Patient-specific modeling of heart and circulation

Theo Arts

t.arts@bf.unimaas.nl

Maastricht University, *Maastricht University Hospital

**University of Tecnology, Eindhoven

The Netherlands

*Tammo Delhaas, Frits Prinzen, Peter Bovendeerd**,

(Biophysics, Pediatric Cardiology, Physiology, Engineering)

Nov, 2005

Tools

Modeling:

- Finite Element Model Electro-Mechanics of the heart
- CircAdapt = whole circulation model
- Incorporation of adaptation with Self-structuring

Disciplines and techniques:

- Biomedical Engineering: MRI-tagging (strain mapping), imaging, modeling
- Physiology: Animal experiments
- Cardiology, Pediatrics, Radiology: Patients

The group

Requirements

1) Bedside-fast simulation and parameter fit
2) Parameters:
   - Minimum number
   - Crucial
   - Easy/Non-invasively obtainable

Strategy

1) Patient identification
   - Most likely state after each added information
   - Simulation of current state
2) Simulation of treatment → Outlook: Selection of best

A Start:

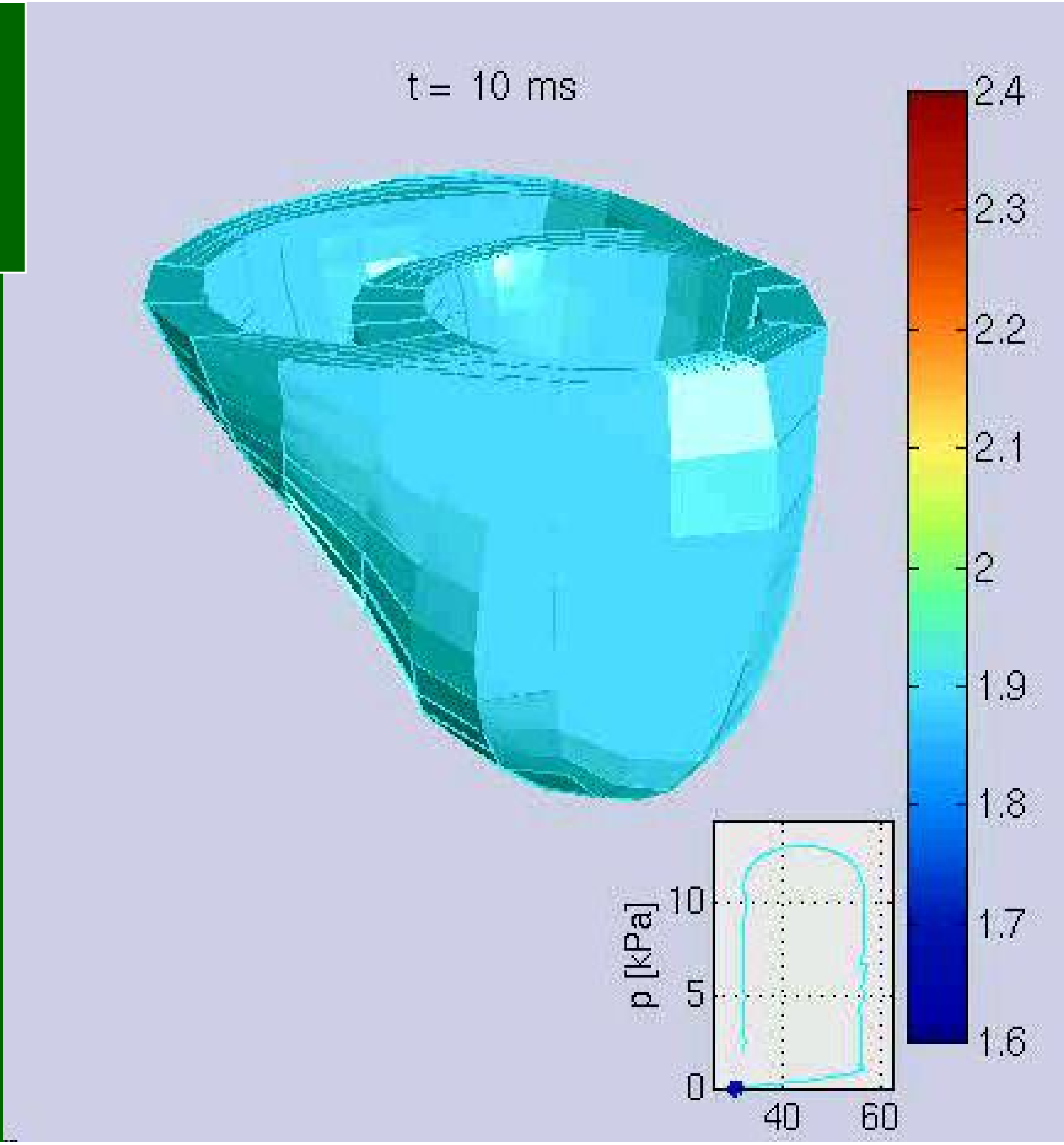
Aortic stenosis: Cardiac deformation

Fetal circulation / birth

FEM of LV+RV
- paced, depolarization wave with pacing
- full cardiac cycle
(R. Kerckhoffs, 2003)

left+right ventricle movie:

sarcomere length & pV



Parameter reduction: Self-structuring by simulation of adaptation

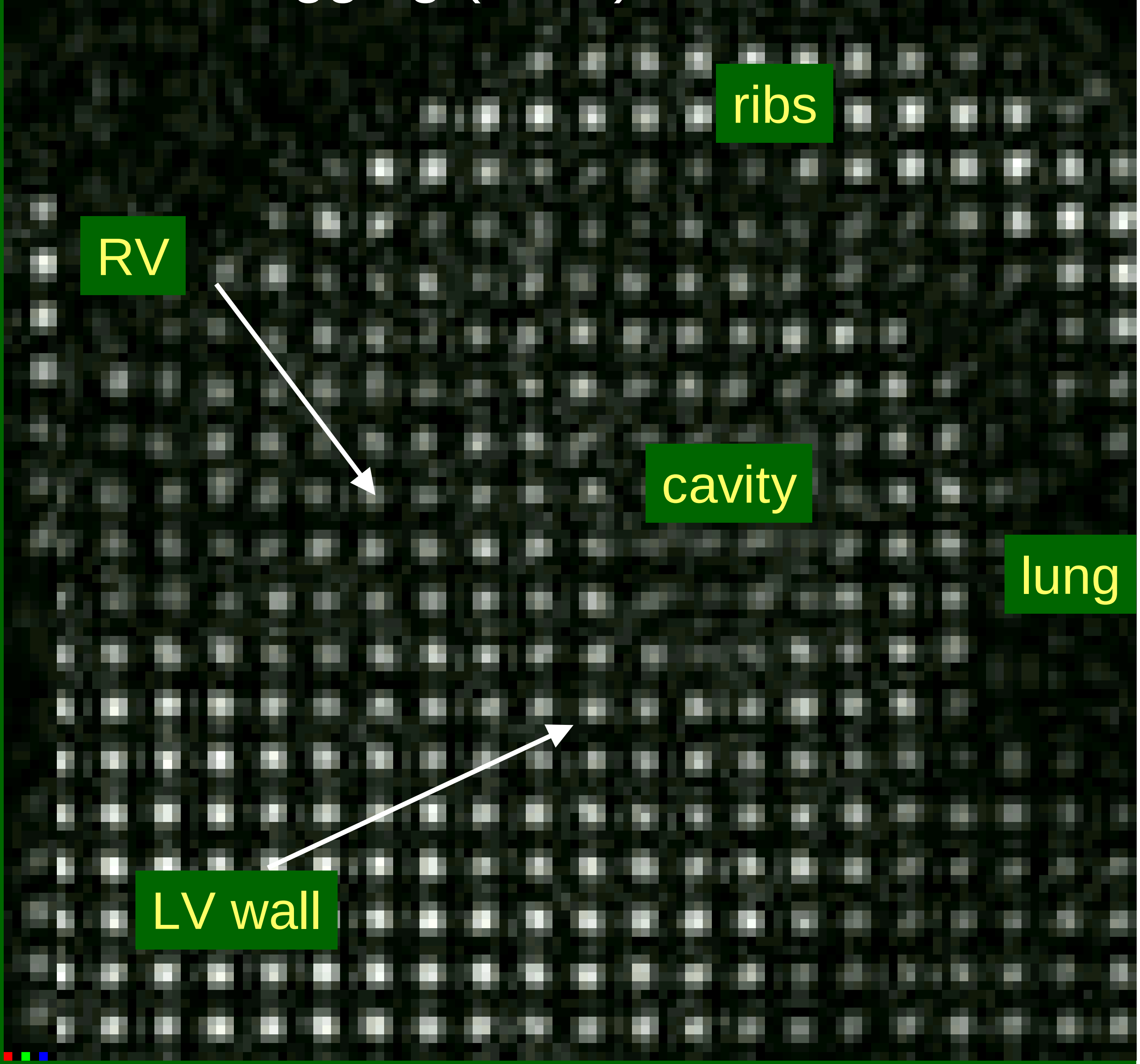
Modeling structure of myocardial wall with adaptation:

- mechanical load determines
  - wall mass
  - fiber orientation
  - sheet orientation
- size and shape of blood vessels

Enormous reduction of number of parameters

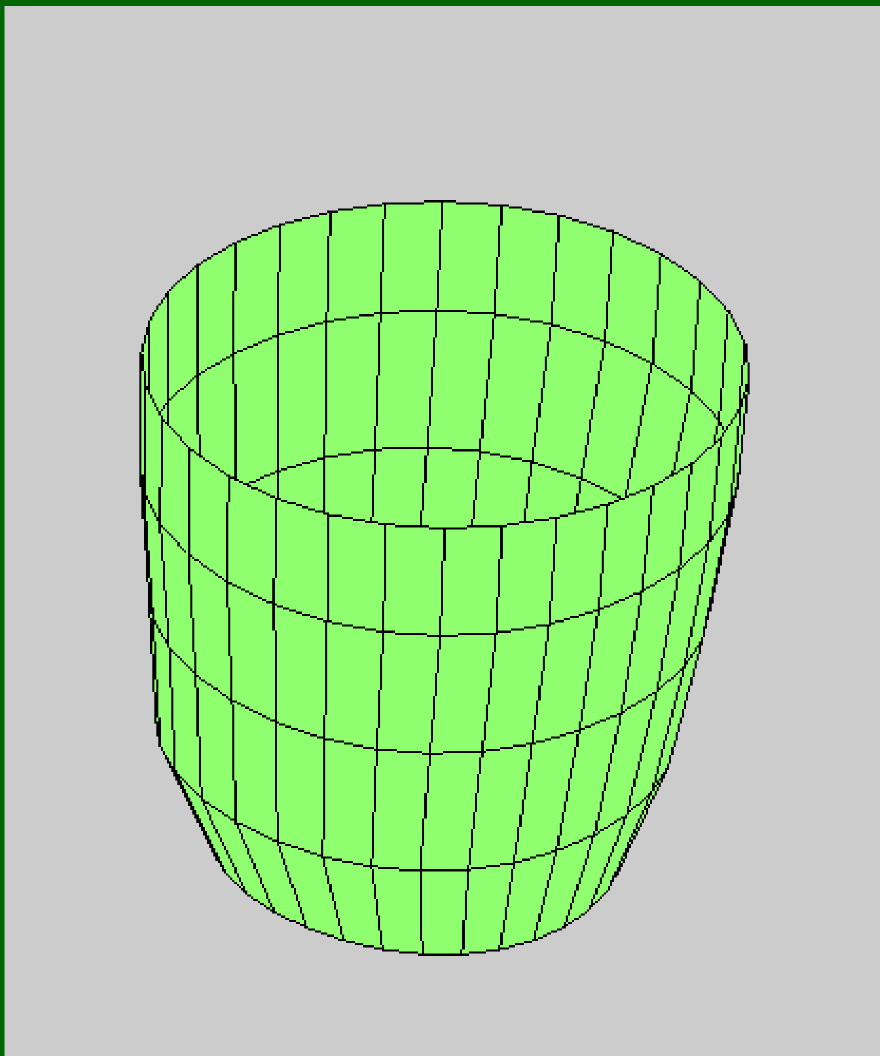
Especially in pathology: Generally 1 basic cause of pathology, followed by physiological adaptation processes.

