

Twin hearts: dalla matematica, alla ricerca, al paziente

26 June 2026

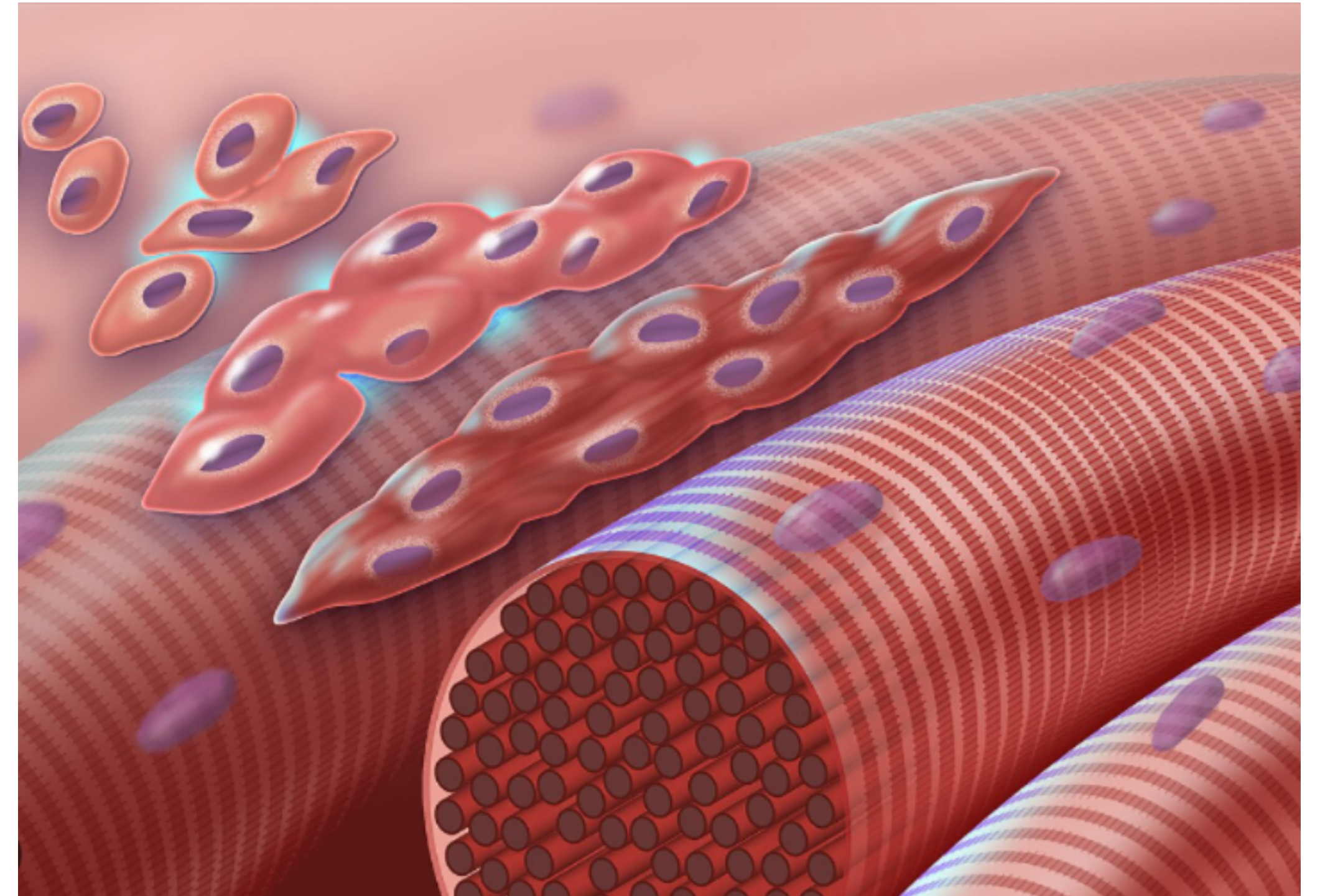
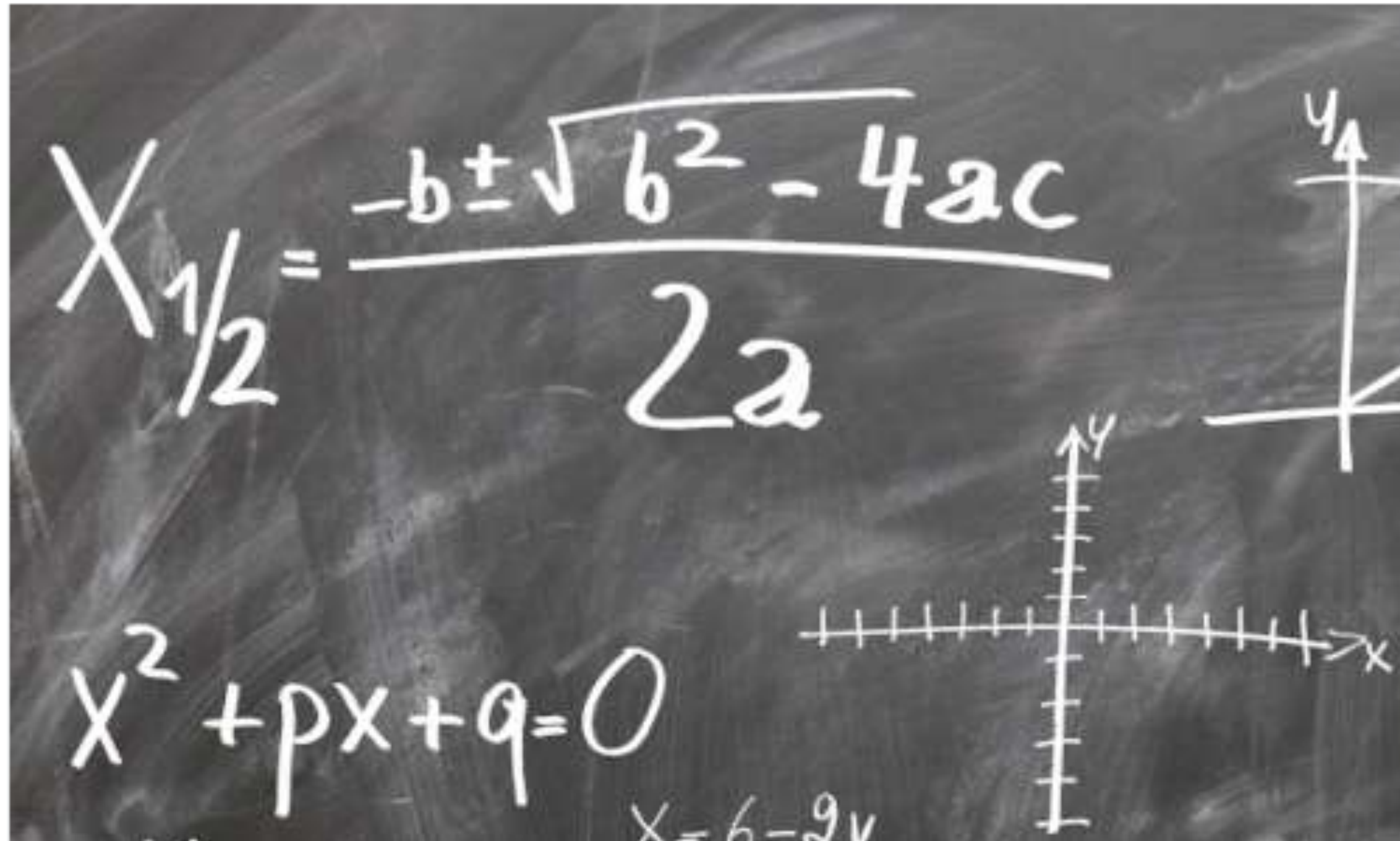
Ing. Francesco Viola, Ing. Roberto Verzicco, Dr. Paolo Ciancarella and Dr. Alessandro Ricci

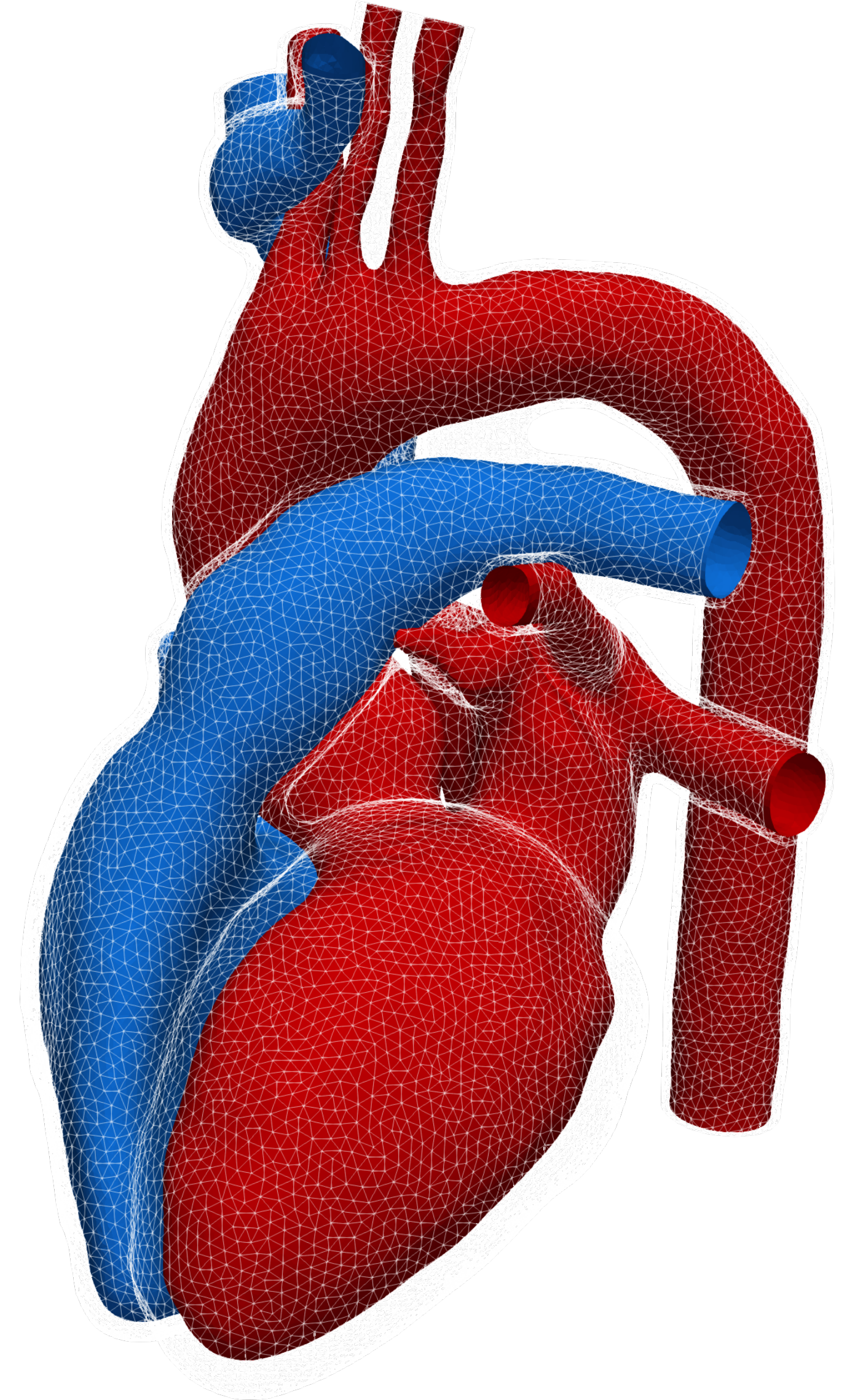
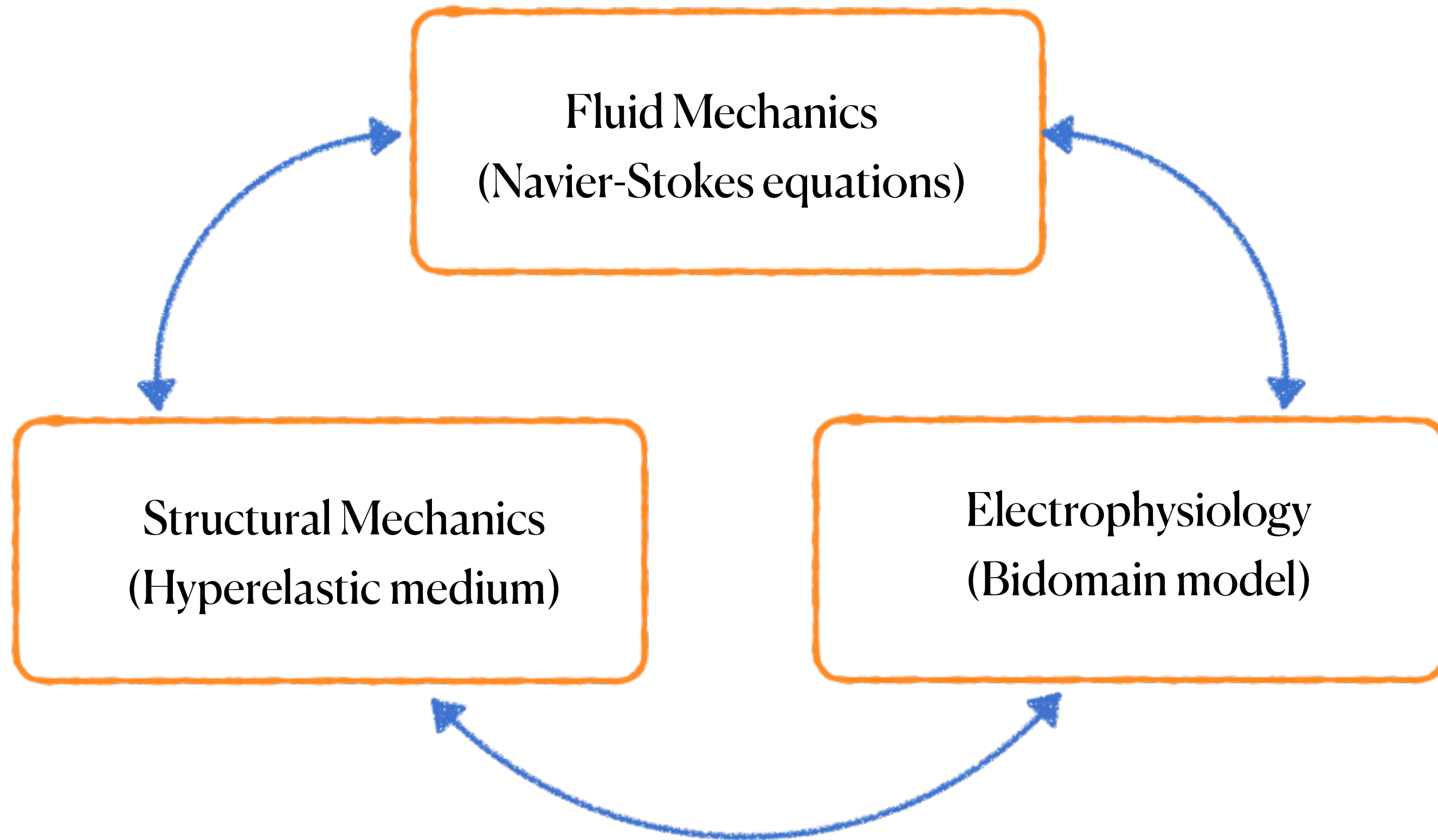


CARDIOTRIALS
(StG) 101039675









Contents lists available at [ScienceDirect](https://www.sciencedirect.com)

Computer Physics Communications

www.elsevier.com/locate/cpc

FSEI-GPU: GPU accelerated simulations of the fluid–structure–electrophysiology interaction in the left heart[☆]

Francesco Viola^a, Vamsi Spandan^b, Valentina Meschini^c, Joshua Romero^d, Massimiliano Fatica^d, Marco D. de Tullio^e, Roberto Verzicco^{f,g,a,*}

PHYSICAL REVIEW FLUIDS **8**, 100502 (2023)

Invited Articles

**High-fidelity model of the human heart:
An immersed boundary implementation**

Francesco Viola^{1,2}, Giulio Del Corso^{1,*} and Roberto Verzicco^{1,3,4,†}

¹GSSI, Viale Francesco Crispi 7, 67100 L'Aquila, Italy

²INFN–Laboratori Nazionali del Gran Sasso, 67100 Assergi (AQ), Italy

³Physics of Fluids Group, University of Twente, 7500AE Enschede, The Netherlands

⁴Dipartimento di Ingegneria Industriale, University of Rome 'Tor Vergata,' Rome 00133, Italy

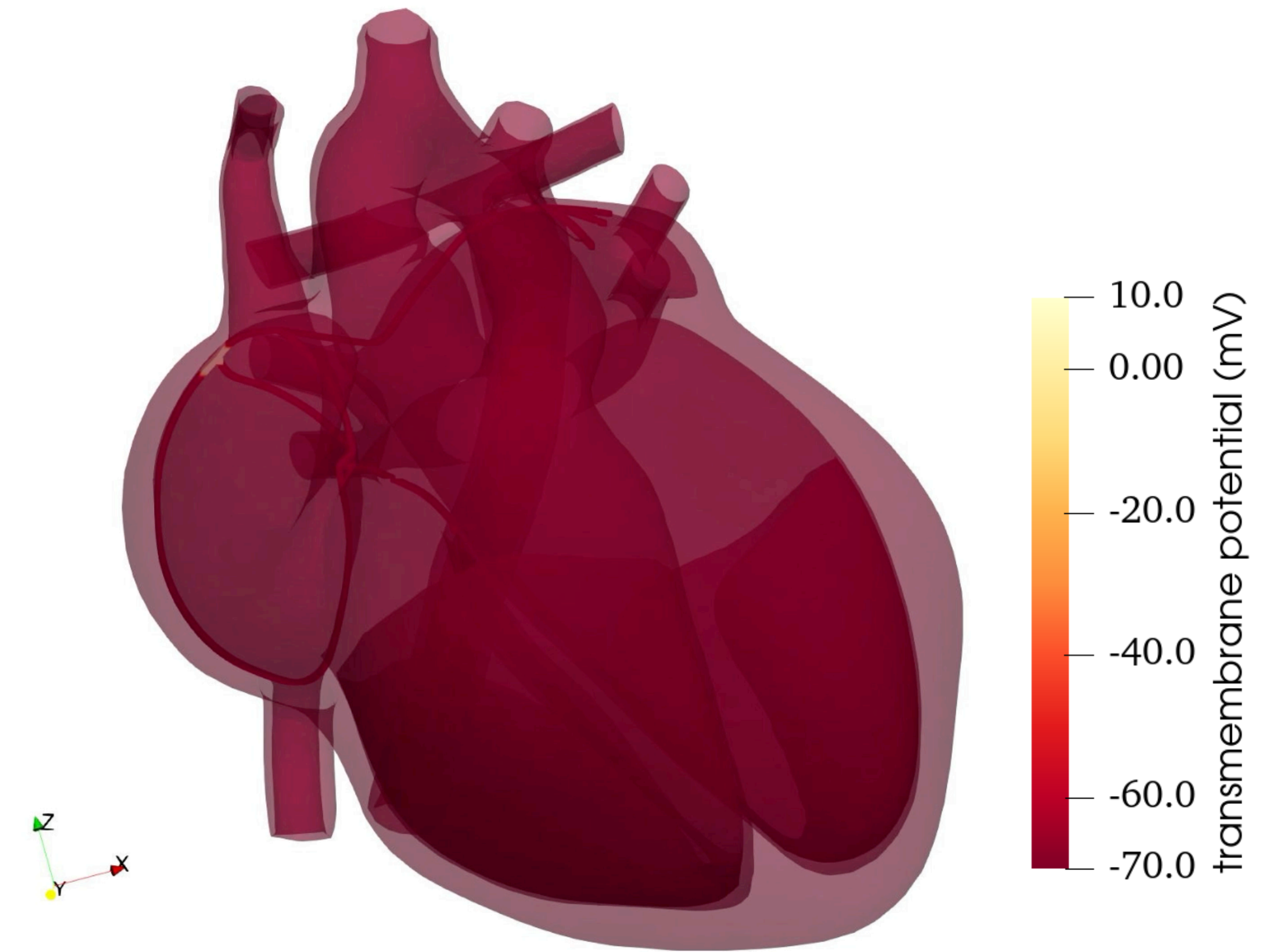
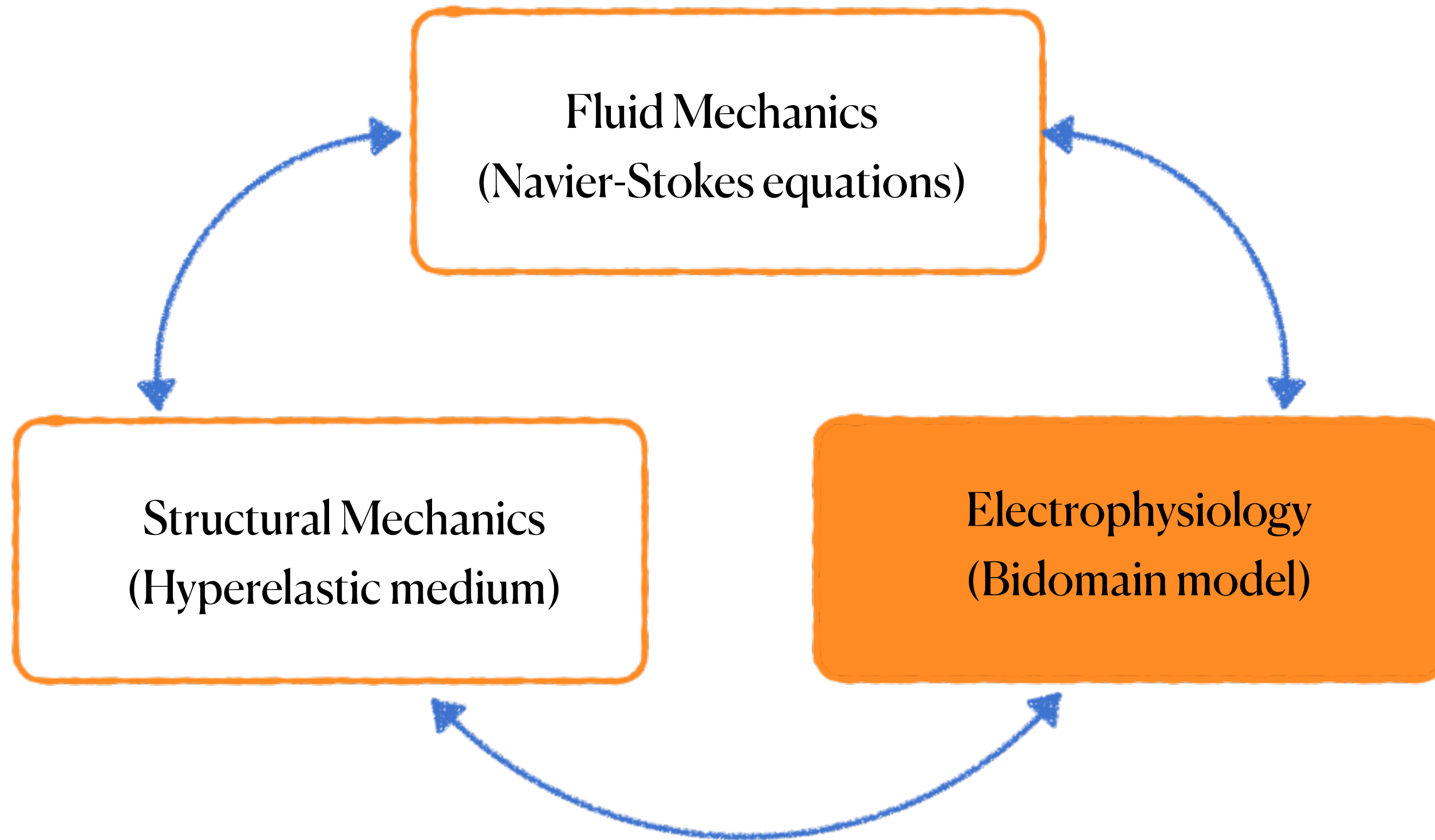
GPU-accelerated Fluid-Structure-Electrophysiology interaction modelling

Bidomain equations

$$C_m \frac{\partial V}{\partial t} - \nabla \cdot (M_i \nabla V + M_i \nabla u) = I_{ion}(V, s) + I_s$$

$$\chi I_a = \nabla \cdot (M_i \nabla V + (M_i + M_e) \nabla u)$$

$$\frac{\partial s}{\partial t} = F(V, s)$$



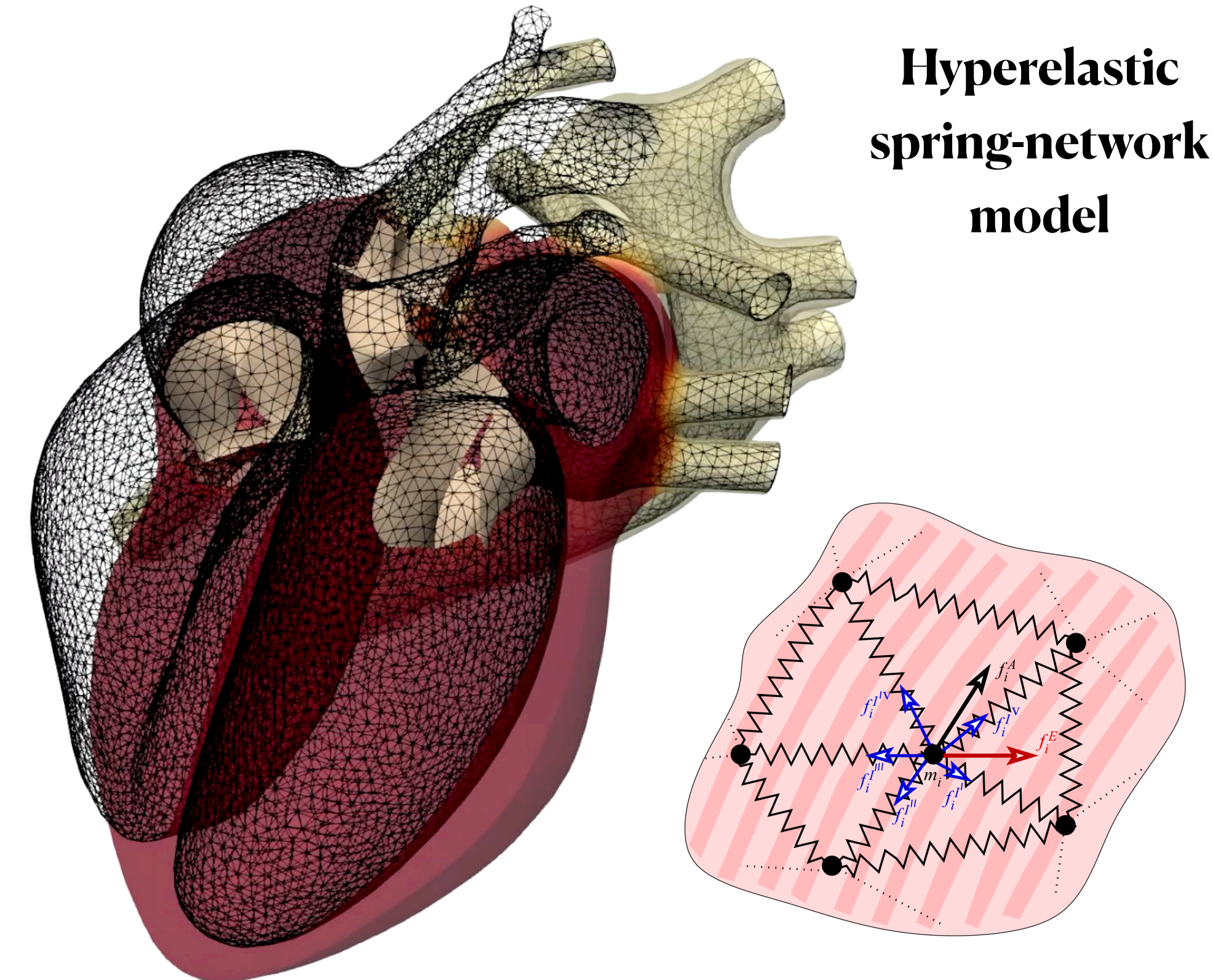
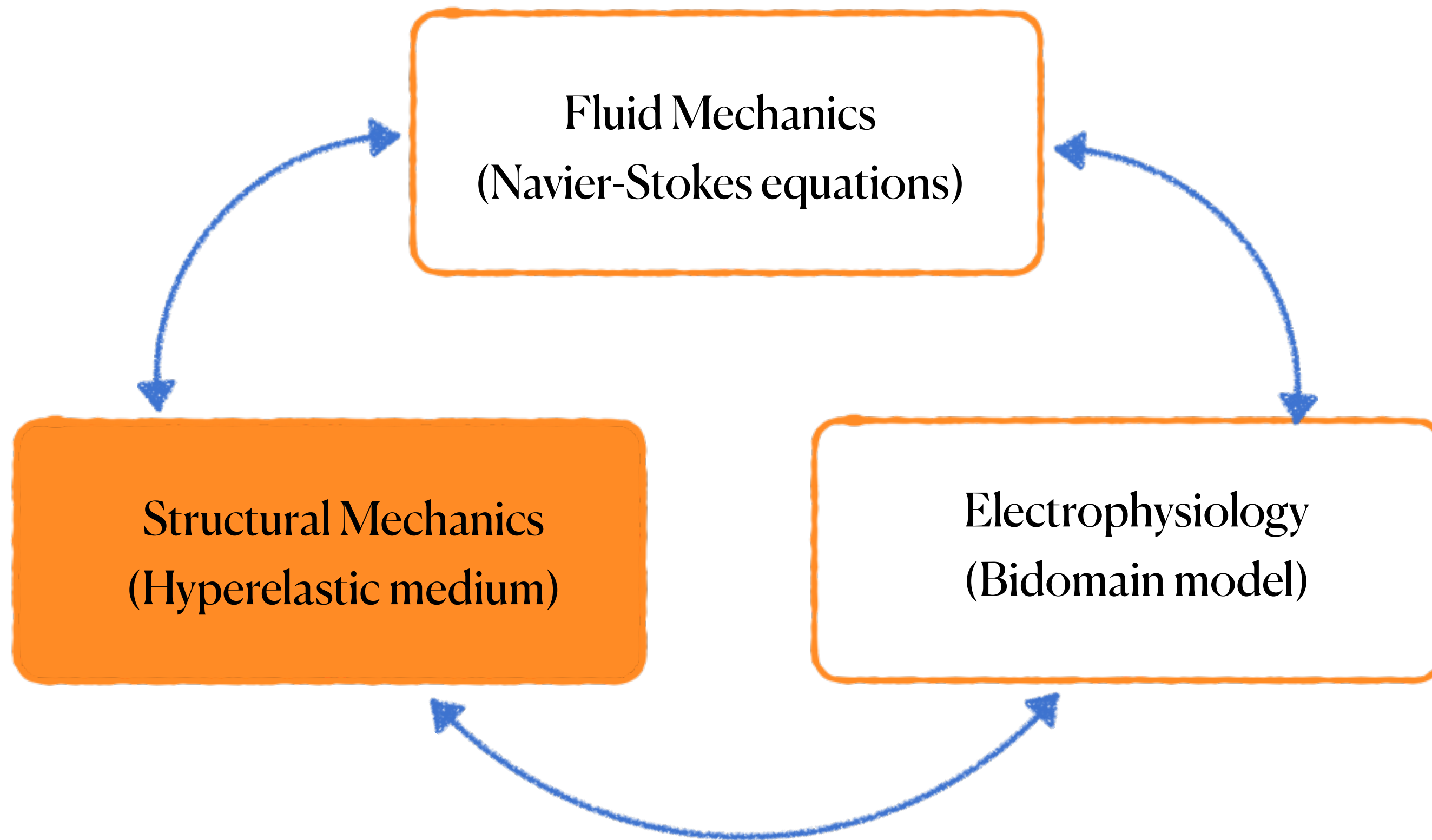
L. Tung, *Abi-domain model for describing ischemic myocardial dc potentials*, Ph.D. thesis, Massachusetts Institute of Technology, 1978.

