S4 Details on the electro-mechanical simulations

For the electro-mechanical simulations we followed a similar pipeline as in [1]. We ran EP simulations with the reaction-eikonal model [2], in which the depolarisation is ruled by the eikonal equation:

$$\|\nabla t_a\|_{\mathbf{V}} = 1, \tag{1}$$

where t_a is the wavefront arrival time and $\mathbf{V}$ is the (symmetric positive definite) tensor containing the squared wave velocities associated to the fibre ($\mathbf{f}_0$), sheet ($\mathbf{f}_0$) and sheet normal ($\mathbf{n}_0$) directions.

As an initial condition, we stimulated points in the endocardium of the mesh, using the UVC. We set the constraint on parameter $Z \leq 0.33$, to limit the stimulation to the bottom part of the ventricles simulating the outbreak of the His-Purkinje network terminals. To simulate the early endocardial activation [3], a fast endocardial conduction (FEC) layer was labelled in the meshes, where we set up a higher conduction velocity. The layer covered the 70% of the bottom apicobasal length ($Z \leq 0.7$ in UVC) and is 1 element thick in the endocardium. These parameters are related to the most dense part of the Purkinje network [4]. The only conductive tissue was the ventricular myocardium - activation was not simulated in the atria. See table A for the values of all the EP parameters. Details on the solver and numerical scheme for the EP have been published previously elsewhere [2]. The solver has also been validated in N-version benchmark studies [5].

Parameter	**Value (reference)**
CV (fibre)	0.8 m/s [6]
CV (transfibre)	0.23 m/s [7]
CV (FEC)	5.6 m/s [8]
Depolarisation threshold	-60 mV [9]

Table A. Parameter values for the reaction-eikonal model (used only in the ventricles). CV stands for conduction velocity and FEC for fast endocardial conduction layer.

We simulated the large deformation mechanics in a Lagrangian reference system [10]. To simulate stresses in the reference configuration we used the model

$$\mathbf{S}_{\text{pas}} = 2\frac{\partial \Psi(\mathbf{C})}{\partial \mathbf{C}} \tag{2}$$

where $\mathbf{S}_{\text{pas}}$ is the passive component of the second Piola-Kirchhoff stress tensor and $\mathbf{C}$ is the right Cauchy-Green deformation tensor.

The ventricular myocardium was modelled as a hyperelastic transversely isotropic material with Guccione's strain energy function [11]:

$$\mathbf{\Psi}(\mathbf{C}) = \frac{\kappa}{2}(\ln J)^2 + \frac{a}{2}\left(e^Q - 1\right), \tag{3}$$

$$Q = b_f E_{ff}^2 + b_t\left(E_{ss}^2 + E_{nn}^2 + 2E_{sn}^2\right) + 2b_{fs}\left(E_{fs}^2 + E_{fn}^2\right), \tag{4}$$

with

$$E_{ij} = \mathbf{i}_0 \cdot \overline{\mathbf{E}}\mathbf{j}_0, \tag{5}$$

$$\overline{\mathbf{E}} = \frac{1}{2}\left(\overline{\mathbf{C}} - \mathbf{I}\right), \tag{6}$$

$$\overline{\mathbf{C}} = J^{-2/3}\mathbf{C} \tag{7}$$

being $\overline{\mathbf{E}}$ the modified isochoric Green-Lagrange strain tensor and J the determinant of the Jacobian matrix of the deformation gradient tensor, for $\mathbf{i}, \mathbf{j} = \mathbf{f}, \mathbf{s}, \mathbf{n}$. See table B for the values of the parameters of this constitutive law.

Parameter	Value
a	1.7 kPa
b_f	8
b_{fs}	4
b_t	3
κ	1000 kPa

Table B. Parameter values for the Guccione's law in the ventricle, extracted from [12].

