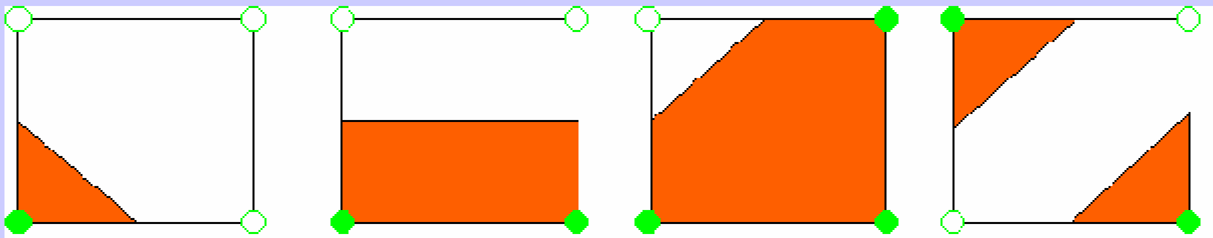
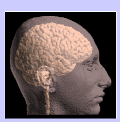
Extraction d'Isocontour (2)

- Calcul de l'isocontour à l'intérieur de chaque pixel :



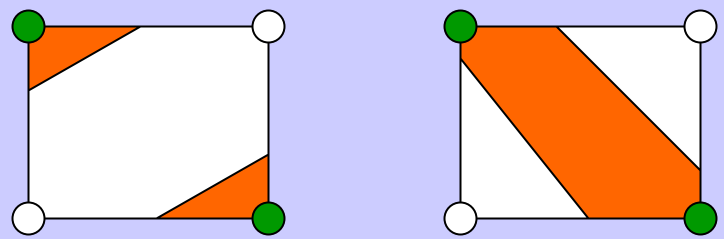
- Algorithme :

Calcul des intersections → Détermination des pixels intersectés → Calcul de l 'isocontour pour chaque pixel

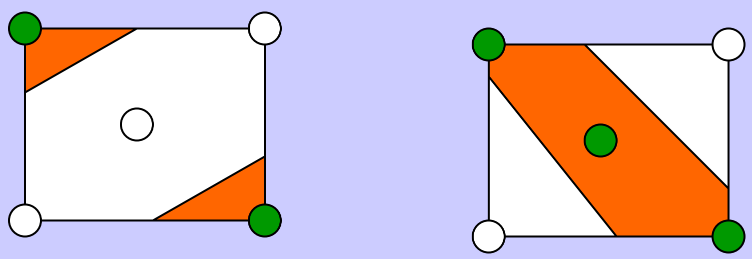


Extraction d'Isocontour (3)

- Ambiguité : 2 choix sont possibles:

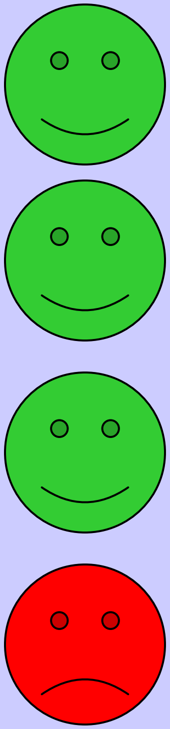


- On utilise l'un ou l'autre ou la moyenne des intensités au centre du pixel :

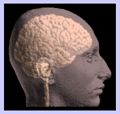


Extraction d'Isosurface (1)

- On procède de même :

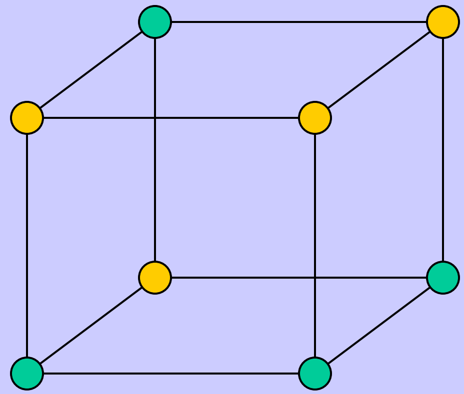


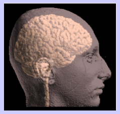
  - Dualité voxel/point
  - Calcul de l'intersection de l'isosurface avec chaque arête
  - Détermination des voxels qui intersecte l'isosurface
  - Calcul de l'intersection de l'isosurface avec chaque voxel



Extraction d'Isosurface (2)

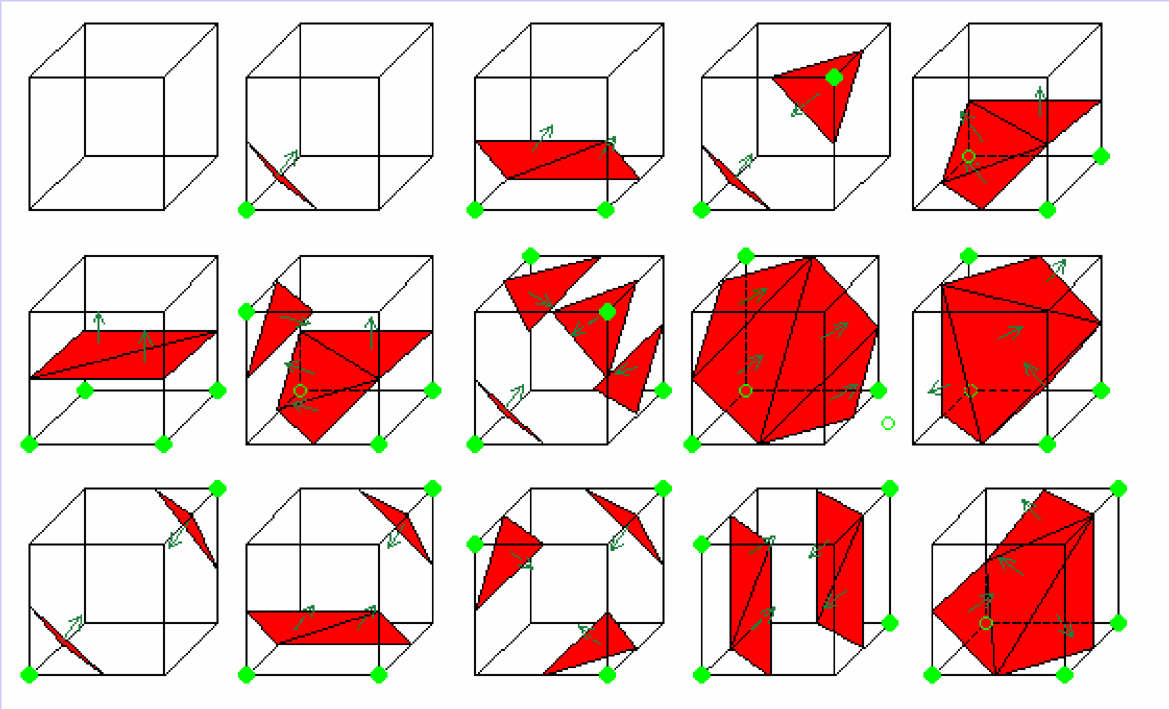
- Calcul de l'intersection de l'isosurface dans chaque voxel :
  - Intersection = ensemble de polygones 3D (non plan)
  - A priori 8x8=256 possibilités





Extraction d'Isosurface (3)

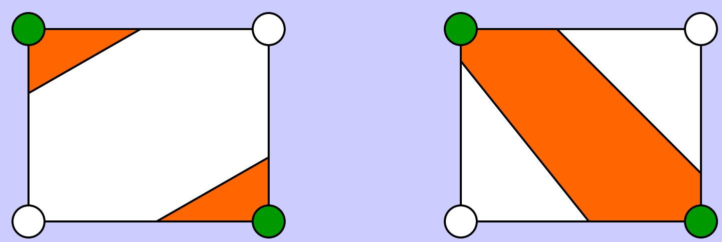
- Algorithme des « Marching Cubes » par Lorensen & Cline (1987)
- Par symmétrie, on peut se ramener à 15 cas

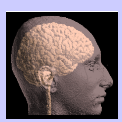


Extraction d'Isosurface (4)

- Utilisation d'une table de hachage :
  - on code sur 1 octet la configuration
  - on associe la configuration avec un tableau ainsi que la symétrie à appliquer (un bit)
- Problème $\Rightarrow$ on peut avoir des trous :

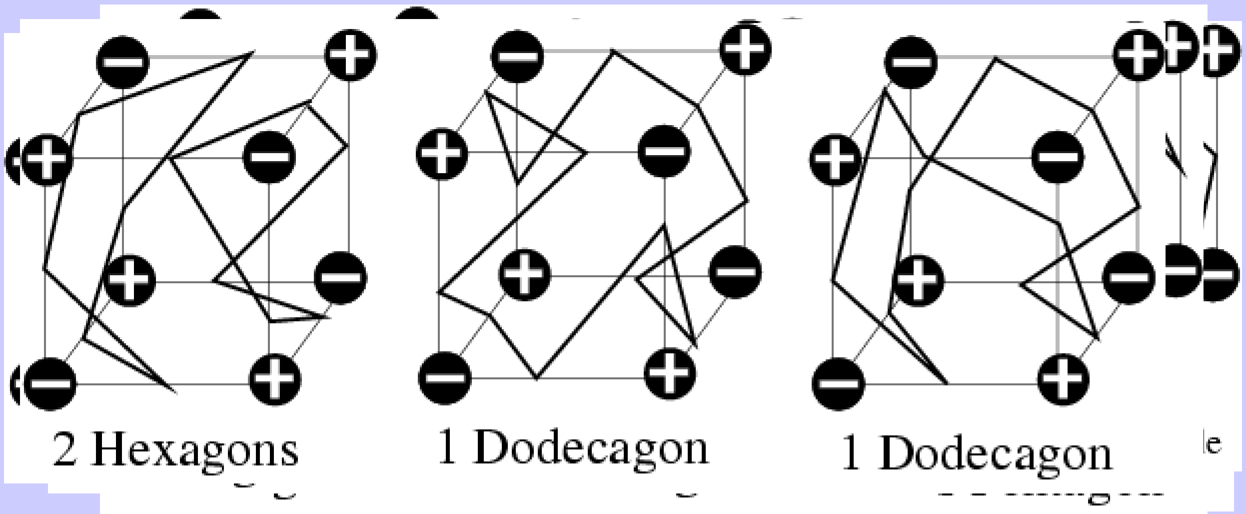
Une même face peut être discrétisée différemment suivant le voxel auquel elle appartient





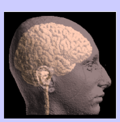
Extraction d'Isosurface (5)