Magnetic Resonance Tagging (MRT)

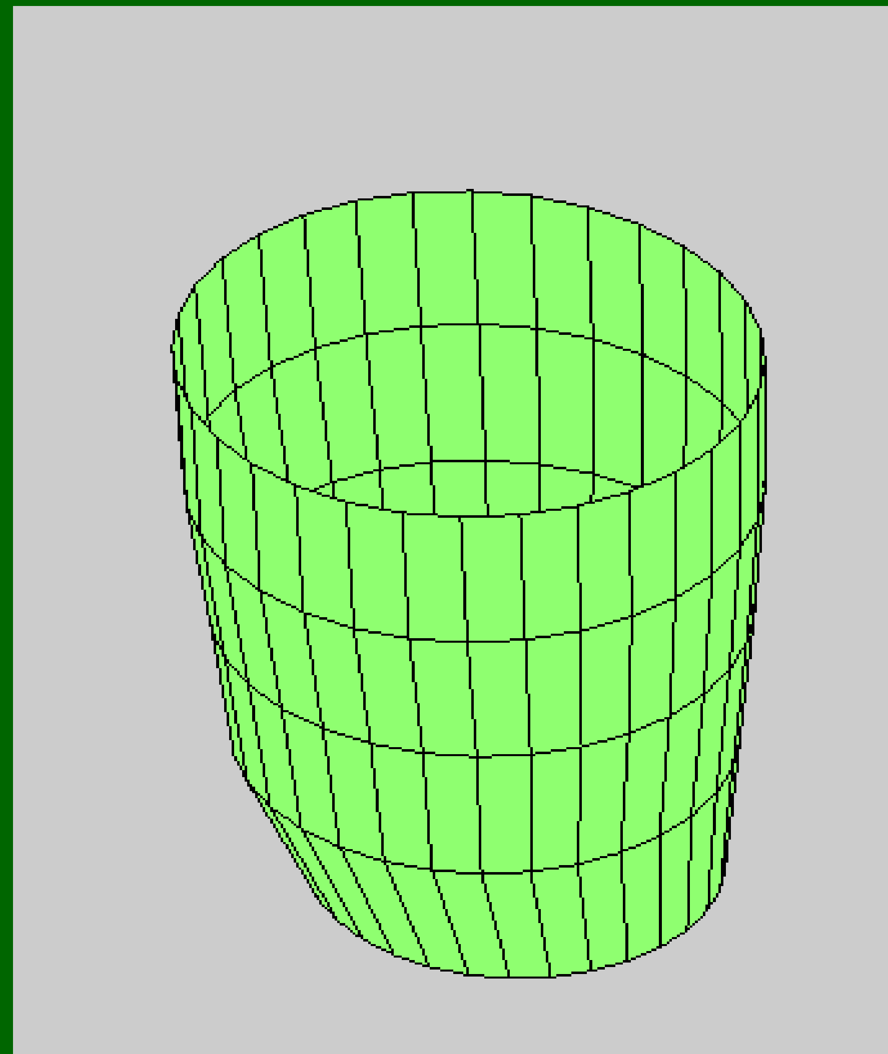


Rotation and deformation of the heart can be quantified

Mri-Software: Midwall motion and circumferential strain

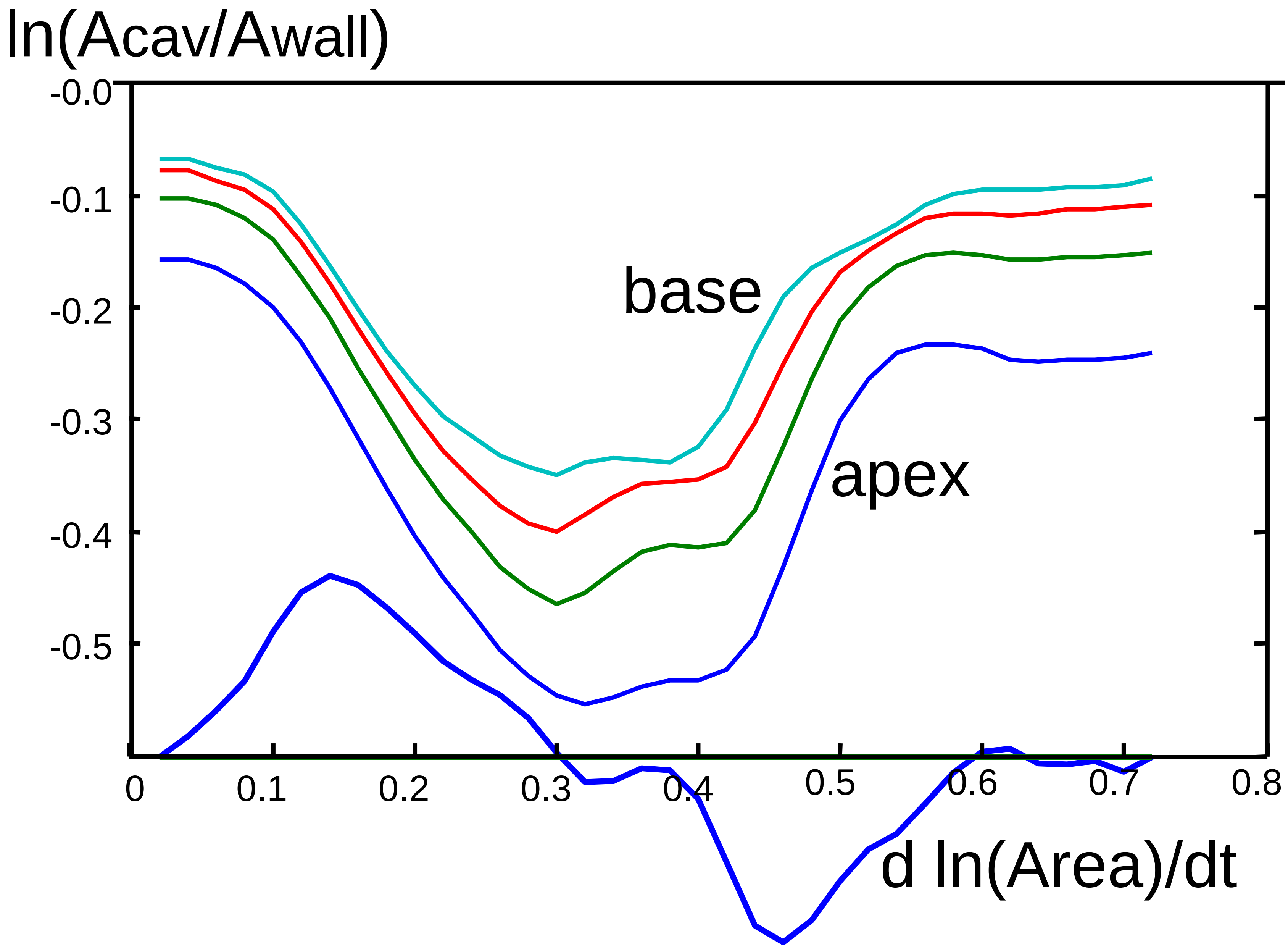


normal