The remaining tissues were modelled as non-contracting neo-Hookean materials [13] following the strain energy function [1]:

$$\mathbf{\Psi}(\mathbf{C}) = \frac{\kappa}{2}(J-1)^2 + \frac{c}{2}(\mathrm{tr}(\overline{\mathbf{C}}) - 3), \tag{8}$$

where tr is the trace and c and κ the material parameters. The values of κ and c for the valves was set to 1000 kPa (as an arbitrary high value) to enforce material incompressibility. See table C for the values of all the parameters of this constitutive law.

Parameter	Value [reference]
κ	1000 kPa
c_{atria}	7.45 kPa [14]*
c_{veins}	7.45 kPa [14]*
c_{aorta}	26.66 kPa [15]*
c_{PA}	3.7 kPa [16]*
c_{valves}	1000 kPa

Table C. Parameter values for the Neo-Hookean material model used in the atria and veins. * extracted as average values of the reference cited.

To simulate the active stress of the ventricles triggered by electrical activation, we used the phenomenological activation-based Tanh Stress model [17, 18]. See table D for the values of the parameters of this constitutive law. Details on the solver and numerical scheme for the EM have been published previously elsewhere [19]. The solver has also been validated in N-version benchmark studies [20].

Parameter	Value)
EM delay	20 ms
Peak isometric tension	120 kPa
Time constant relaxation	50 ms
Time constant contraction	50 ms
Duration of transient	550 ms

Table D. Parameter values for the Tanh strees active tension model (only applied to the ventricles). All the values were hand-tuned.

The preload of the ventricles was modelled via constant atrial pressures, while the afterload was modelled with two three-element Windkessel models, one for the systemic circulation and one for the pulmonary circulation [21]. For the values of the initial pressures see table E.

Parameter	Value (reference)
LV endocardial pressure	1.6 kPa [22]
RV endocardial pressure	0.8 kPa [23]
Aortic pressure	77 mmHg [24]
Pulmonary artery pressure	17.4 mmHg [25]

Table E. Parameter values for initial pressures.

The value of the backward resistances for all the valves were set to either 1000 or 10000 mmHg/mL/s as arbitrary high values to avoid backflow or regurgitation. The forward resistances of the valves have been set to 0 mmHg/mL/s for the aorta and pulmonary artery to achieve numerical convergence and an EF in a healthy range. In the case of the mitral and tricuspid valves, a value of 0.05 mmHg/mL/s was set to allow for physiological filling. Similar values were also used in [26].

Parameter	Value [reference]
Aortic valve: serial resistor	0.03 mmHg s/mL [27]
Aortic valve: parallel resistor	0.63 mmHg s/mL [27]
Aortic valve: capacitor	5.16 mL/mmHg [27]
Mitral valve peak forward flow resistance	0.05 mmHg/mL/s
Mitral valve peak backward flow resistance	1000 mmHg/mL/s
Aortic valve peak forward flow resistance	0 mmHg/mL/s
Aortic valve peak backward flow resistance	10000 mmHg/mL/s
Pulmonary valve: serial resistor	0.015 mmHg s/mL [28]
Pulmonary valve: parallel resistor	0.1008 mmHg s/mL [28]
Pulmonary valve: capacitor	21.156 mL/mmHg [28]
Tricuspid valve peak forward flow resistance	0.05 mmHg/mL/s
Tricuspid valve peak backward flow resistance	1000 mmHg/mL/s
Pulmonary valve peak forward resistance	0 mmHg/mL/s
Pulmonary valve peak backward resistance	10000 mmHg/mL/s

Table F. Parameter values for the 3-elements Windkessel model used for the circulatory system (and valves).

The right pulmonary veins and the superior vena cava were assigned omni-directional spring boundary conditions, to allow spring-like motion within them [29, 30]. To achieve a physiologically plausible motion with downward motion of the atrioventricular plane and limited motion of the apex, we simulated the pericardium applying springs (Robin boundary conditions) normally to the epicardial surface [26].

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