- Plusieurs solutions :
  - prendre en compte le centre de chaque face (6 bit supplémentaires)



Extraction d'Isosurface (5)

- Problèmes supplémentaires :
  - Nécessité de trianguler chaque polygone :
    - ajout d'arête entre sommets
    - ajout d'un sommet central
  - Grande quantité de triangles générés :
    - simplification des surfaces ultérieures
    - utilisation d 'approches hiérarchiques
  - Possibilité d'utiliser des grilles simpliciales



Segmentation

2. Modèles déformables

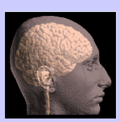
Modèles Deformables

- 2 idées principales:
  - Modèle Déformable est un récipient pour stocker de l’information a priori sur la géométrie et l’apparance de structures anatomiques
  - Définition des frontières fondée sur l’intensité et le gradient d’intensité

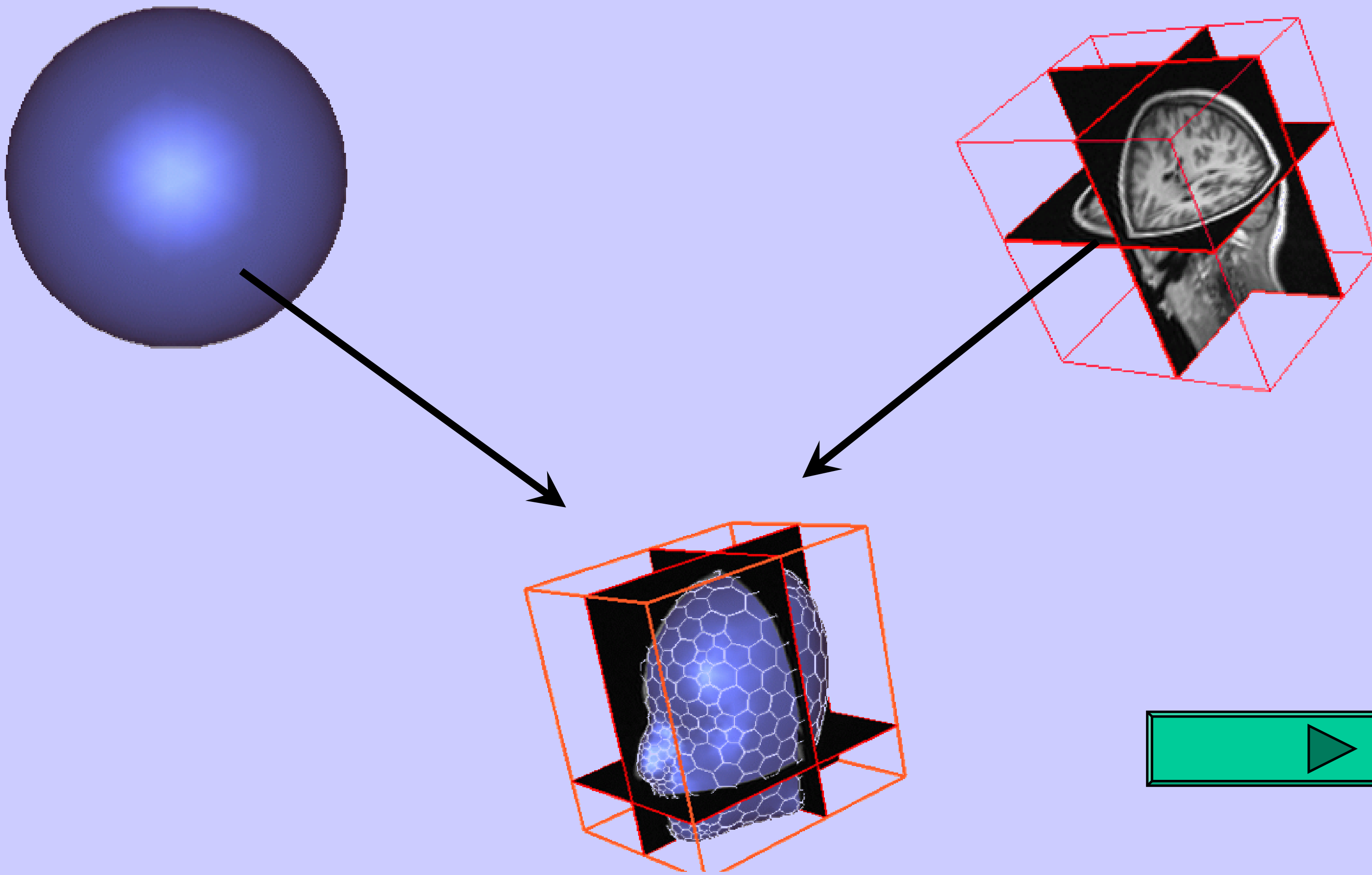
Segmentation d'images à l'aide de modèles déformables

- Un modèle déformable est un récipient pour stocker de l'information a priori sur la géométrie et l'apparence de structures anatomiques
- Deux niveaux de connaissance a priori:

	Faible Connaissance a priori	Grande Connaissance a priori
Forme	Contrainte de continuité C1 ou C2 Initialisation avec formes génériques (sphère, …)	Contrainte de Forme Initialisation avec forme moyenne
Apparence	Utilise information de gradient et/ou intensité	Utilise profils d'intensité ou appariement de blocs



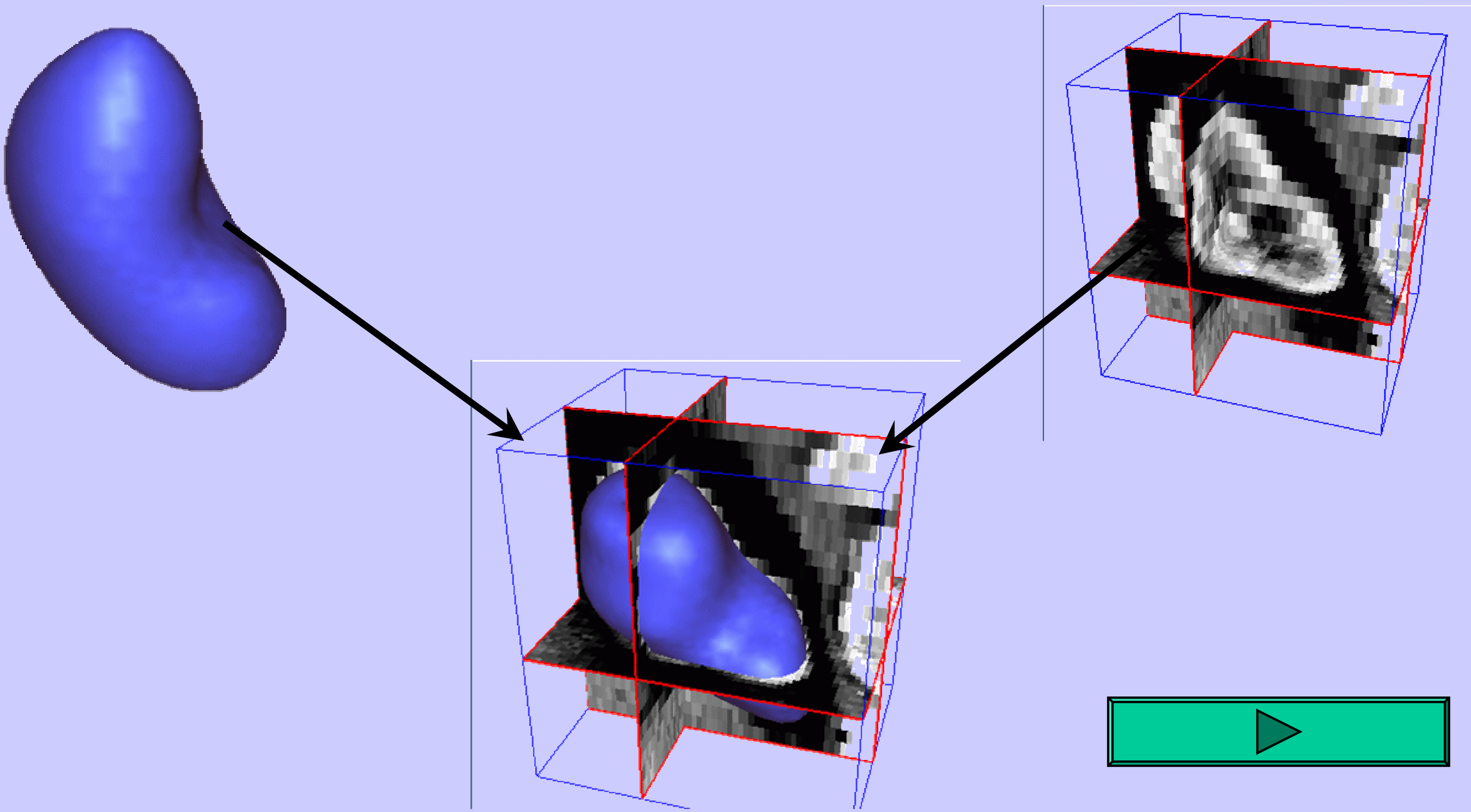
Faible connaissance A Priori



•Valide pour des structures fortement contrastées

•Peut demander une intervention de l’utilisateur

Forte connaissance A Priori



•Valide pour une structure donnée et une modalité donnée

•Plus robuste à l'exception des formes « anormales »

Deformable model

- Snake / active contours
  - Minimisation of a two/three terms energy:

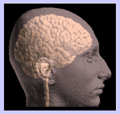
$$E(v(s)) = \int_0^1 \underbrace{E_{in}(v(s))} + \underbrace{E_{im}(v(s))} + \underbrace{E_{con}(v(s))} ds$$

E_{in} : internal energy (regularisation)

$$E_{in}(v(s)) = \alpha(s)\left|v_s(s)\right|^2 + \beta(s)\left|v_{ss}(s)\right|^2$$