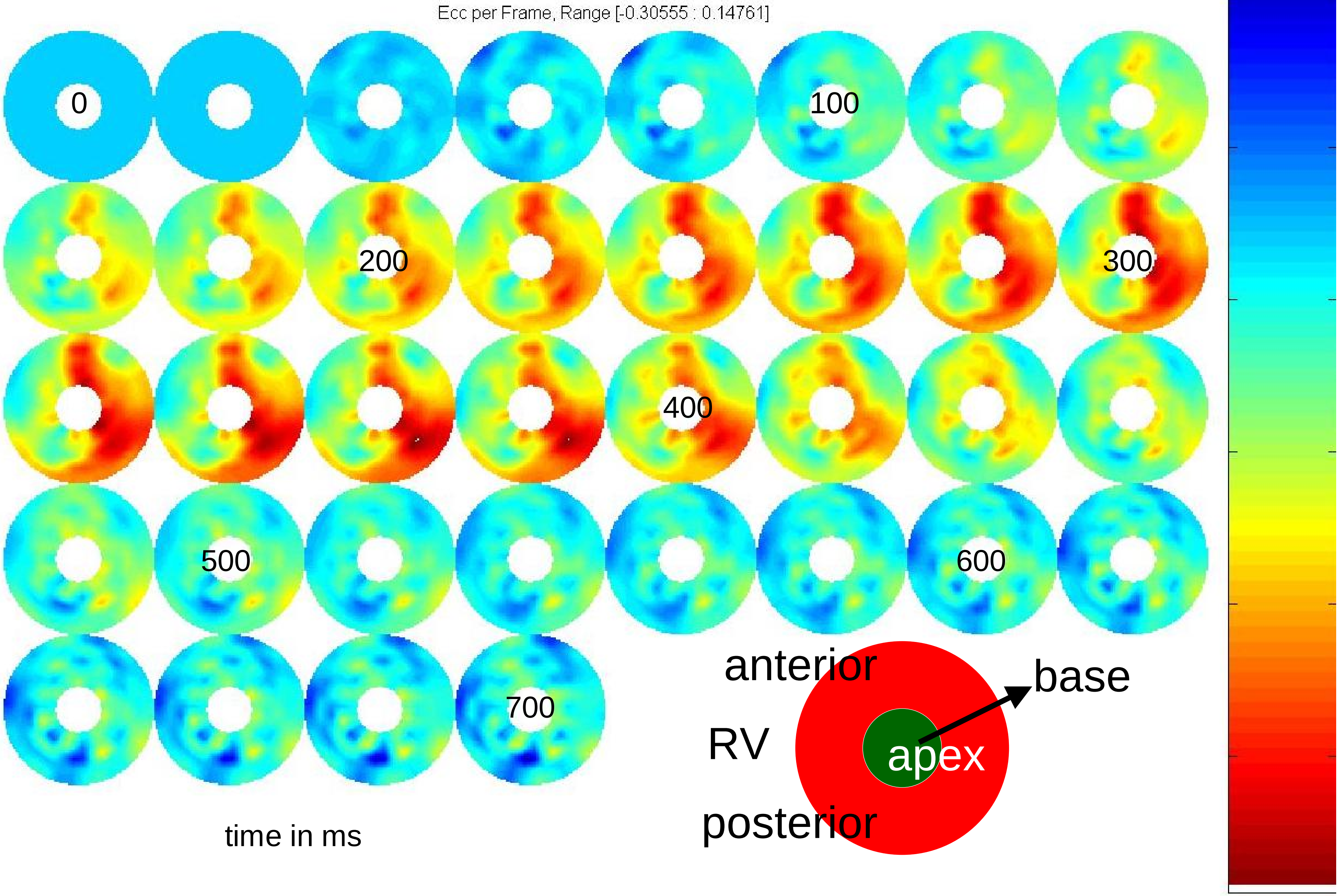


infarction

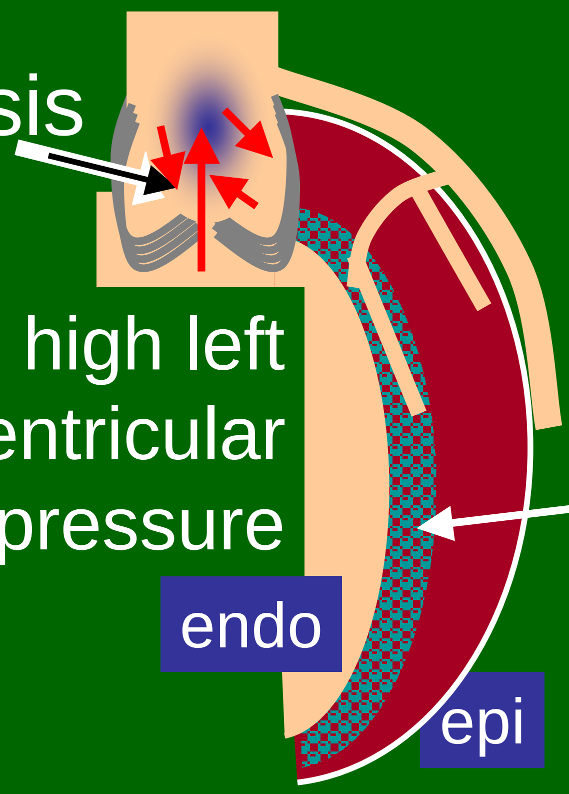
Cavity area and d~/dt



LV circumferential Strain

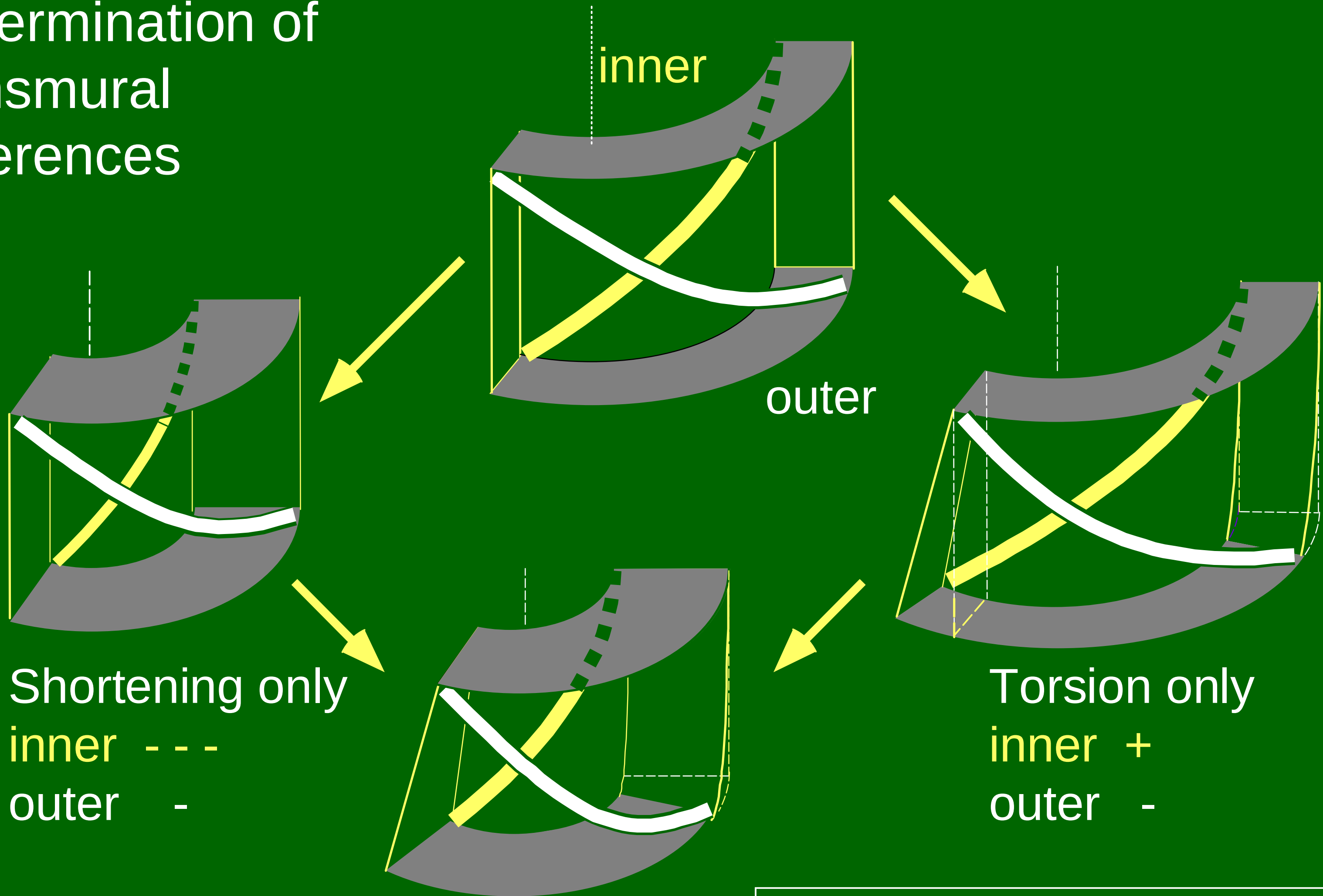


Aortic stenosis → Subendocardial dysfunction



Wall segment model (cylindrical)