F. V. G. Del Corso and R. Verzicco *High-fidelity model of the human heart: An immersed boundary implementation*, PRF 2023

GPU-accelerated Fluid-Structure-Electrophysiology interaction modelling

Hyperelastic, orthotropic material $\rightarrow$ Fung type constitutive model for the strain energy density

$$W = \frac{c}{2}(e^Q - 1), \quad Q = \alpha_f \epsilon_{ff}^2 + \alpha_s \epsilon_{ss}^2 + \alpha_n \epsilon_{nn}^2$$

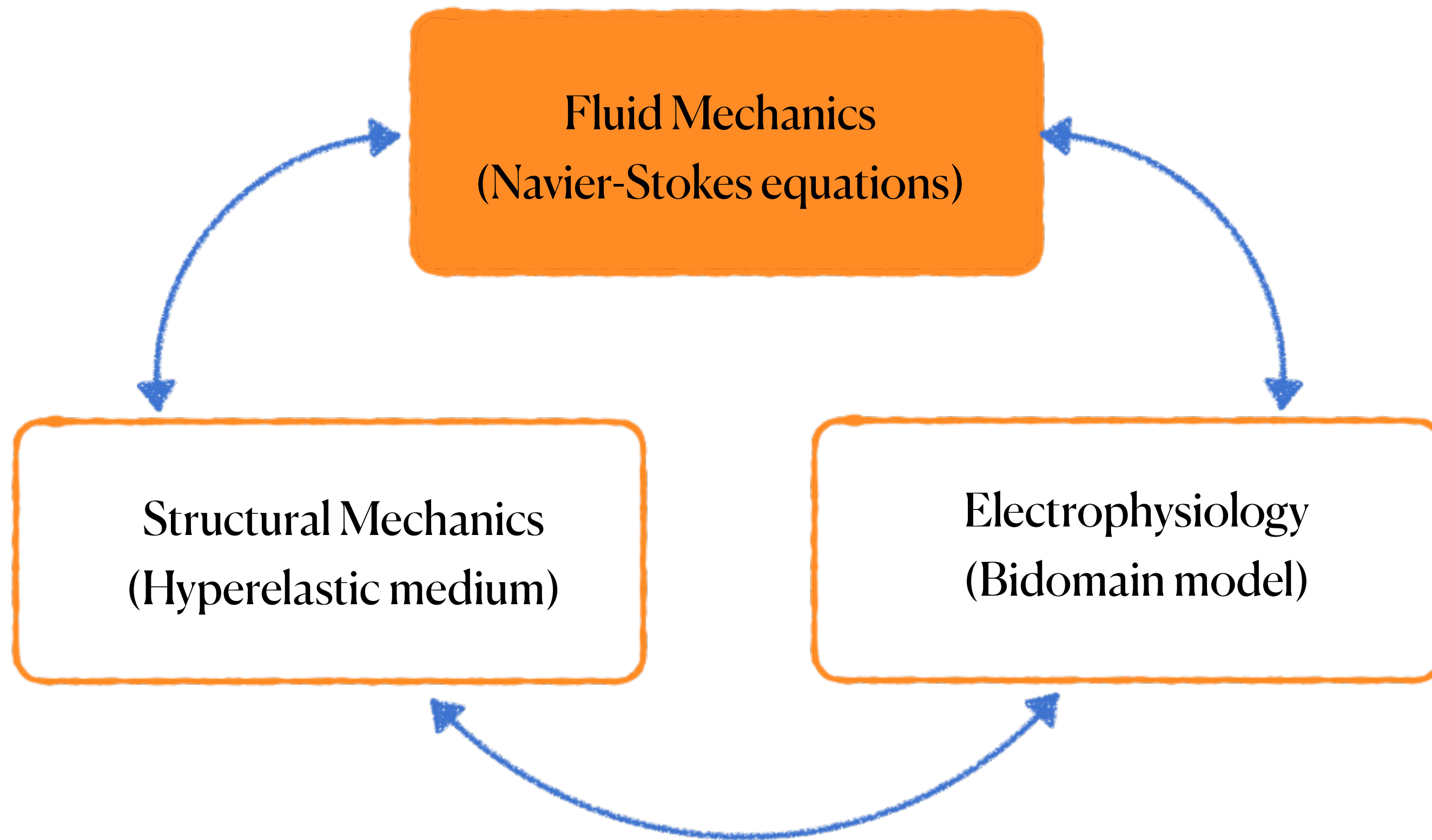


Discrete network of concentrated masses
connected by (hyper)-elastic elements

P.E. Hammer, et al. *Mass-spring model for simulation of heart valve tissue mechanical behavior*, Ann. Biomed. Eng. 2011

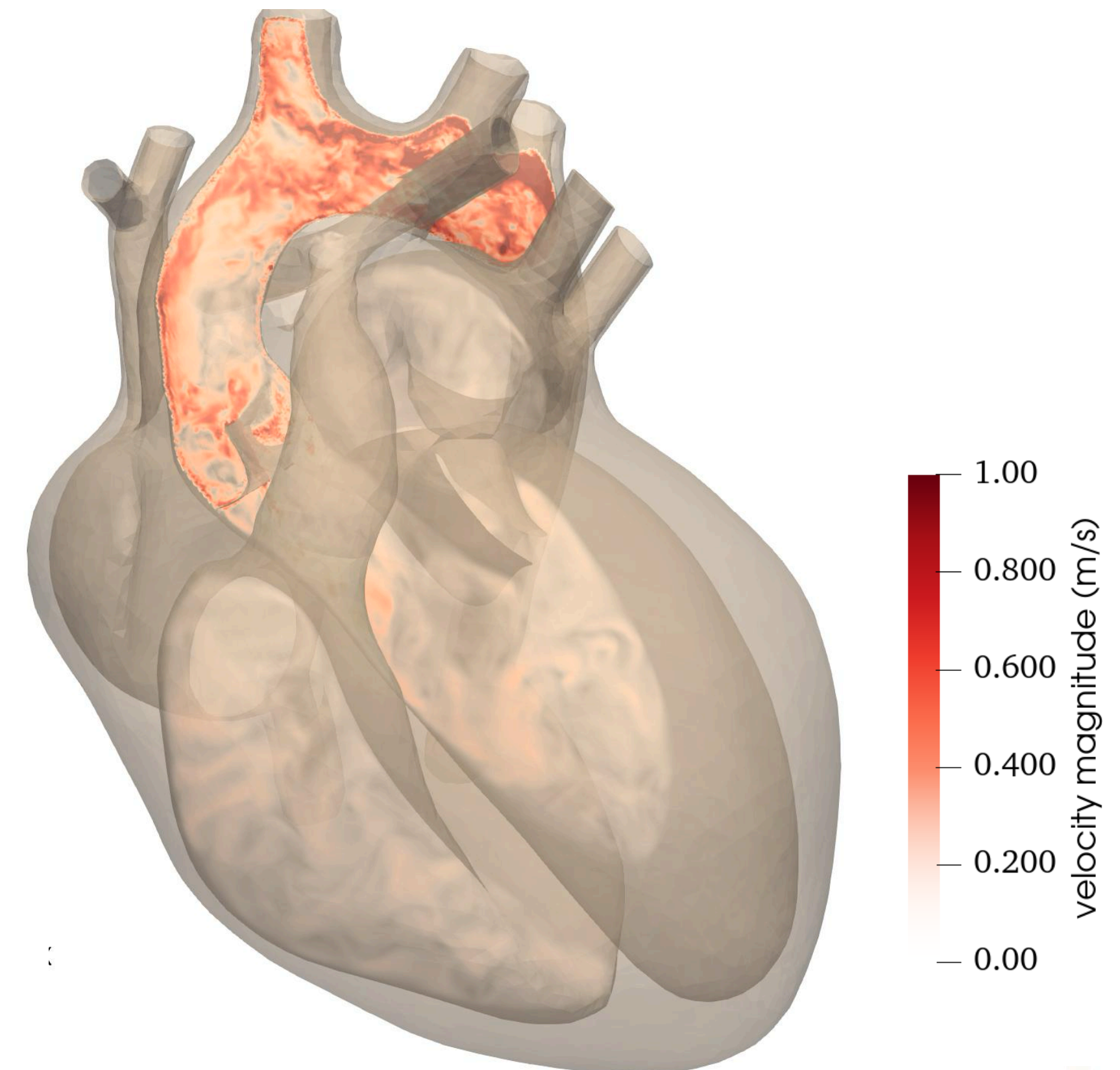
F. V. G. Del Corso and R. Verzicco *High-fidelity model of the human heart: An immersed boundary implementation*, PRF 2023

GPU-accelerated Fluid-Structure-Electrophysiology interaction modelling



Navier-Stokes equations (DNS)

$$\frac{\partial \mathbf{u}}{\partial t} + \nabla \cdot (\mathbf{u}\mathbf{u}) = -\nabla p + \nabla \cdot \boldsymbol{\tau} + \mathbf{f}, \quad \nabla \cdot \mathbf{u} = 0,$$



M. Vanella & E. Balaras *A moving-least-squares reconstruction for embedded-boundary formulations*, JCP 2009

F. V. G. Del Corso and R. Verzicco *High-fidelity model of the human heart: An immersed boundary implementation*, PRF 2023

APPLICATION 1

Optimal cardiac resynchronization therapy: an in-silico investigation using a whole-heart model

www.nature.com/scientificreports

scientific reports



OPEN

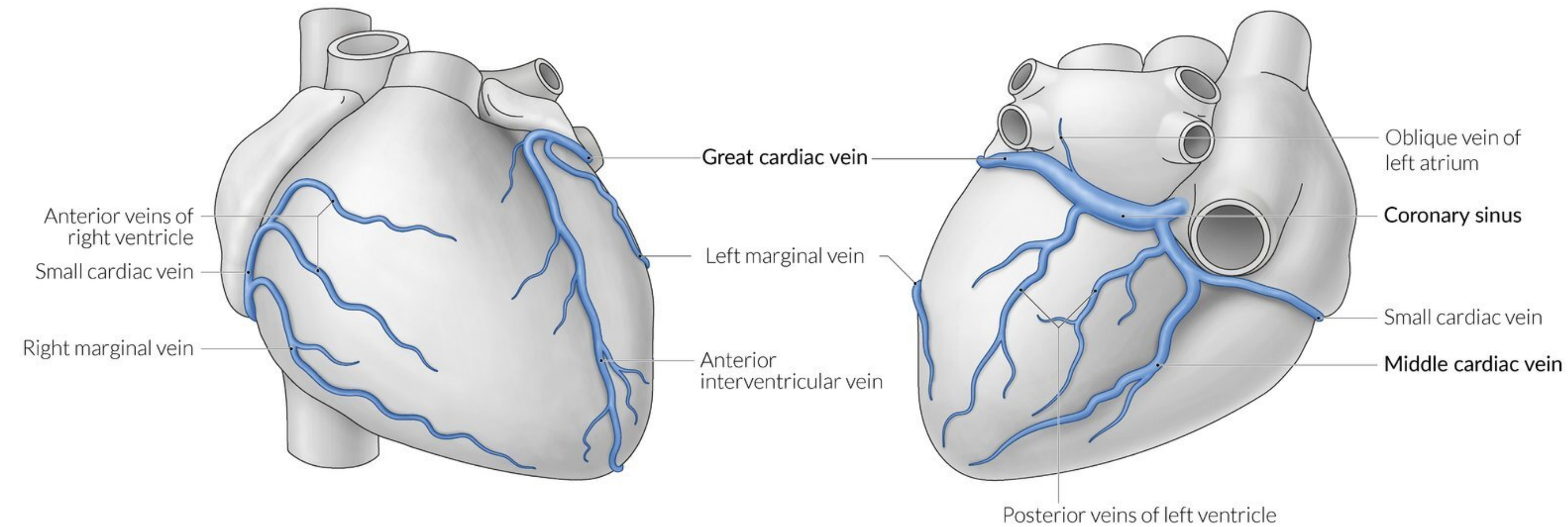
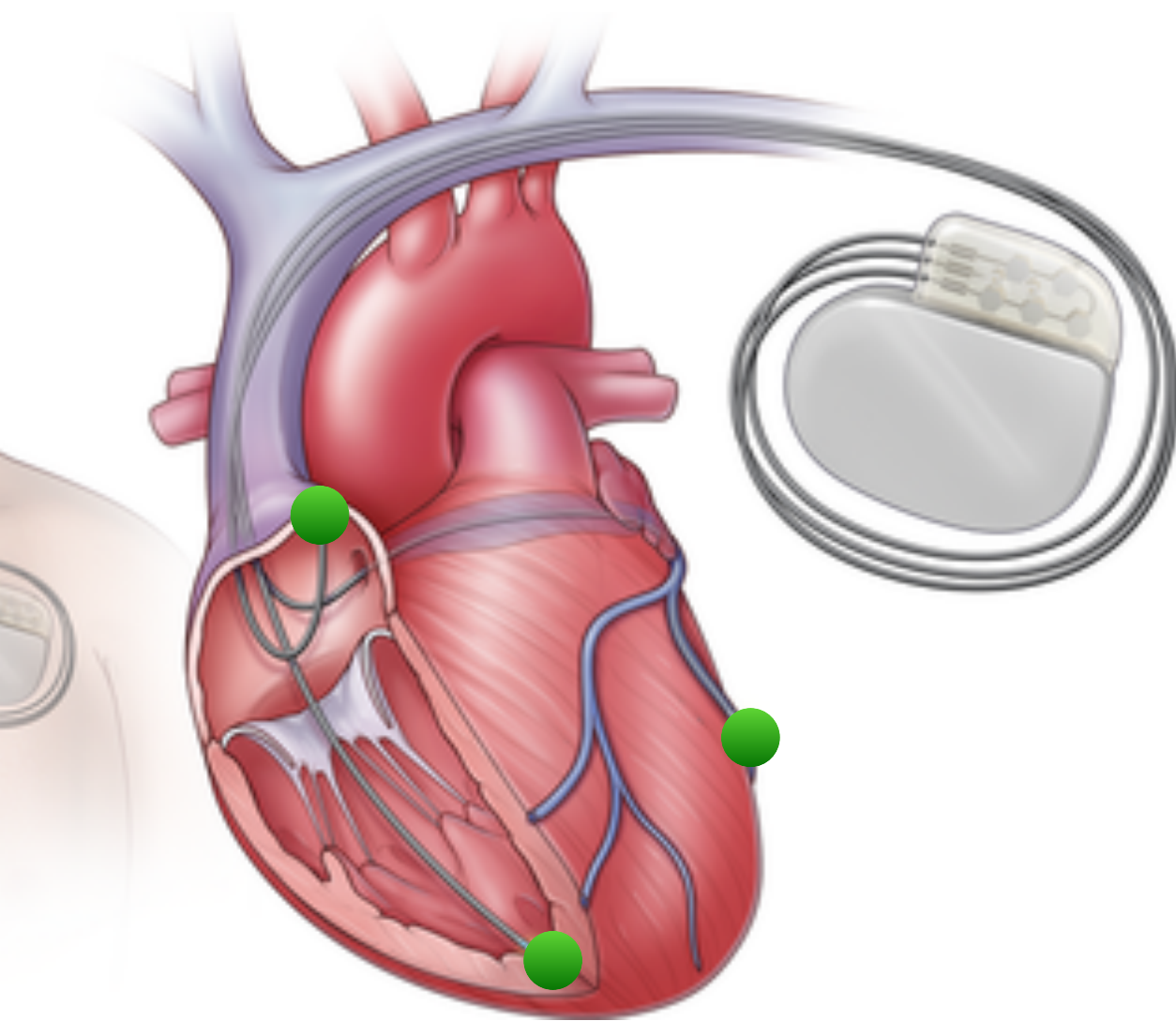
GPU accelerated digital twins of the human heart open new routes for cardiovascular research

Francesco Viola^{1,7}, Giulio Del Corso^{1,6}, Ruggero De Paulis^{2,3} & Roberto Verzicco^{1,4,5}✉

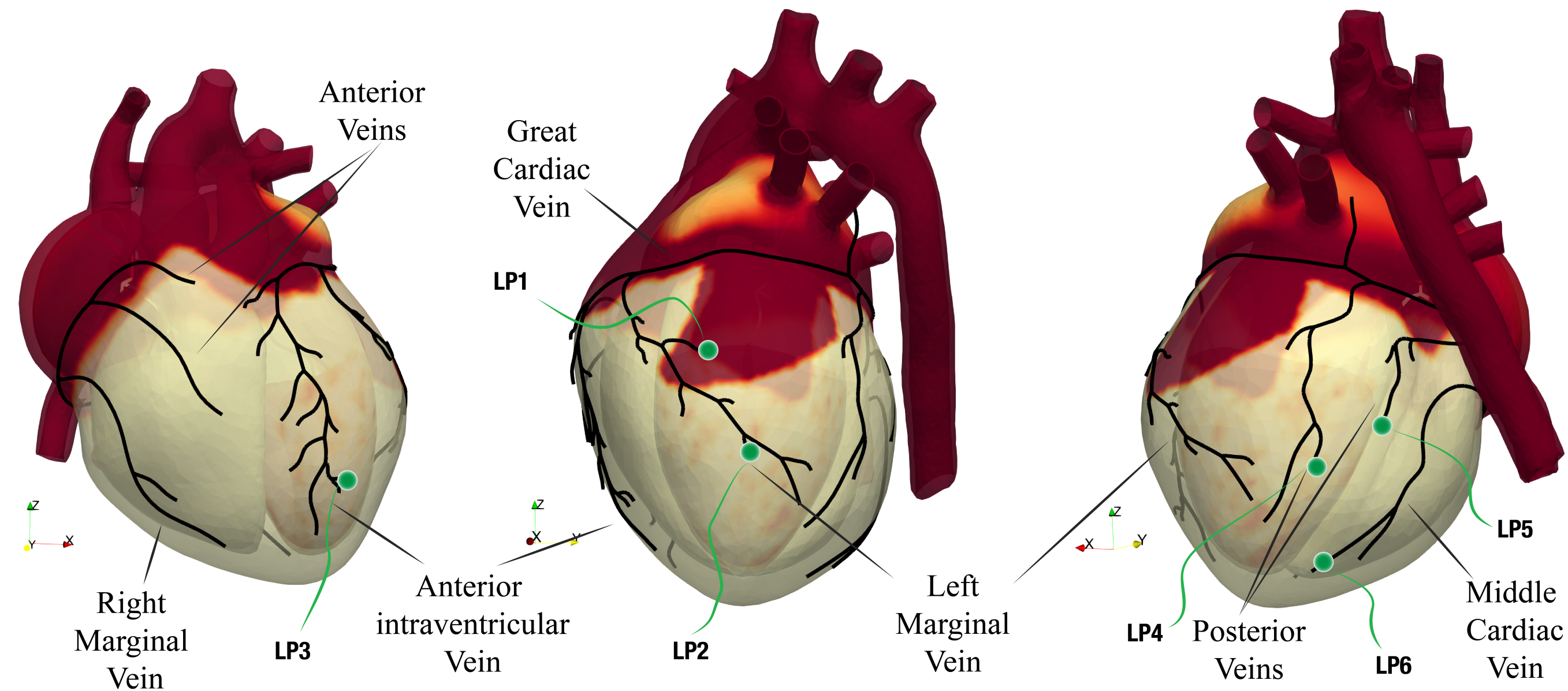
Background and motivation

What is the optimal positioning of the LV lead in CRT?