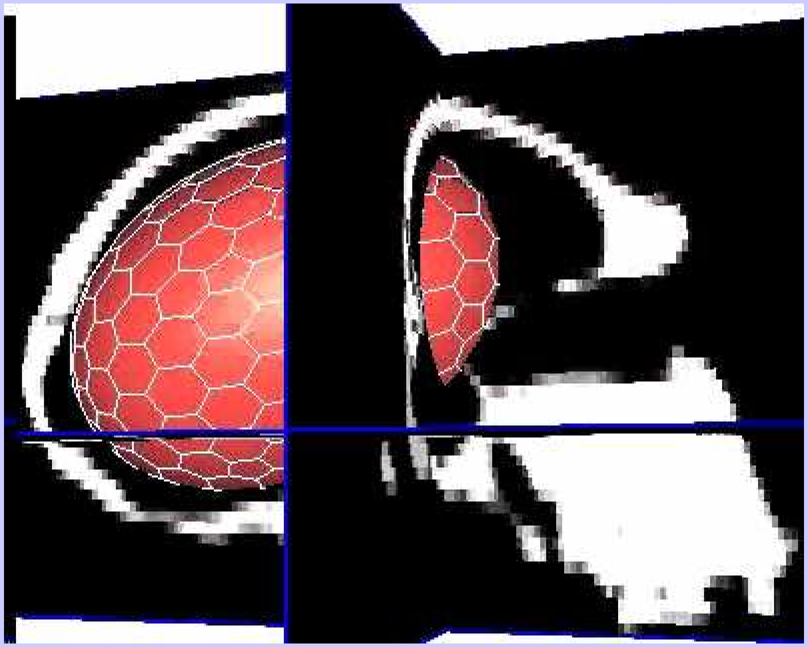
E_{im} : external energy (image)

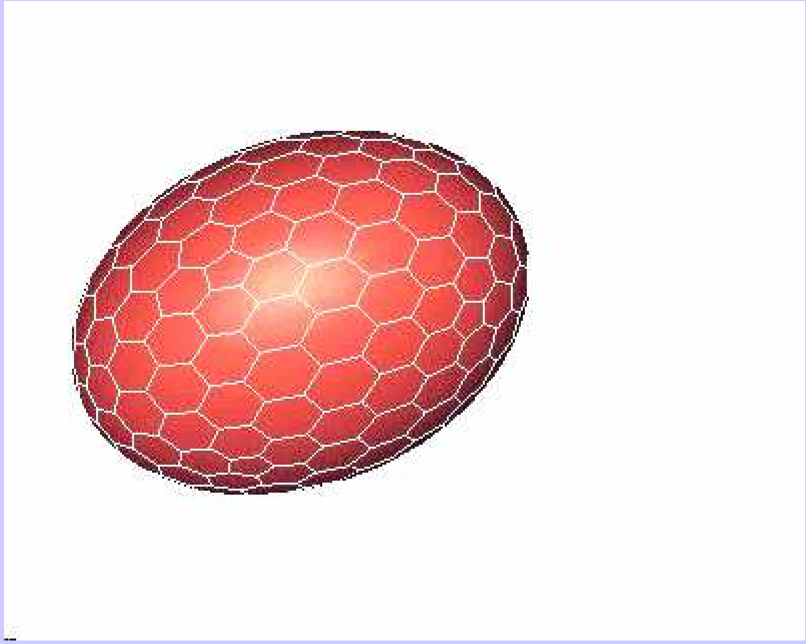
$$E_{im}(v(s)) = -\left|\nabla I(x,y)\right|^2$$



Segmentation: endocranium

CT scan image, Bony structures



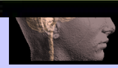


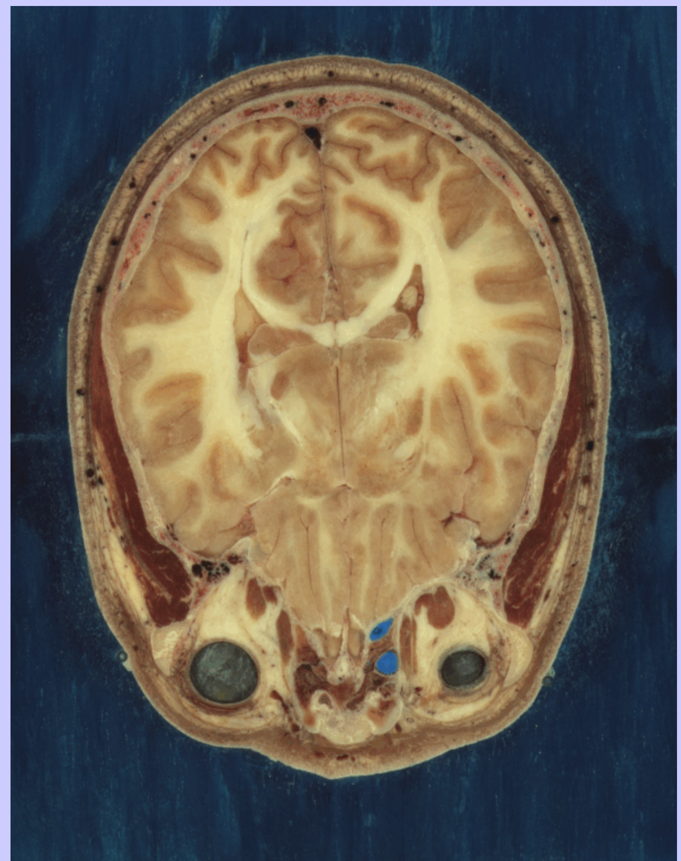
Time of convergence : 13,8 s

model: 1169 cm^3
mold: 1150 cm^3

Visible Human

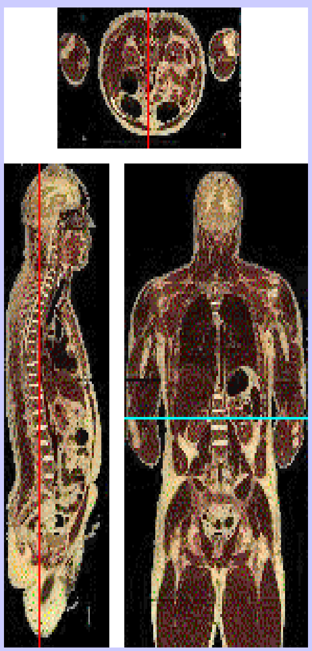
Joseph Paul Jernigan (died August 5th 1993)

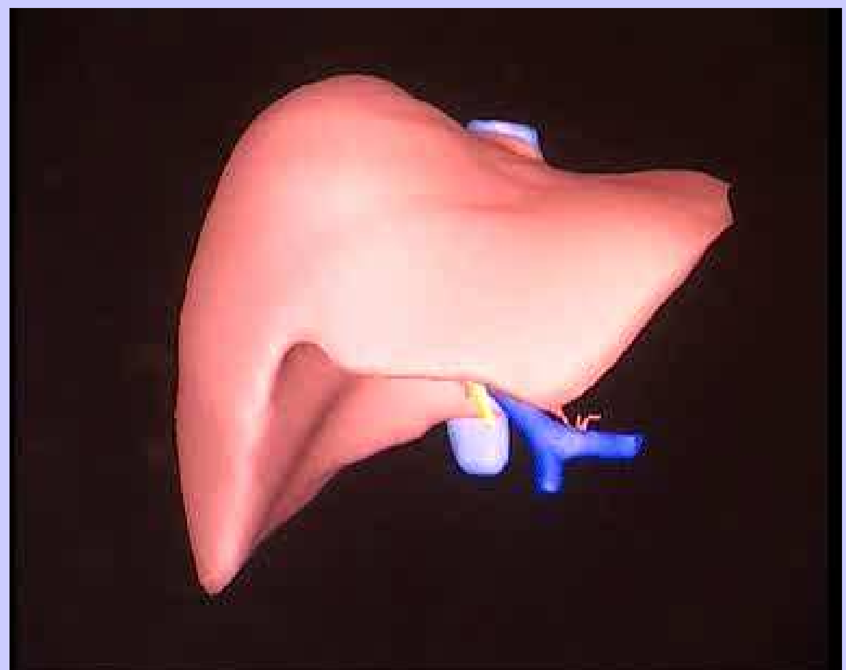


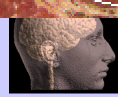


Reconstruction du Foie

Deformation à partir d 'un modèle de référence issu du « *Visible Human Project* »