Determination of transmural differences



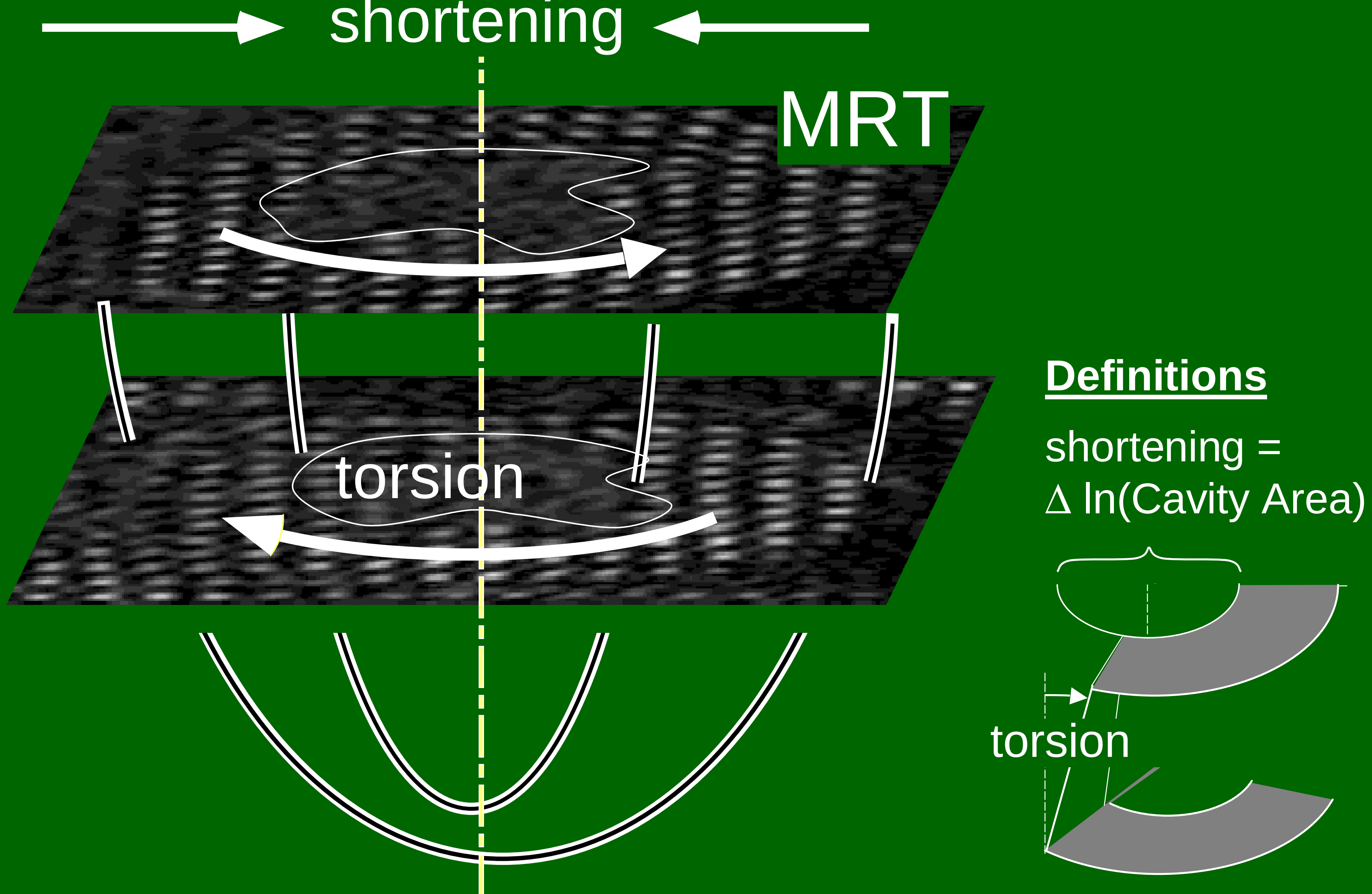
Shortening only
inner - - -
outer -

Torsion only
inner +
outer -

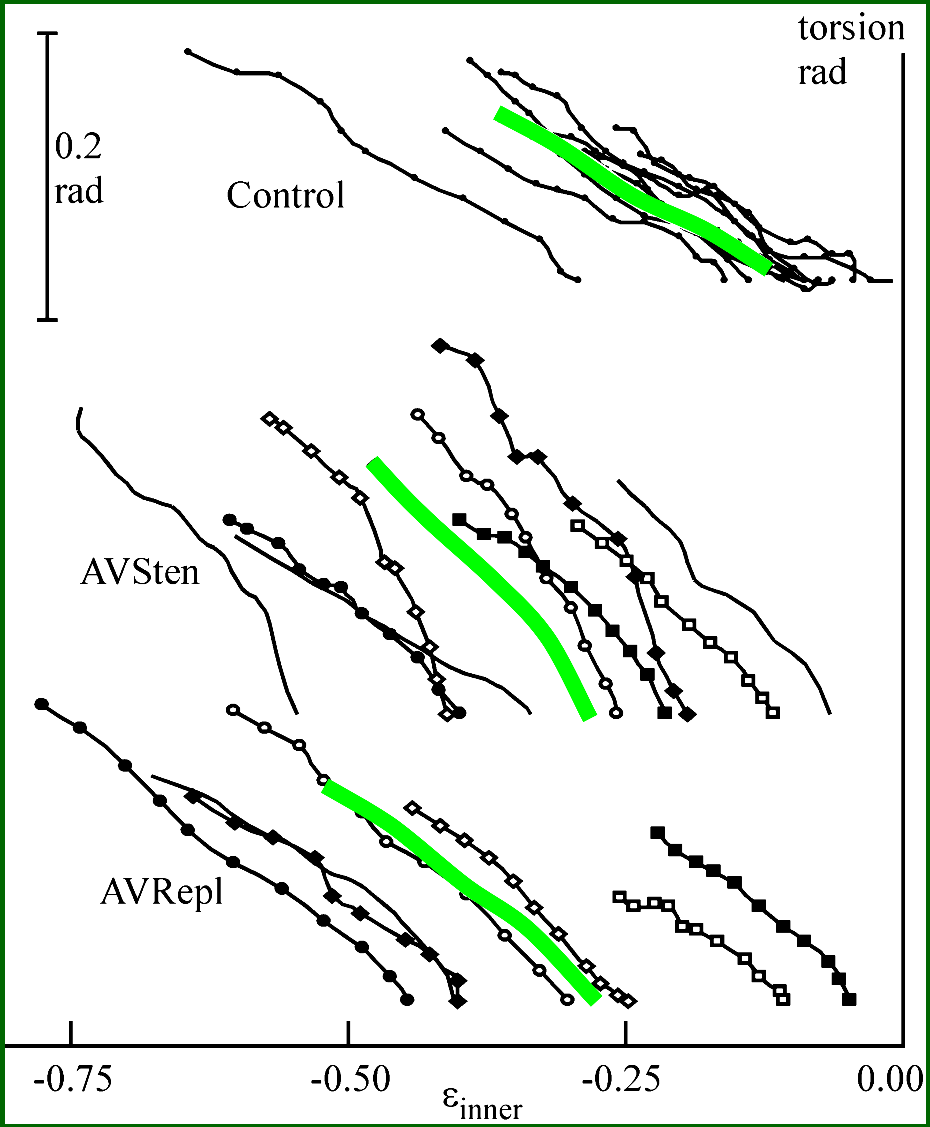
Shortening & Torsion
inner - - outer - -

- Torsion tuned to Shortening
- IF Subendocardial ↓
TSR=(Torsion– / Shortening↓) – –

Torsion/Shortening measurement



TSR in Control and AVS patients



Inner wall strain $\ln(L/L_{Vc=Vw})$

TSR=slope of torsion versus inner wall strain

- Dimensionless

- Species independent

- Expresses transmural difference in contractile function

Control= healthy young

AVSten= Aortic Valve Stenosis

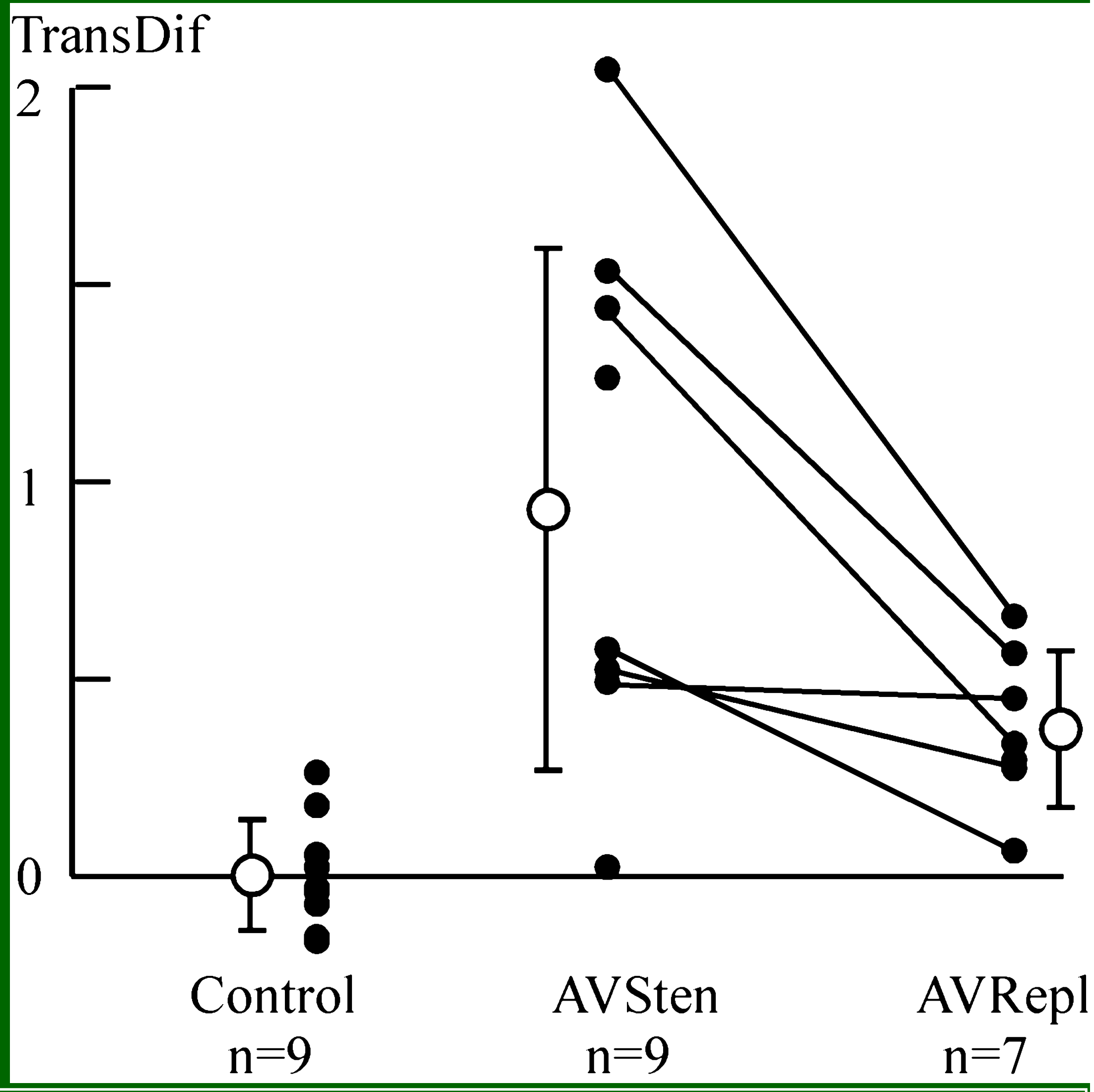
AVRepl= 3 mo after aortic valve replacement

From TSR
to TransDif

Model:

Torsion/Shortening (TSR)
↓
normalized transmural difference in myofiber shortening (TransDif)

TransDif=Difference/Mean



Van der Toorn A et al. Am J Physiol. 2002;283:H1609-1615

CircAdapt model

a. Modeling of circulation
- Lumped model in modules:
  chambers, tubes, valves

b. Adaptation of modules to load

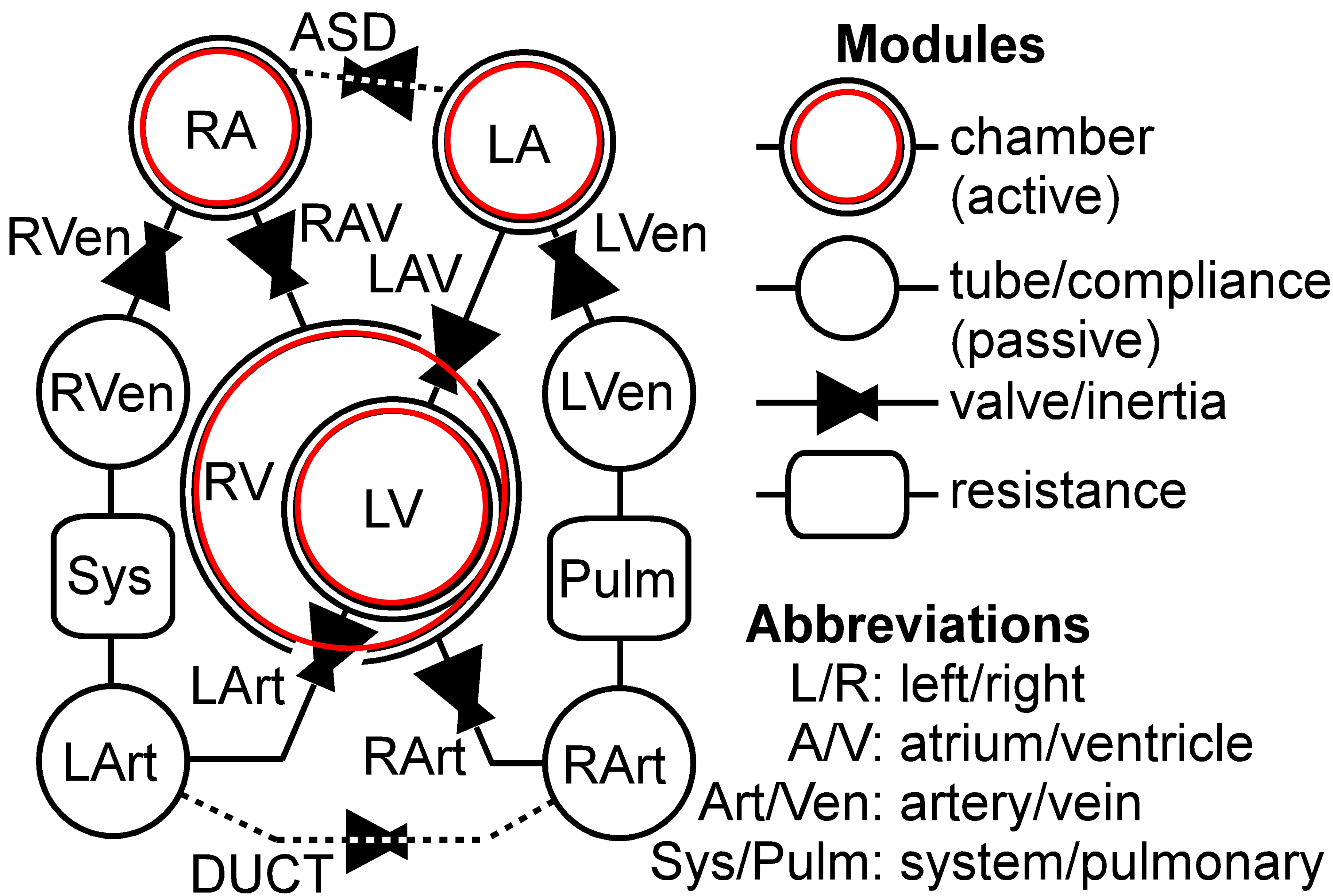
c. Patient specific modeling with adaptation, example