► The LV lead has to be implanted thorough the coronary veins network

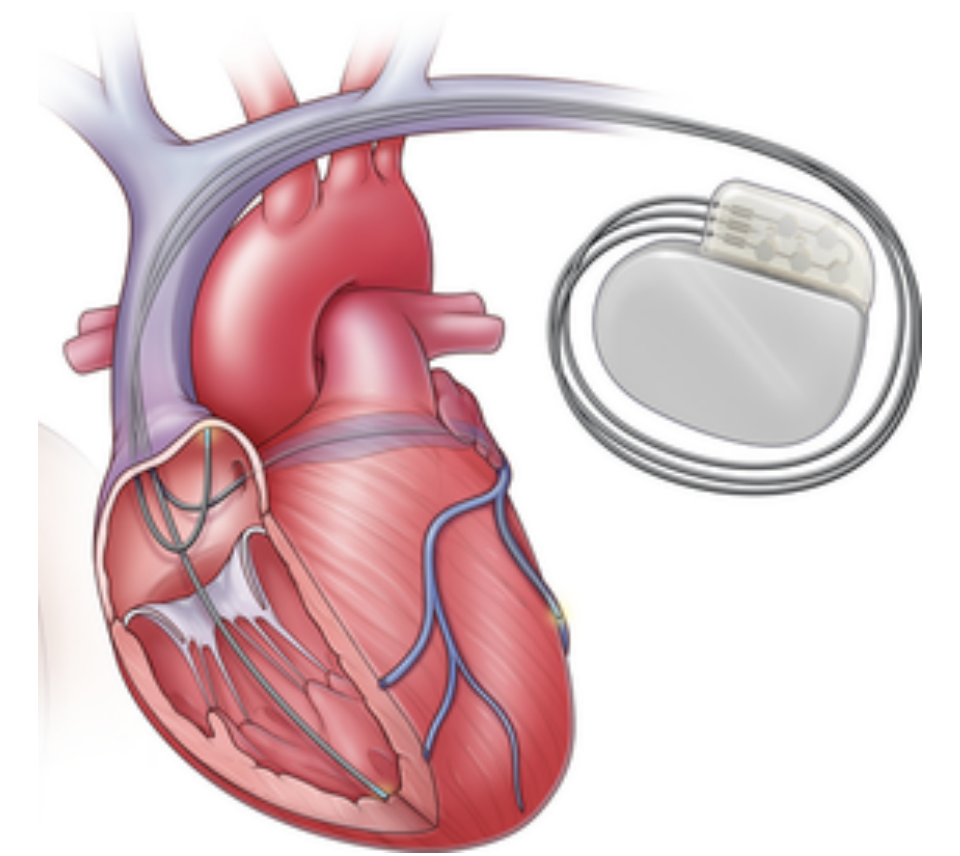


► Optimal CRT positioning: numerical investigation of single-site stimulation

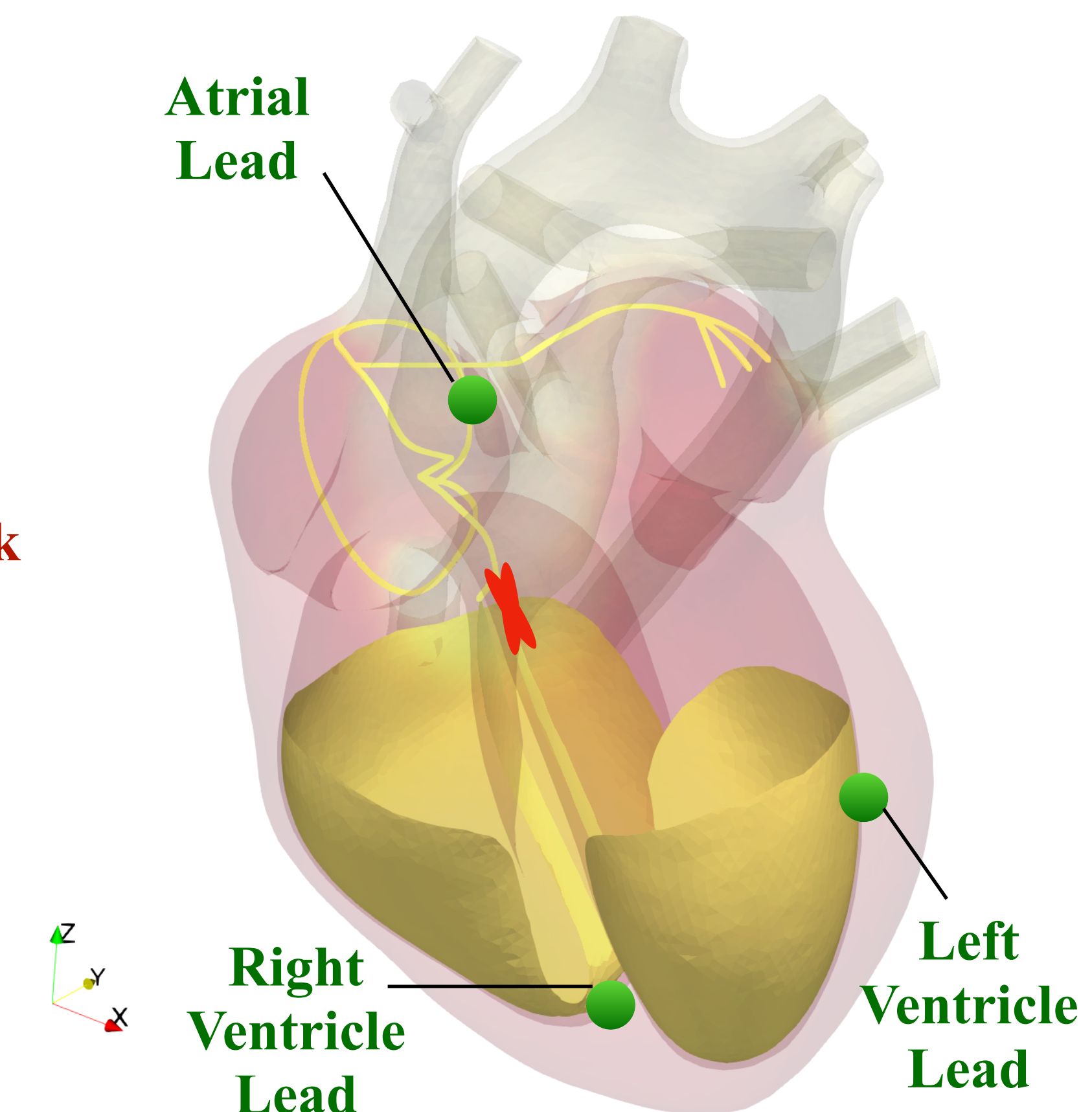
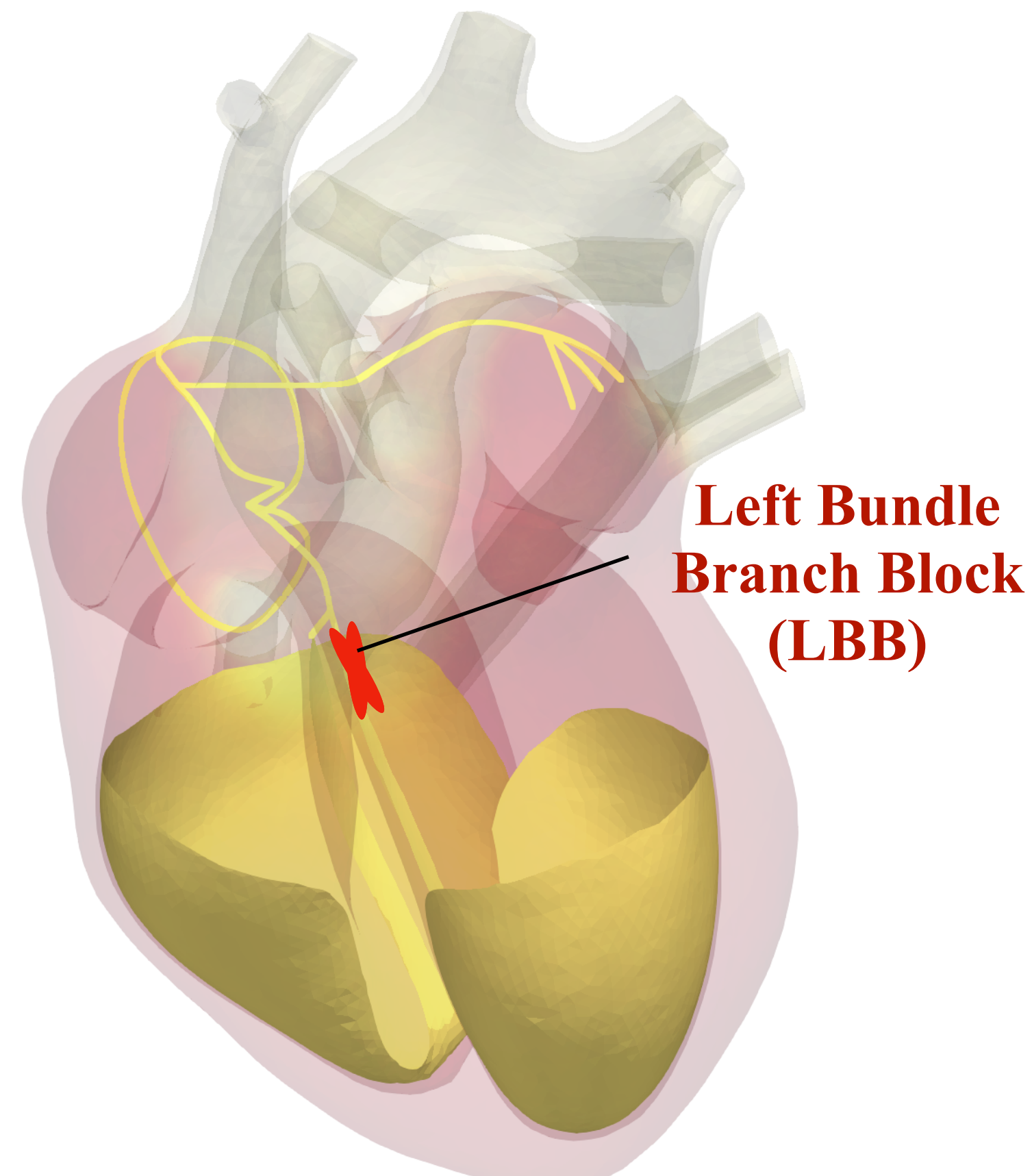
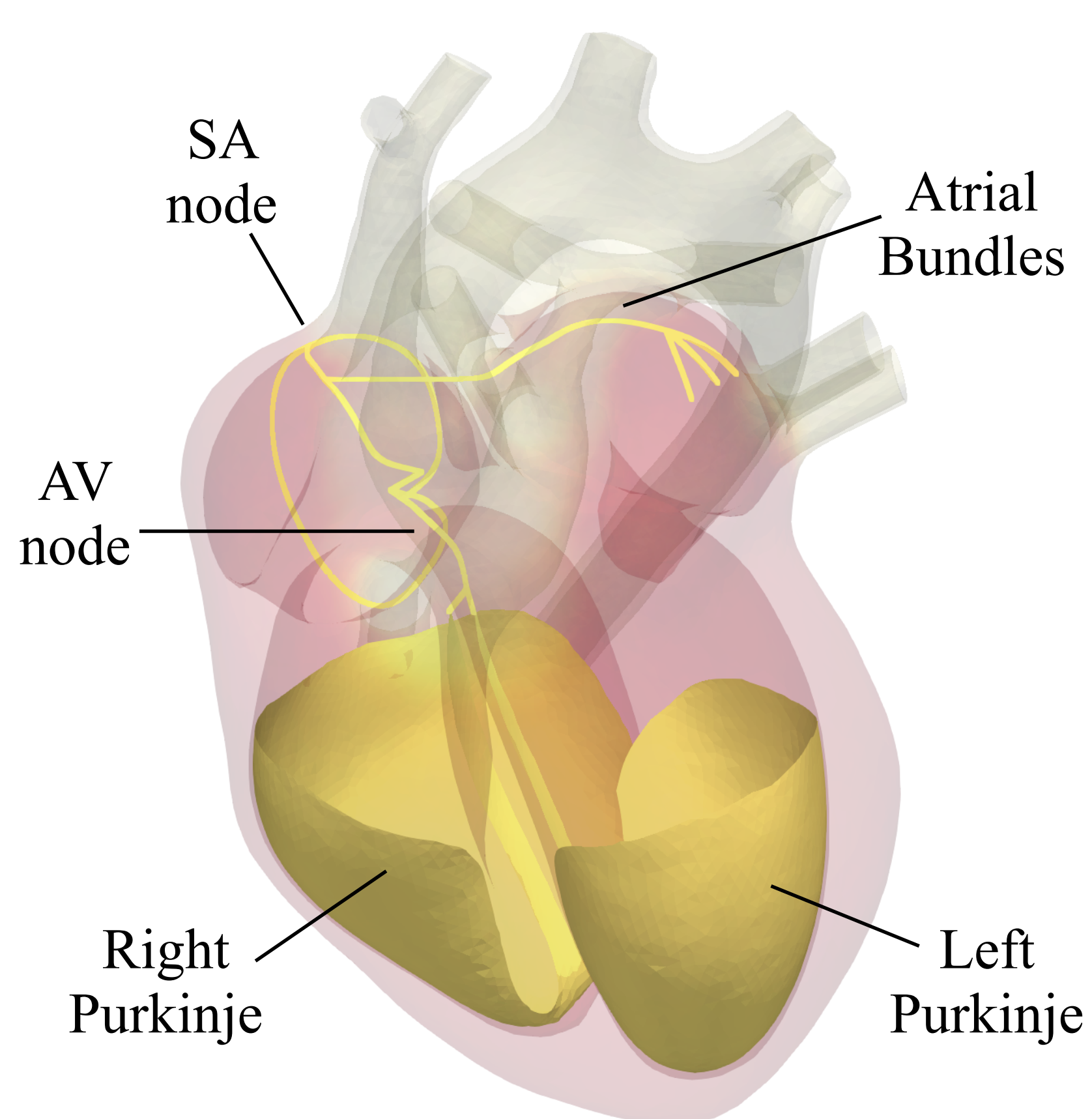


Numerical modelling of CRT

- ▶ In-silico implantation of CRT after a left bundle branch block
- ▶ Electrical disconnection of the left Purkinje network: delayed LV contraction
- ▶ LV lead in the region of last electrophysiology activation



HEALTHY **PATHOLOGY** → LEFT BUNDLE BRANCH BLOCK (LBB) **THERAPY** → RESYNCHRONIZED



Electrophysiology

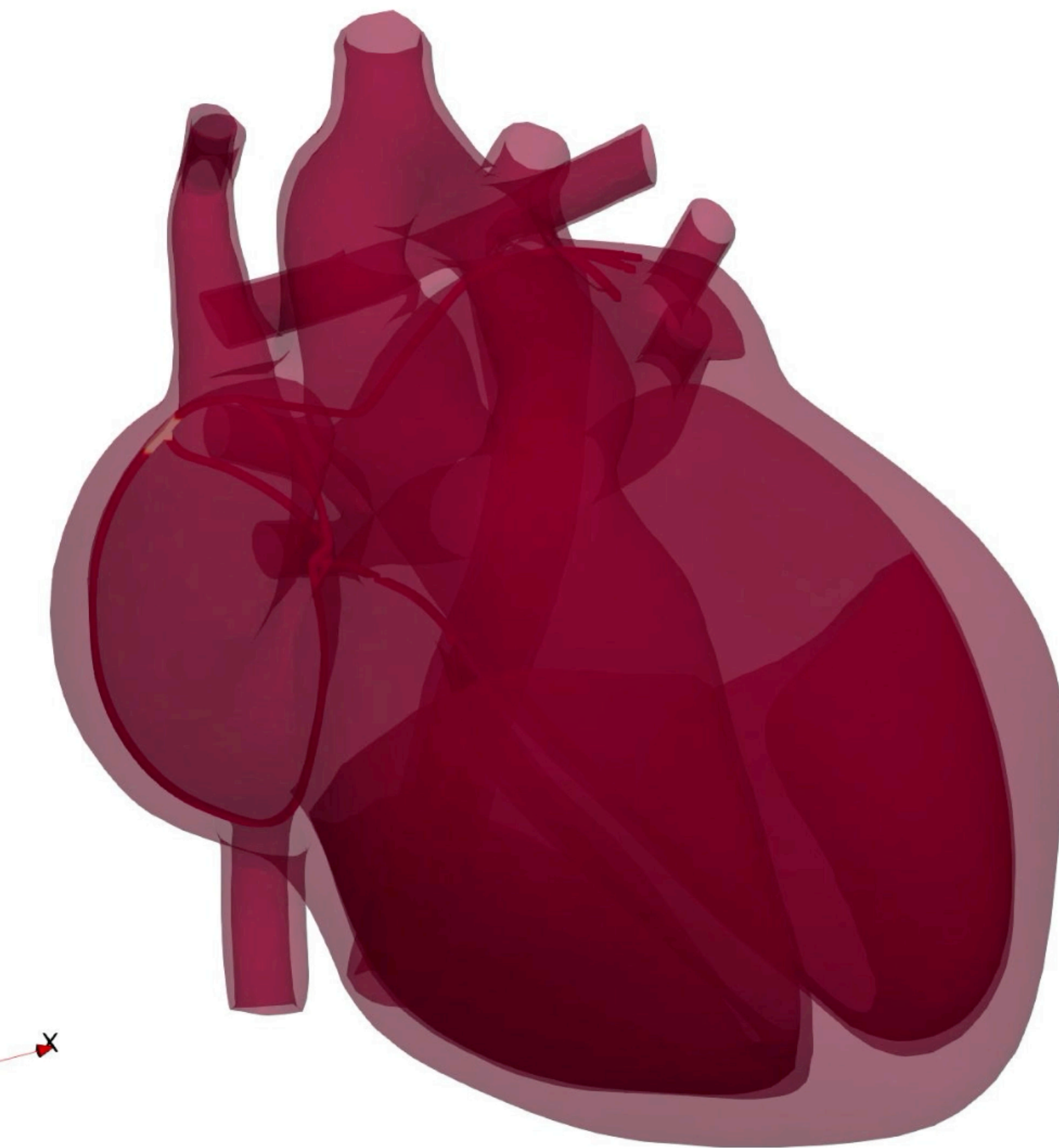
HEALTHY

PATHOLOGY

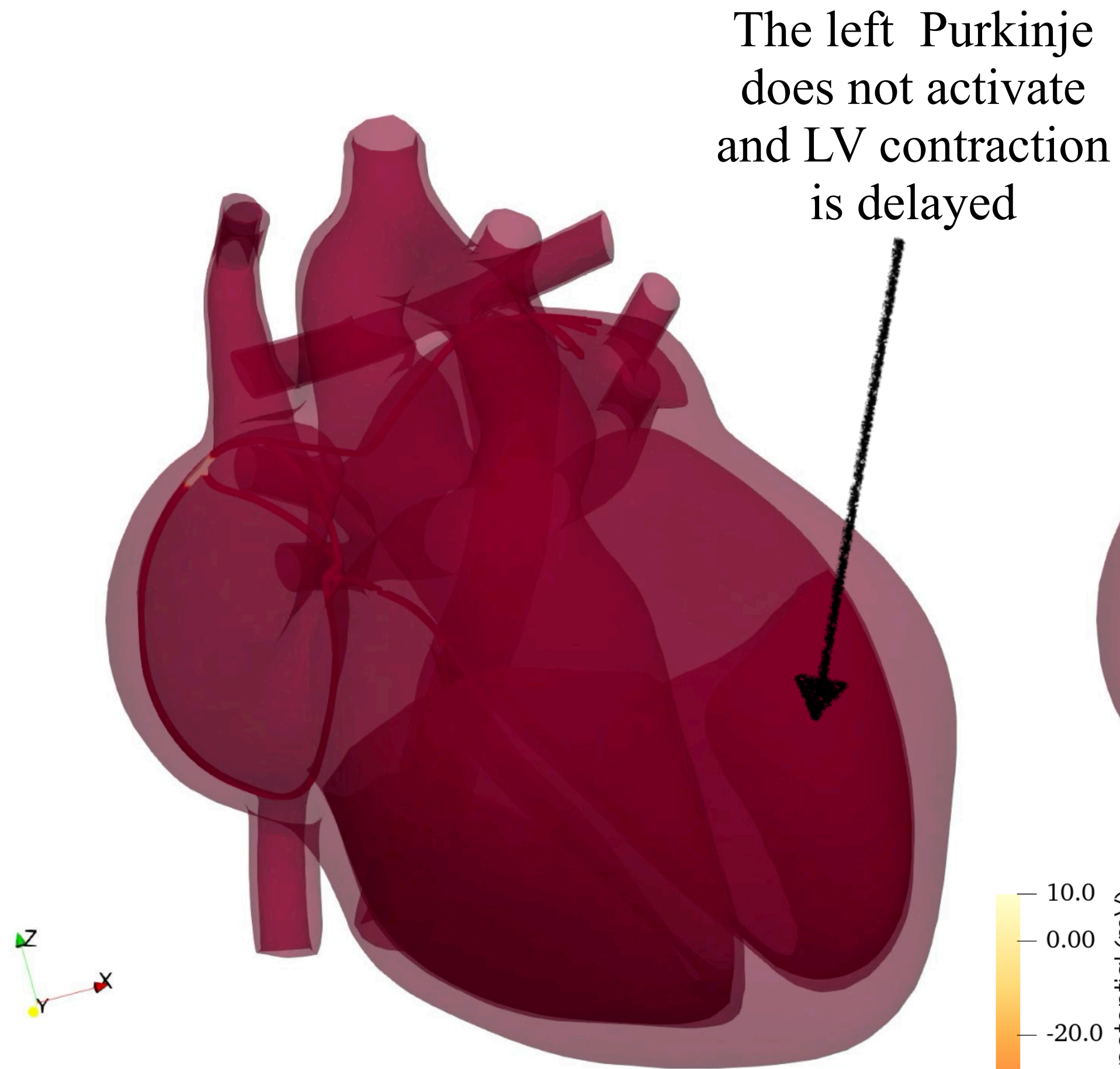
LEFT BUNDLE BRANCH BLOCK (LBB)