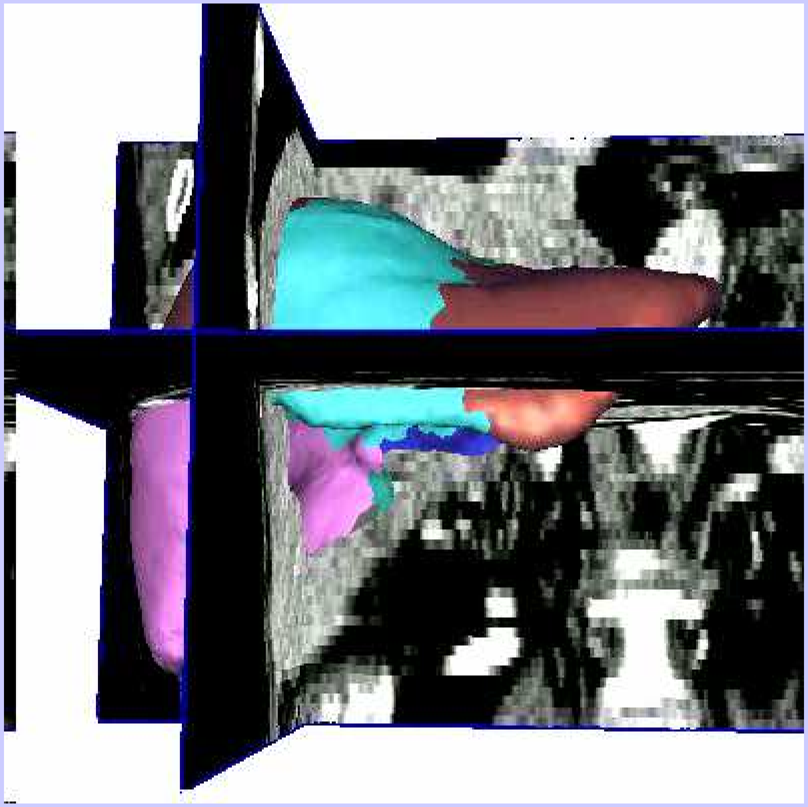


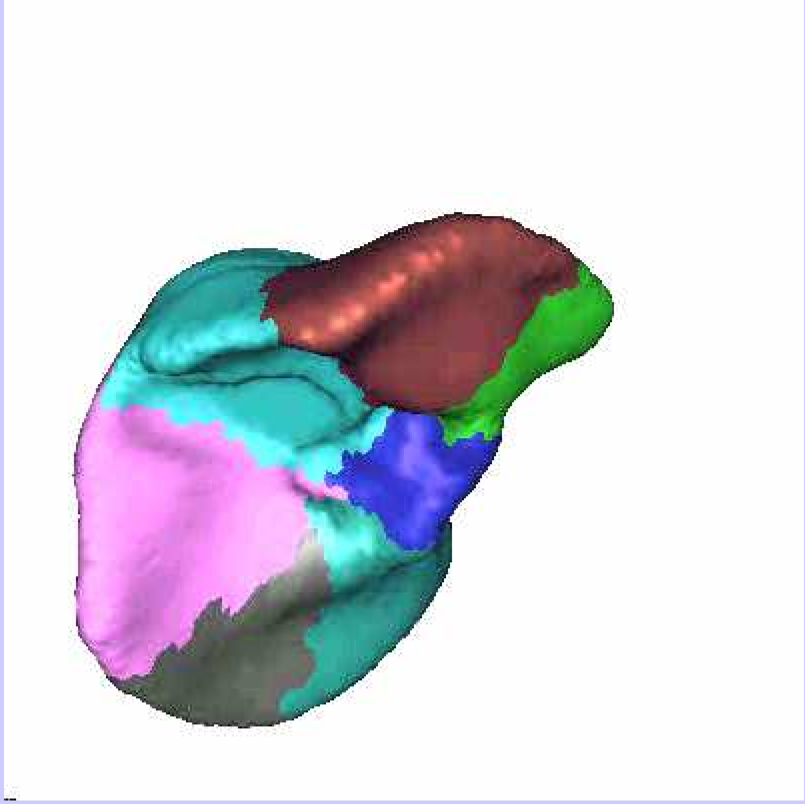




Segmentation: foie

Image scanner de l'IRCAD, extraction du foie

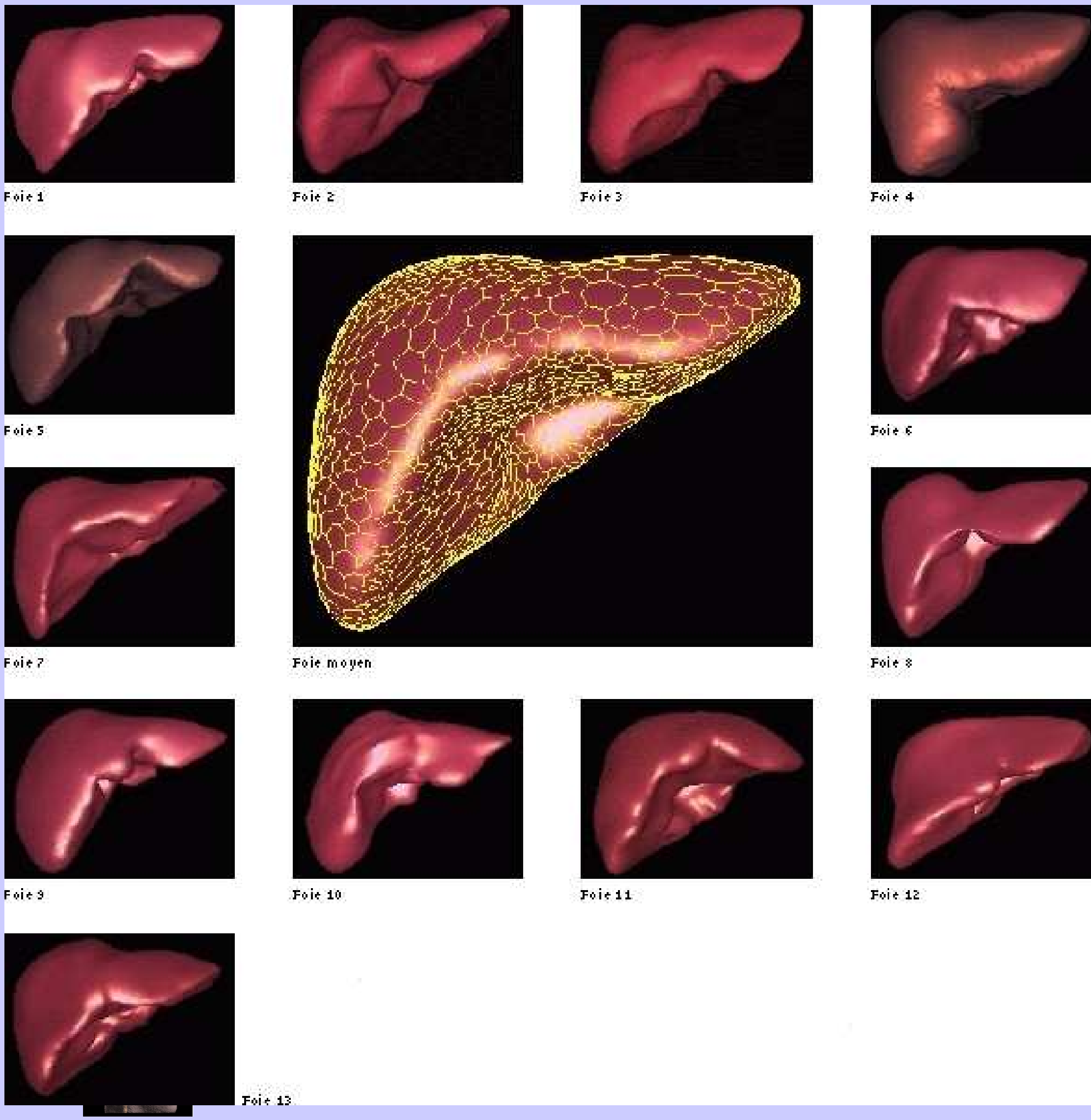




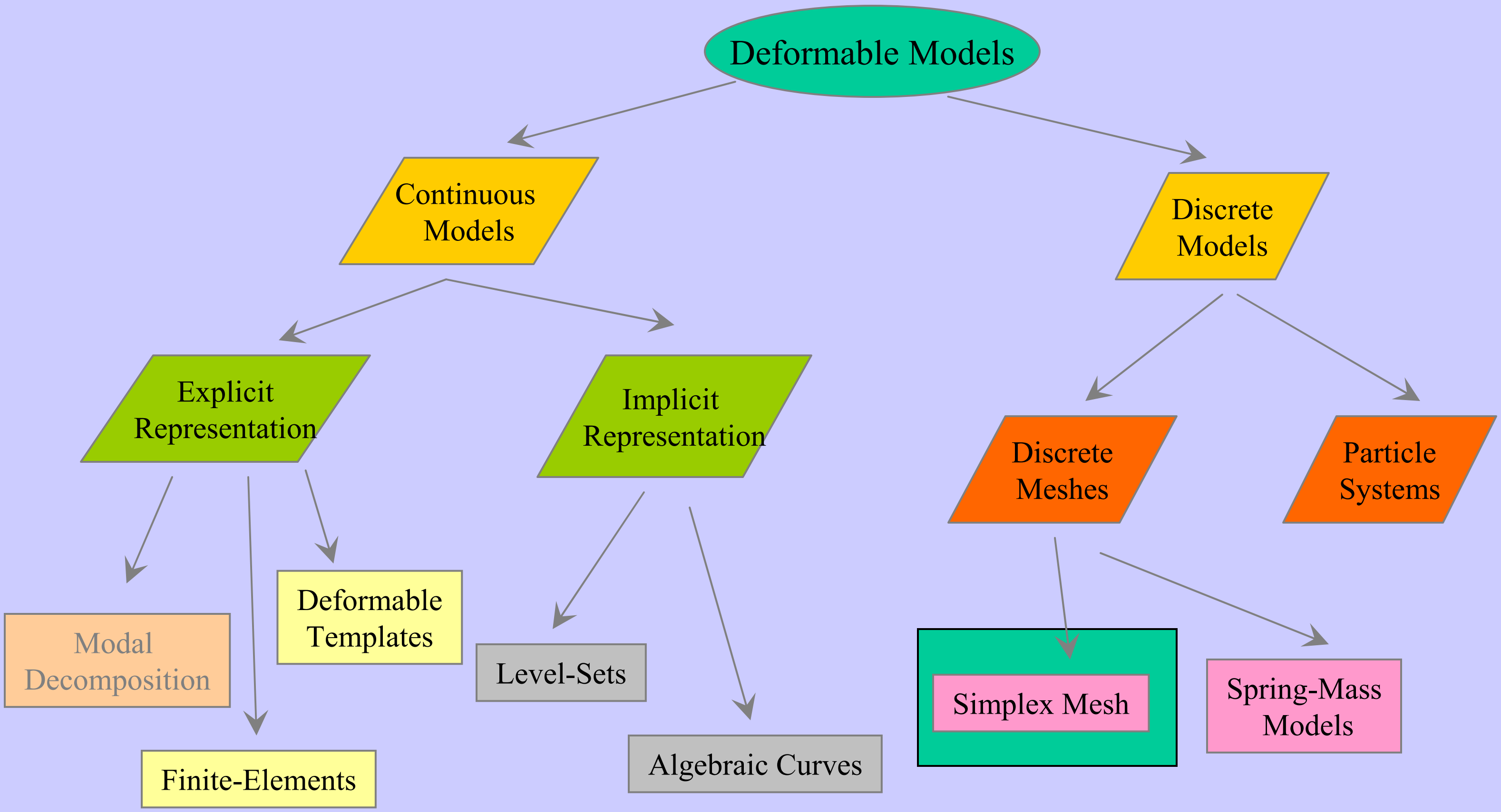
Temps de convergence: 2 mn 12 s

Extraction des segments de Couinaud

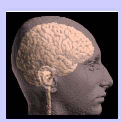
Variabilité anatomique du Foie



Deformable Model Geometry (3)



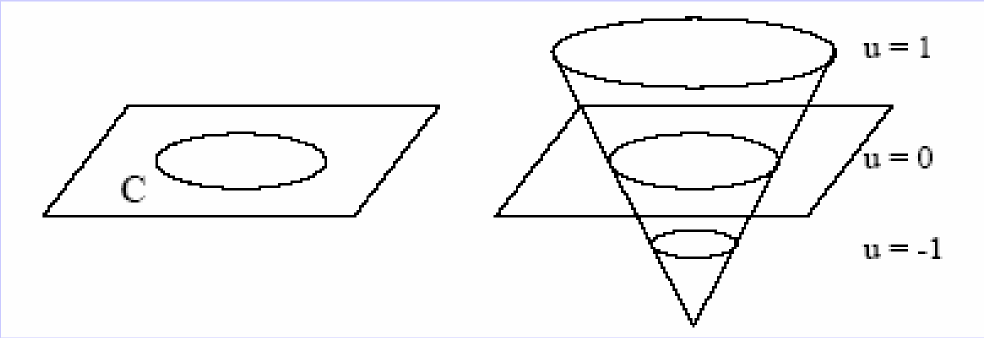
[Montagnat2001]