Search:

Google + keyword 'CircAdapt'

hit 'AJP': Arts T et al. *Am J Physiol*. 2005;288:H1943-H1954
hit 'Biophysics': source code

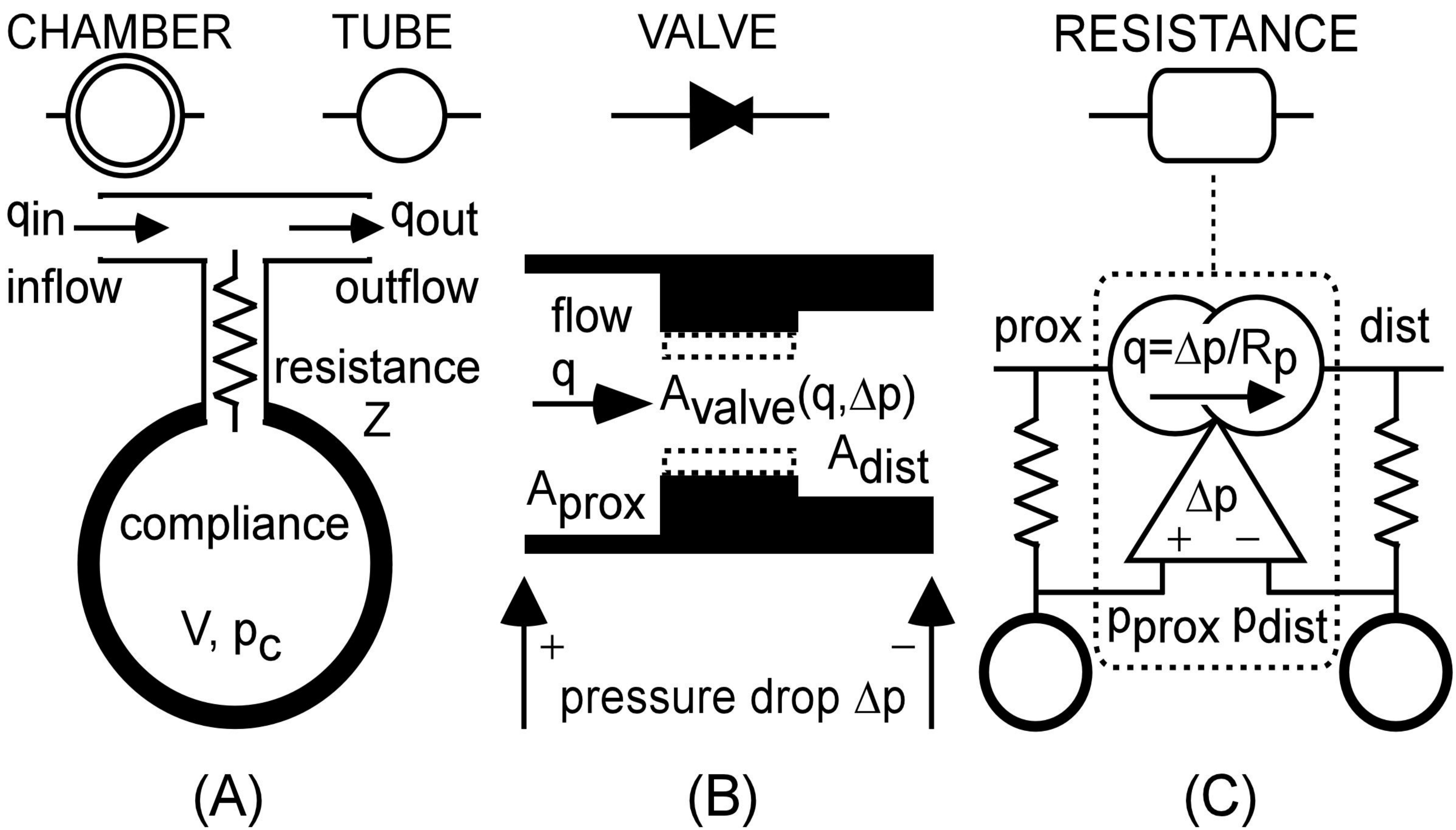
Circulation in modules



Abbreviations

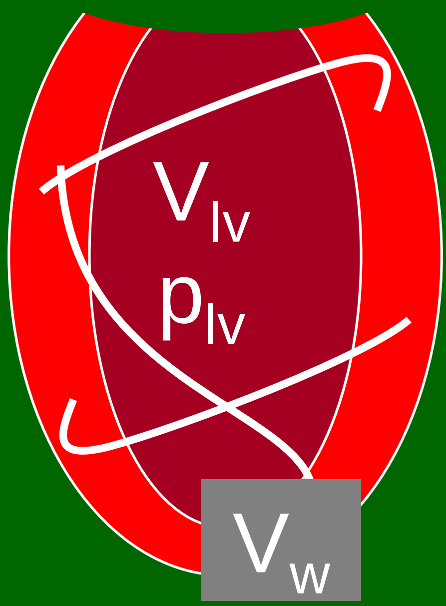
L/R: left/right
A/V: atrium/ventricle
Art/Ven: artery/vein
Sys/Pulm: system/pulmonary

Plumbing modules

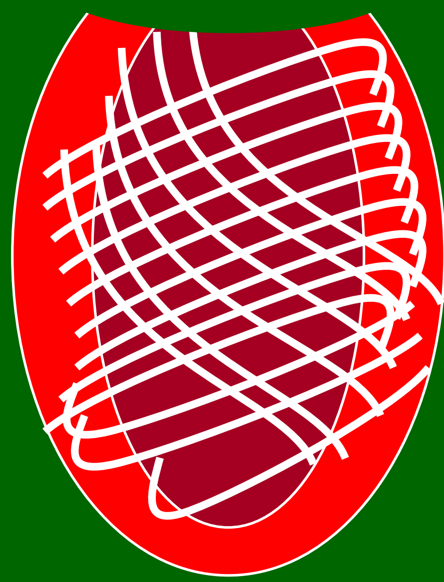


1-fiber model of a thick-walled cavity (chamber or blood vessel)

ventricle (LV)