THERAPY

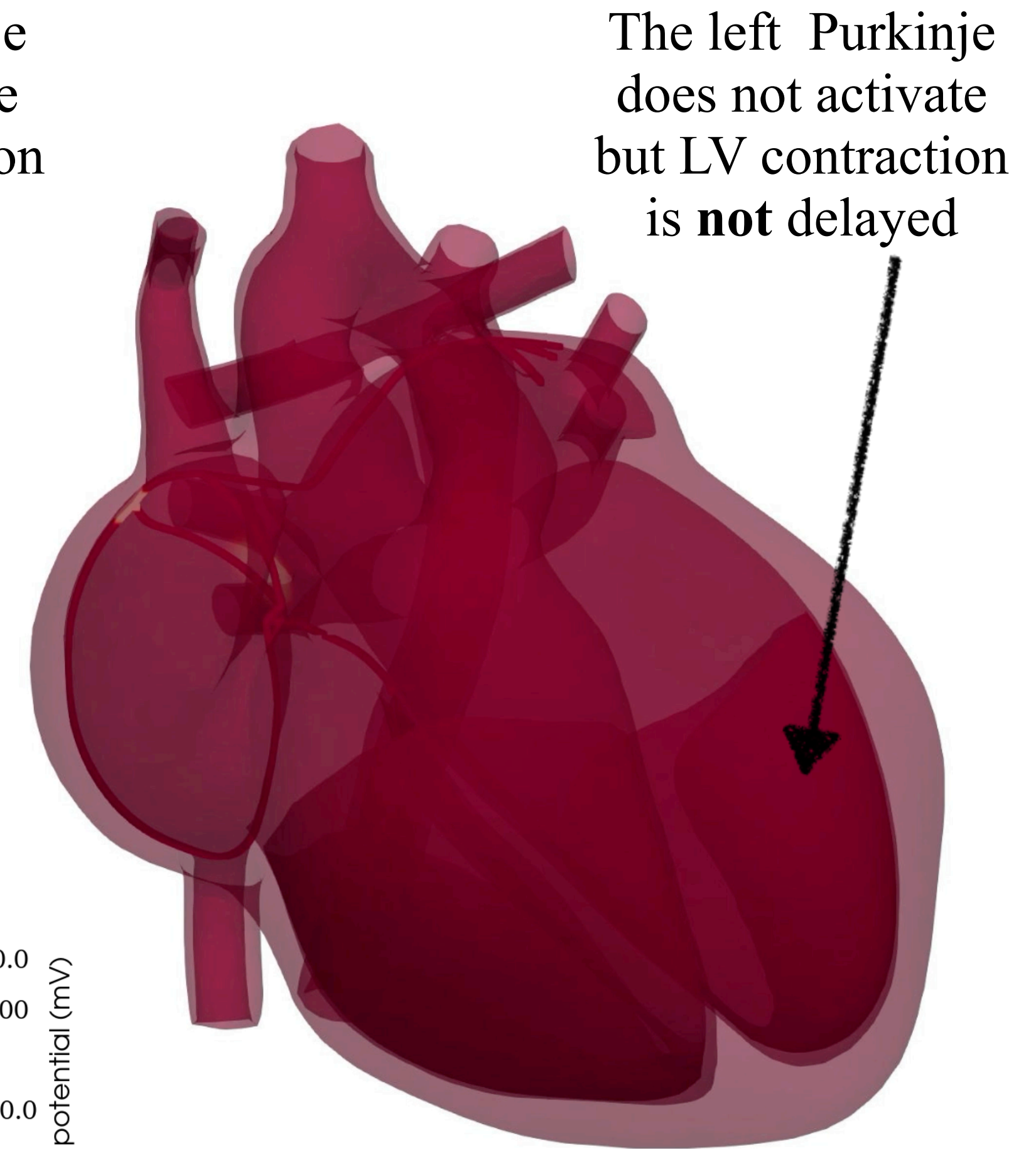
RESYNCHRONIZED



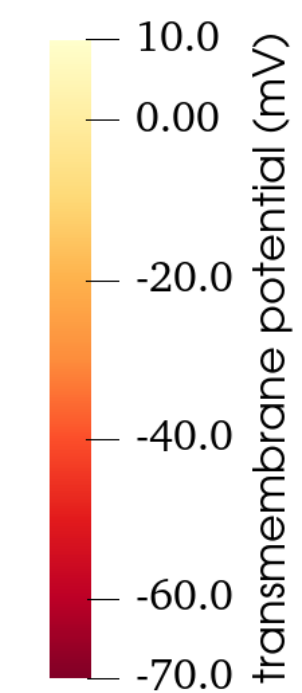
time/10



time/10

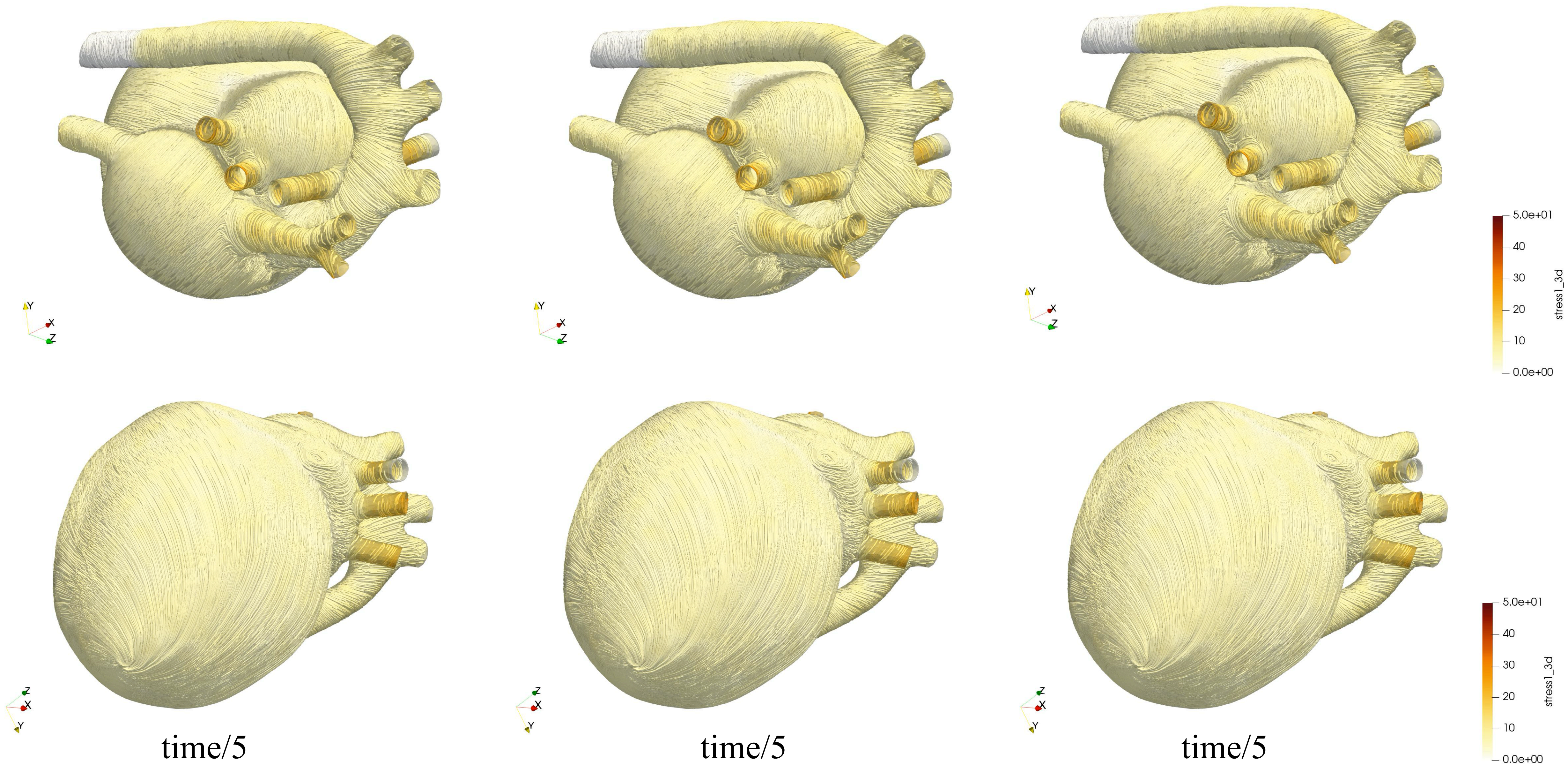


time/10



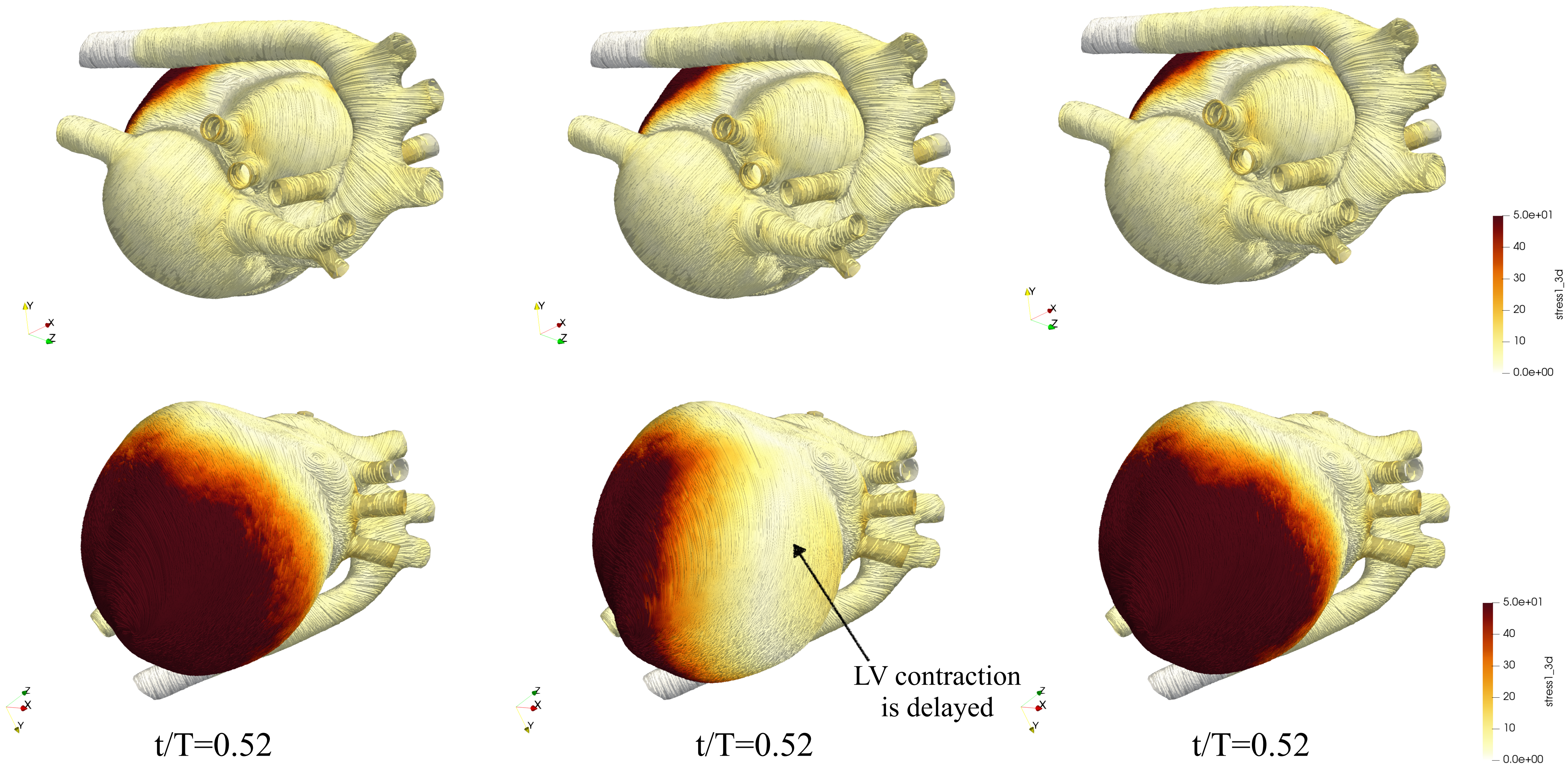
Internal tissue stresses

HEALTHY **PATHOLOGY** LEFT BUNDLE BRANCH BLOCK (LBB) **THERAPY** RESYNCHRONIZED

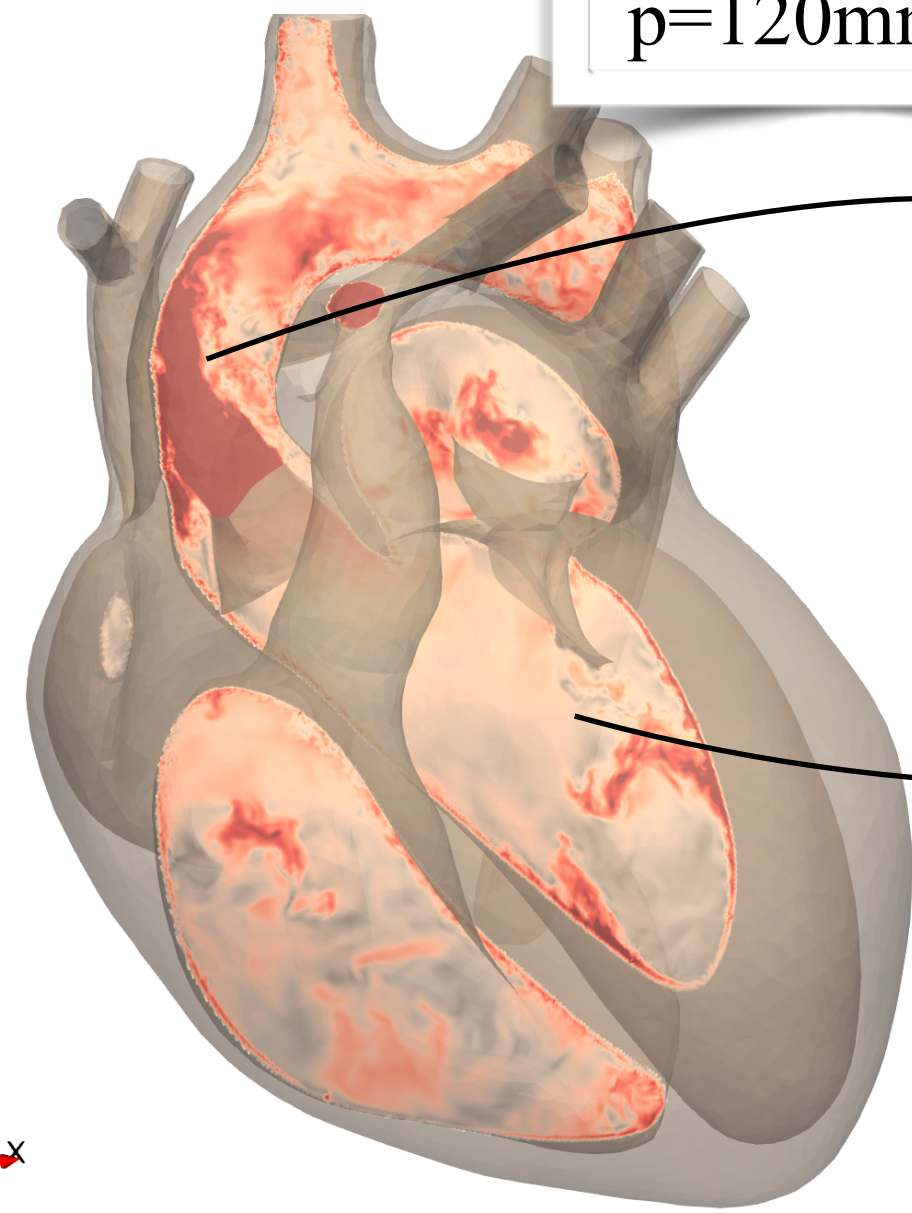


Internal tissue stresses

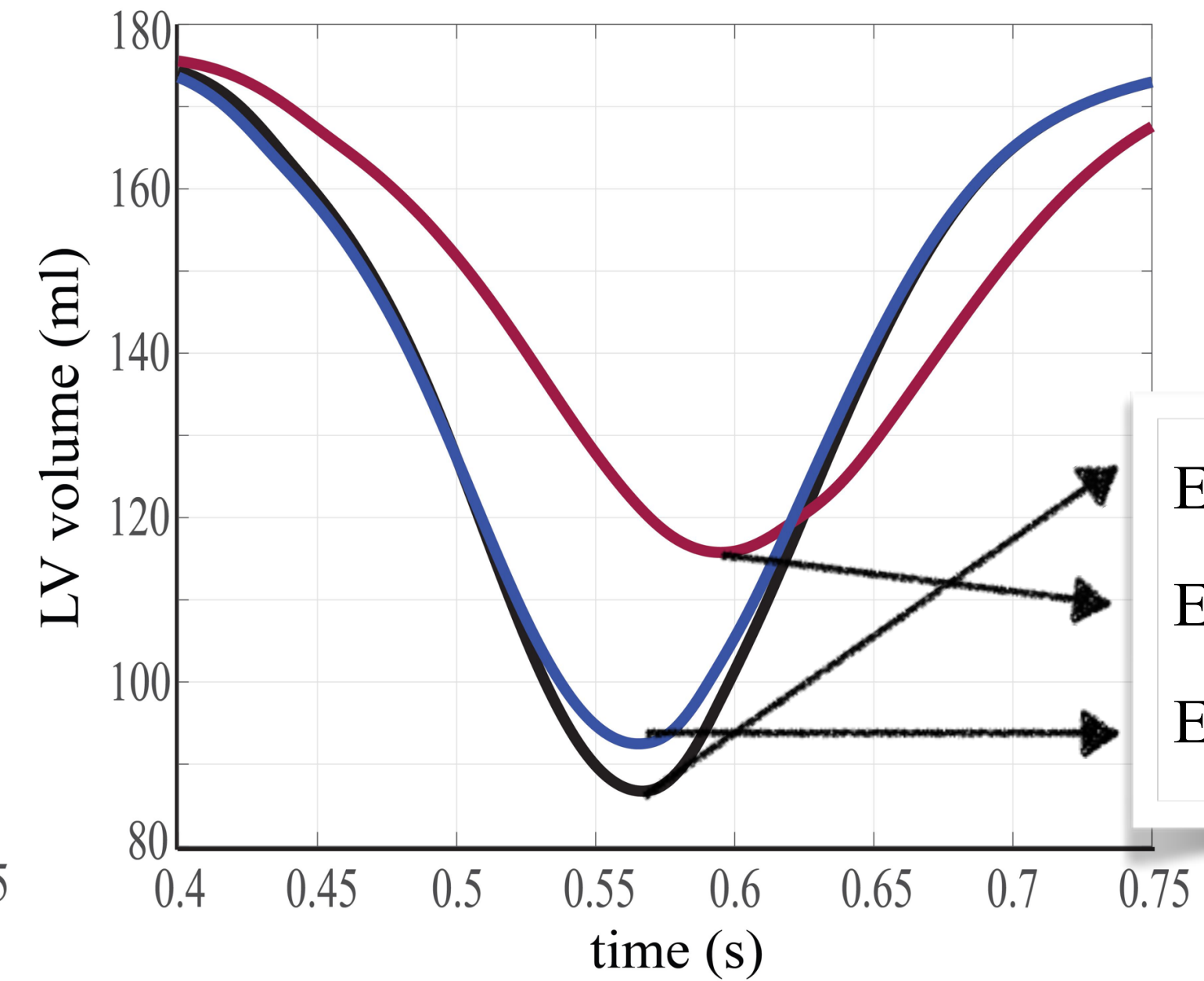
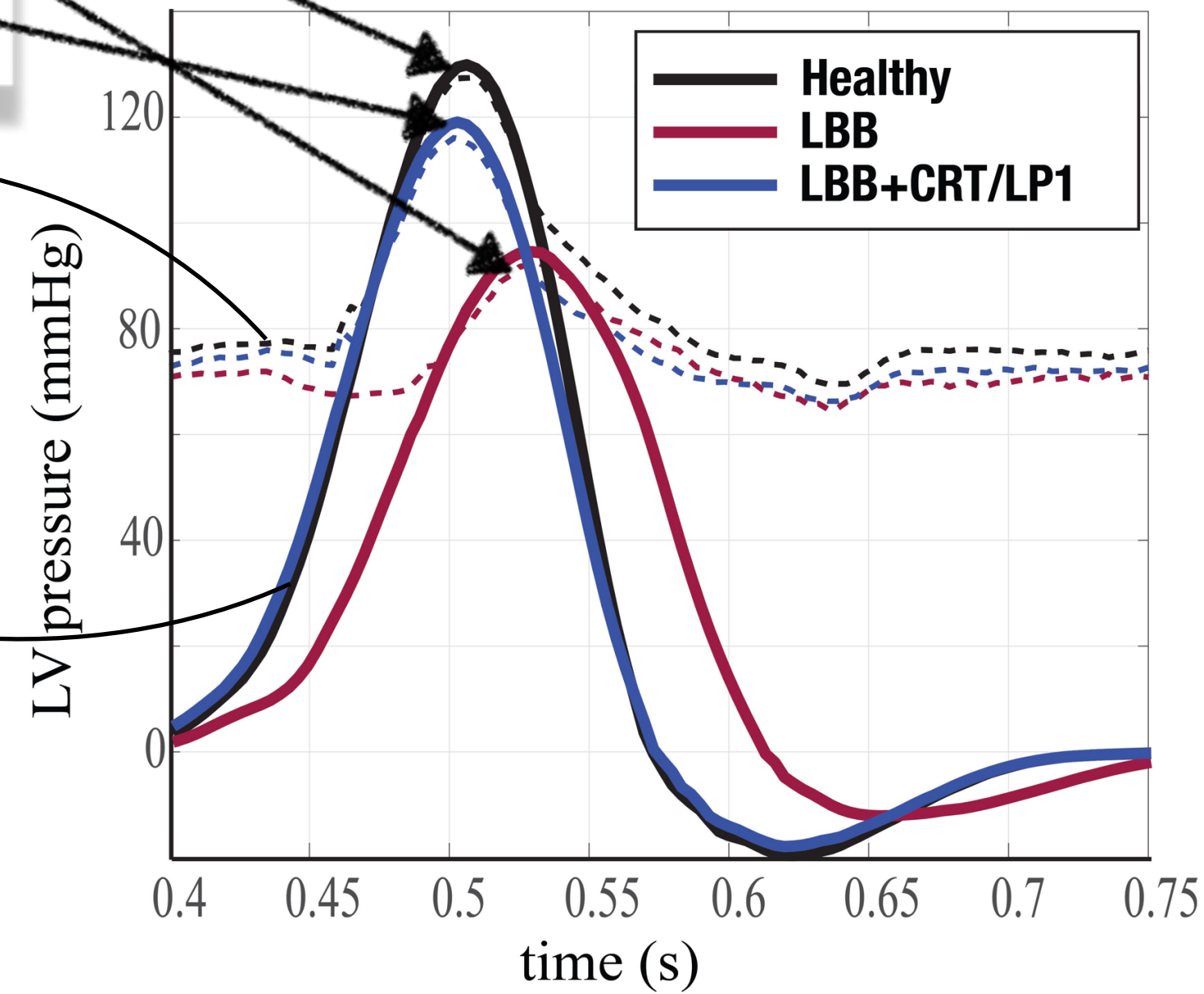
HEALTHY **PATHOLOGY** LEFT BUNDLE BRANCH BLOCK (LBB) **THERAPY** RESYNCHRONIZED



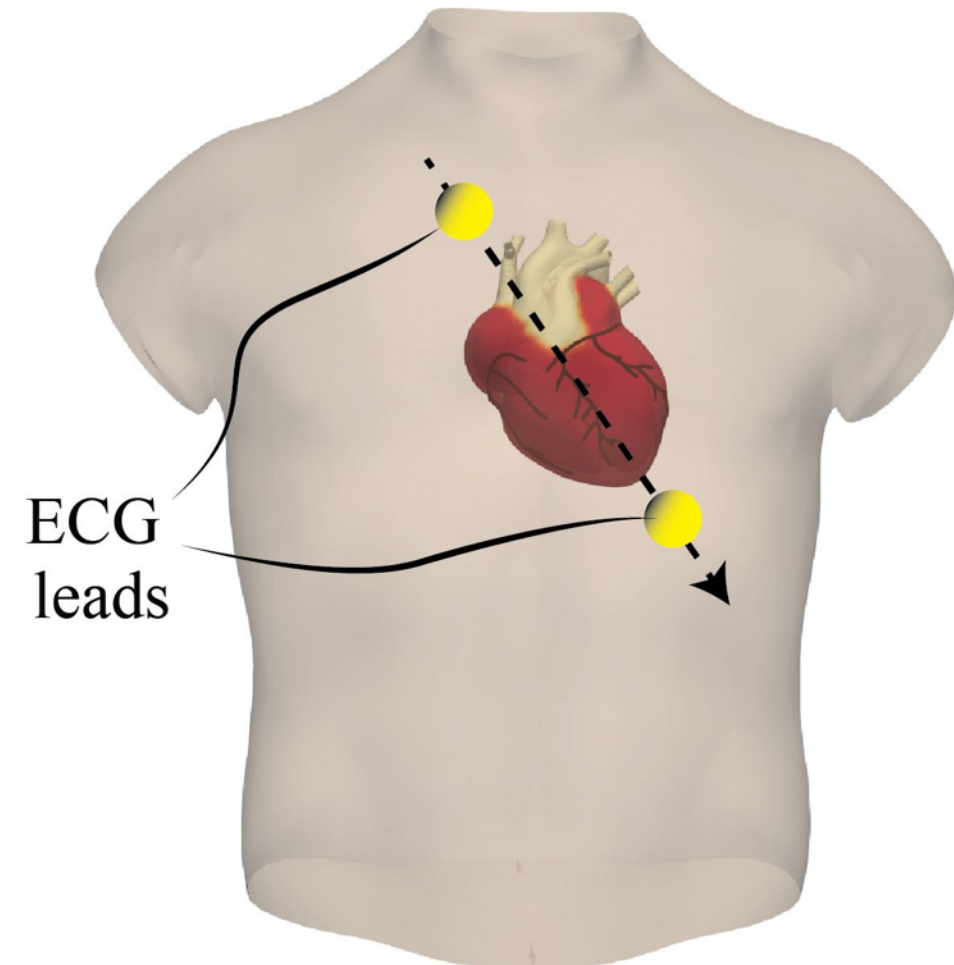
Hemodynamics



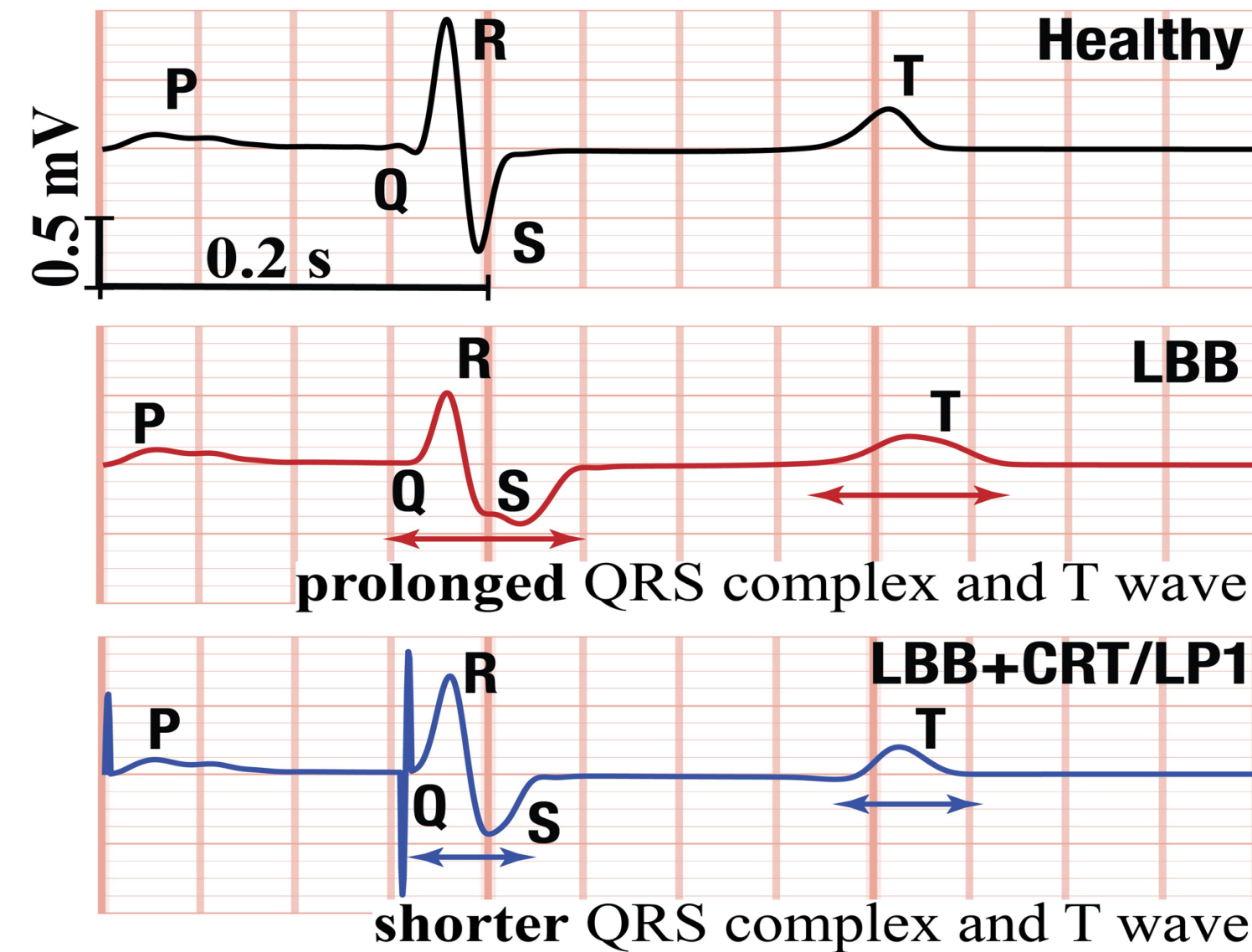
p=130mmHg
p=95mmHg
p=120mmHg



EF=51%, CO=5.4l/min
EF=34%, CO=3.6l/min
EF=48%, CO=5.0l/min



ECG leads



Ejection fraction:

$$EF = \frac{V_{max} - V_{min}}{V_{max}}$$

Cardiac output:

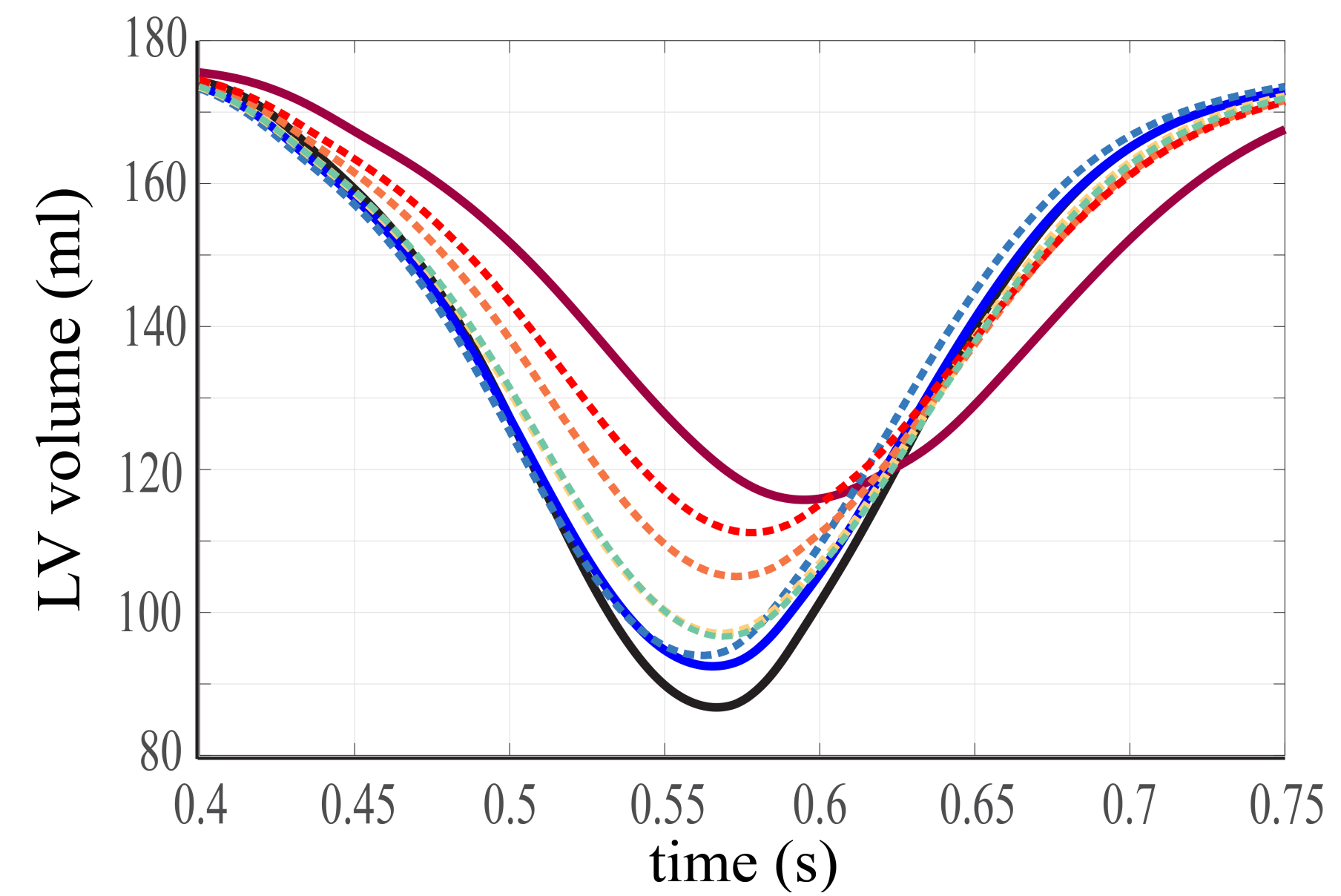
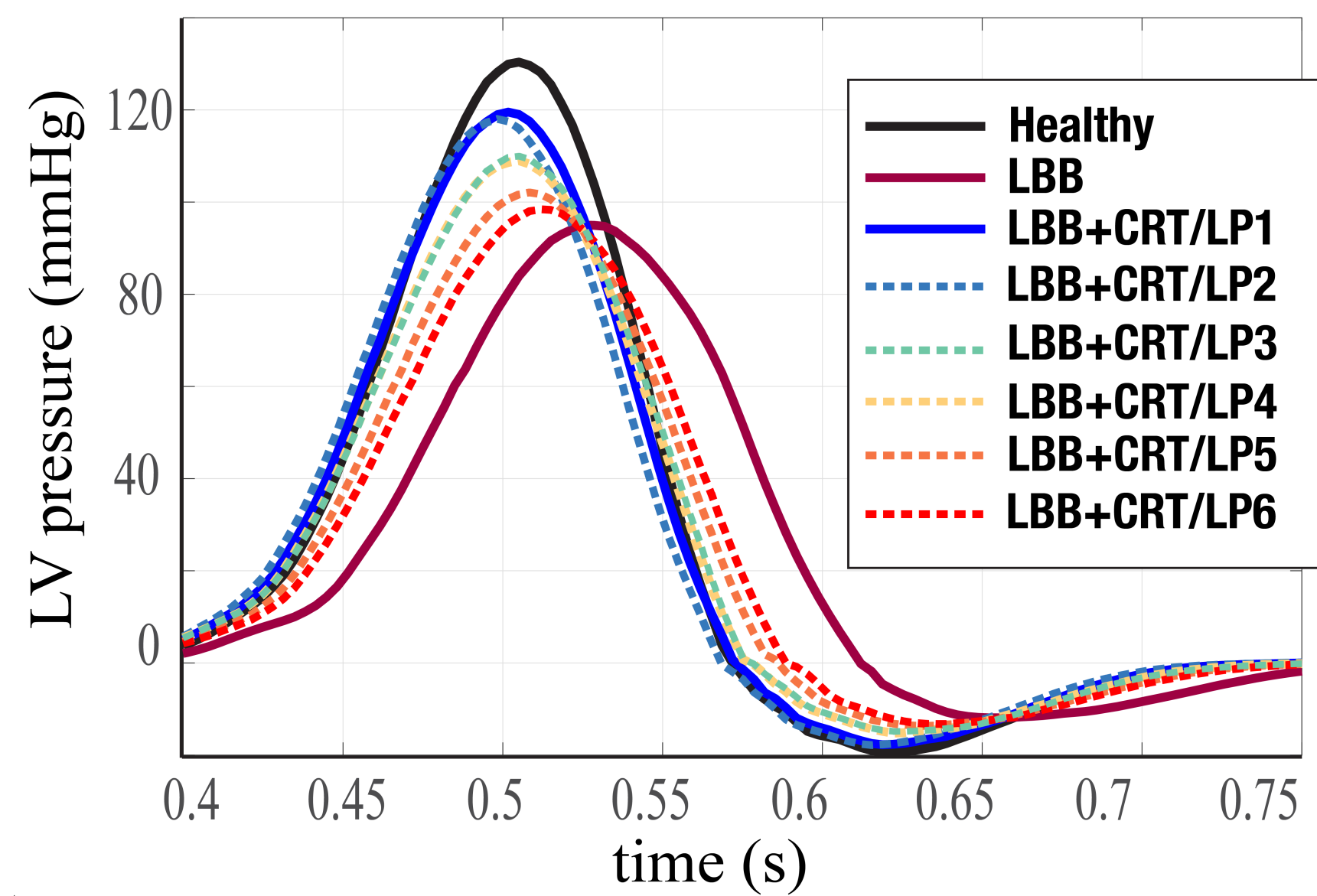
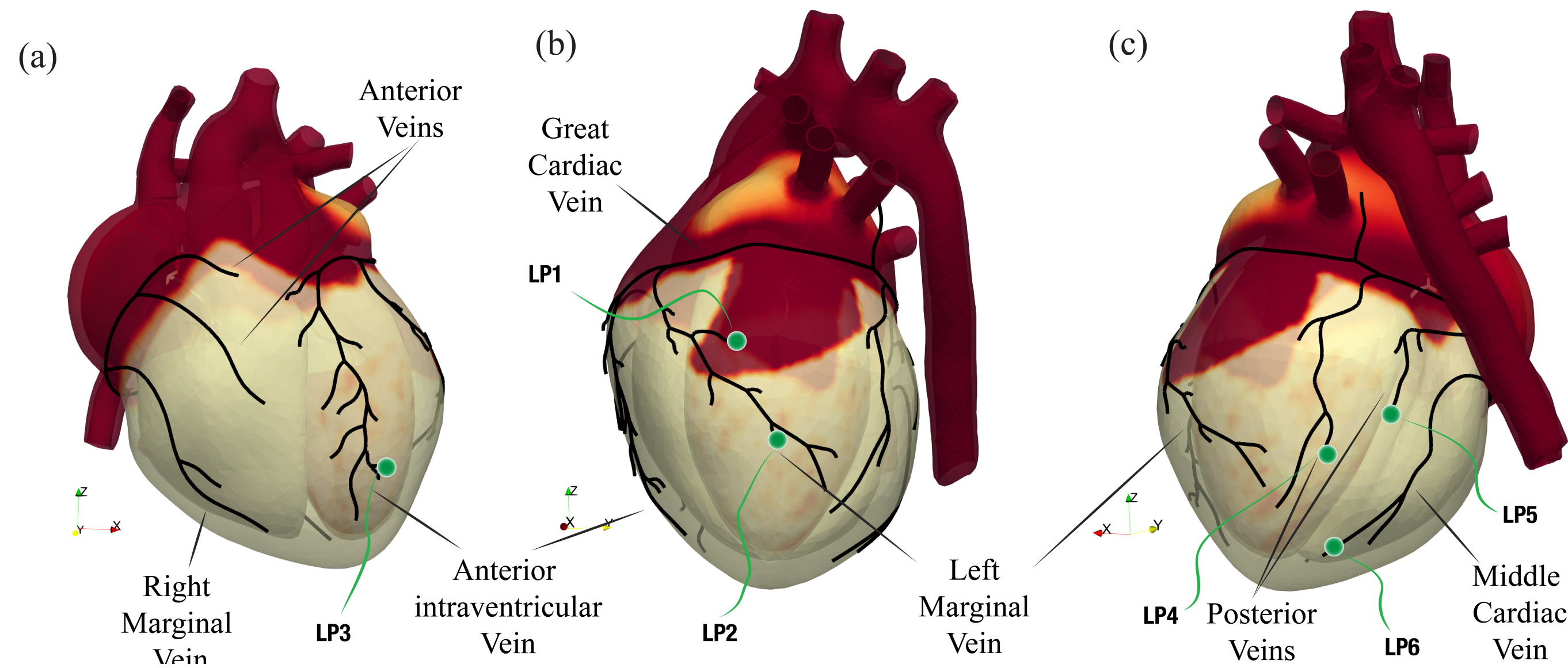
$$CO = (V_{max} - V_{min})HR$$

Heart rate

Pseudo-ECG:

$$V_s(\mathbf{x}_s, t) = -K \int_{\Omega_{heart}} \nabla v(\mathbf{x}, t) \cdot \nabla \left(\frac{1}{||\mathbf{x} - \mathbf{x}_s||} \right) d\mathbf{x},$$

A small in-silico trial...