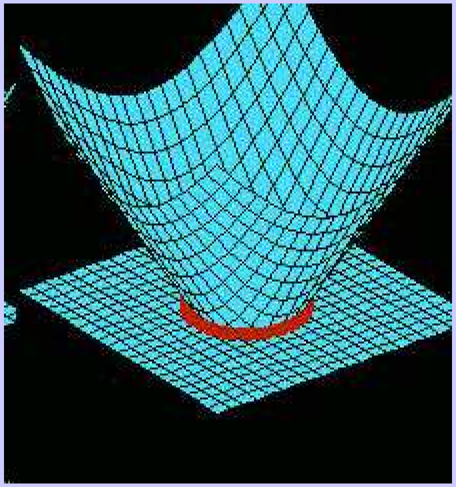


Deformable models

- Level-sets
  - Curve/Surface C (in $\Re^2$/ $\Re^3$) that corresponds to an iso-level of a surface/hypersurface (in $\Re^3$/ $\Re^4$)

$$C(t) = \{(x, y) \mid u(x, y, t) = 0\} \quad u_t + F \left| \nabla u \right| = 0$$

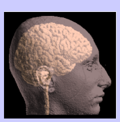




  - Implicit representation

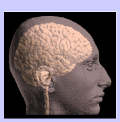
Main difficulties in segmentation algorithms

- Ill-posed problem
  - Boundaries between structures may not be seen on images
  - Strong variability between experts for validation
- Most algorithms are dependent on the acquisition protocole and image modality
- Robustness required in the presence of pathologies



Perspectives (2)

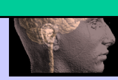
- Applications of segmentation :
  - Diagnosis
    - demand for very fast and automated algorithms with degree of confidence
  - Planning - Prediction -Prevention
    - demand for accurate but potentially not fully automated algorithms combined with high quality meshing
  - Clinical Research
    - demand for automated and accurate algorithm for use with large database (grid computing)



Perspectives (3)

- Segmentation techniques is more and more split between :
  - <u>Registration techniques :</u>
    - registration with a anatomical/physical/physiological model
    - registration with a set of images (data fusion)
  - <u>Low-level techniques :</u>
    - anisotropic filtering, watershed, mathematical morphology

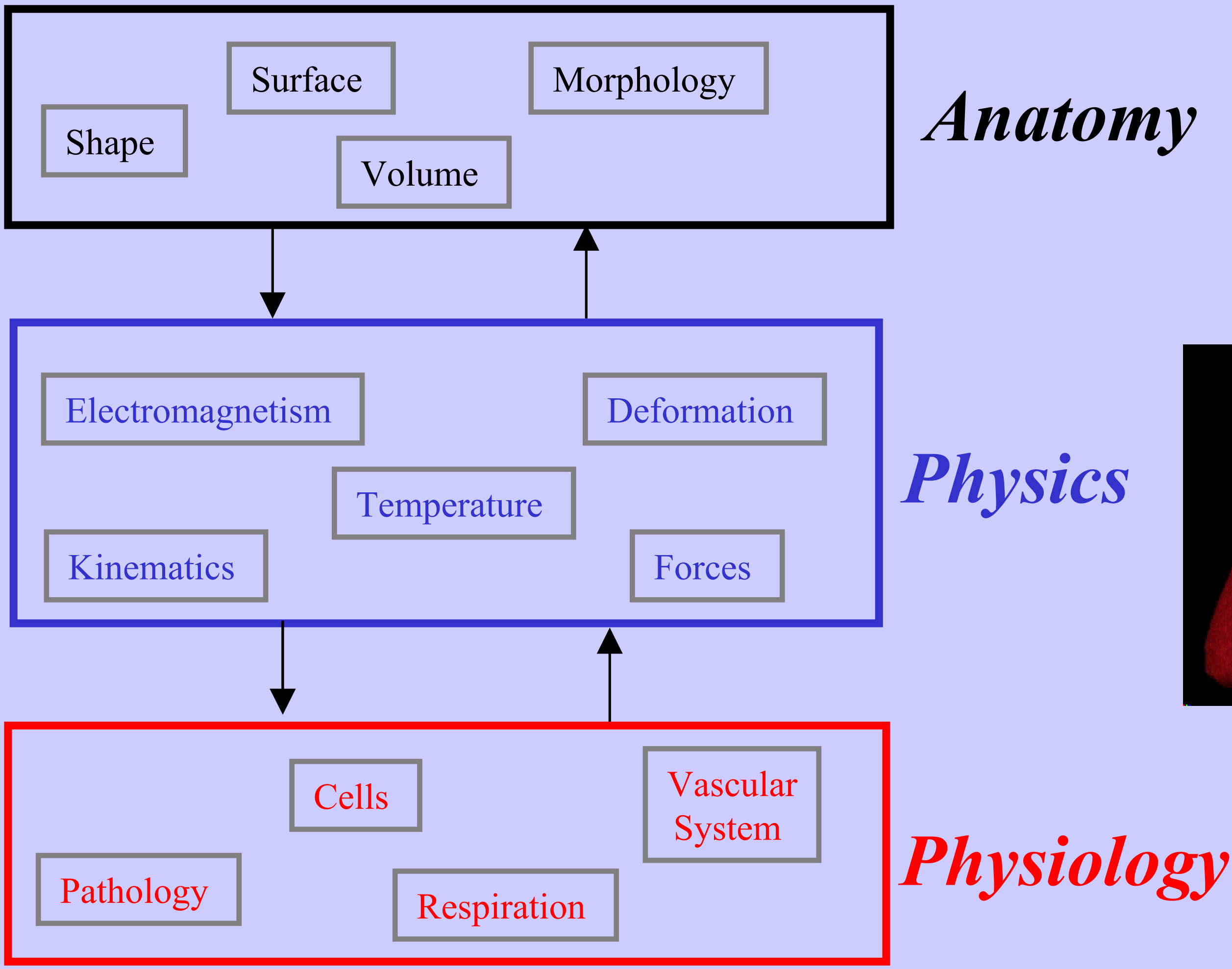
Need to define a unifying framework

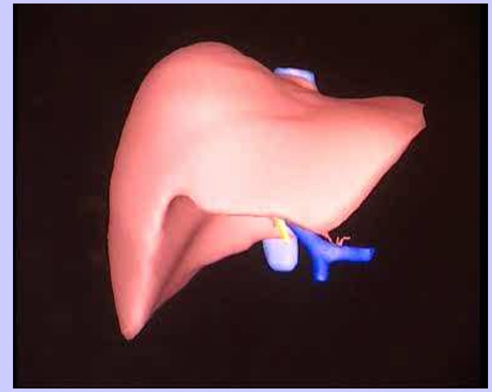


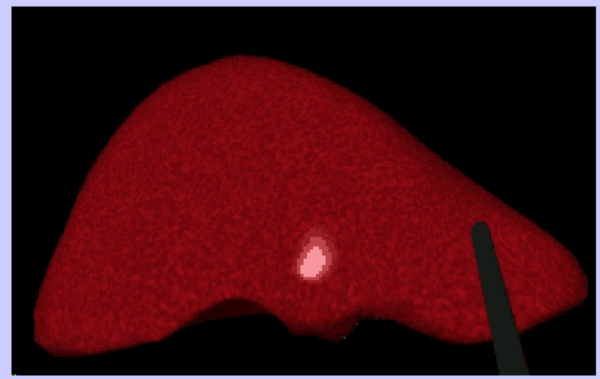
Modélisation

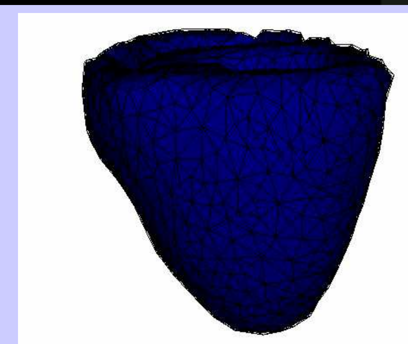
3. Computational Models of the Human Body