is wrapped in 1 myofiber



$$\sigma_f = p_{lv}\left(1 + 3\,V_{lv}/V_w\right)$$

$$\Delta\varepsilon_f = \tfrac{1}{3}\,\Delta\ln\left(1 + 3\,V_{lv}/V_w\right)$$

σ_f= myofiber stress

$\Delta\varepsilon_f$= myofiber strain

p_{lv}= LV pressure

V_{lv}= LV volume

V_w= wall volume

FEM model confirms: Shape is practically irrelevant

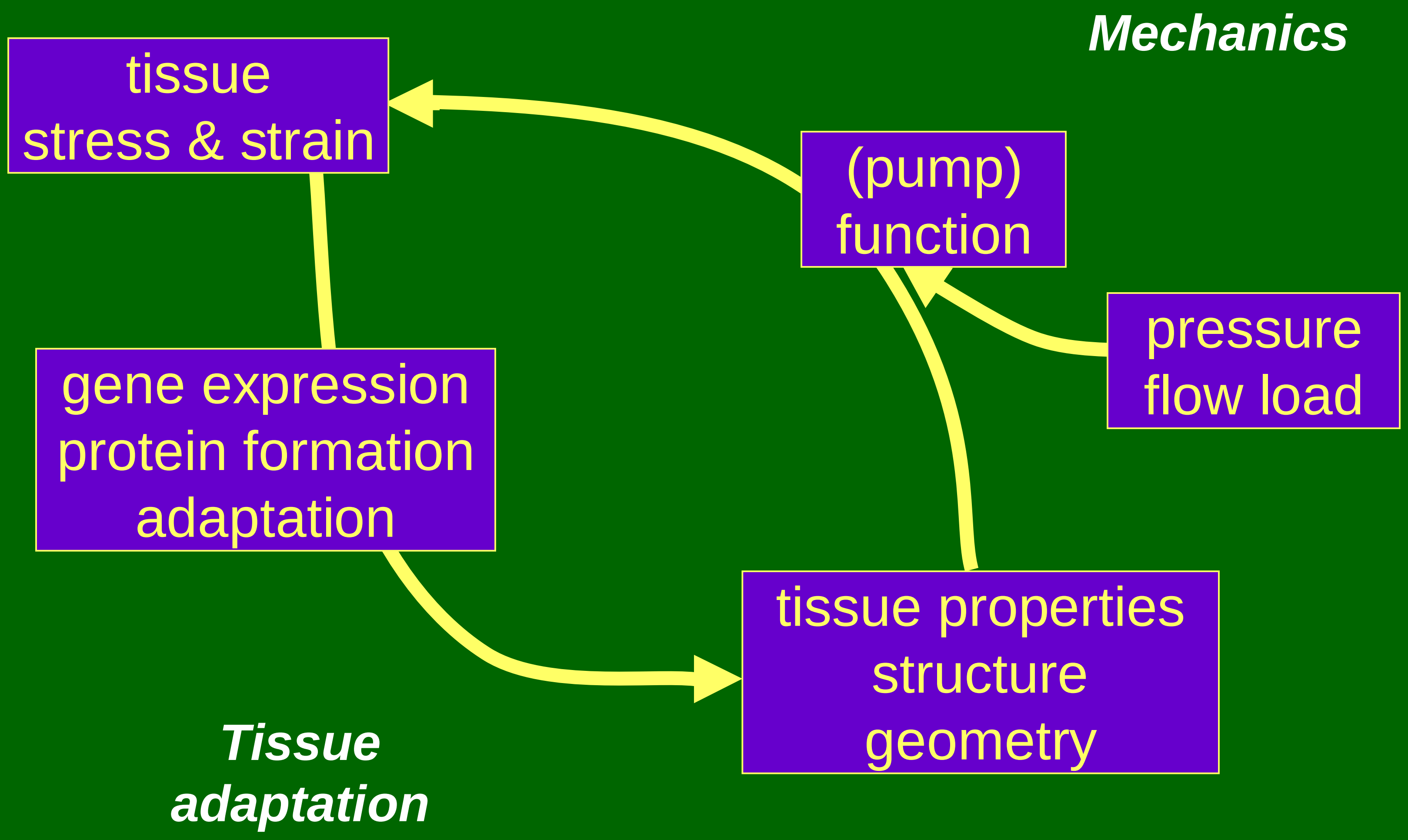
CircAdapt model

a. Modeling of circulation
- Lumped model in modules:
  chambers, tubes, valves

b. Adaptation of modules to load

c. Patient specific modeling with adaptation, example

Cardiovascular adaptation to mechanical load



Laws of adaptation

Adaptation of Cavity

- Contractility → Hypertrophy
- Deformation → Dilatation

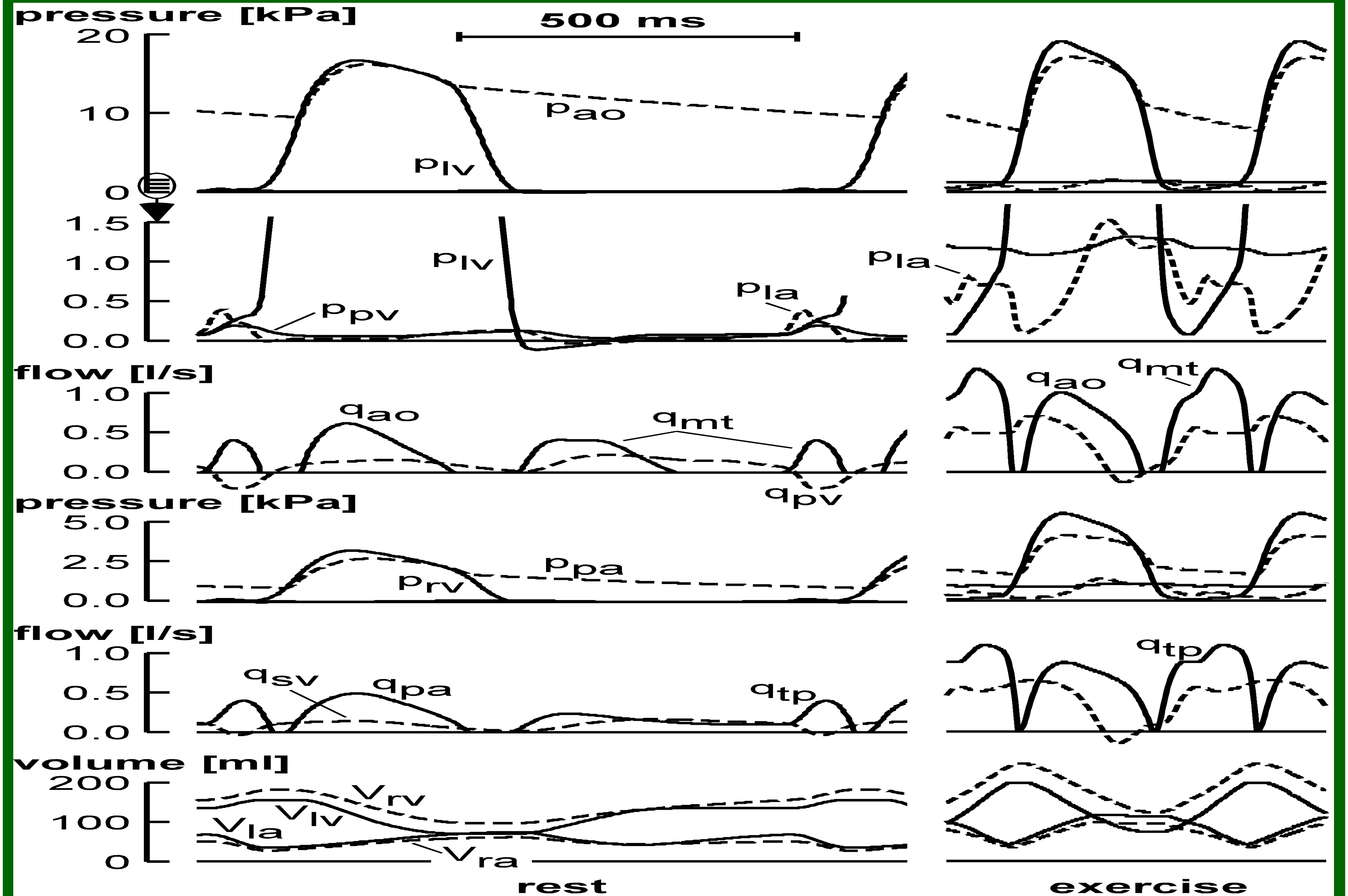
Adaptation of Blood vessel

- Shear stress → Diameter ↑
- Wall stress → Wall thickness ↑

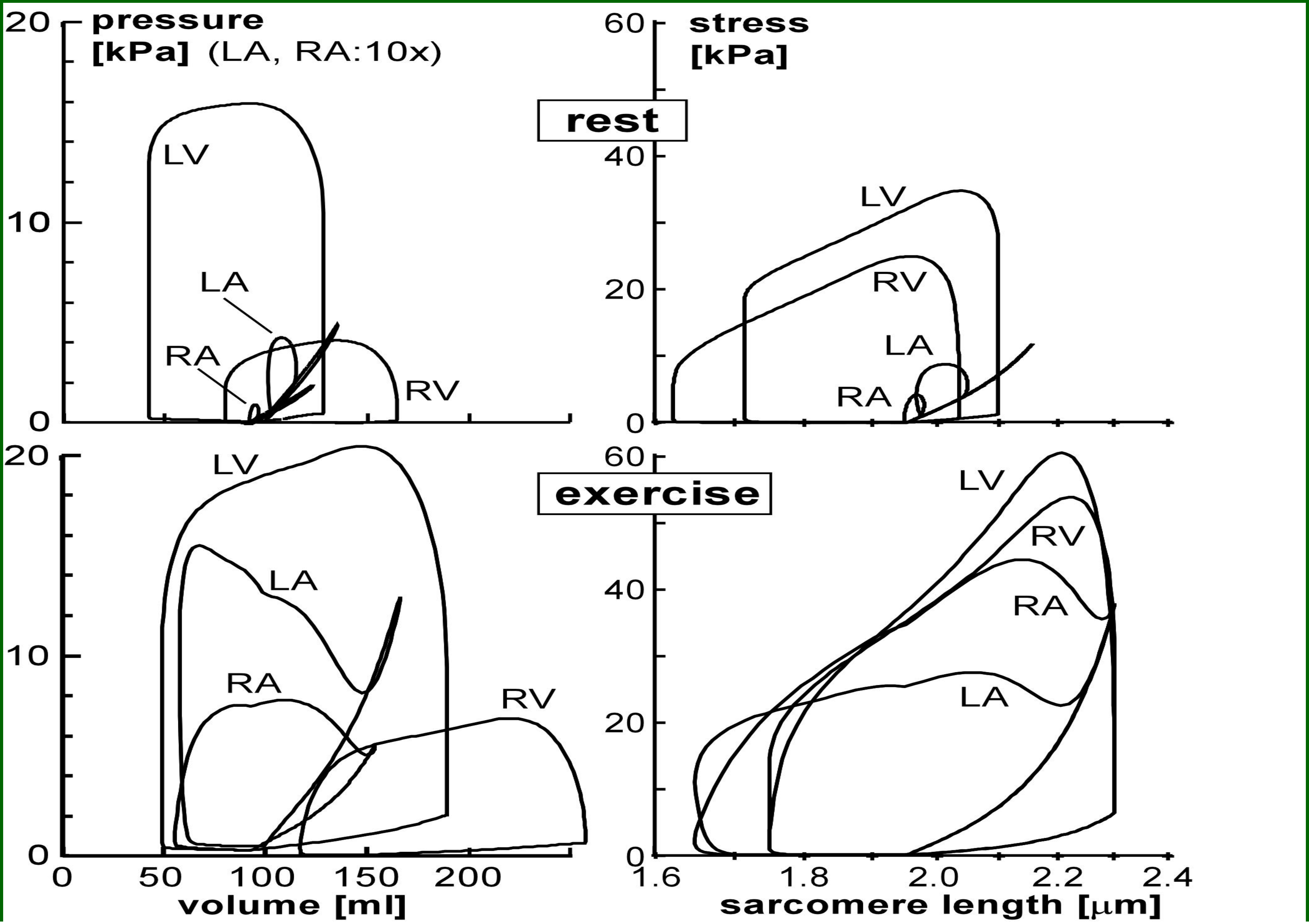
Input to CircAdapt

value	SI-unit	description
		hemodynamics at rest
12.2	kPa	mean blood pressure (=92 mmHg)
100	ml/s	mean systemic blood flow
0.850	s	cycle time (=1/HR)
1.5	kPa	blood pressure drop over lungs
		tube properties
60	s^{-1}	wall shear rate for adaptation
565	kPa	maximum wall stress for adaptation
5.0	m/s	accidental velocity of body impact
7.0	-	exponent of wall material stiffness
		sarcomere properties
2.3	µm	L_s at beginning of ejection for adaptation
1.8	µm	L_s at end of ejection for adaptation
93	kPa	isometric peak active stress at $L_{s,be,adapt}$
4.8	kPa	passive stress at $L_{s,be,adapt}$
		changes with exercise
400	ml/s	mean systemic blood flow
0.350	s	cycle time (=1/HR)