APPLICATION 2

- **In-silico testing medical devices**

Investigation of the hemodynamics features past **bioprosthetic valves**: effects of the valve **denlovmnt location**

PHYSICAL REVIEW FLUIDS **10**, 090501 (2025)

Hemodynamic effects of intra- and supra-deployment locations for a bioprosthetic aortic valve

Martino Andrea Scarpolini ^{1,2} Giovanni Vagnoli ¹ Fabio Guglietta ^{3,2}
Roberto Verzicco ^{1,3} and Francesco Viola^{1,*}

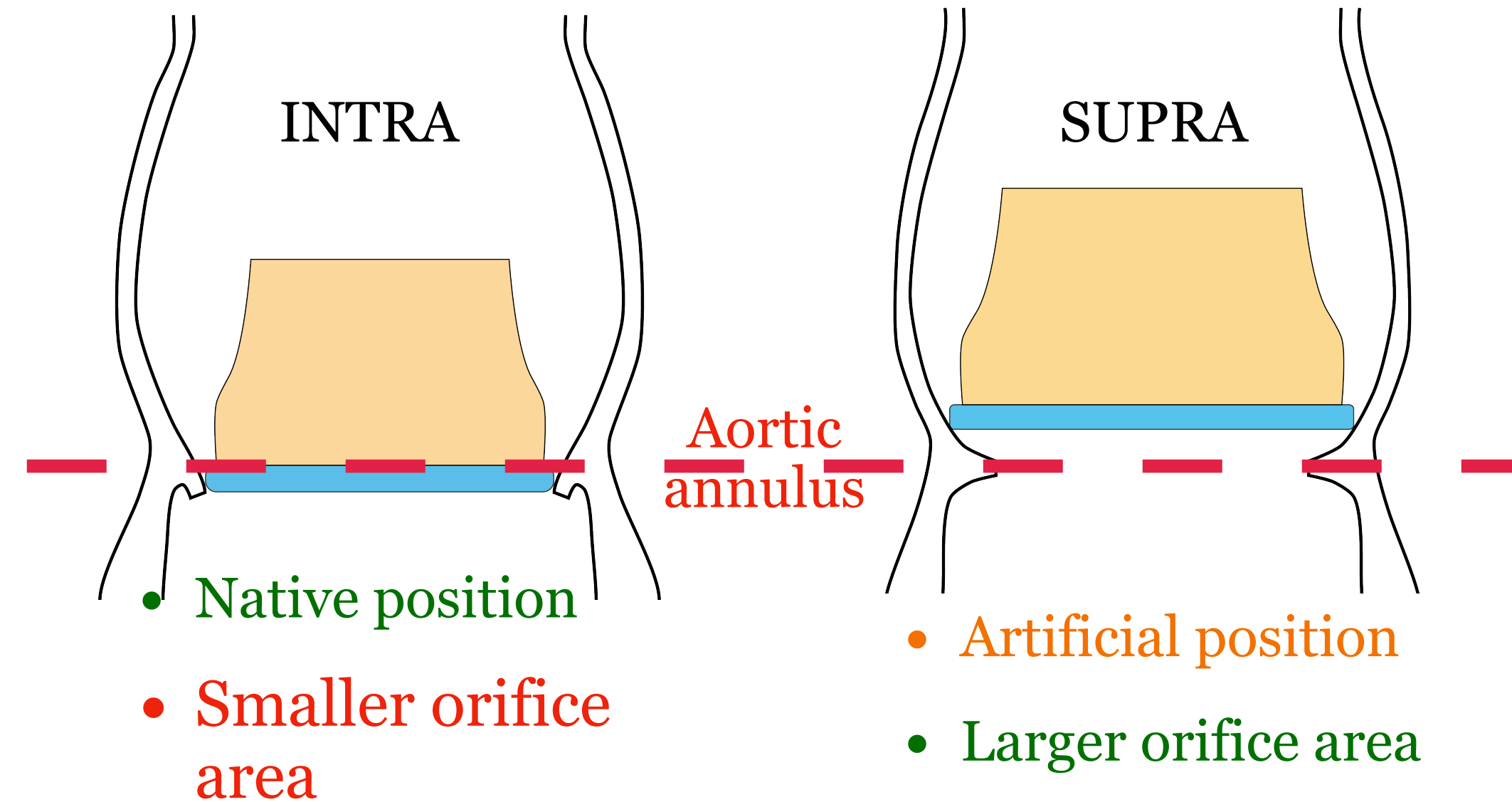
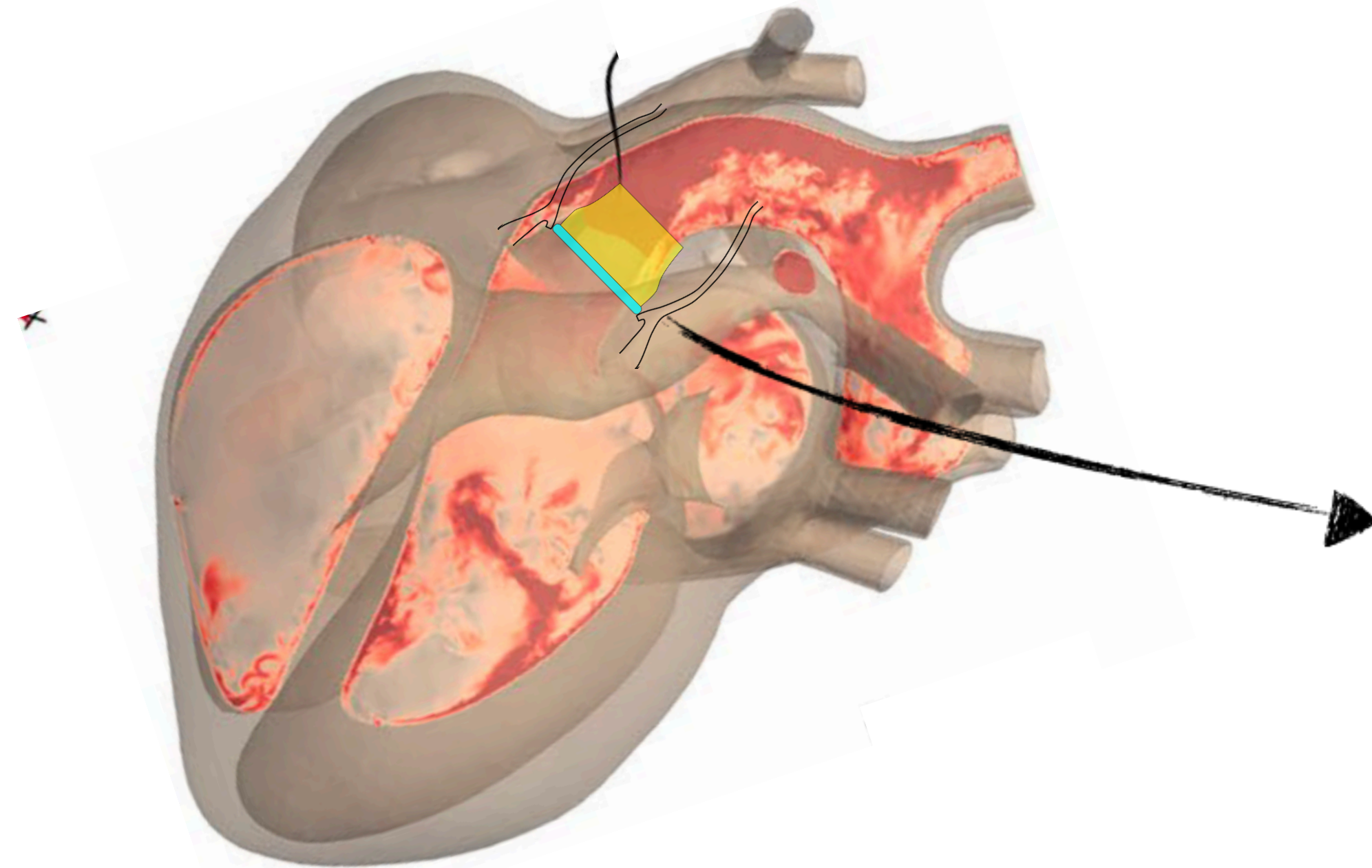
¹*Gran Sasso Science Institute (GSSI), L'Aquila, Italy*

²*INFN Tor Vergata, Rome, Italy*

³*Tor Vergata University of Rome, Rome, Italy*

Aortic Valve replacement: two possibilities

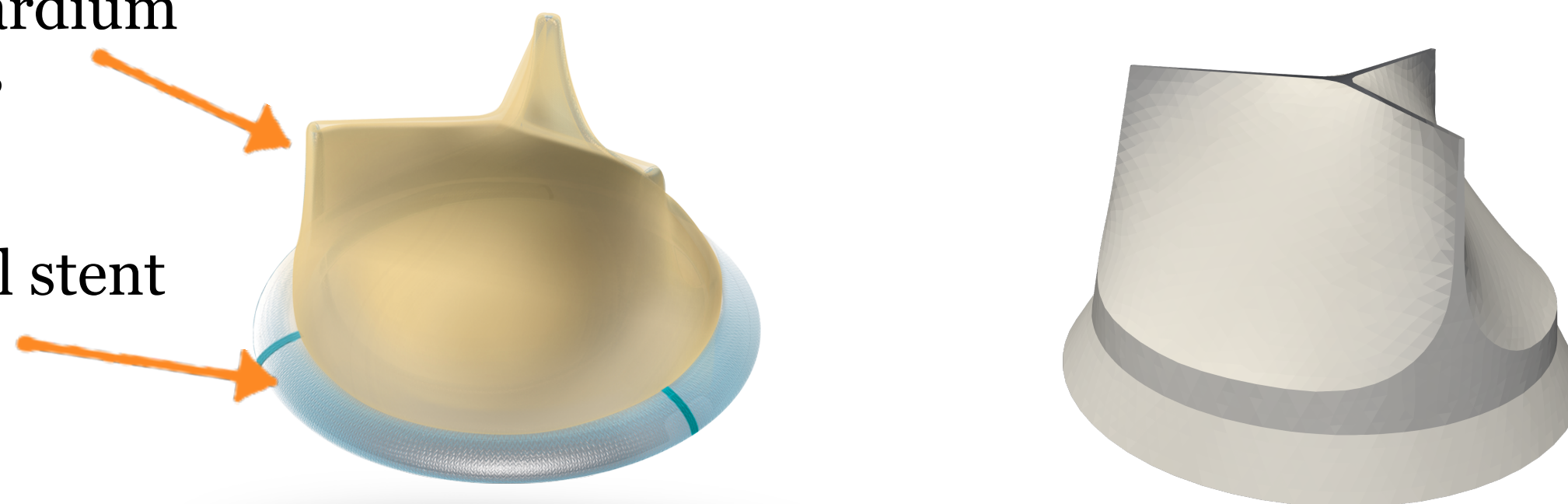
Two possible valve placement



LivaNova Crown PRT

Bovine pericardium
leaflets

Rigid internal stent



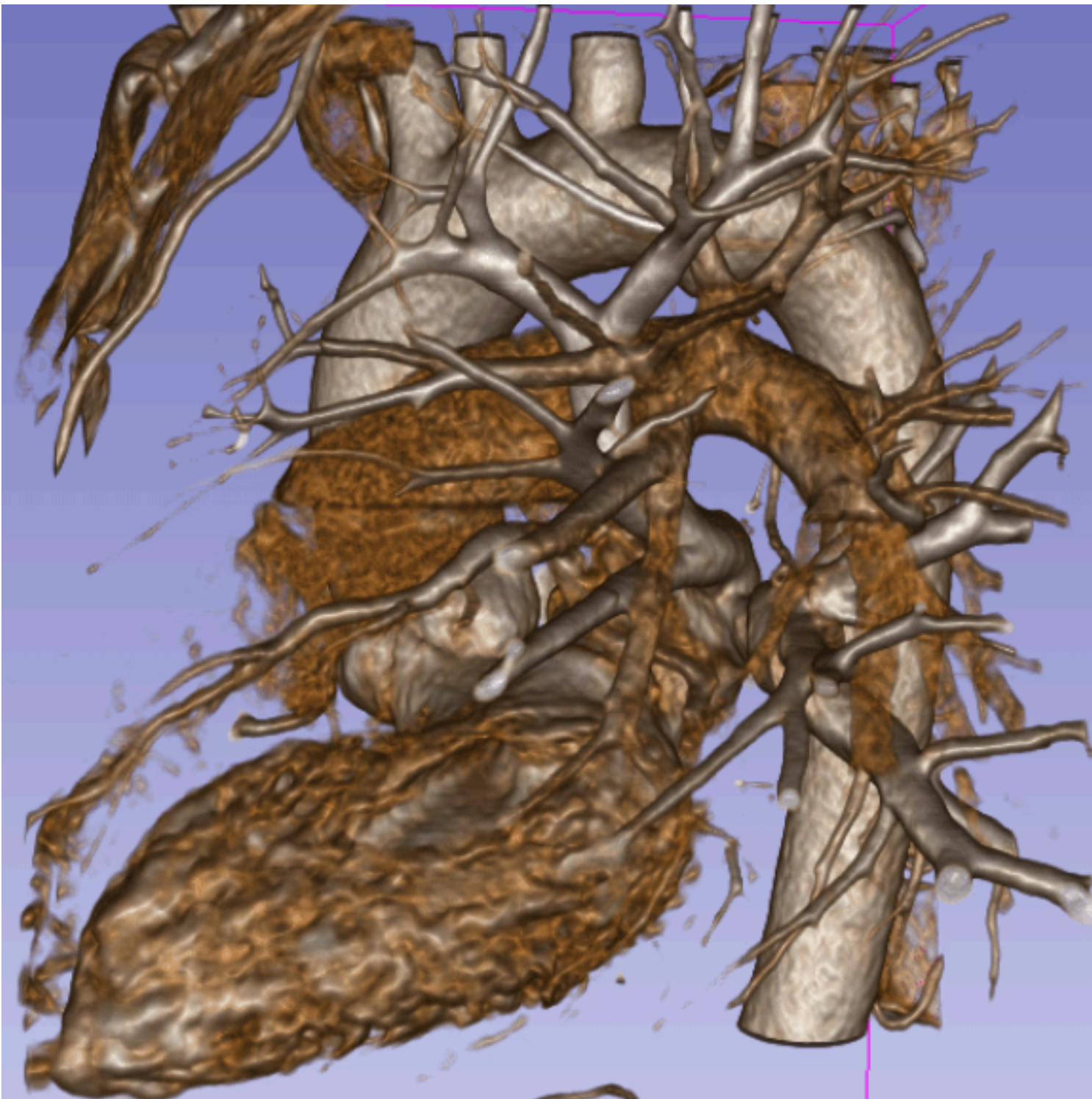
Research/clinical question:

LivaNova 19mm intra-
Or
LivaNova 21mm supra-
annular?

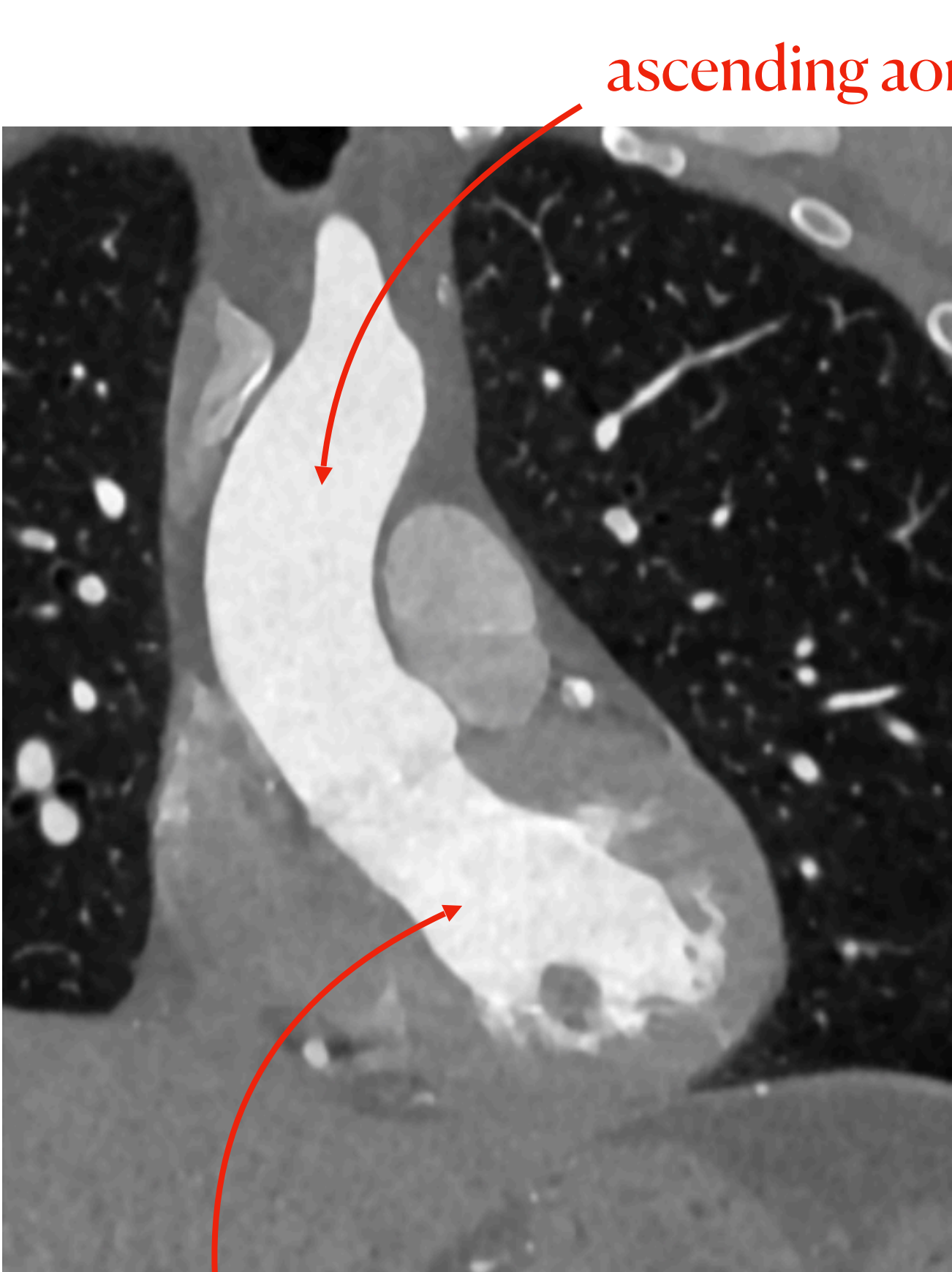
Kim, S. H., et al. (2019). Interactive cardiovascular and thoracic surgery, 28(1), 58-64.

Vahidkhah, K., & Azadani, A. N. (2017). Journal of biomechanics, 58, 114-122.

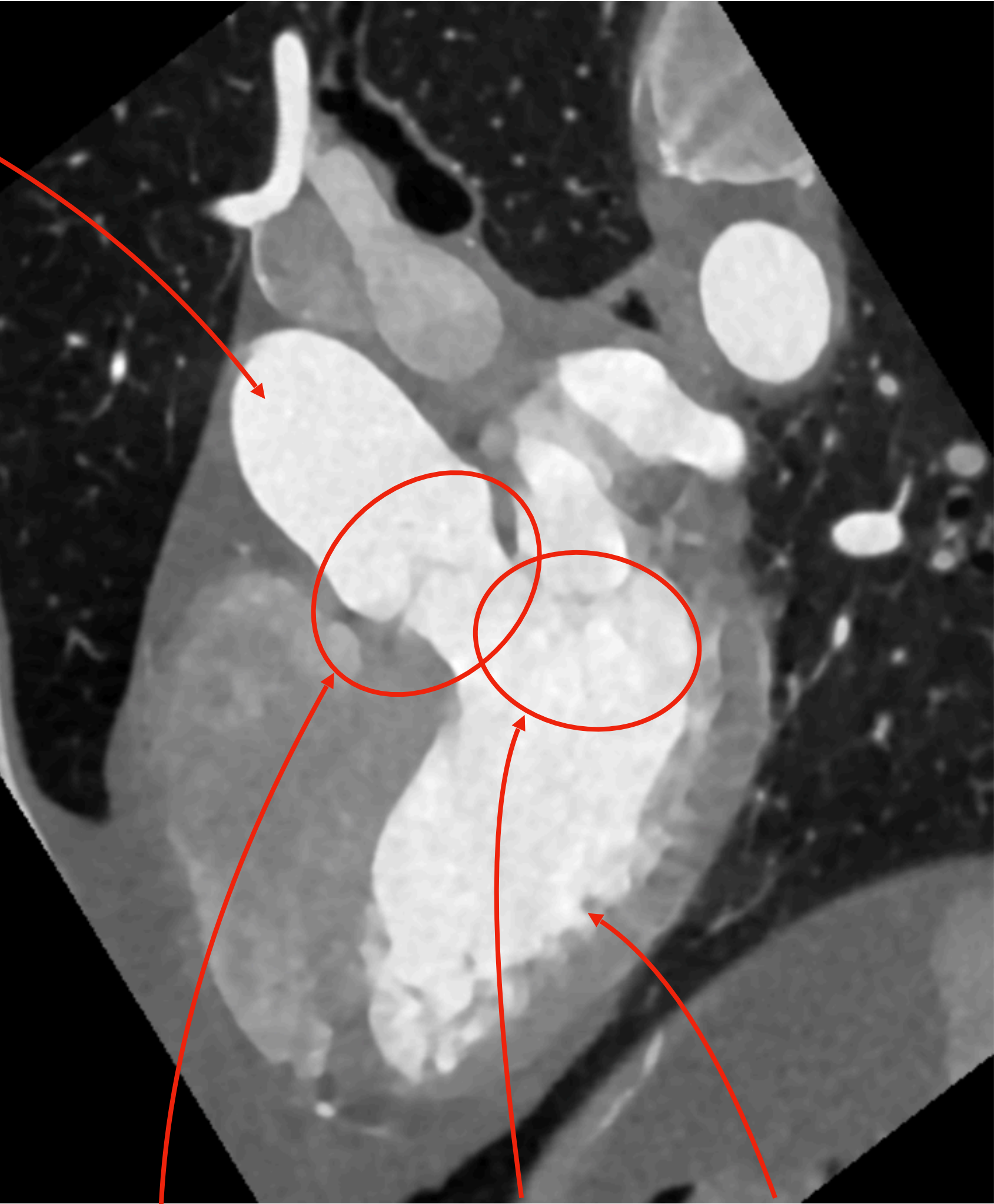
Starting from imaging data (dynamic CT scan) of a patient



CT rendering



left ventricle



aortic valve

mitral valve

trabeculae

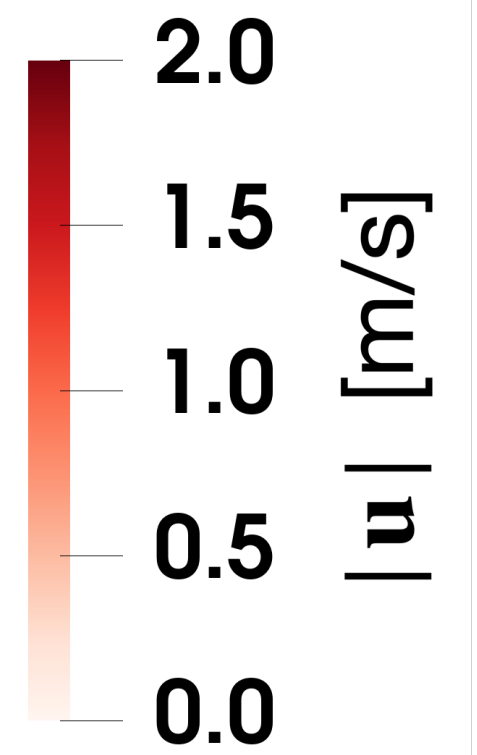
Hemodynamics

Systole

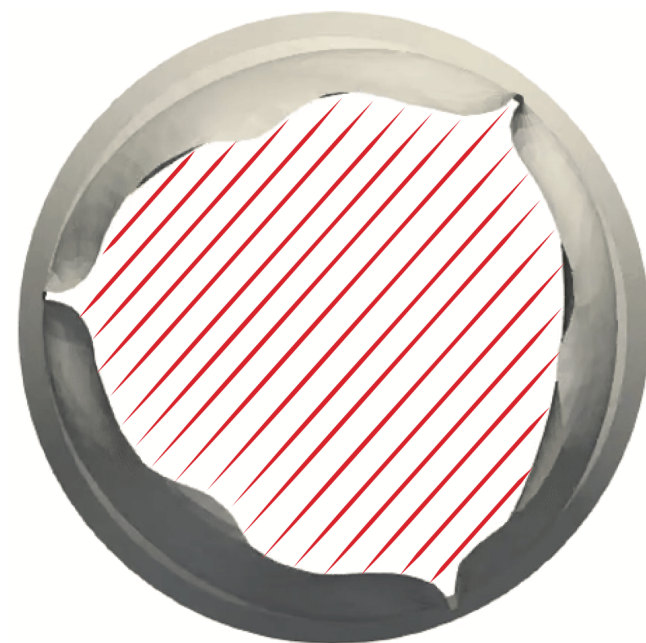
Diastole

Aortic jet

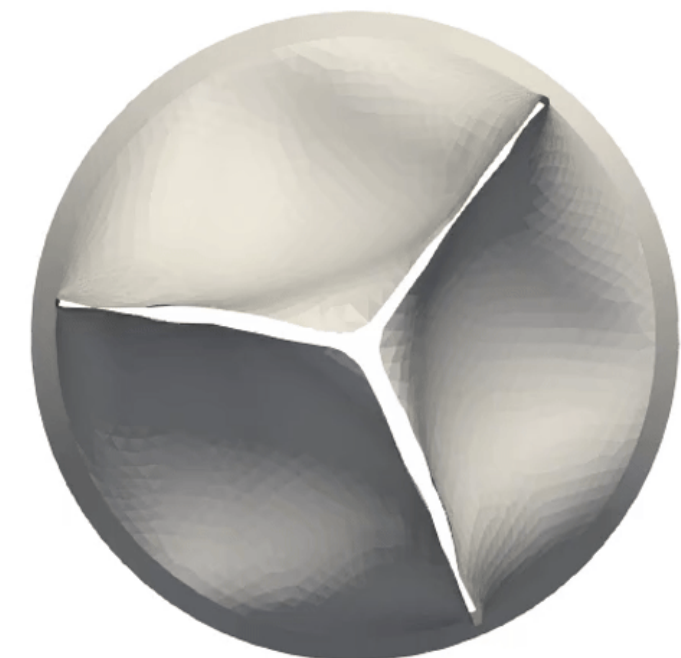
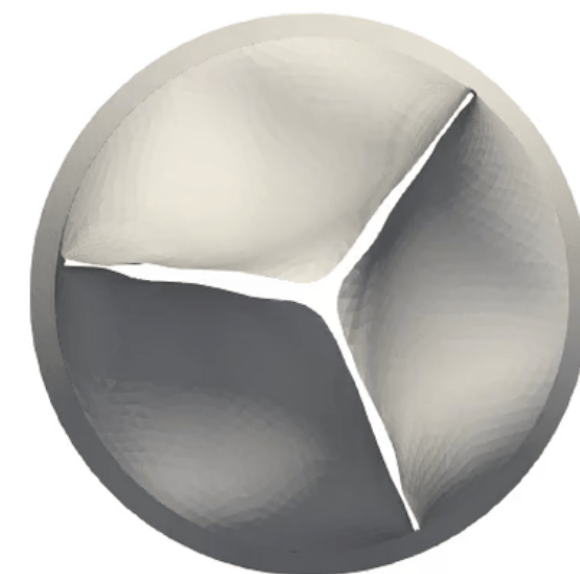
Mitral jet