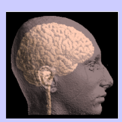
Modeling Levels



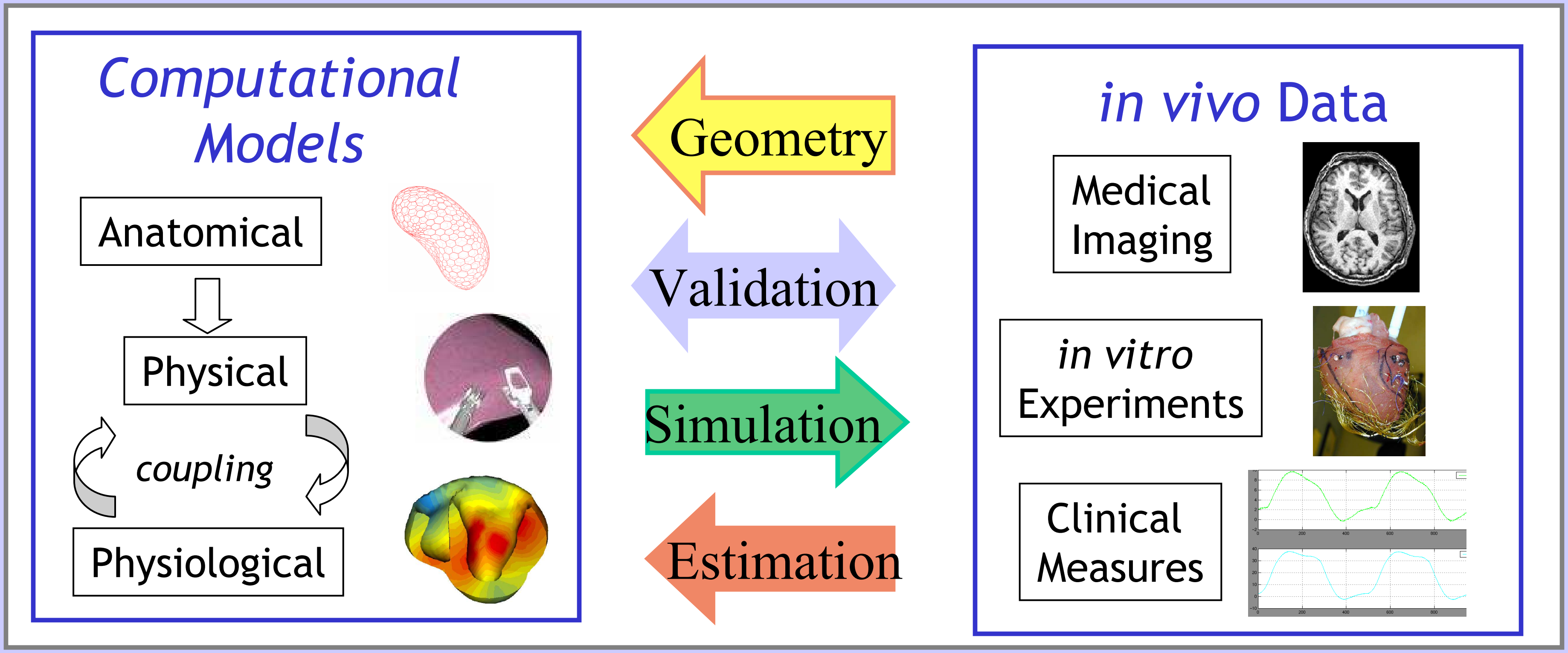








Modeling and Imaging

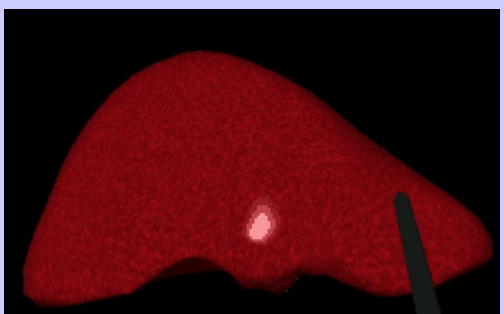


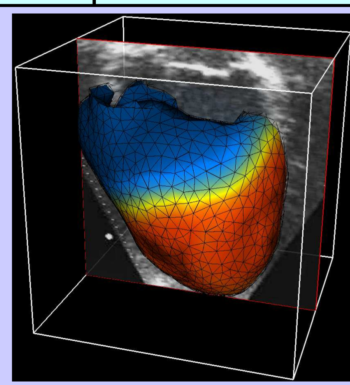
Mesh construction from medical images

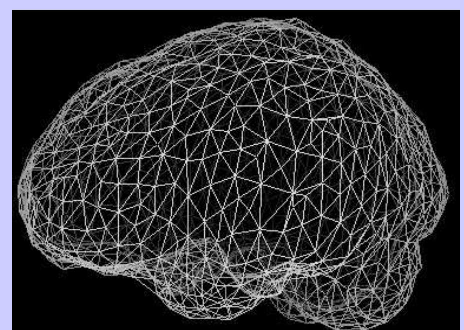
Example of Computational Models of the Human Body

- 3 applications of human modeling

Application Type	**Organ**	**Real/Time Constraint**	**Image Interaction**	**Physiological Modeling**
Surgery Simulation	Liver	Yes	Validation	No
Diagnosis & Therapy Planning	Heart	No	Image Segmentation	Yes
Prediction of outcome	Brain	No	Validation	No





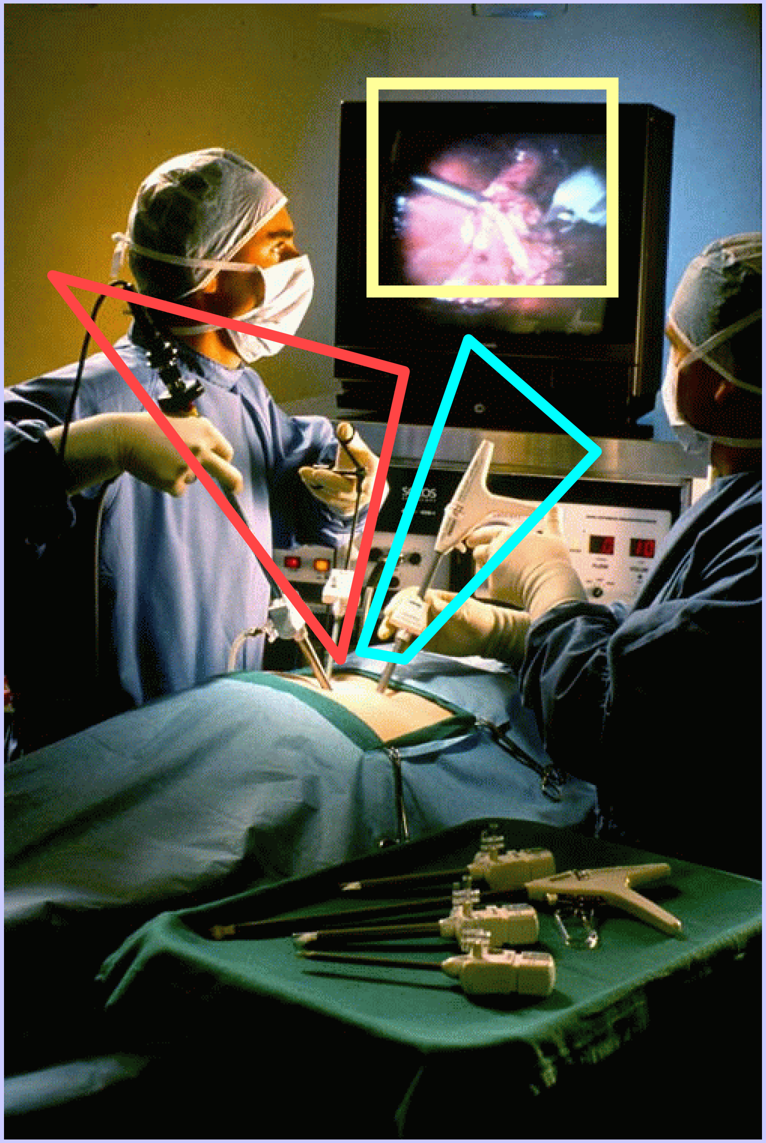


Surgery Simulation

Acknowledgments:

- Clément Forest, Guillaume Picinbono, Stéphane Cotin, Jean-Christophe Lombardo, Nicholas Ayache
- INRIA projects member of the AISIM collaborative action (Imagis,
- Sharp,Macs)
- IRCAD

Need for Training



Hand-eye Synchronisation

Camera being manipulated by an assistant

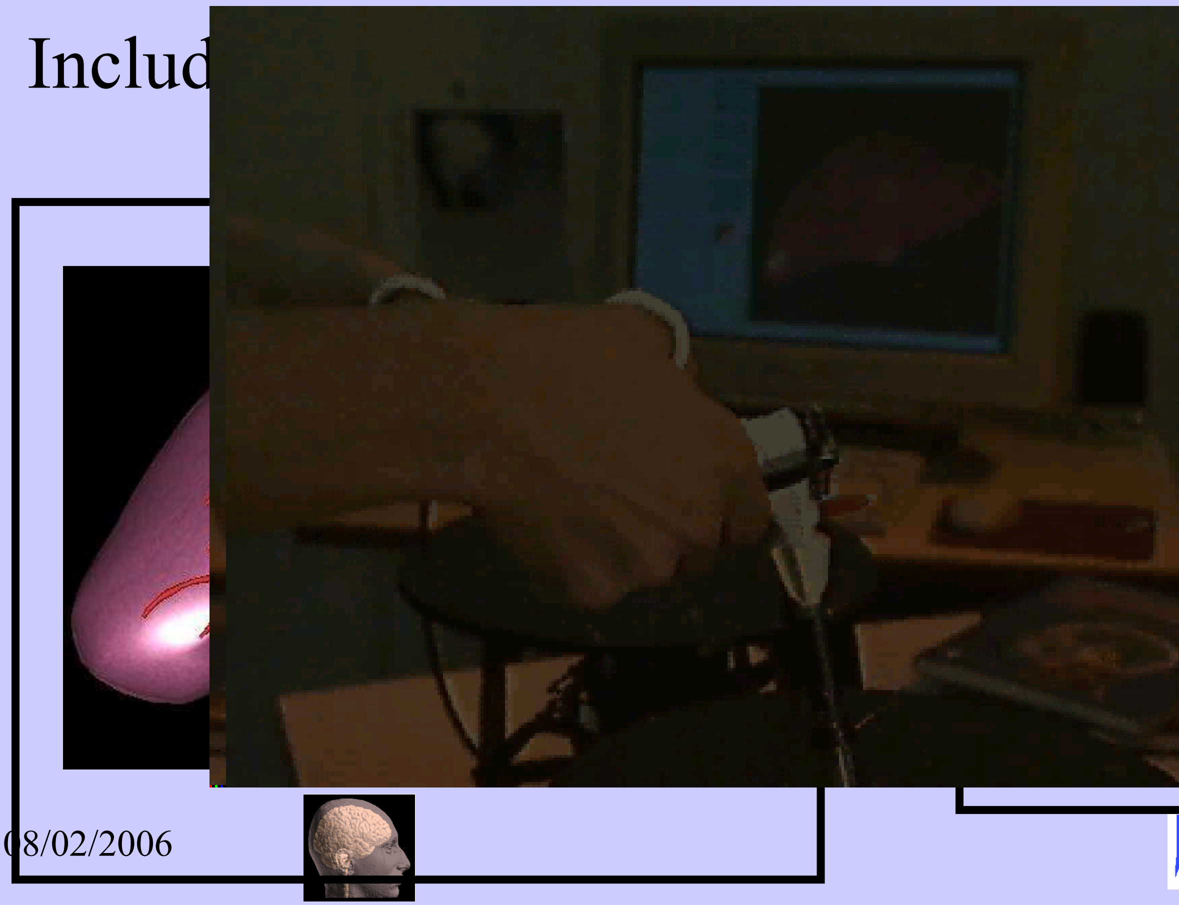
Long instruments going through a fixed point in the abdomen

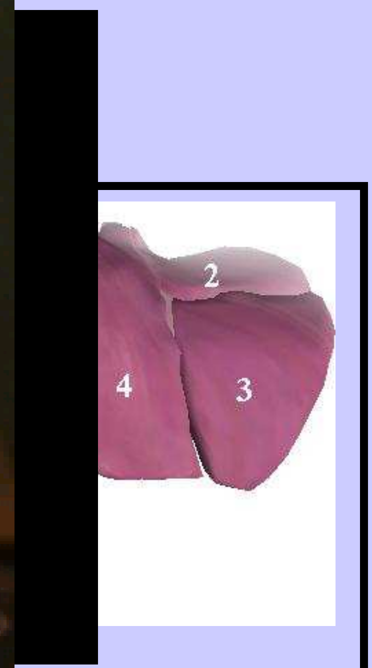
INRIA SOPHIA ANTIPOLIS

EPIDAURE SIMULATION

[Cotin, 1997] [Picinbono, 2001] [Forest 2003]