Simulations after adaptation



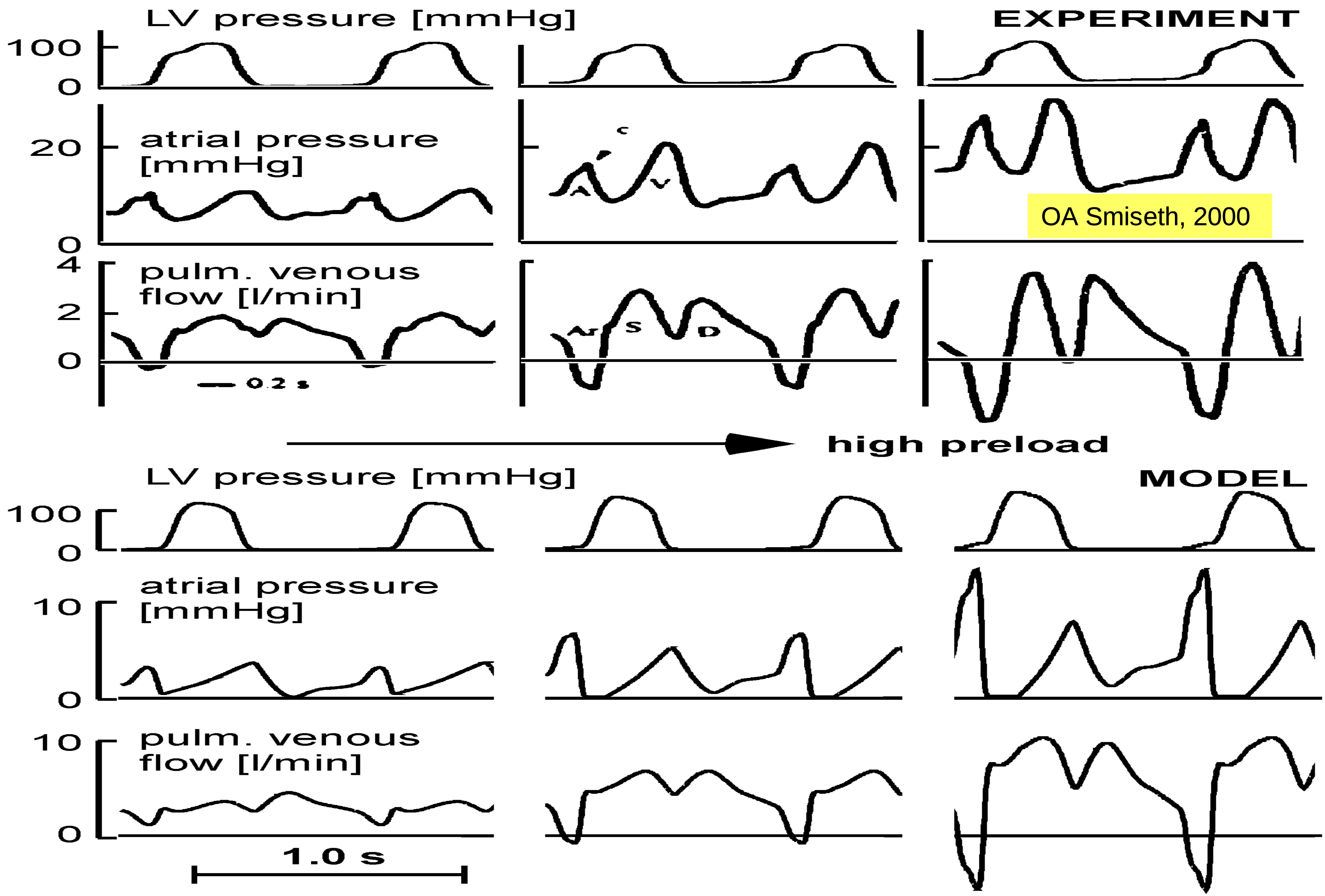
Pressure-Volume & Stress-Strain Loops



Calculated parameters= Result

description		adult value	fetus value	SI-unit
pulmonary vein	operating pressure (p_0)	46.2	68.1	Pa
	length compliance segment (l)	0.228	0.067	m
	wave impedance (Z)	0.427	3.77	$MPa \cdot m^{-3}s$
	wall area (A_{wall})	23.6	4.02	mm^2
	lumen area ($A_{cav,0}$)	535	71.9	mm^2
left atrium	wall volume (V_{wall})	8.47	0.234	ml
	$L_{s,v0}$ (Eq. 19)	0.703	0.542	µm
	passive stiffness G_{pas} (Eq. A1.4)	5.02	3.12	kPa/µm
mitral valve	lumen area	1297	287	mm^2
left ventricle	wall volume	113	3.12	ml
	$L_{s,v0}$ (Eq. 19)	1.24	1.00	µm
aortic valve	lumen area	519	115	mm^2
aorta	operating pressure	12.3	8.53	kPa
	length compliance segment	0.41	0.193	m
	wave impedance	7.87	16.9	$MPa \cdot m^{-3}s$
	wall area	254	69.2	mm^2
	lumen area	506	195	mm^2
system	resistance	122	342	$MPa \cdot m^{-3}s$
systemic veins	operating pressure	54.3	166	Pa
	length compliance segment	0.41	0.258	m
	wave impedance	0.57	2.00	$MPa \cdot m^{-3}s$
	wall area	23.4	18.1	mm^2
	lumen area	522	217	mm^2
right atrium	wall volume	2.64	0.198	ml
	$L_{s,v0}$ (Eq. 19)	0.526	0.512	µm
	passive stiffness G_{pas} (Eq. A1.4)	5.58	7.11	kPa/µm
tricuspid valve	lumen area	1297	398	mm^2
right ventricle	wall volume	76.4	3.65	ml
	$L_{s,v0}$ (Eq. 19)	0.881	0.985	µm
pulm. valve	lumen area	519	159	mm^2
pulm. artery	operating pressure	1.62	8.79	kPa
	length compliance segment	0.228	0.258	m
	wave impedance	3.12	21.7	$MPa \cdot m^{-3}s$
	wall area	126	56.1	mm^2
	lumen area	499	153	mm^2
pulm. system	resistance	15.0	3420	$MPa \cdot m^{-3}s$

Preload variation: Pulmonary venous flow



CircAdapt model

a. Modeling of circulation
- Lumped model in modules:
  chambers, tubes, valves

b. Adaptation of modules to load

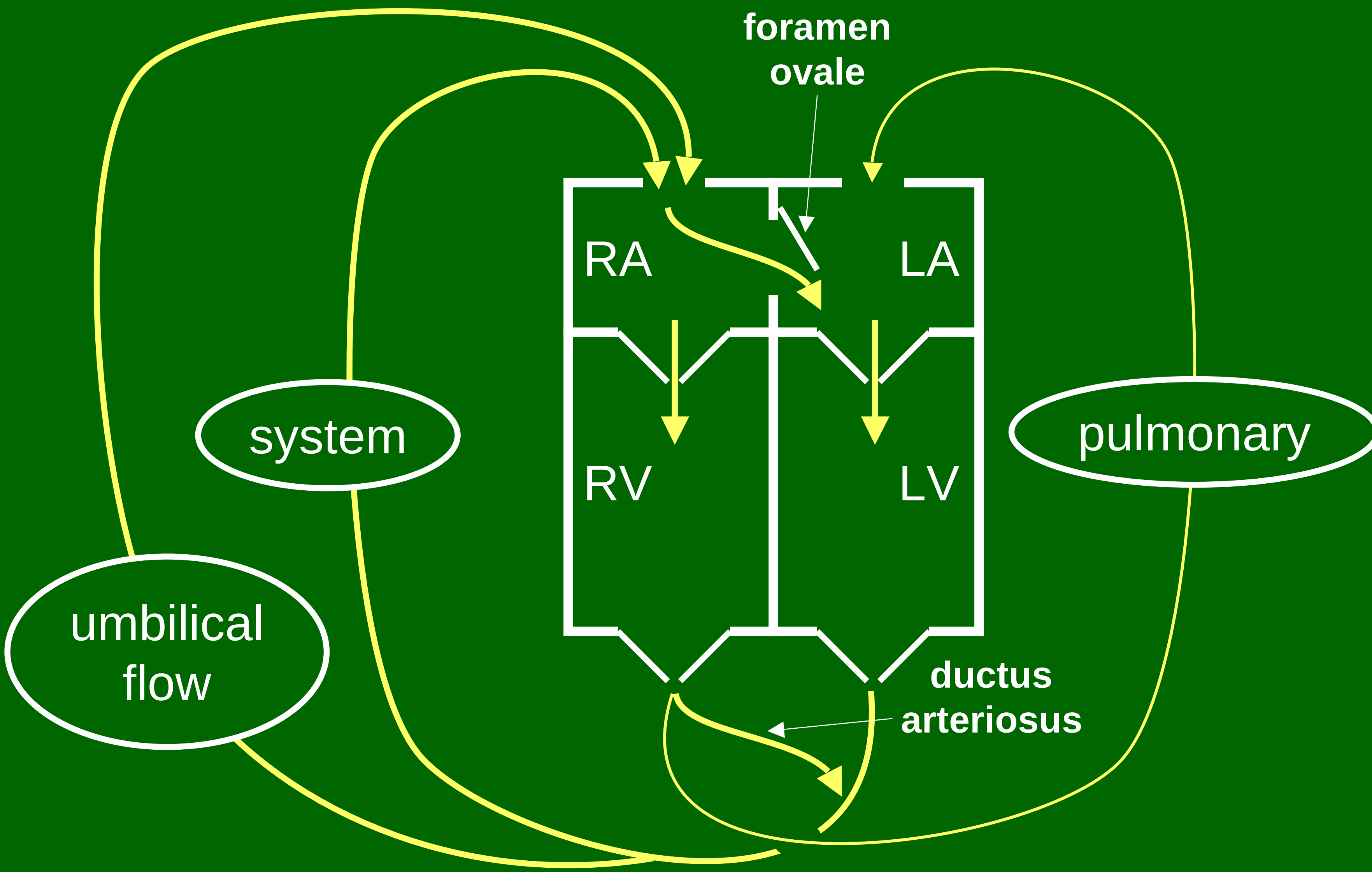
c. Patient specific modeling with adaptation, example of fetus and birth

Search:

Google + keyword ‘CircAdapt’

hit ‘AJP’: Arts T et al. *Am J Physiol*. 2005;288:H1943-H1954
hit ‘Biophysics’: source code

Fetal circulation



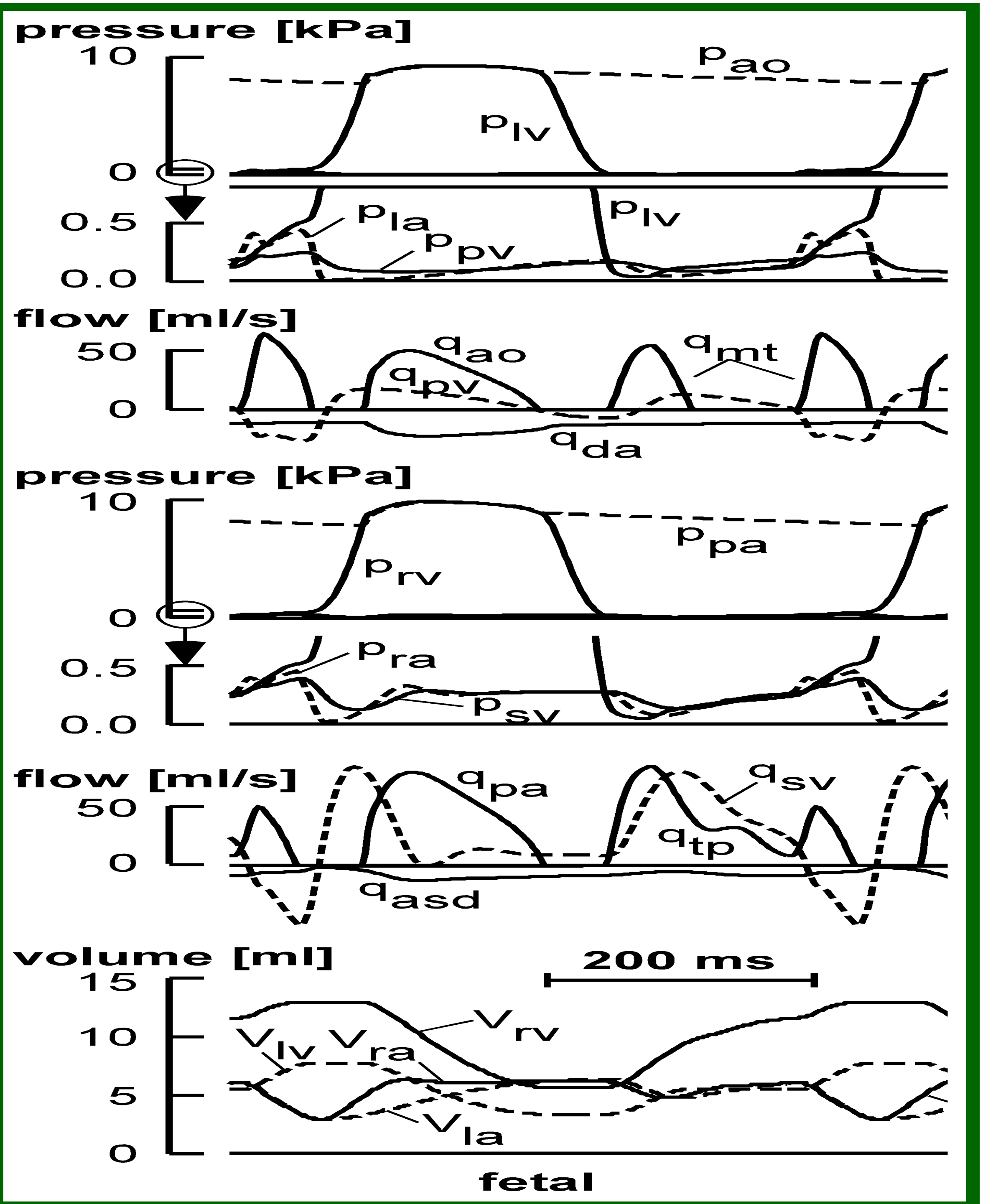
Modification to fetal/newborn circulation