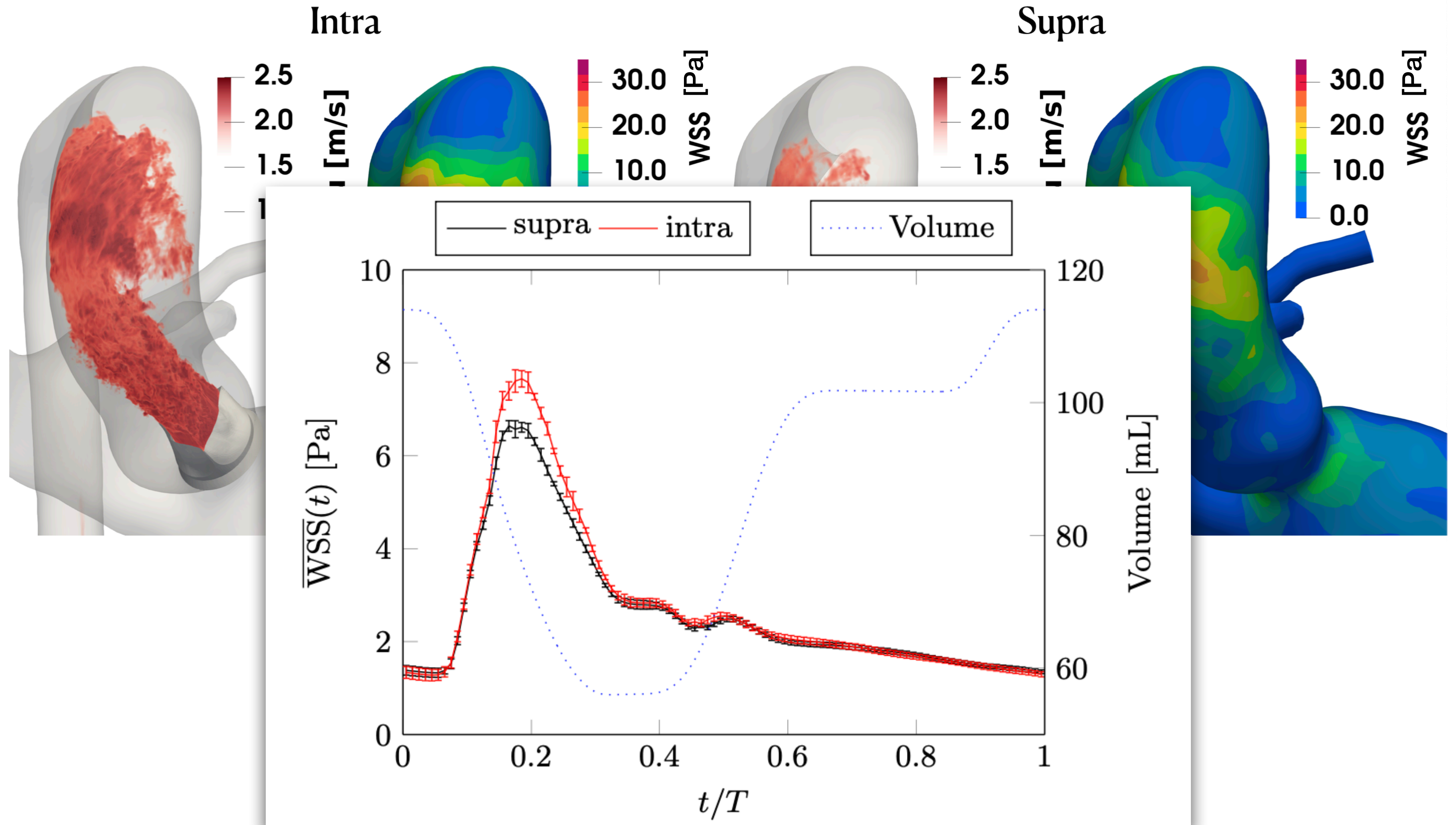
GOA =



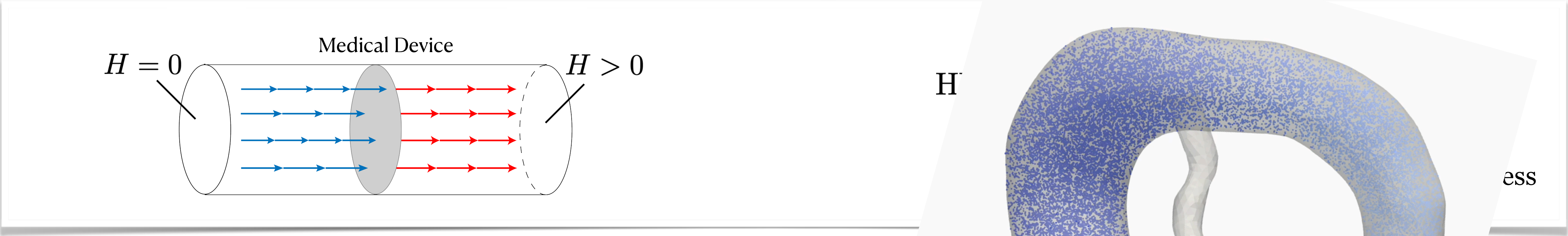
$$\text{EOA} = \frac{Q_{AV}}{u_{jet}}$$



Wall shear stress



Hemolysis

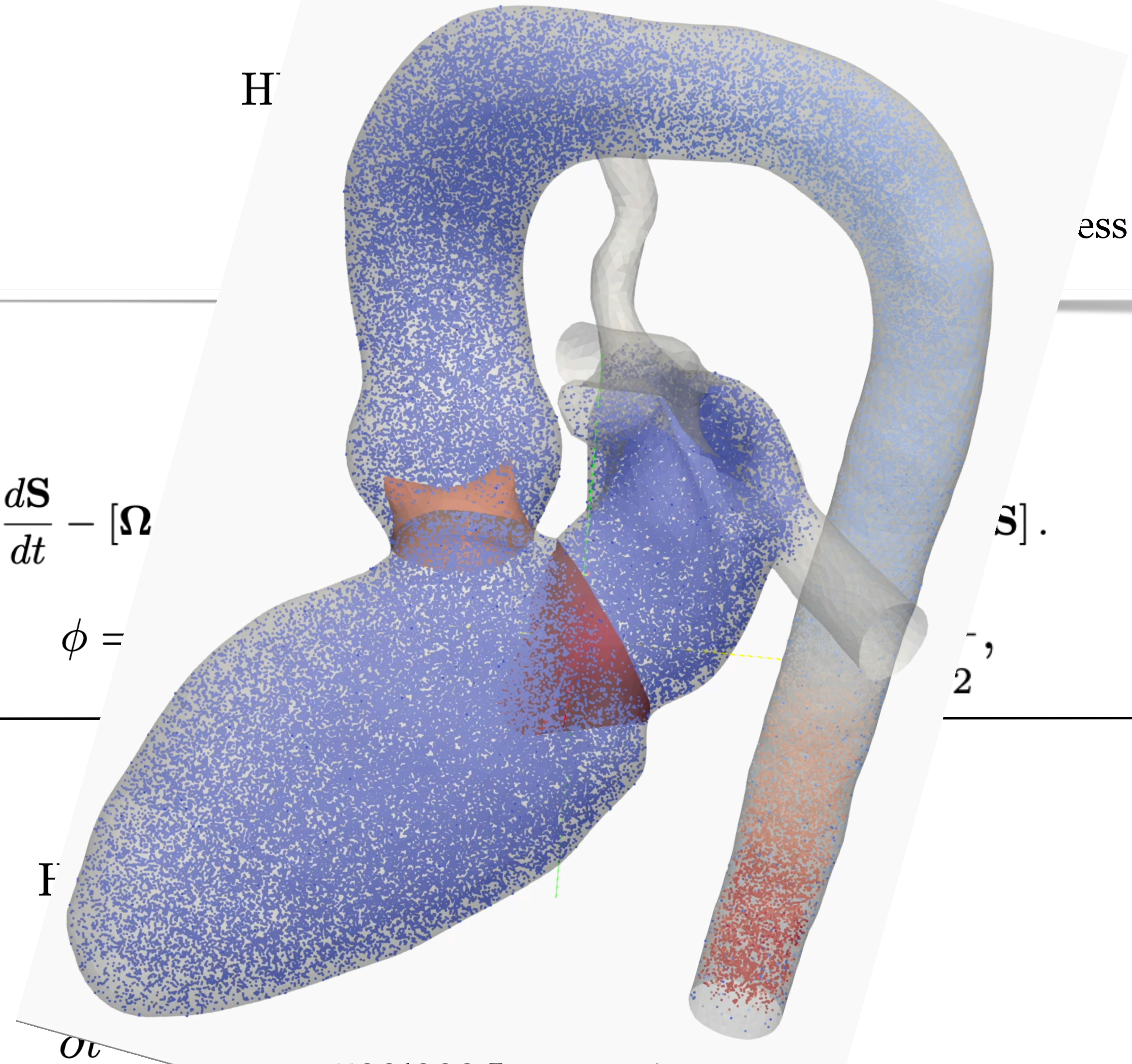


Stress based

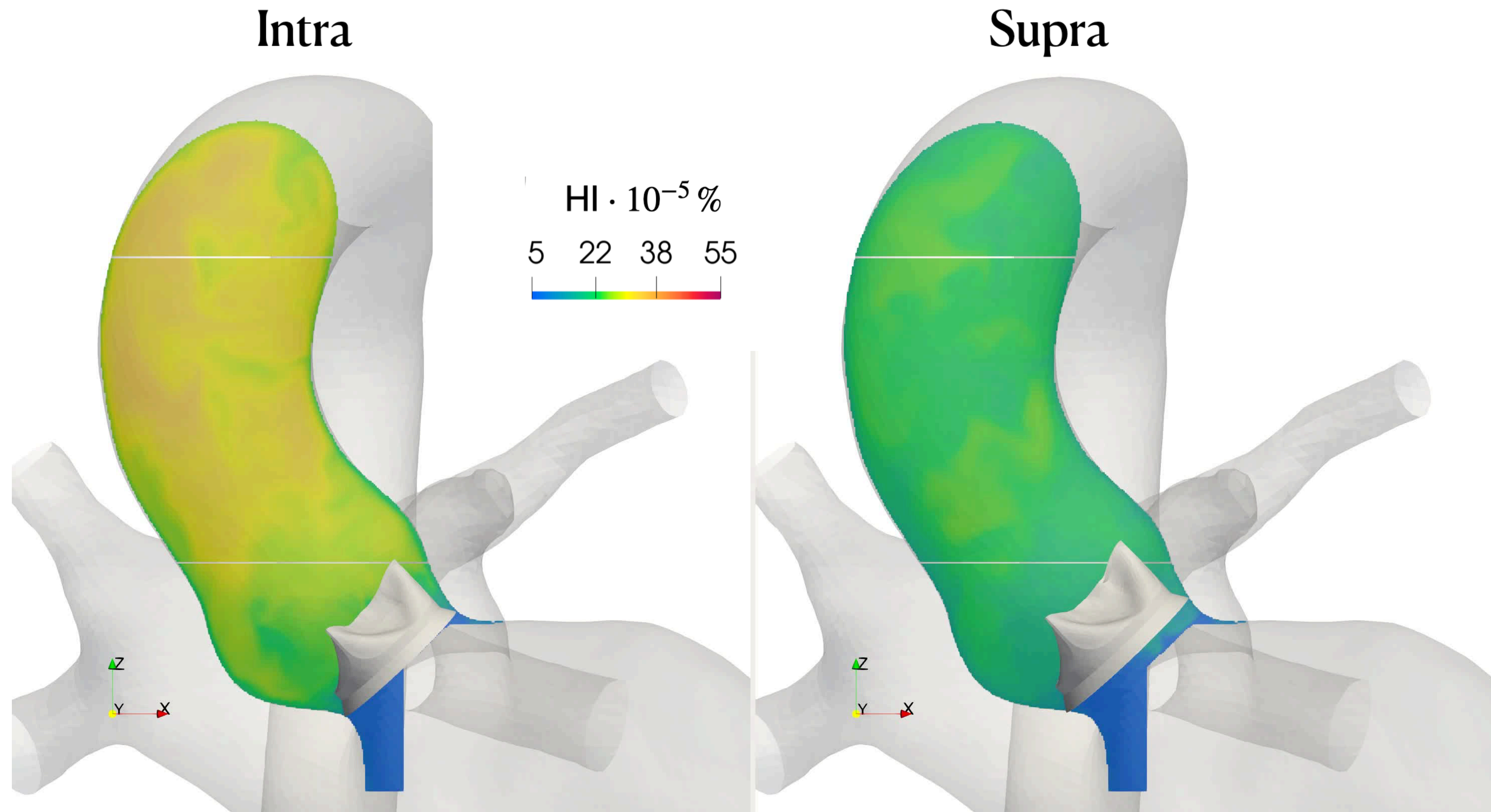
$$\tau_{eq} = \sqrt{\frac{1}{2} \boldsymbol{\tau} : \boldsymbol{\tau}} = \sqrt{\frac{1}{2} \text{tr}(\boldsymbol{\tau}^2)}.$$

Lagrangian

$$\begin{cases} \frac{d\mathbf{X}_i}{dt} = \mathbf{u}(\mathbf{X}_i(t), t) , \\ \mathbf{X}_i(t_{i0}) = \mathbf{X}_{i0}, \end{cases}$$

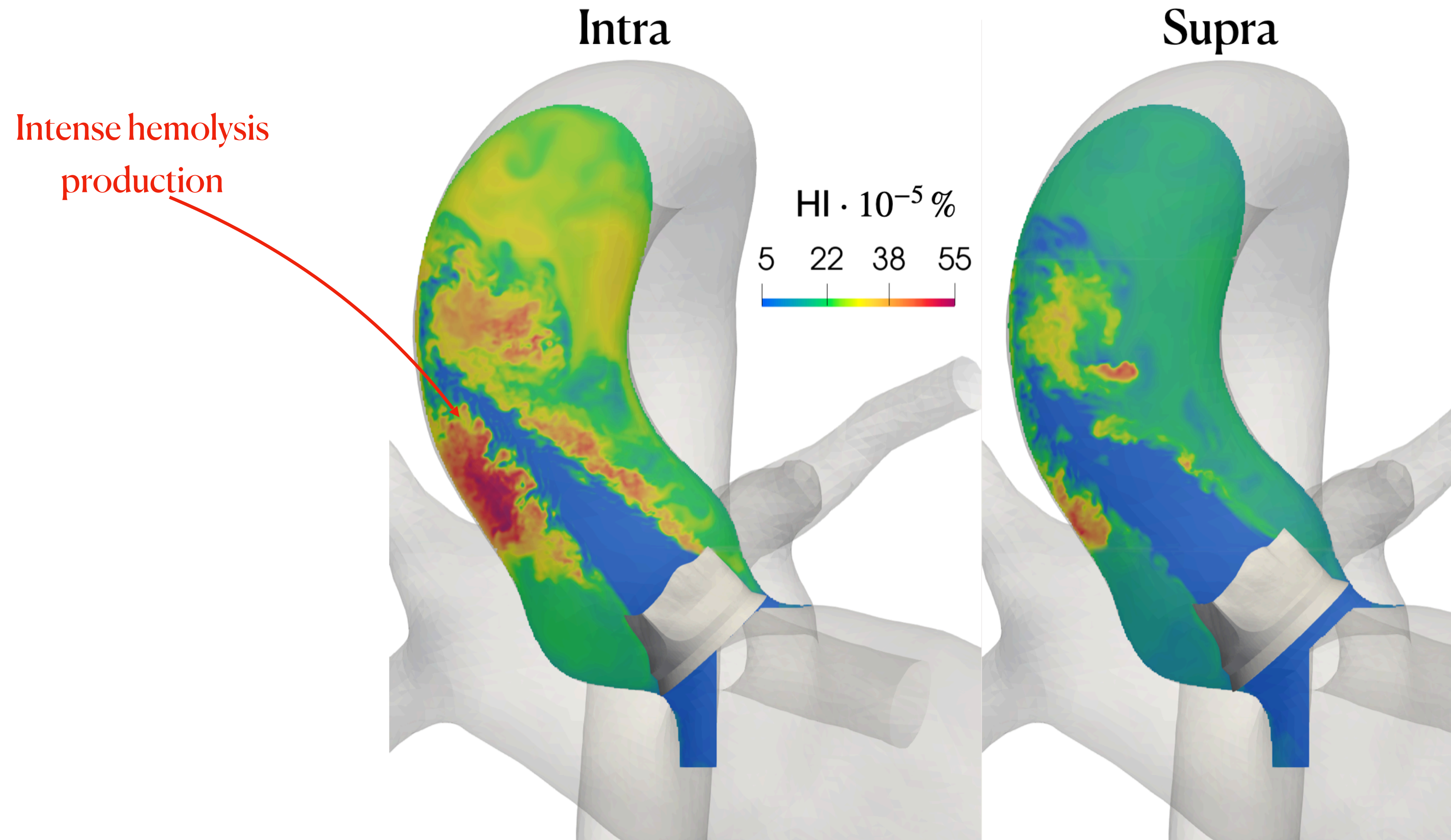


Eulerian hemolysis



$$\frac{\partial H_L}{\partial t} + \mathbf{u} \cdot \nabla H_L = C^{1/\alpha} \tau_{eq}^{\beta/\alpha} (1 - H_L)$$

Eulerian hemolysis



$$\frac{\partial H_L}{\partial t} + \mathbf{u} \cdot \nabla H_L = C^{1/\alpha} \tau_{eq}^{\beta/\alpha} (1 - H_L)$$



A data-driven computational methodology for assessing ventricular ablation procedures

Biomechanics and Modeling in Mechanobiology (2026) 25:58
<https://doi.org/10.1007/s10237-026-02073-7>

ORIGINAL PAPER



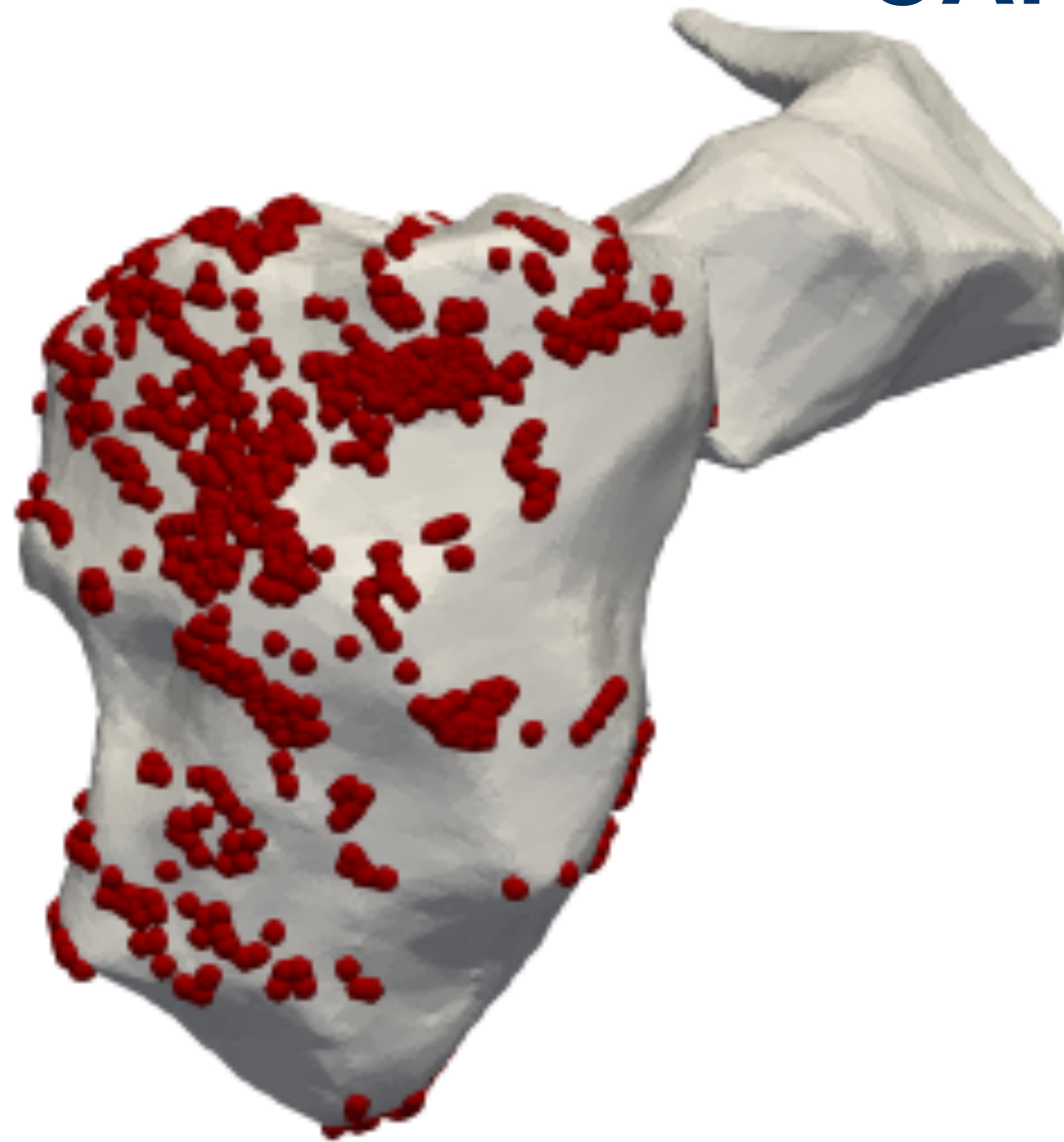
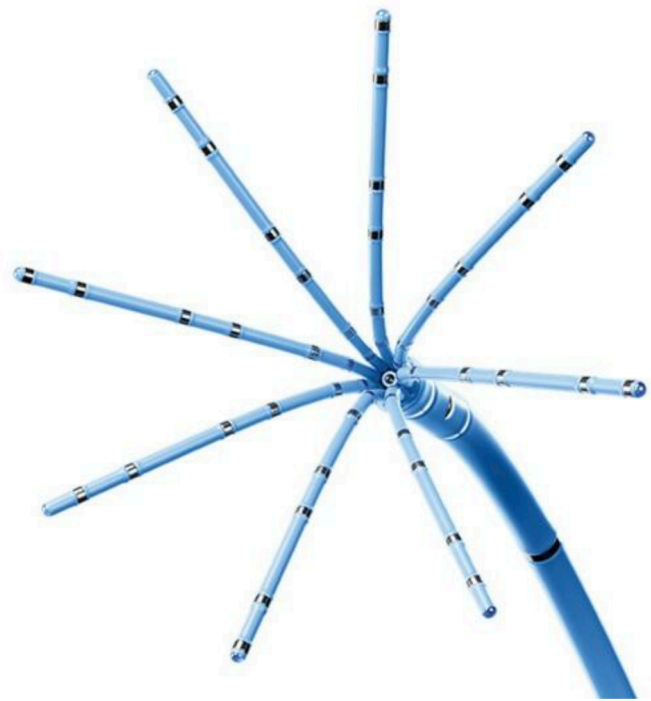
A data-driven computational methodology for assessing ventricular ablation procedures

Filippo Caruso Lombardi^{1,6} · Anna Crispino² · Bich Lien Nguyen³ · Nicola Galea⁴ · Alessandro Loppini⁵ ·
Simonetta Filippi² · Francesco Viola^{1,6} · Alessio Gizzi²

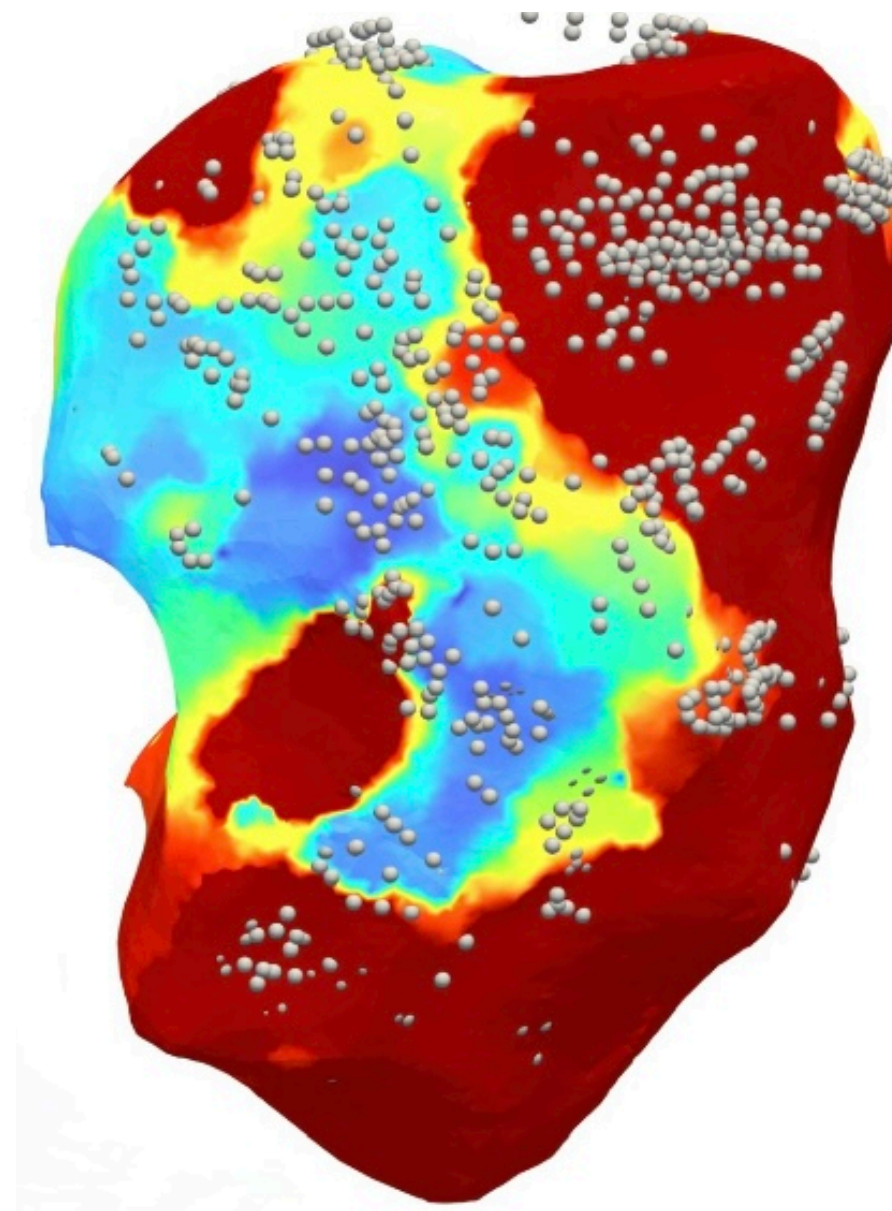
Received: 19 September 2025 / Accepted: 7 April 2026
© The Author(s) 2026

Modelling and predicting the outcome of Ventricular Ablation

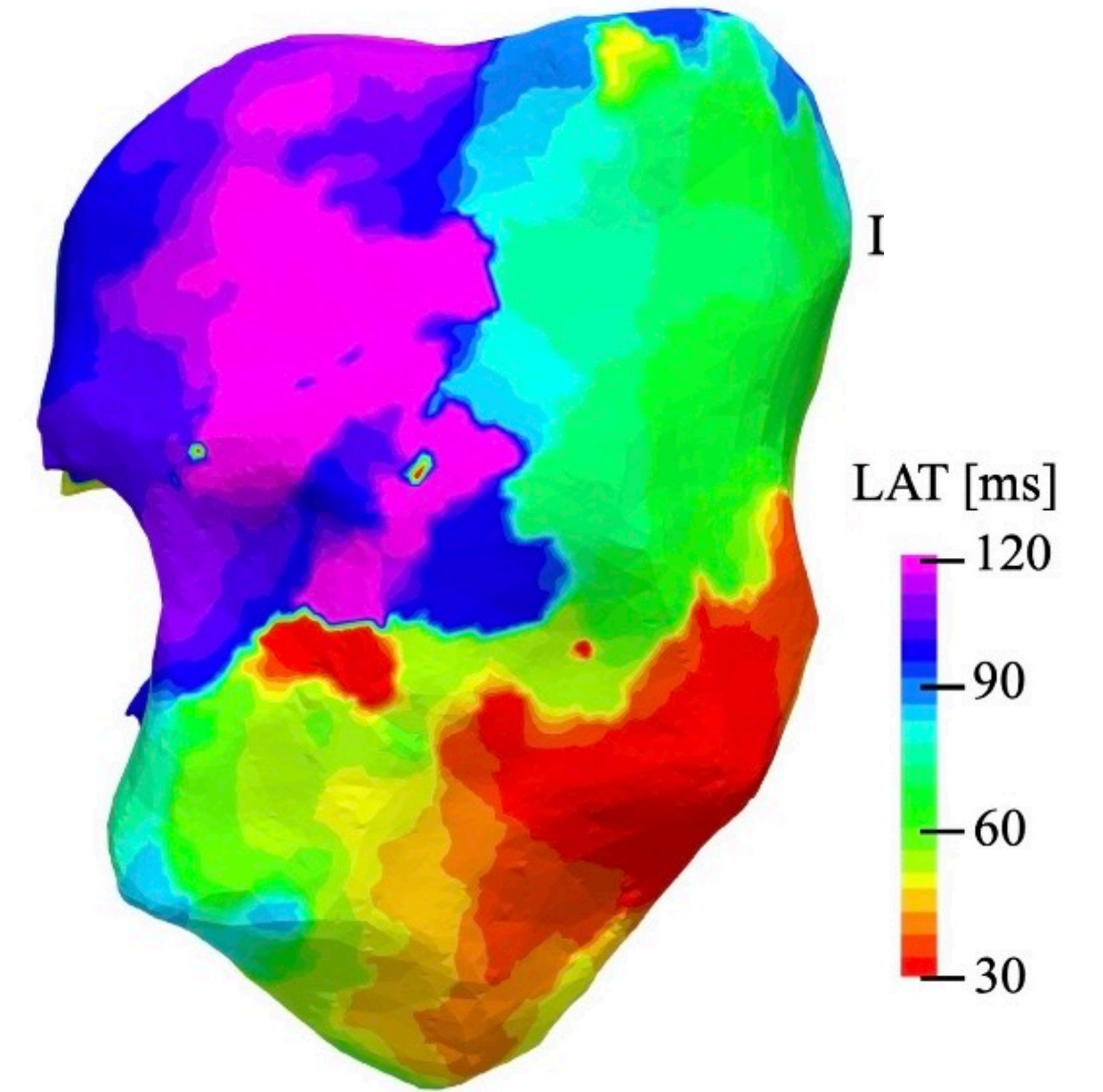
CARTO 3 System



779 EGMs

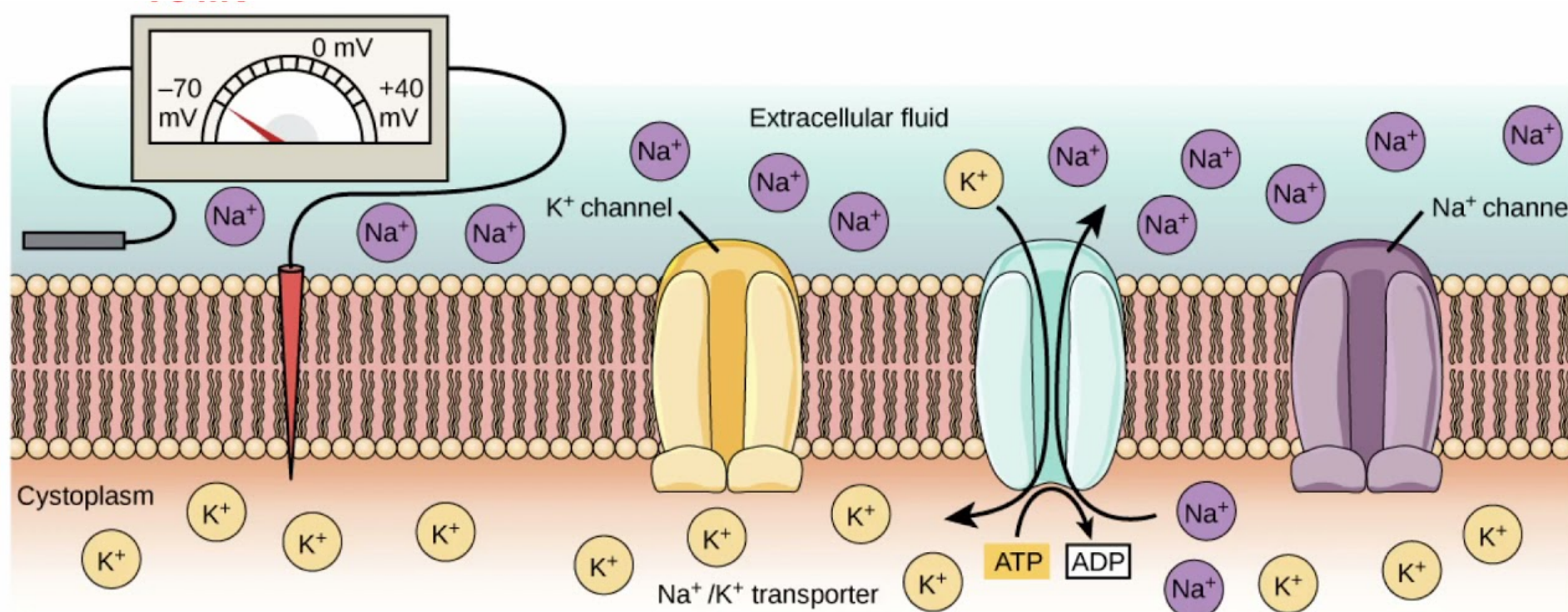


Voltage Map



Local activation time

The **cardiac electrophysiology** is solved via the **MONODOMAIN MODEL**:



$$\begin{cases} \frac{\partial U_m}{\partial t} = \nabla \cdot (\mathbf{D}(F(x)) \nabla U_m) + I_{\text{ion}}(U_m, \phi) + I_s(x, t), \\ \frac{d\phi}{dt} = G(U_m, \phi, F(x)), \quad \text{in } \Omega \end{cases}$$