- Hepatectomy Simulation by laparoscopy
- Includ

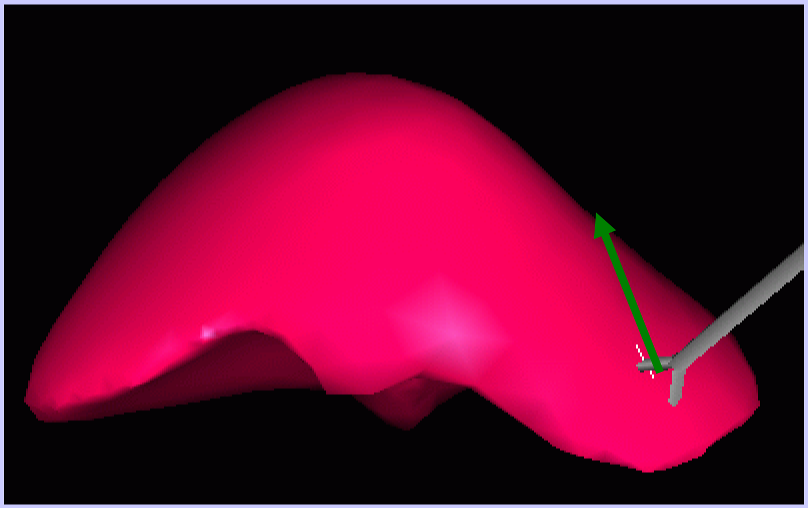




aud Segments
ce *[Soler, 1998]*)

Simulator Workflow

Position



Force

Collision

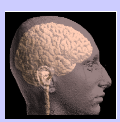
Contact

Deformation

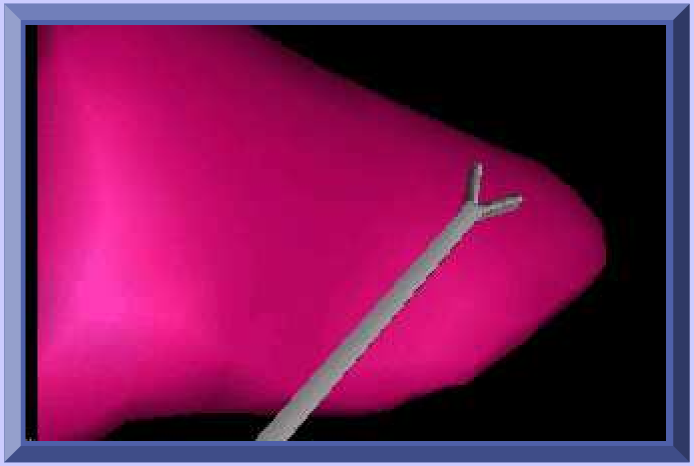
Force

Different Technical Issues

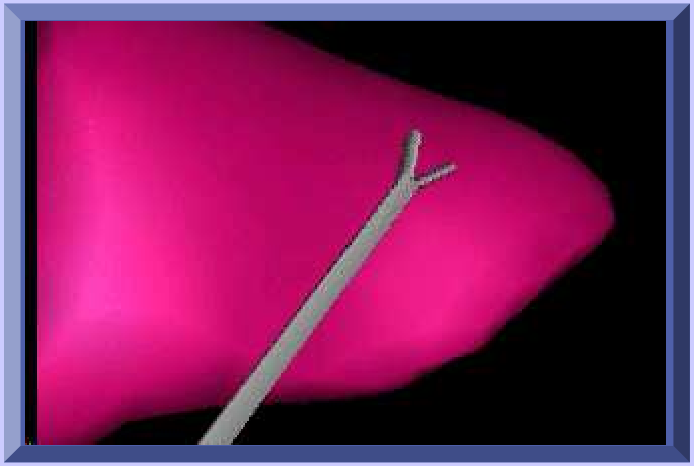
- Mesh Reconstruction from Images
- Soft Tissue Modeling
- Tissue Cutting
- Collision Detection
- Contact Modeling
- Surface Rendering
- Haptic Feedback



Modeling basic surgical gesture



Gliding



Gripping

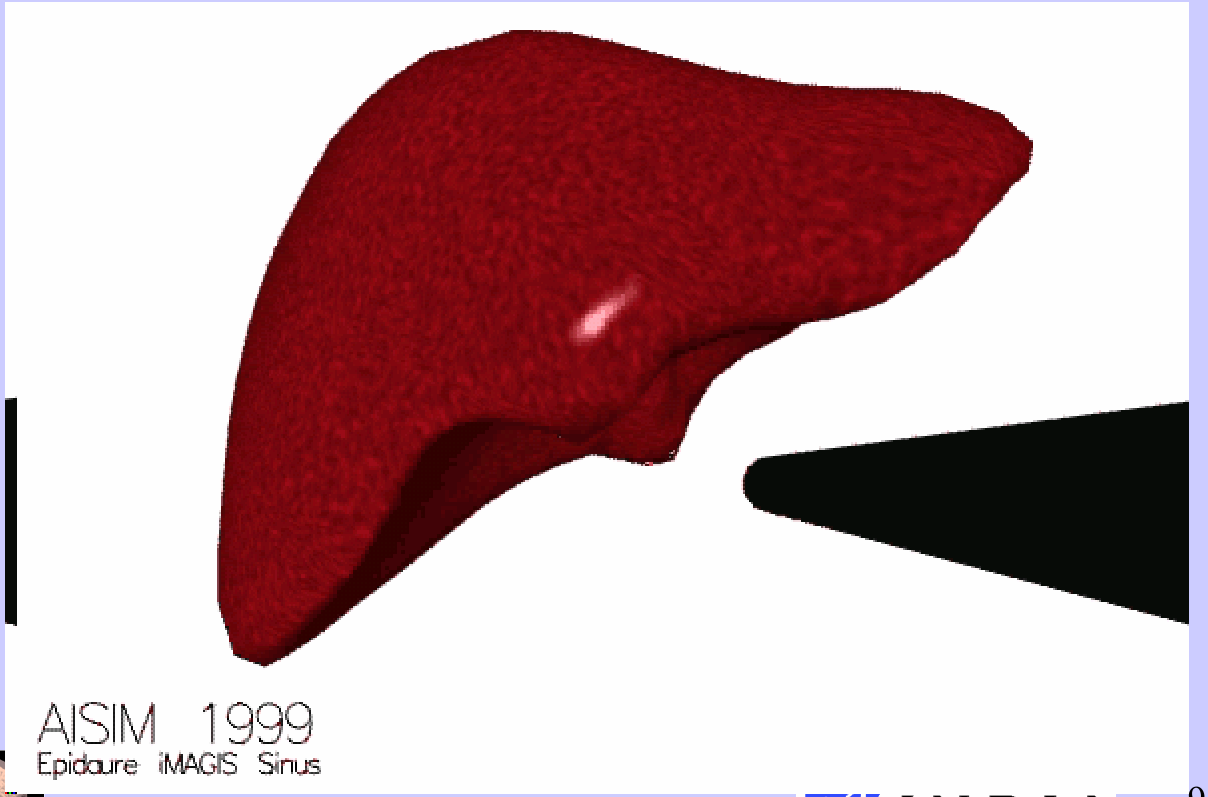
Cutting (pliers)

Cutting (US)

A Family of Models (1/3)

Pre-computed Linear Elastic Model

- Very Fast
- No topology changes

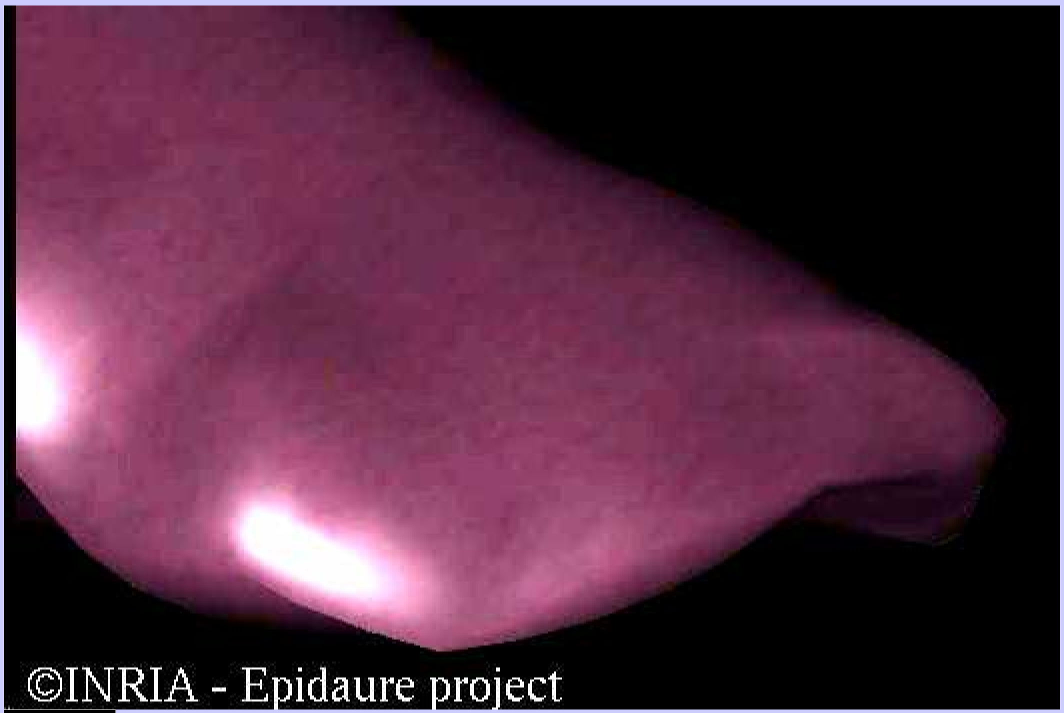


A Family of Models (2/3)

Elastic Model

« Tensor-Mass »

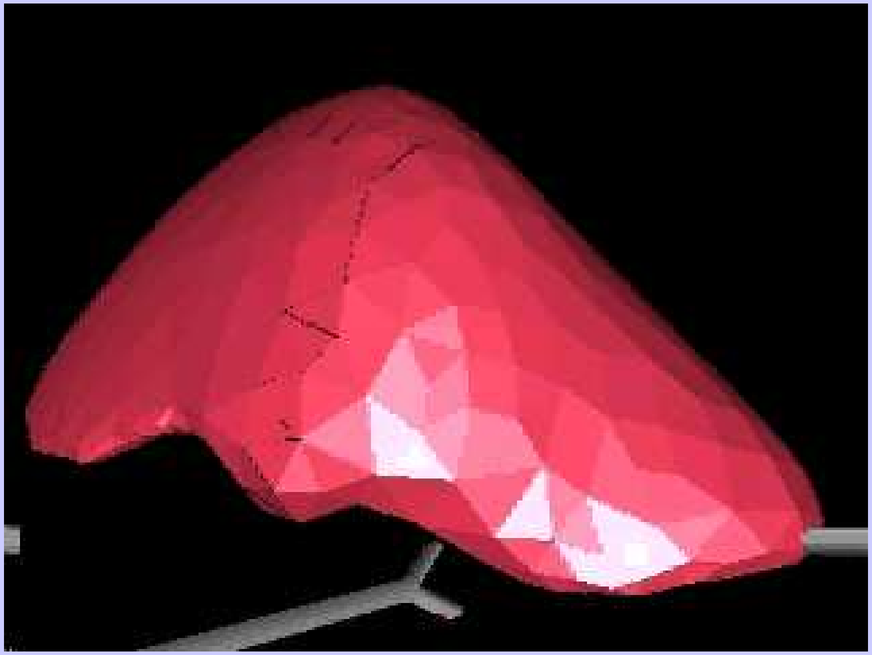
- Topology Changes
- Limited to small displacements