		fetal hemodynamics
8.5	kPa	mean blood pressure (64 mmHg)
25	ml/s	mean systemic blood flow
0.417	s	cycle time (=1/HR)
10 p_0/q_0		fetal pulmonary peripheral resistance
20	mm^2	DUCT area
18	mm^2	ASD area
6	mm^2	ASD area (flow reversal after birth)

		changes in newborn
8.8	ml/s	mean systemic blood flow
3.0	kPa	blood pressure drop over lungs

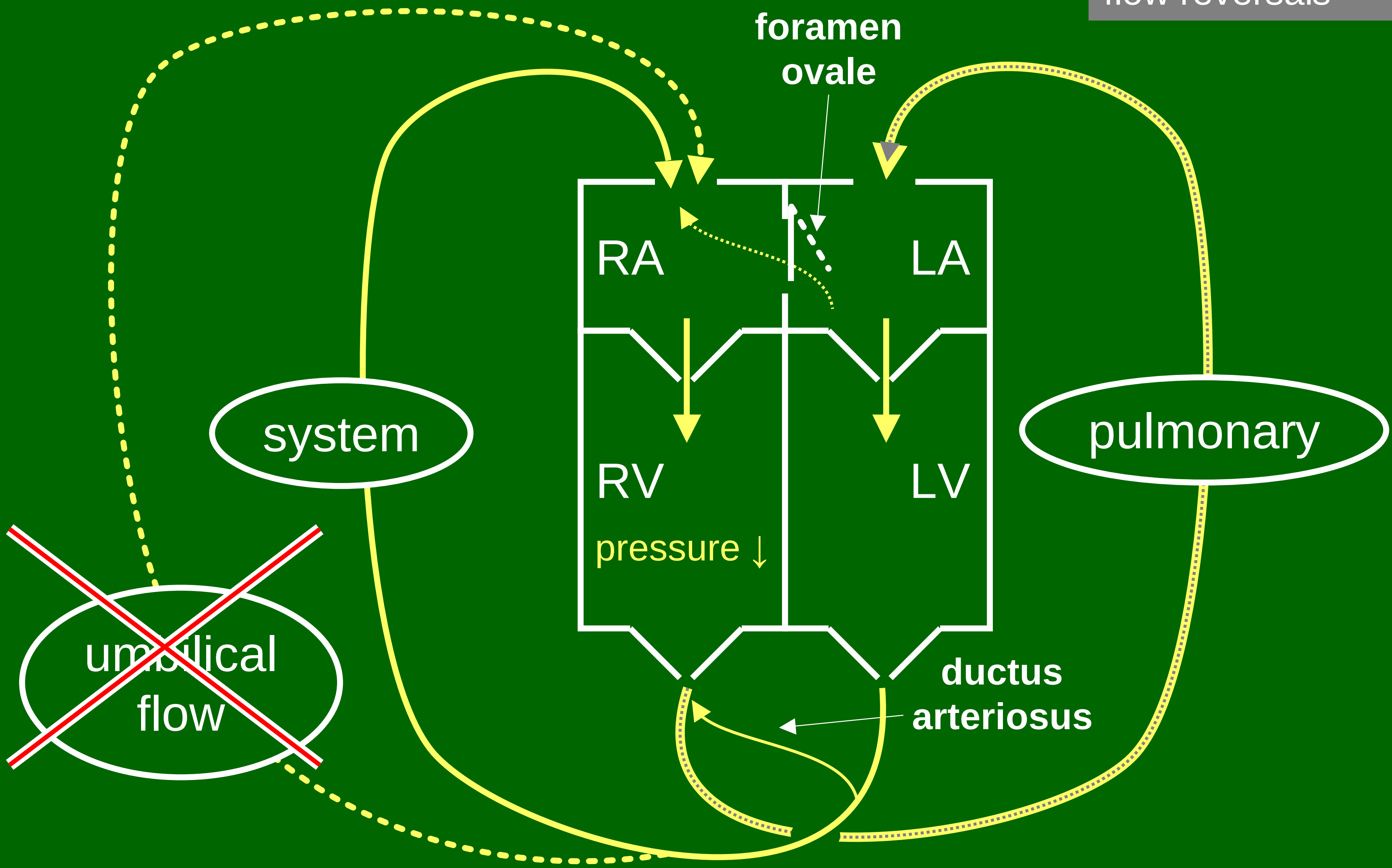
Fetal hemodynamics

Left < Right



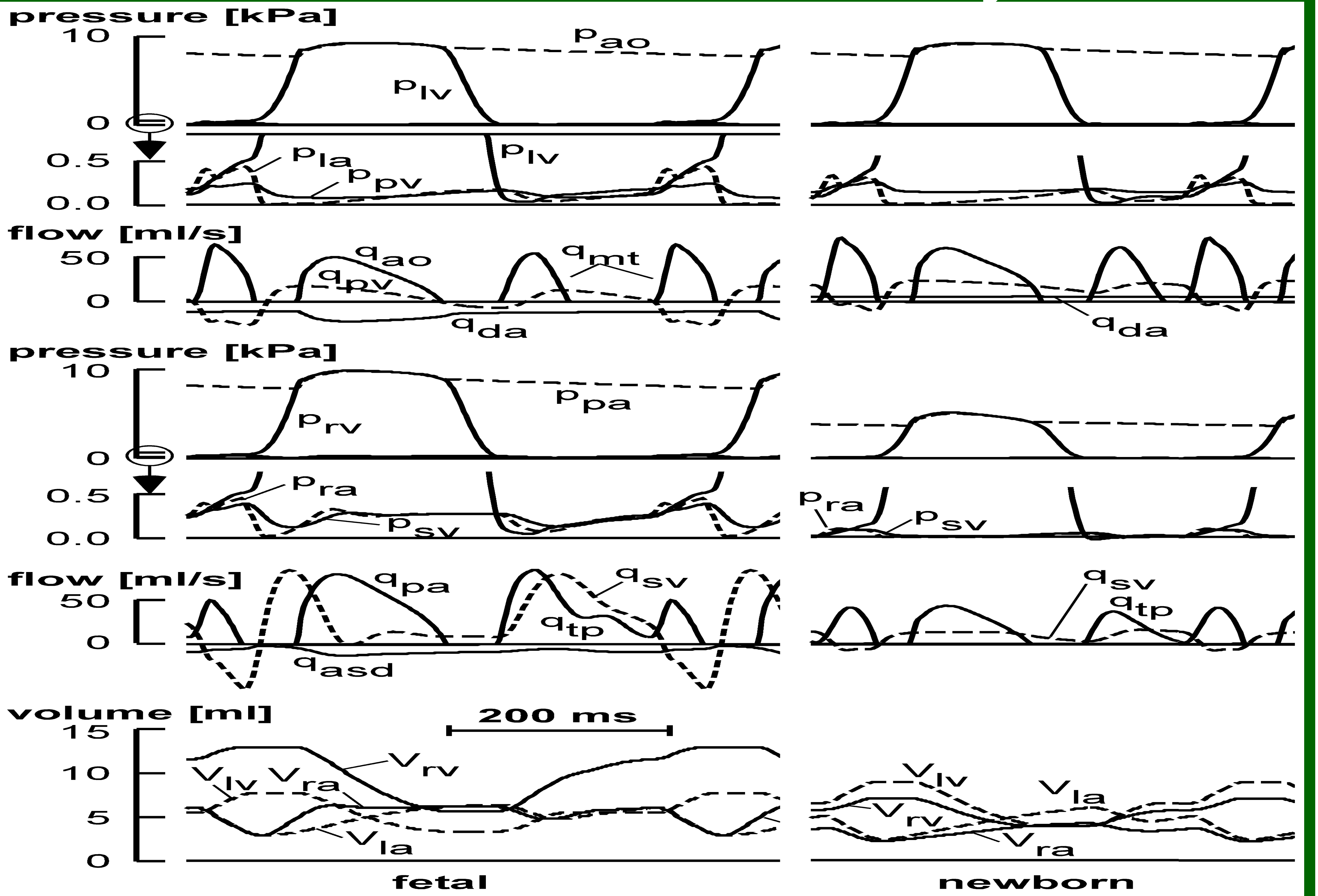
Modification to newborn= Birth

umblical flow stop
pulmonary flow ↑
prv ↓
flow reversals



DEMO BIRTH+→

Fetal and Newborn hemodynamics



Promising applications

1. Patient specific modeling
2. “Non-invasive catheterization”

Pressure difference over:

- membrane → pressure transducer
- valve with inertia
  - doppler velocity and acceleration
  - mass ~ dynamic membrane
    - → pressure transducer

Performed applications

Measurements:

- Max and Min arm blood pressure → arterial stiffness
- 2DE → LV wall volume and stroke volume
- Doppler → Aortic valve opening time, heart rate

Filling in a form (easier than hardware data acquisition)

Result: Non-invasive
Hemodynamics(t)
pV-loop of LV
Stress-strain loop LV myocardium
(hypertrophic stimulus)

Conclusion

- Patient-specific modeling is within reach

- Self structuring by adaptation rules reduces the number of parameters considerably

- Modeling in physiology is of great help to unravel adaptation to mechanical load in the intact organism

- It is not all electrophysiology