Cellular Model

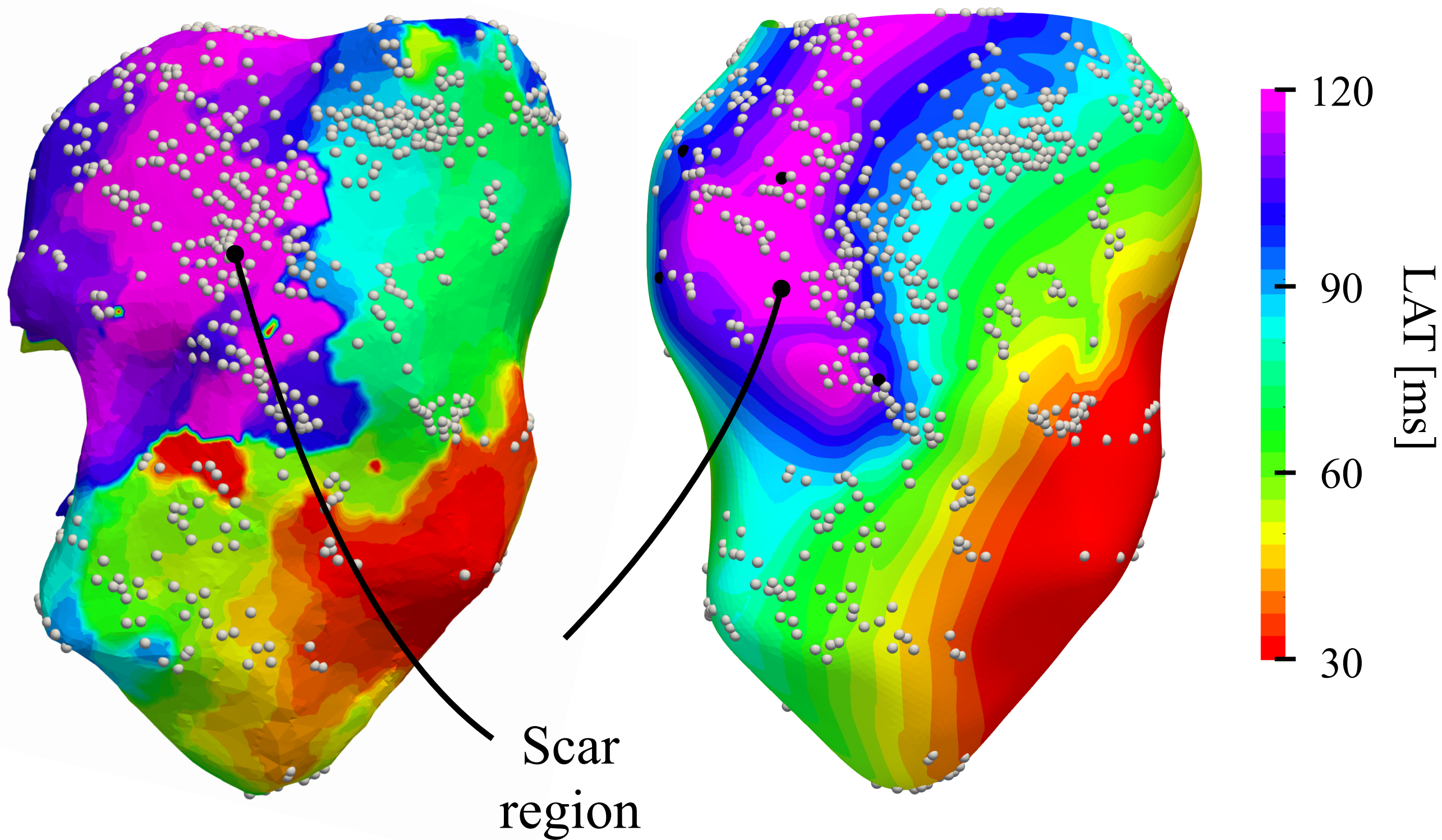
Modelling and predicting the outcome of Ventricular Ablation

Local activation time

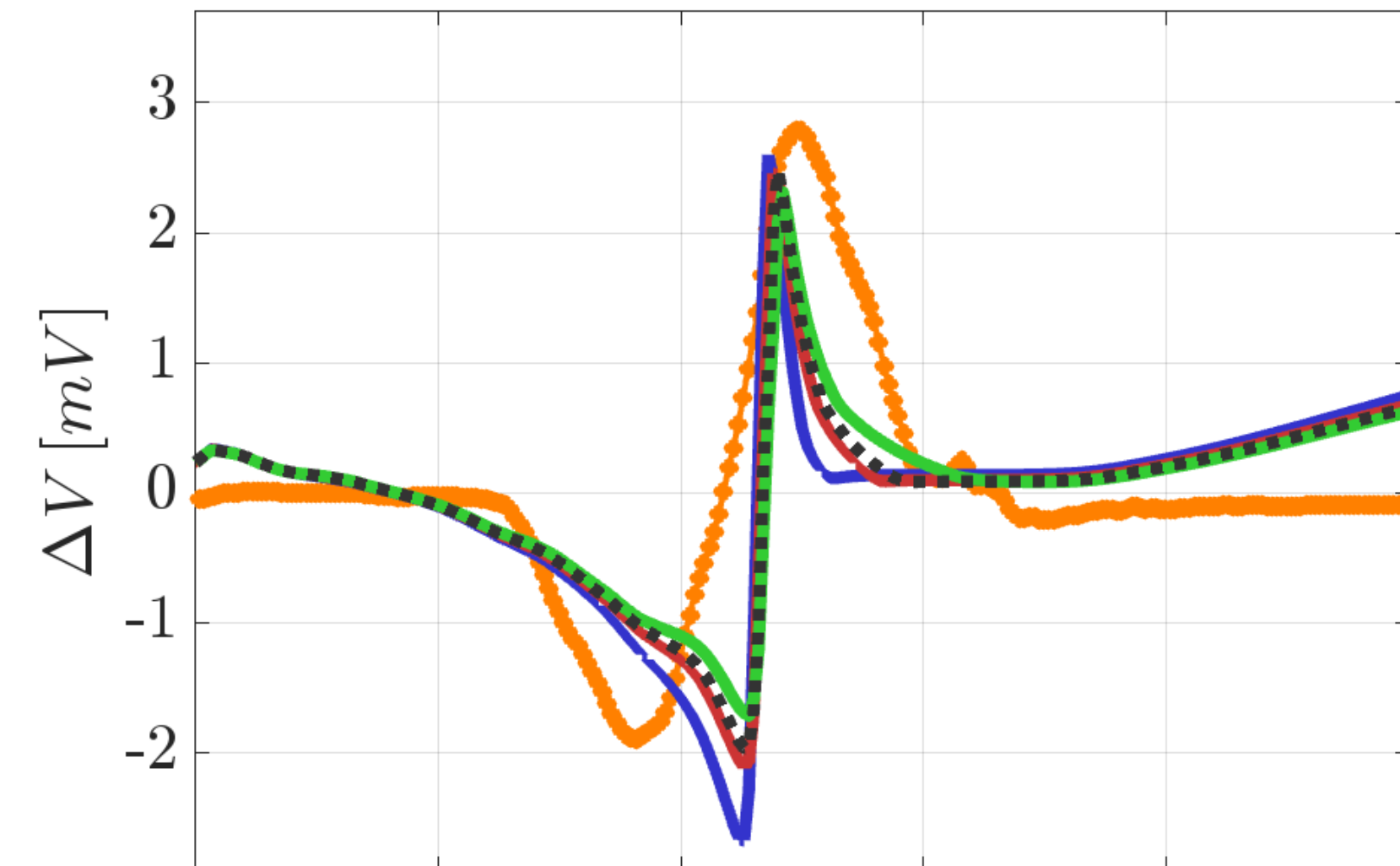
Electrograms

Experimental LATs

In-Silico LATs



$$\phi(\mathbf{R}) = \frac{1}{4\pi\sigma} \int_{\Omega} -\sigma_t \nabla v \cdot \nabla \left(\frac{1}{\mathbf{R}} \right) d\omega$$

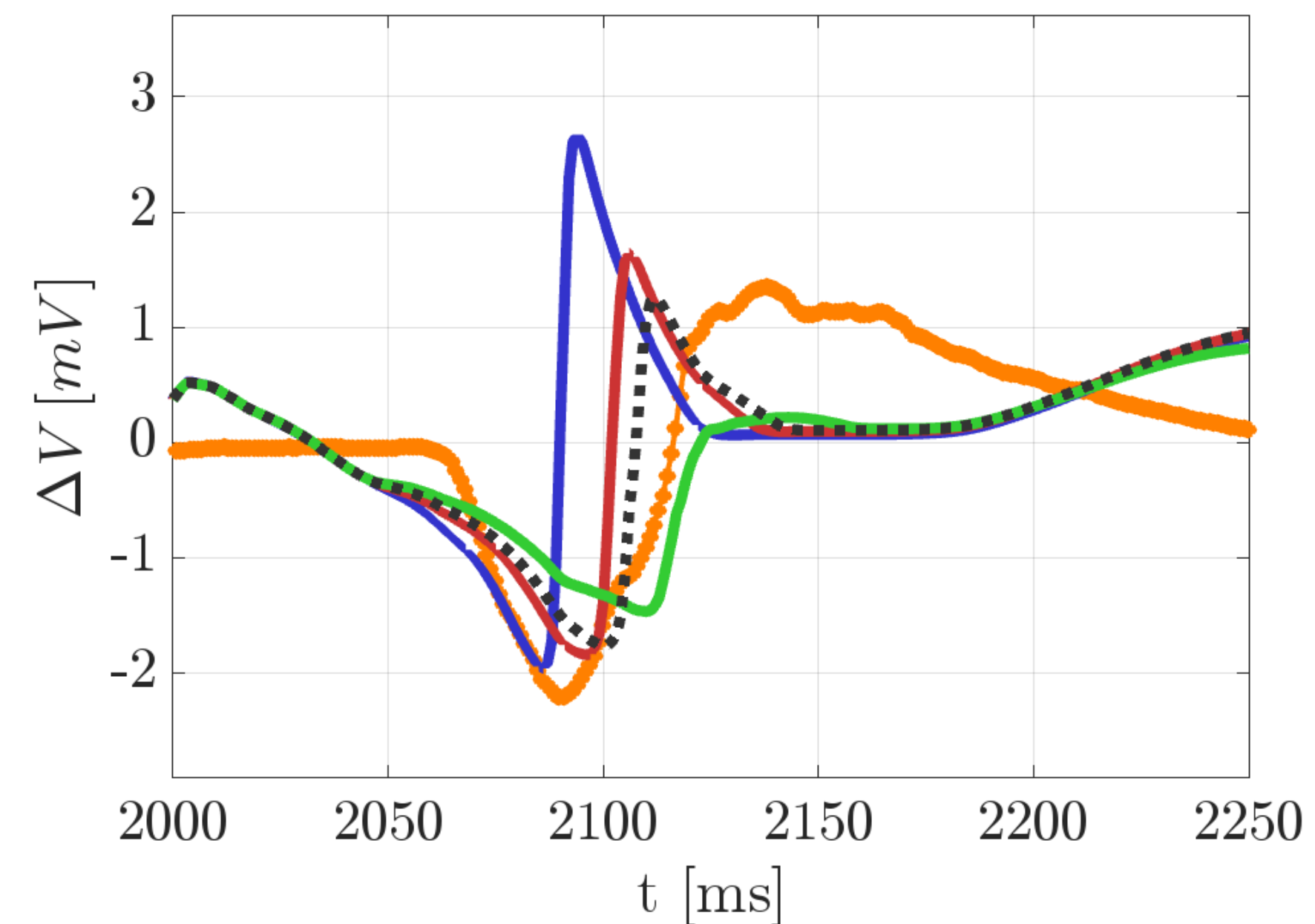


$\alpha = 0.45$

$\alpha = 0.60$

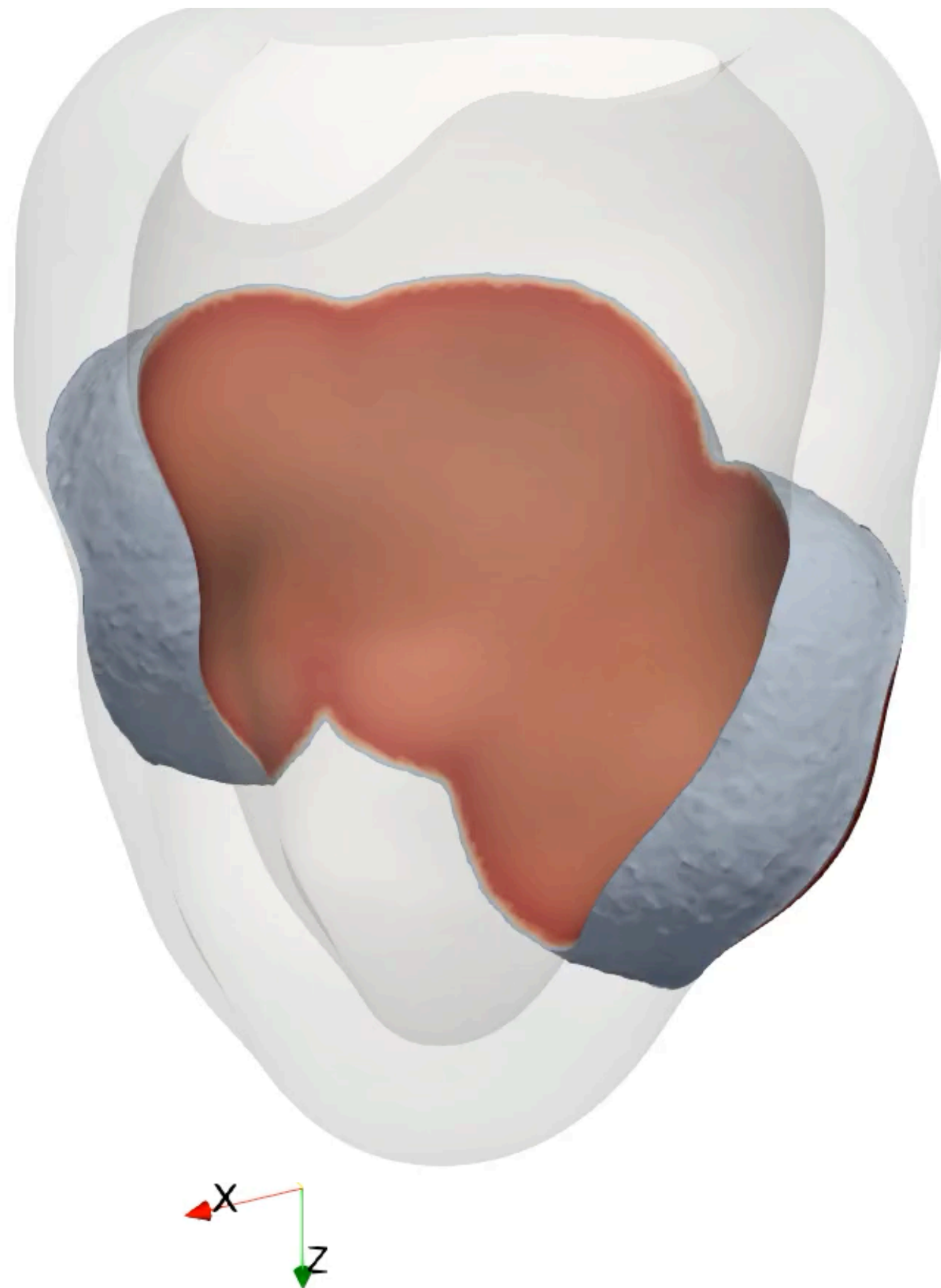
$\alpha = 0.65$

$\alpha = 0.75$

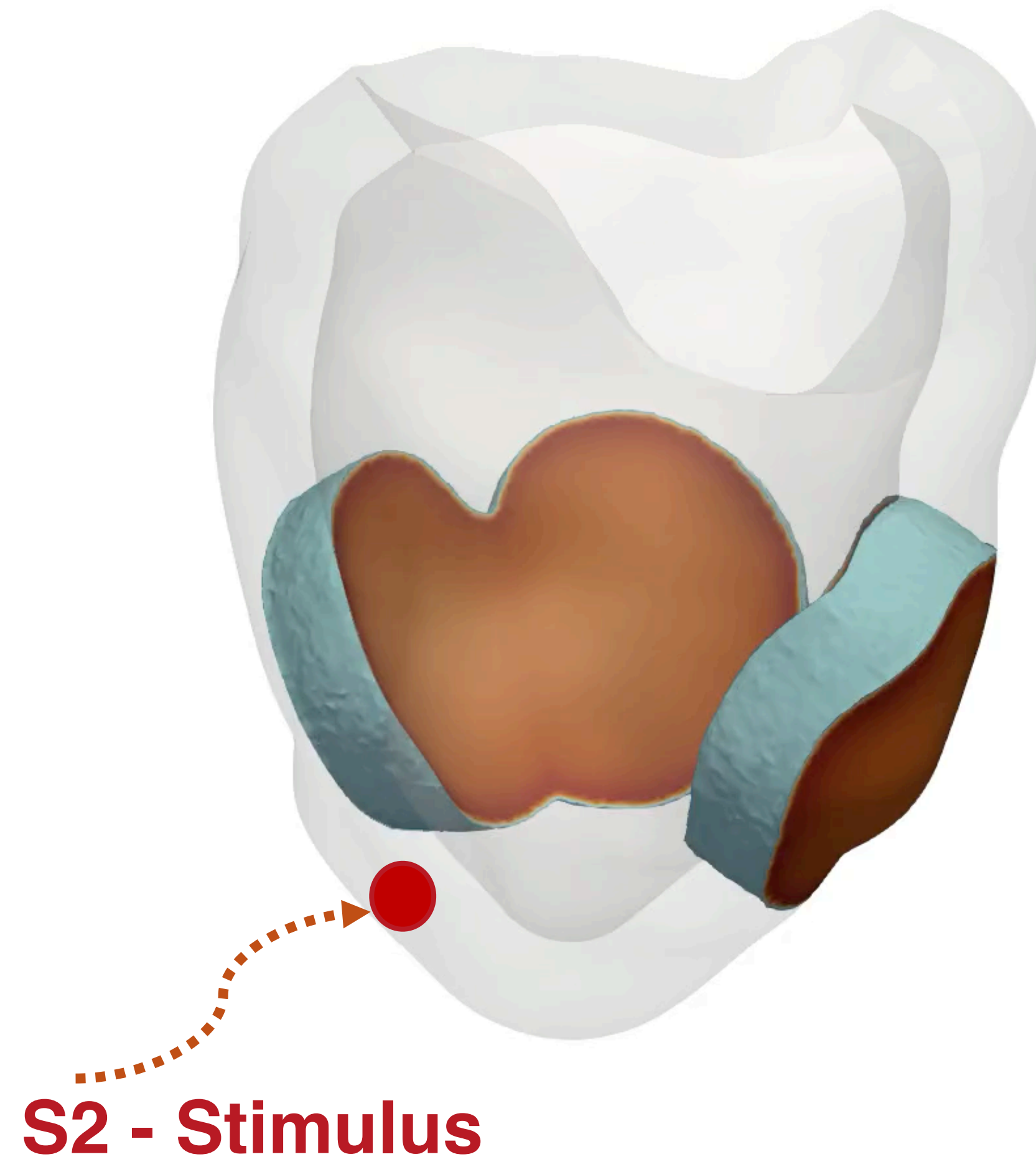


Modelling and predicting the outcome of Ventricular Ablation

**Normal Sinus
Rhythm**

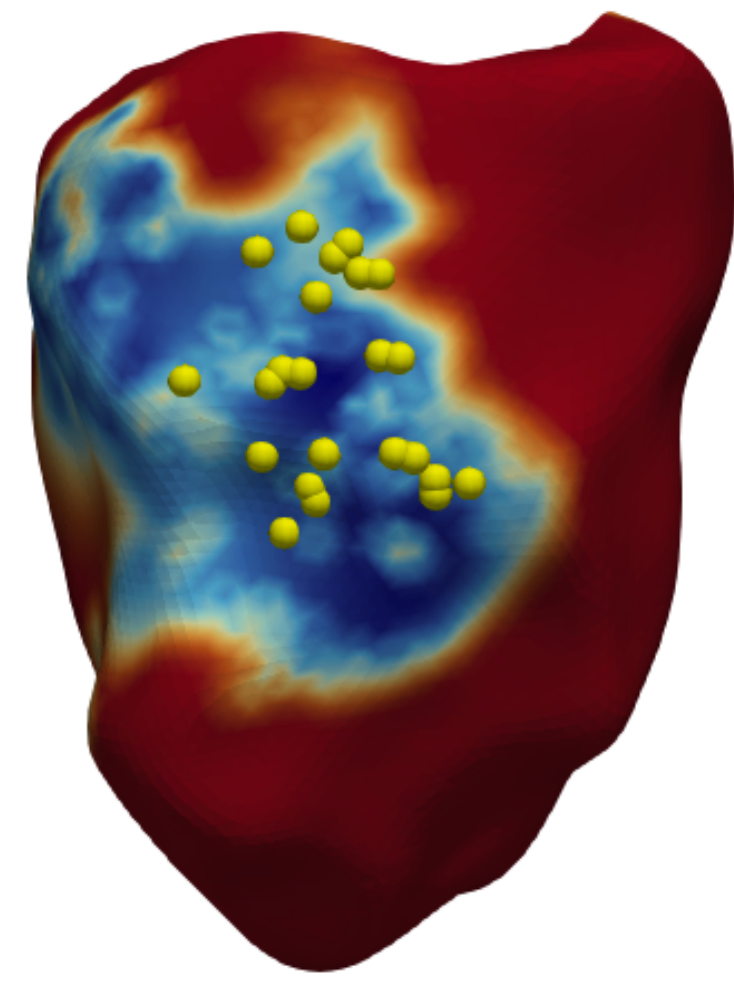
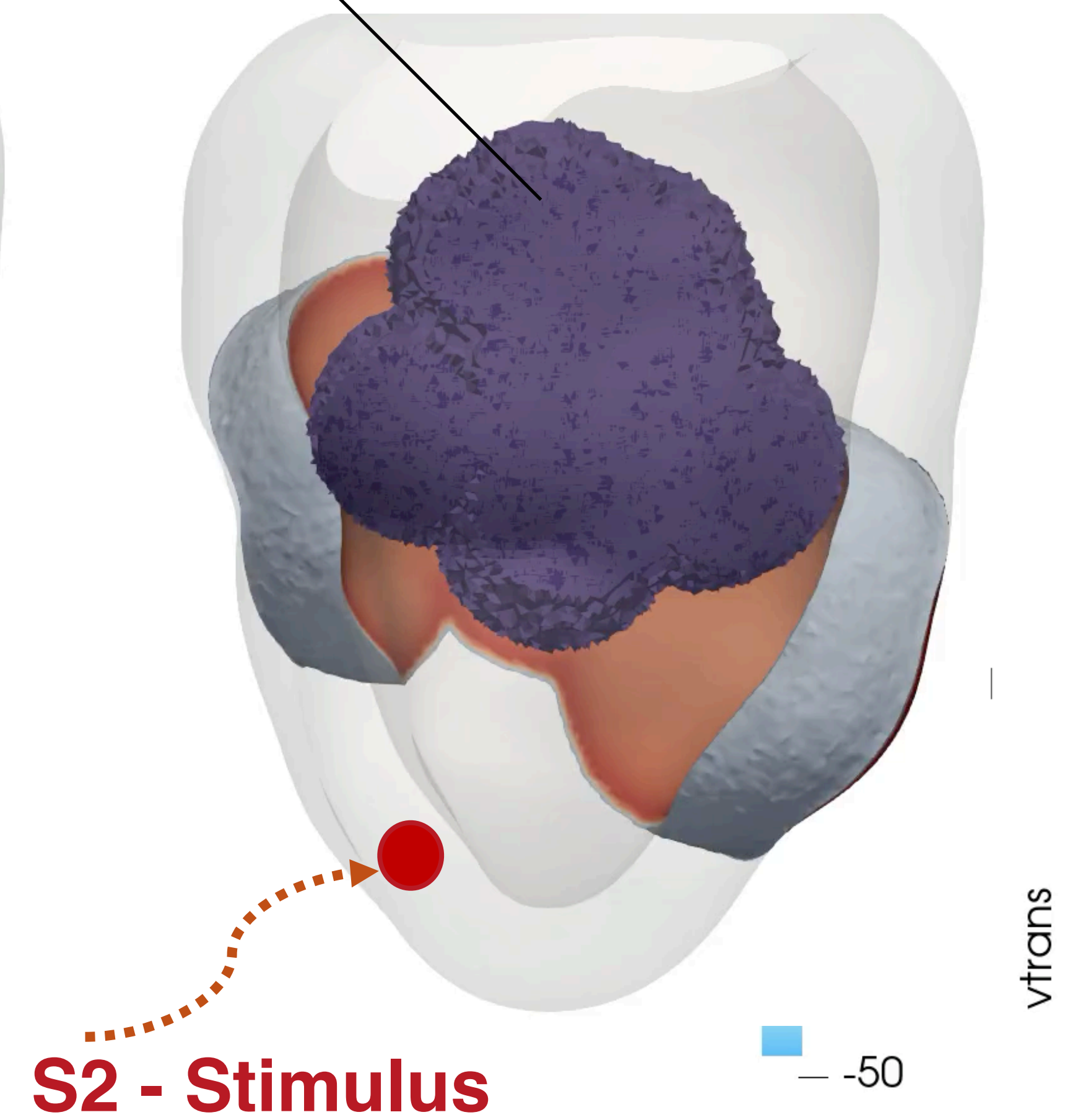


