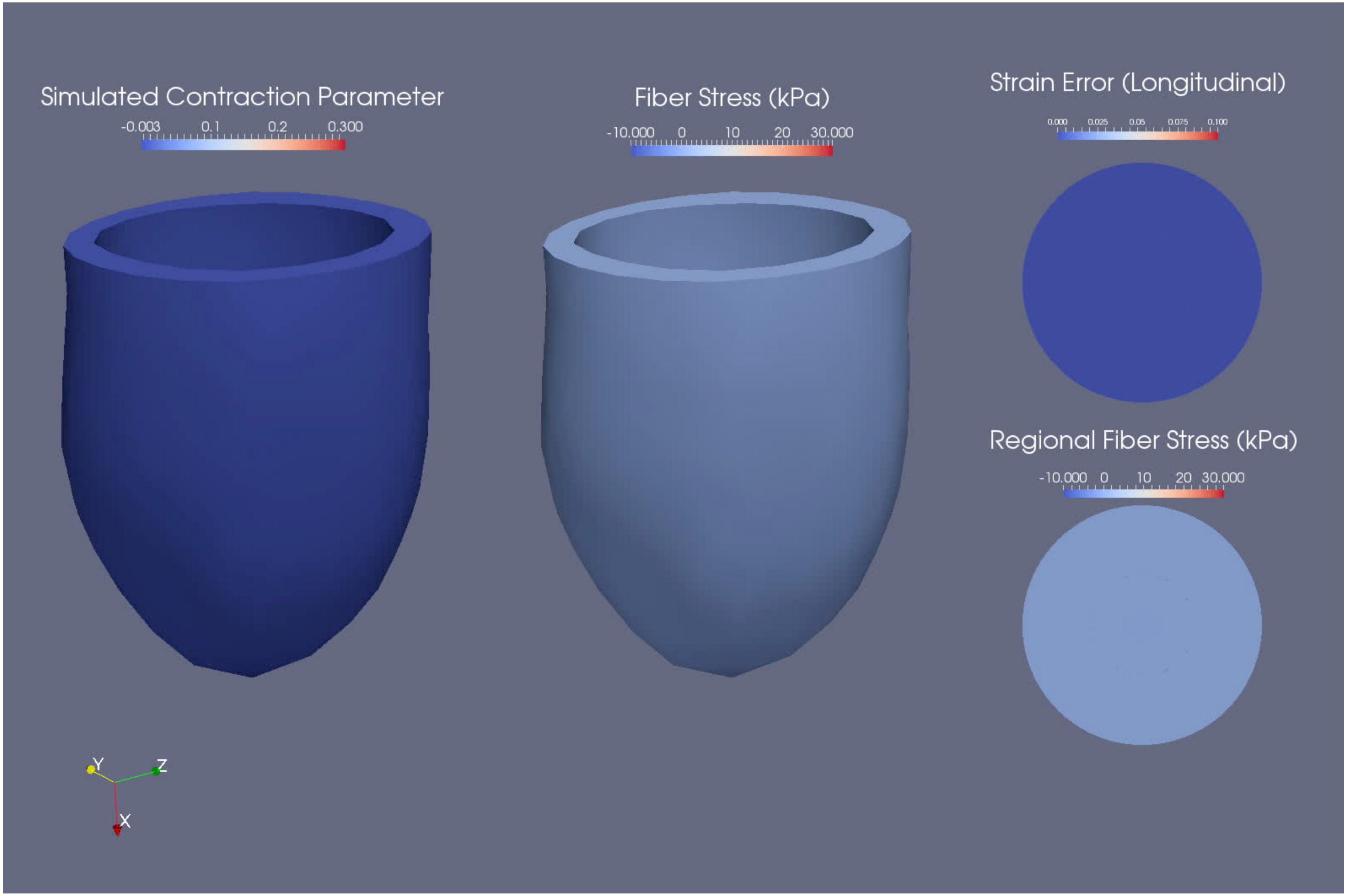
Volume match:



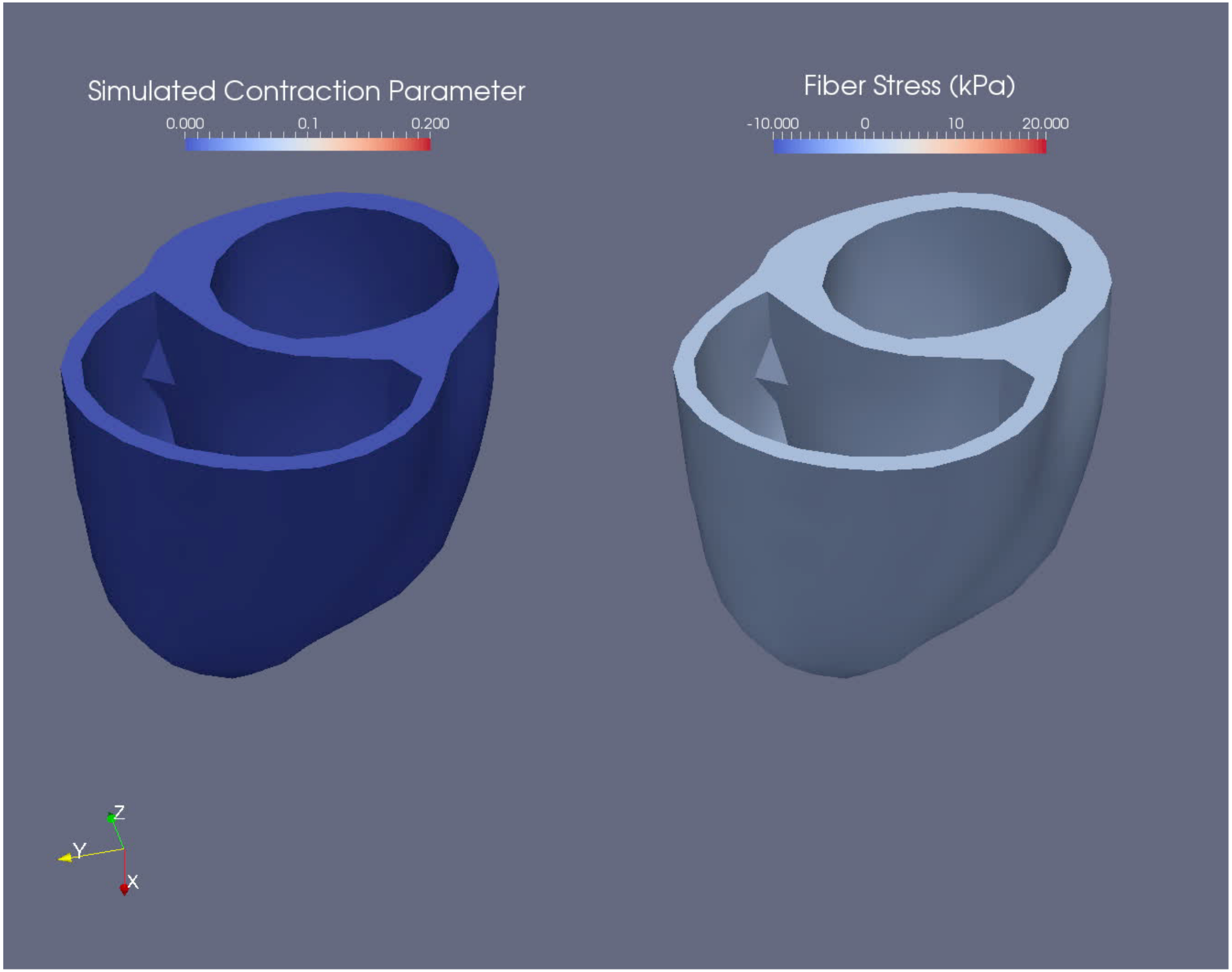
Strain match in regions with high weight:



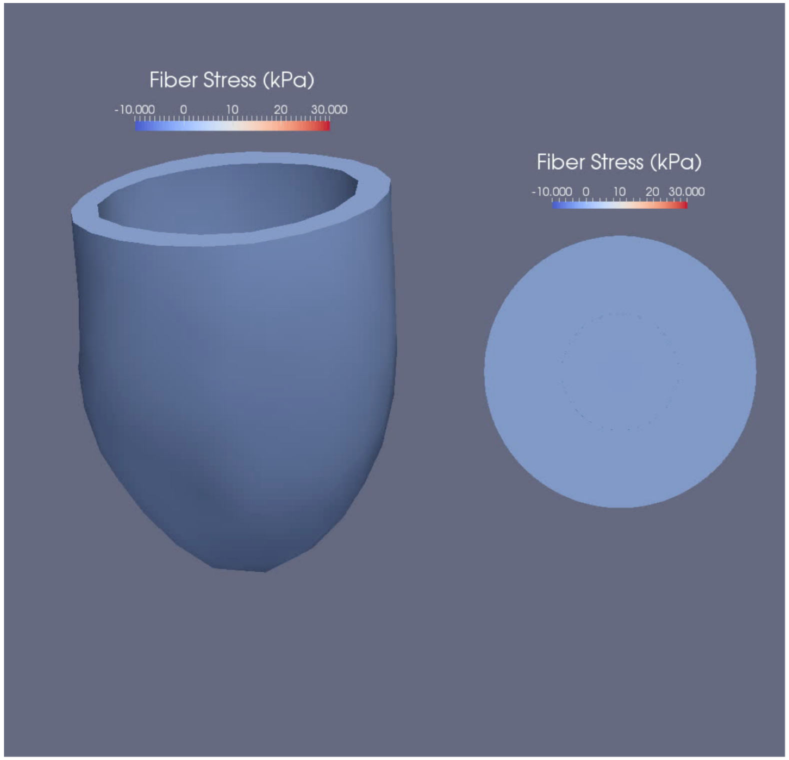
Results with equal weighting of strain and volume



We can also apply this method using a bi-ventricular mesh



In summary, we are able to calibrate a model using clinical measurement, and hence image patient specific stress maps



Acknowledgements

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