**Tachycardic
Rhythm**



24 experimental
ablation points

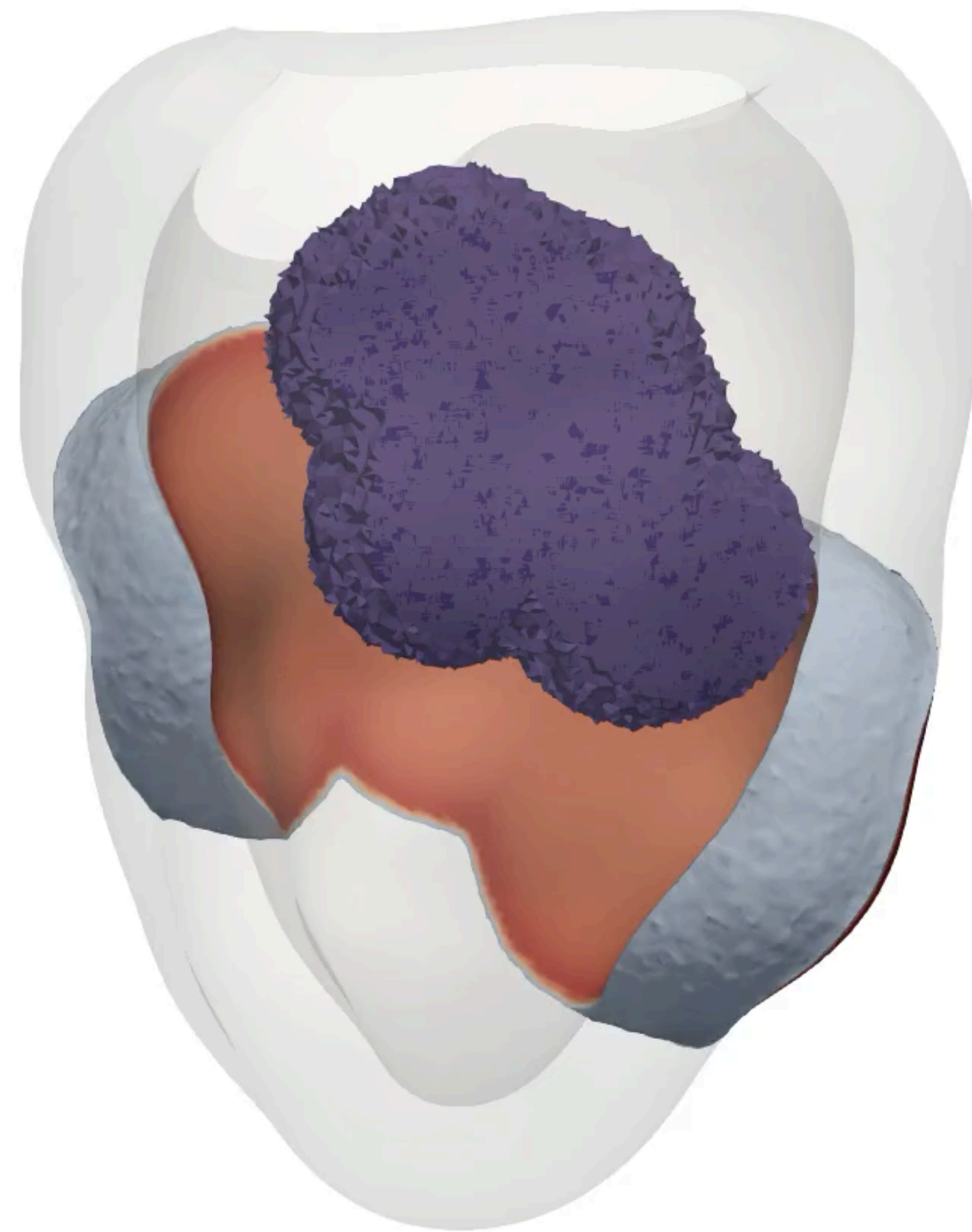
Ablation



Modelling and predicting the outcome of Ventricular Ablation

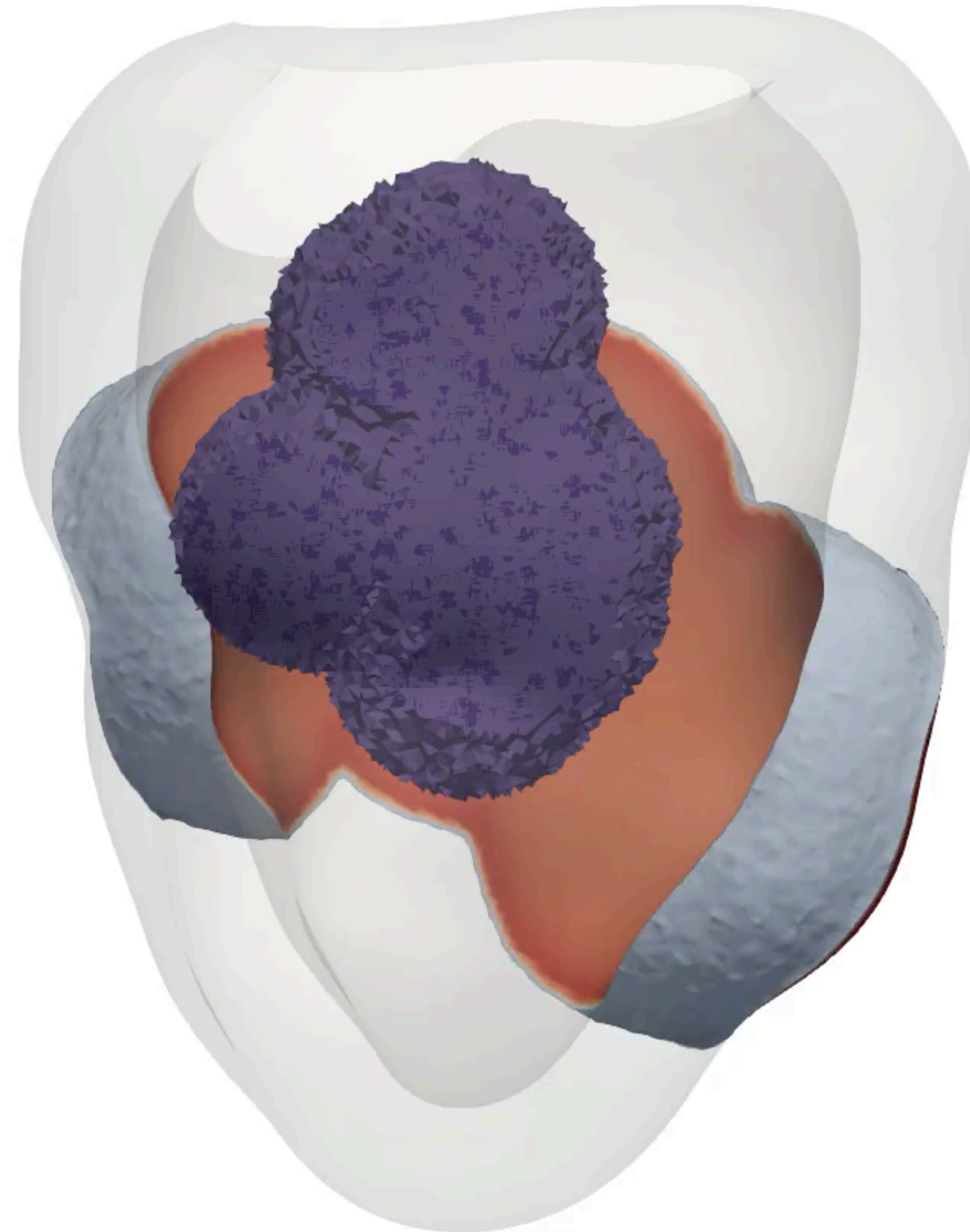
Predict Optimal ablation

Late activation time
(LAT) based ablation



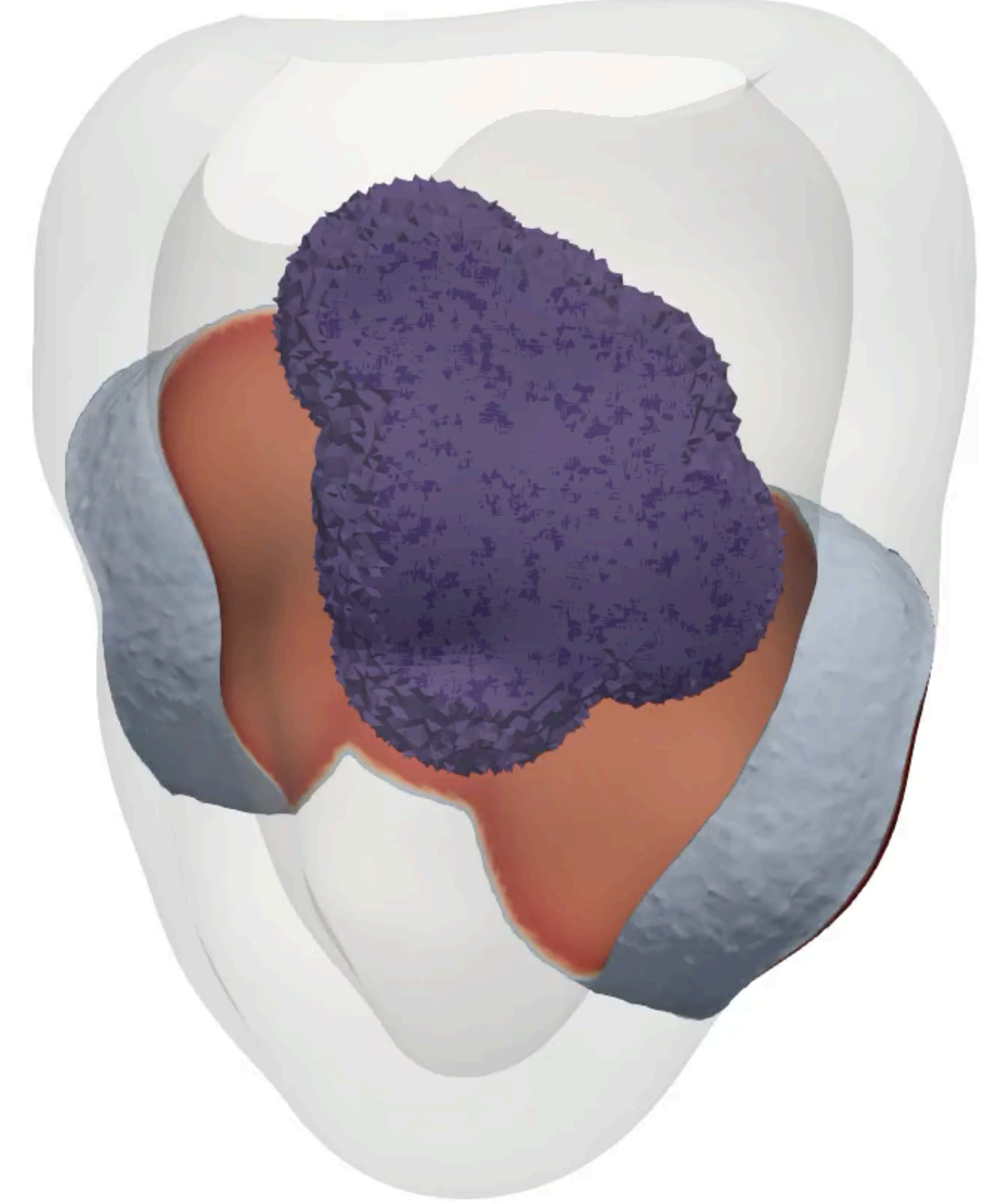
-23.4%

Conduction velocity
(CV) based ablation



-25.2%

Action potential
(APD) based ablation



-22.8%

Reduction
ablated
volume



Statistical shape modelling of the heart and virtual cohorts generation

From Raw Segmentations to Simulation-Ready Cardiac Meshes:
An Automated Framework for Anatomical Reconstruction and Virtual
Cohort Generation

Francesco Fabbri, Martino Andrea Scarpolini, Paolo Ciancarella,
Francesco Tudisco, Roberto Verzicco, Alessandro Ricci, Francesco Viola

In-silico clinical trials

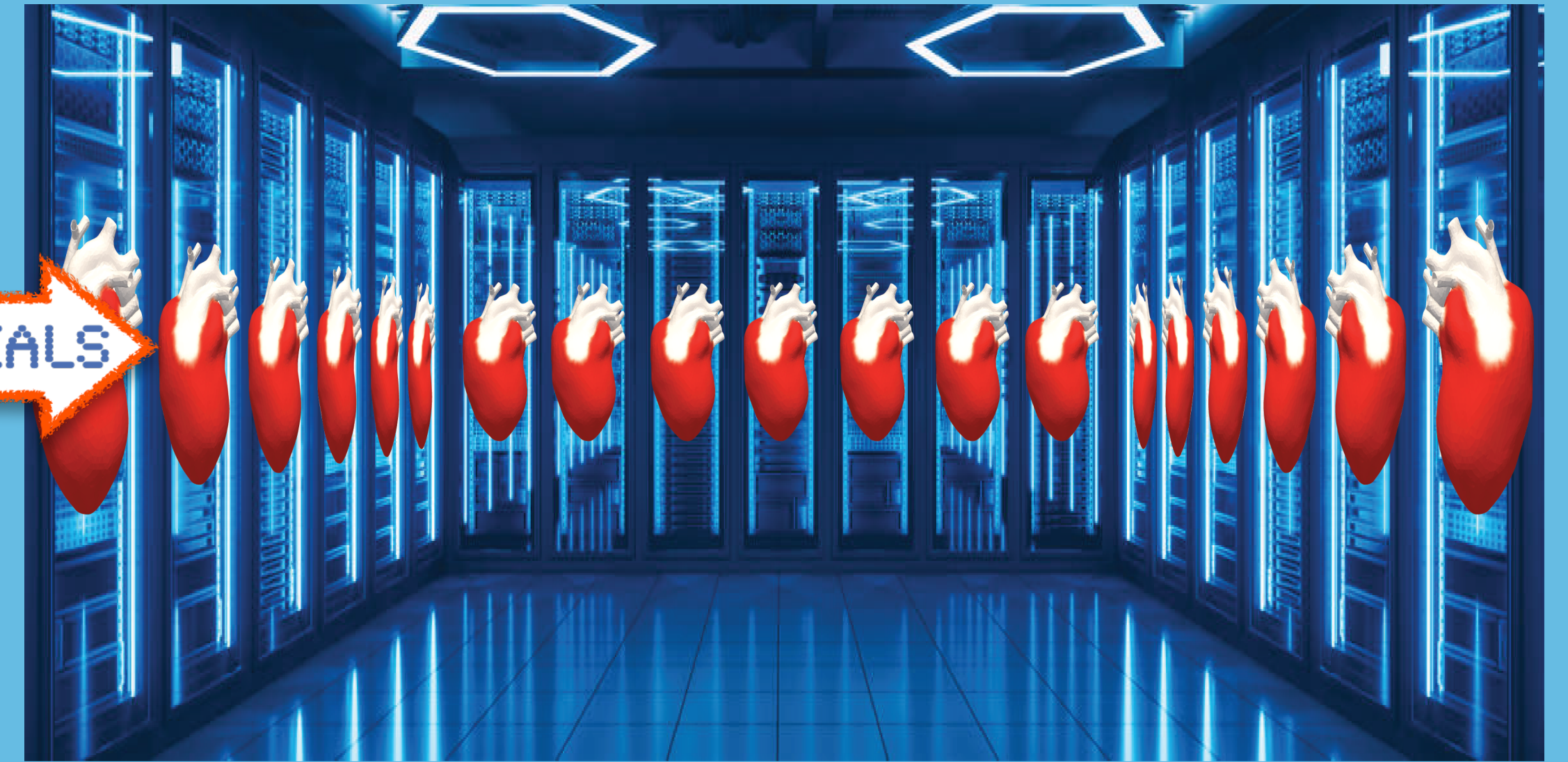
Clinical trials are research studies on human participants to answer specific questions about biomedical interventions such as new drugs or medical devices

Present: In-vivo clinical trials



CARDIOTRIALS

Future: IN-SILICO clinical trials



In-vivo clinical trials



require real patients



no modelization —> more accurate



expensive (due to finite resources)



time-consuming

Only non-invasive measures



In-silico clinical trials



patients are virtual



possibly less accurate, depending on the model



virtually infinite cohort

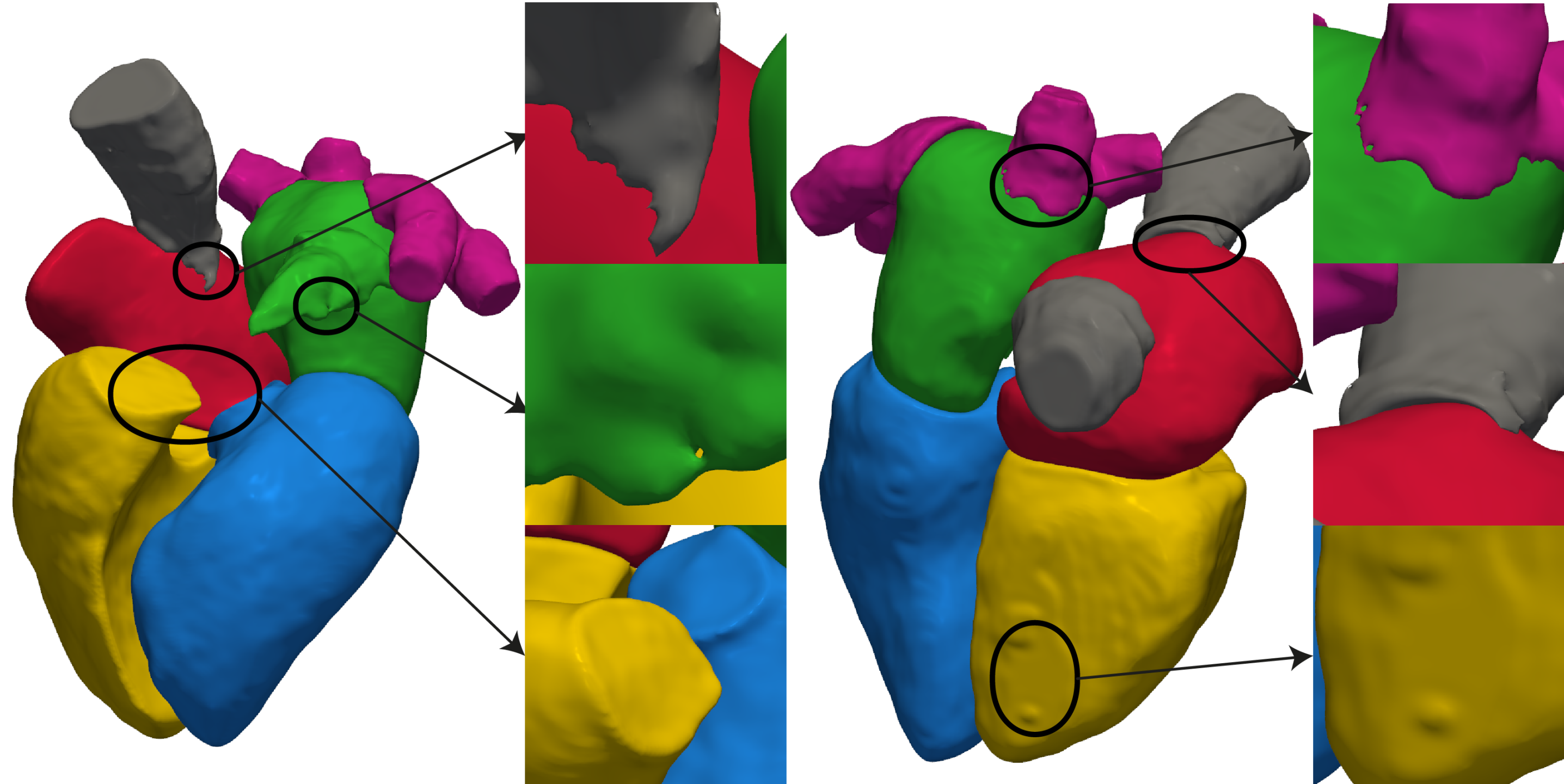


quick (scale well with large computer facilities)

every physical quantity can be measured

From Medical Images to Computational Ready Geometries

Raw medical images (CT-scans, MRI) often come incomplete, with inhomogeneous quality and plenty of artifacts



Existing segmentation tools (made for visualization or diagnosis) produce geometric and topological defects (holes, discontinuities, artificial closures) and lack cohort-level consistency.

Considerable work is needed to produce a computational ready geometry (order of two weeks).

Methods: Registration-Deformation

- A set of **affine transformation** aligns the template and anatomical meshes,

$$\mathcal{R}(\mathcal{M}_{\mathcal{T}}, \{\mathcal{M}_o^k\}_{k=1}^m) \rightarrow \{\mathcal{M}_o^k\}_{k=1}^m$$

- Chamber-based Chamfer functional:**

$$\min_{\mathbf{x}_{\mathcal{T}} \in V_{\mathcal{T}}} \Phi(\mathcal{M}_{\mathcal{T}}, \mathcal{M}_o^k) = \min_{\mathbf{x}_{\mathcal{T}} \in V_{\mathcal{T}}} \sum_{c=1}^4 d_{\text{CD}}^c(\mathcal{M}_{\mathcal{T}}^{(c)}, \mathcal{M}_o^{k,(c)})$$

Optimization is carried out by means of gradient descent:

$$\mathbf{x}_{\mathcal{T}} \leftarrow \mathbf{x}_{\mathcal{T}} - \eta(I + \gamma L)^{-1} \frac{\partial \Phi}{\partial \mathbf{x}_{\mathcal{T}}},$$

The Laplacian enables vertex displacements to be smoothed over the entire mesh

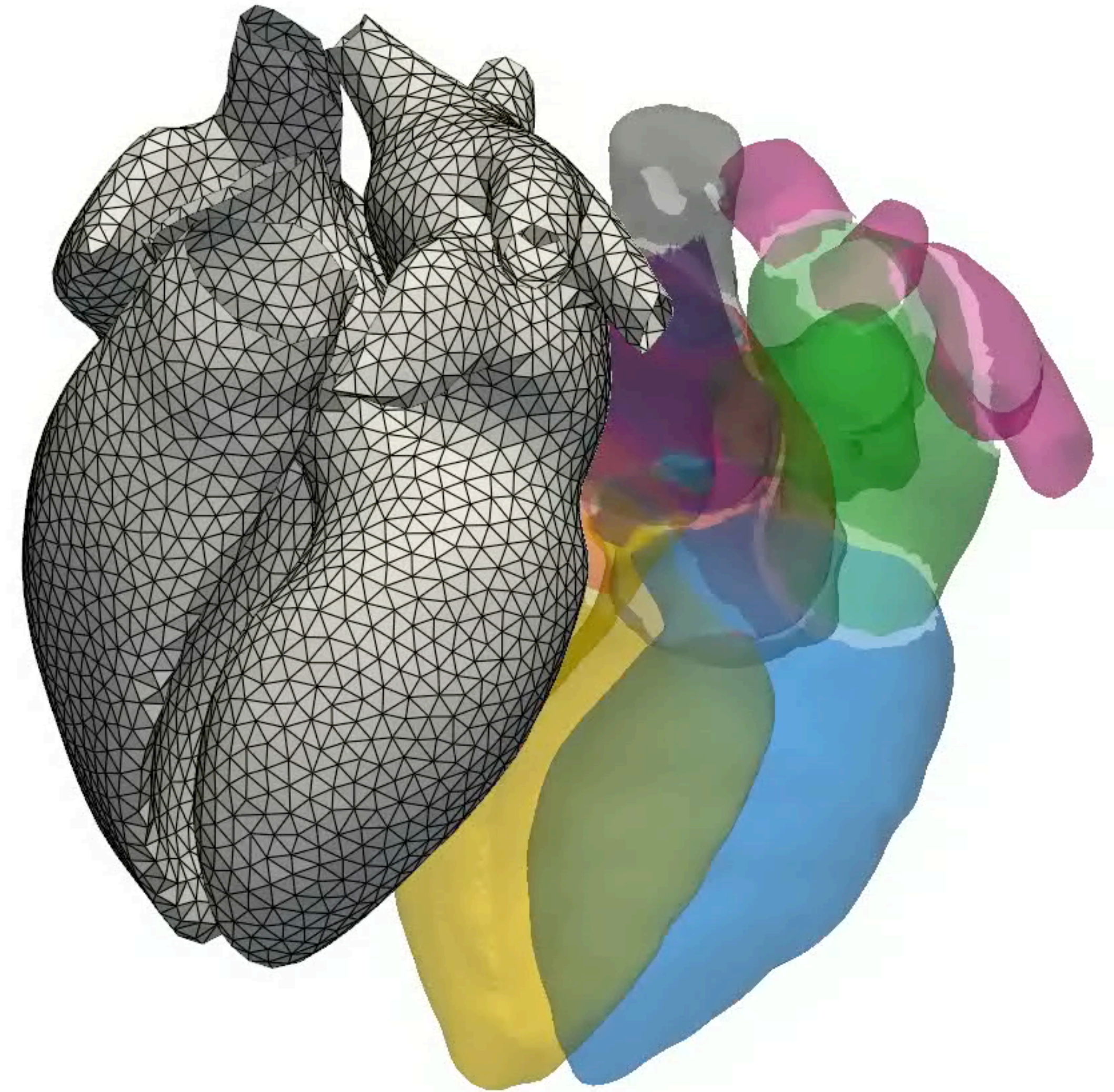
$$d_{\text{CD}}(\mathcal{M}_{\mathcal{T}}, \mathcal{M}_o) = \frac{1}{|V_{\mathcal{T}}|} \sum_{\mathbf{x}_{\mathcal{T}} \in V_{\mathcal{T}}} \min_{\mathbf{x}_o \in V_o} \|\mathbf{x}_{\mathcal{T}} - \mathbf{x}_o\|_2^2 + \frac{1}{|V_o|} \sum_{\mathbf{x}_o \in V_o} \min_{\mathbf{x}_{\mathcal{T}} \in V_{\mathcal{T}}} \|\mathbf{x}_{\mathcal{T}} - \mathbf{x}_o\|_2^2.$$

Chamfer distance evaluates the geometric shape correspondance.

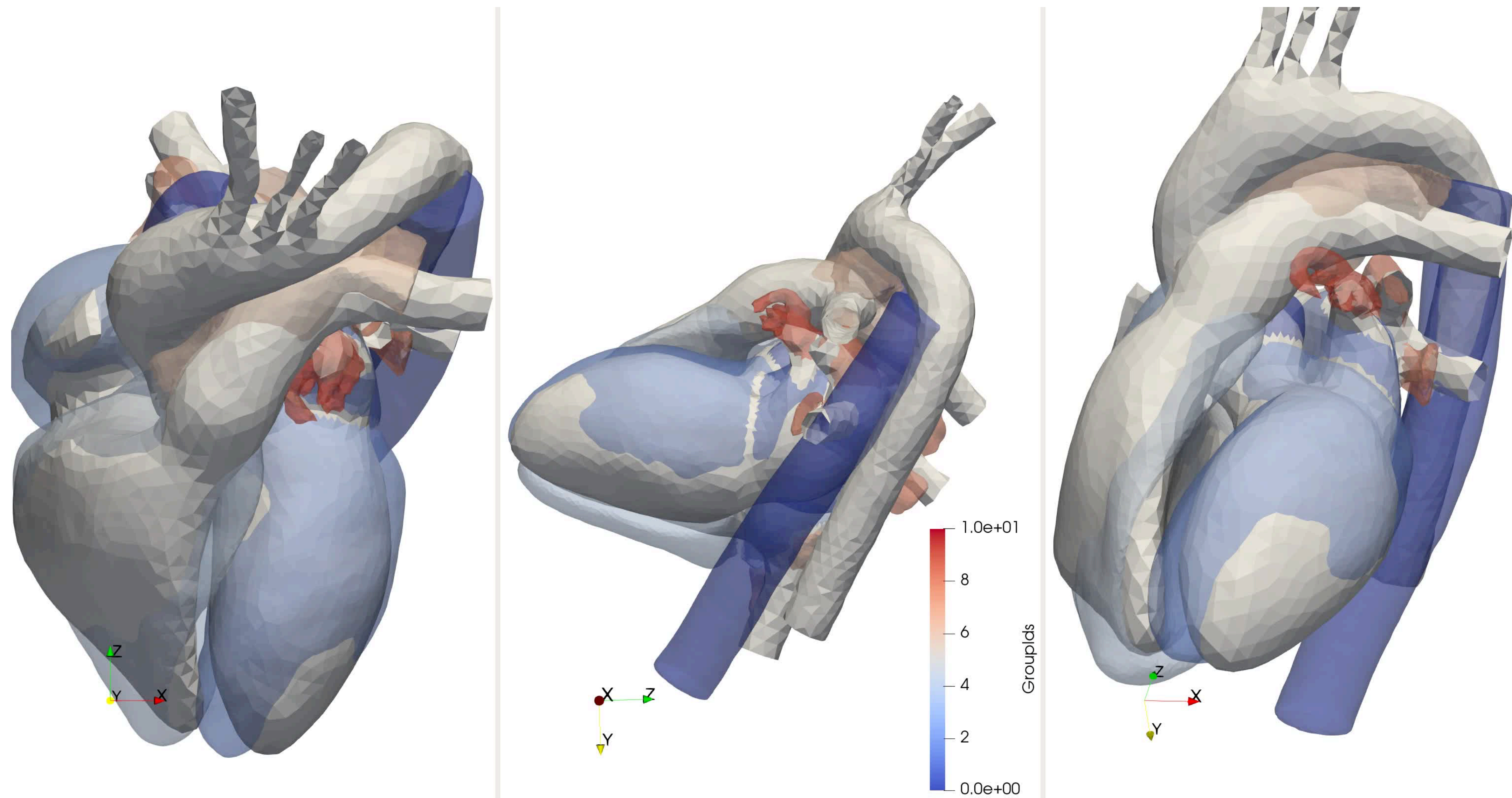
- After deformation, we retrieve the set of **deformed meshes**. Then, the inverse set affine transformation is applied, retrieving the **registered meshes**

$$\min_{\mathbf{x}_{\mathcal{T}} \in V_{\mathcal{T}}} \Phi(\mathcal{M}_{\mathcal{T}}, \mathcal{M}_o^k) \rightarrow \mathcal{M}_d^k \quad \mathcal{R}^{-1}(\mathcal{M}_{\mathcal{T}}, \{\mathcal{M}_d^k\}_{k=1}^m) \rightarrow \{\mathcal{M}_r^k\}_{k=1}^m$$

The template is deformed onto the patient specific model to yield its computational ready geometry (the whole procedure takes 15-30 minutes)



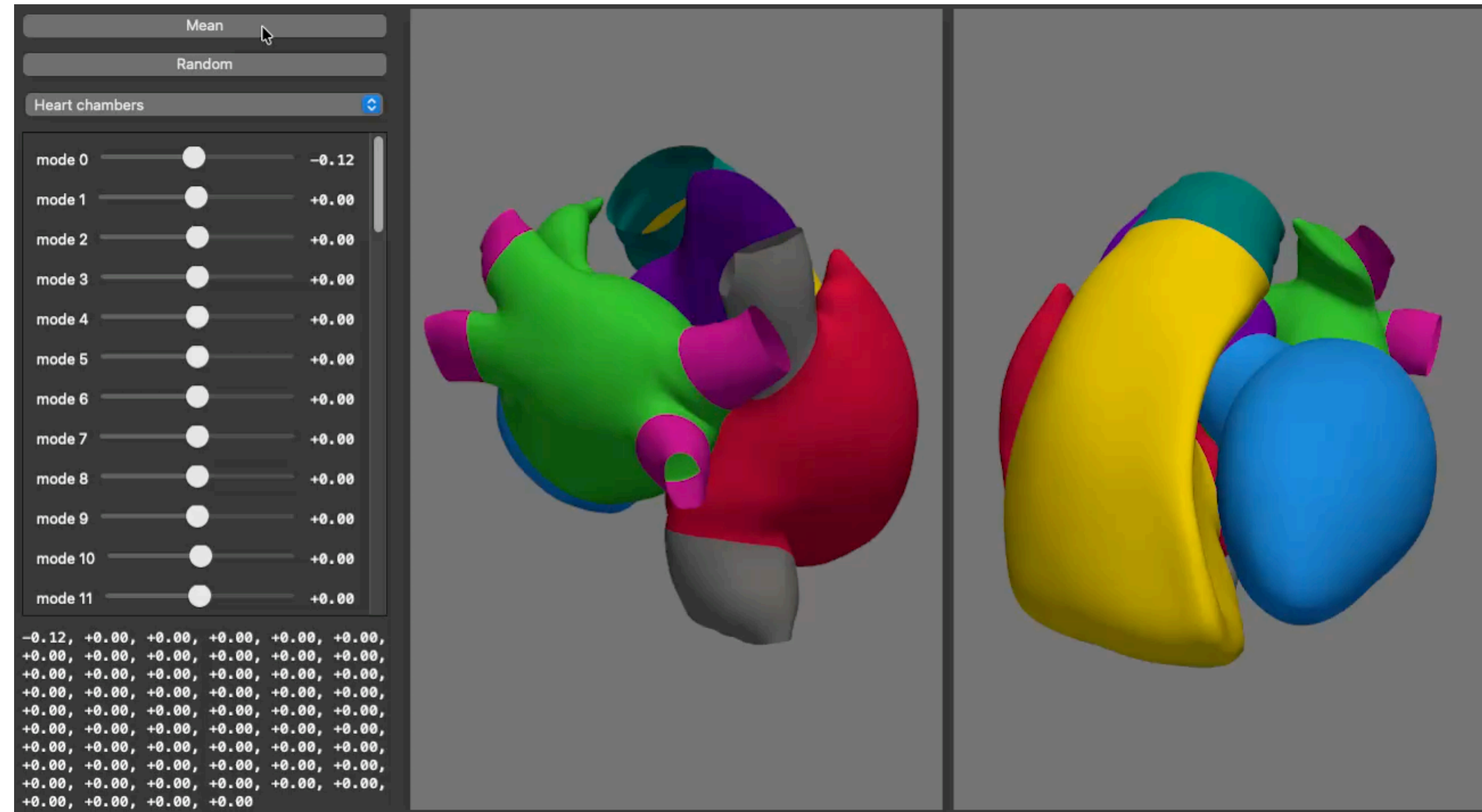
Methods: Registration-Deformation



Exploring the new latent representation

Latent space is “convenient” but less interpretable

- Study the probability distribution of shapes in this new latent space
 - We used a simple Gaussian Mixture model
- Improve the interpretability of the new representation
 - Compute sensitivity of well-known geometrical descriptors with respect to each latent space parameter



Summary

Overview of the Computational Model

- APP1 Optimal **cardiac resynchronization therapy**:
an in-silico investigation using a whole-heart model
- APP2 Investigation of the hemodynamics features past
bioprosthetic valves: effects of the valve deployment
location
- APP3 A data-driven computational methodology
for assessing **ventricular ablation procedures**
- APP4 Statistical shape modelling of the heart and
virtual cohorts generation