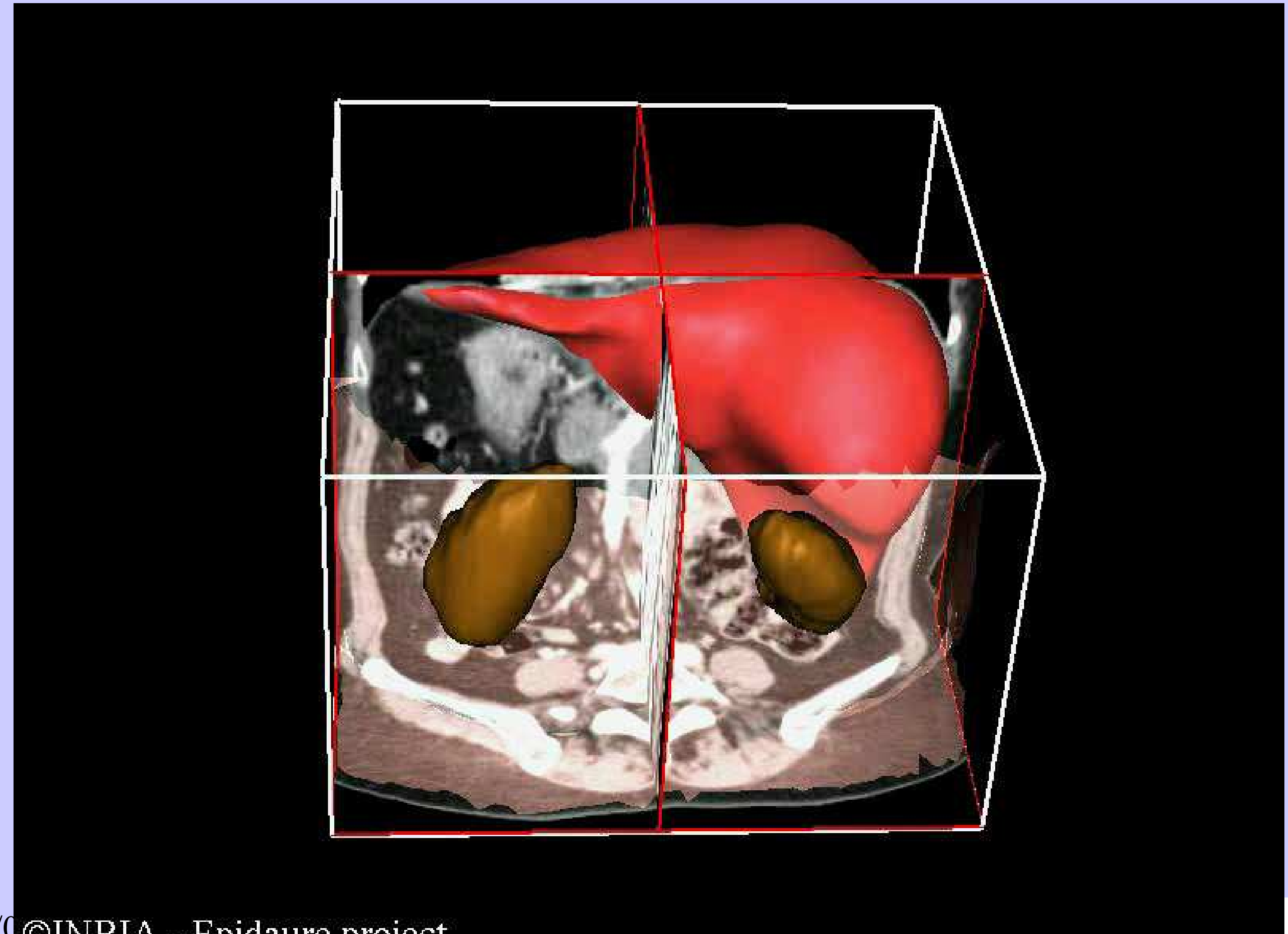
A Family of Models (3/3)

Non-Linear Elastic Model

- Large Displacements
- Small Deformations



Complete Simulation



Cardiac Modeling

CARDIOSENSE3D

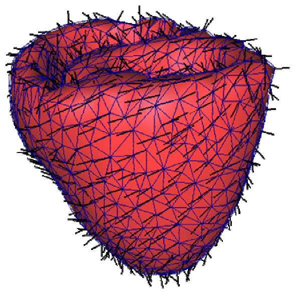
http://www.inria.fr/CardioSense3D

Acknowledgments:

- Maxime Sermesant, Valérie Moreau, Nicholas Ayache
- INRIA projects member of the ICEMA collaborative action (Sosso, Macs, Caiman, University of Nantes),
- NIH (Elliott Mc Veigh), Guy's Hospital (D. Hill)
- Philips Research France

Electromechanical Model of the Heart

Anatomical Model



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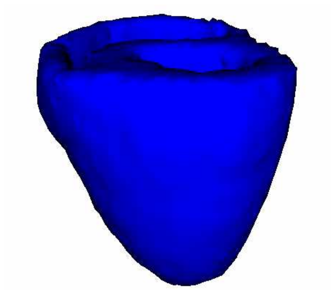
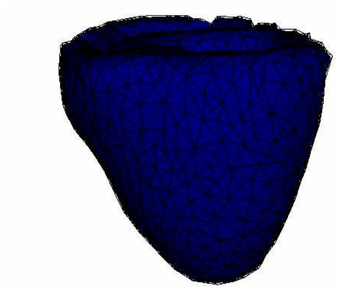
Electrical Model



↓

Electro mechanical Coupling

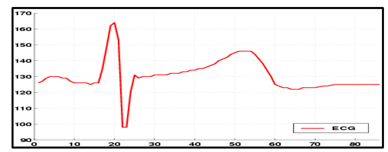


⟷ **Medical Simulator**

pathology simulation
intervention planification

⟷ **Patient Data**

ElectroCardioGram



Parameters

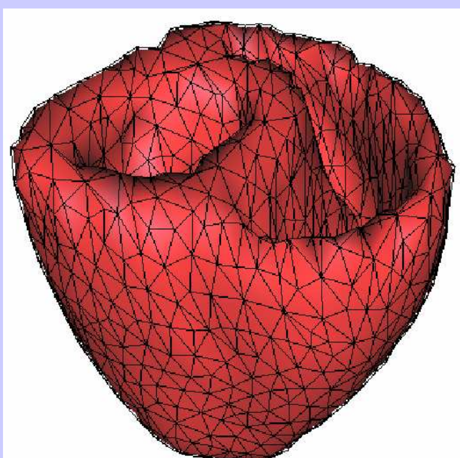
cardiac function :

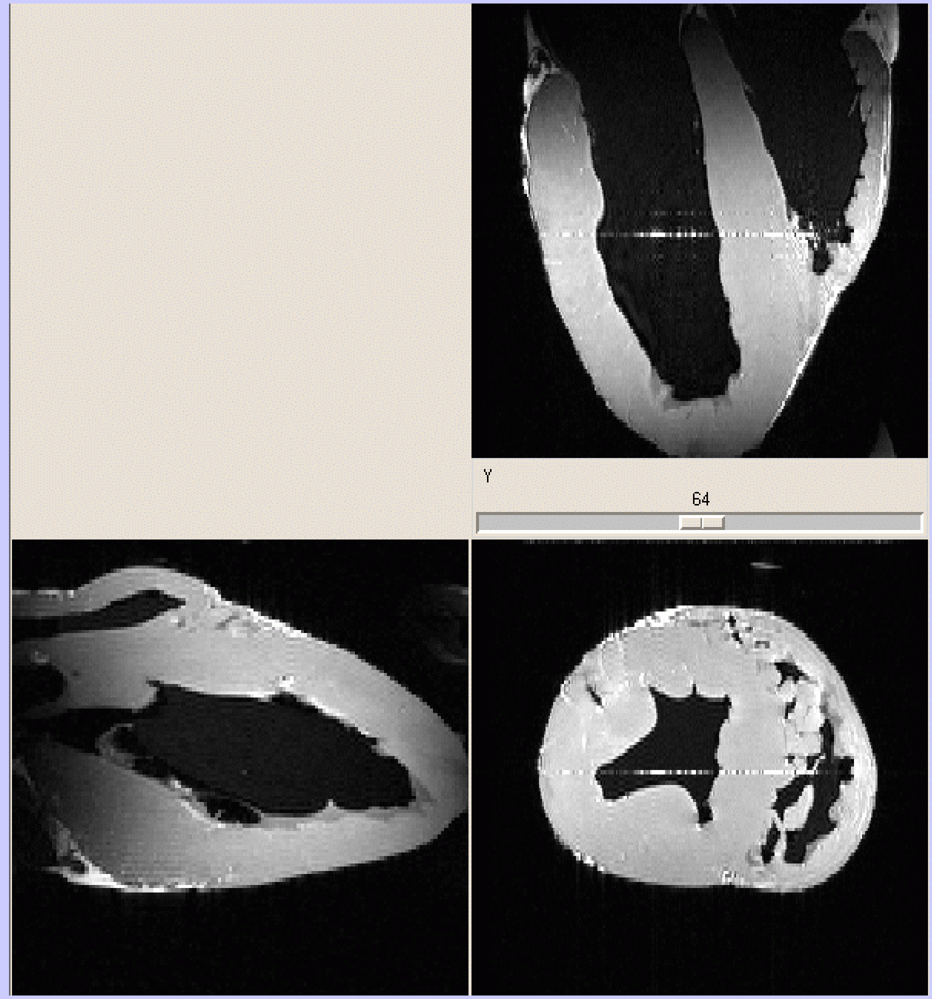
- *ejection fraction*
- *wall thickness*
- *strain, stress*

Myocardium Geometry

- Obtained from High Resolution Post-Mortem MRI of canine heart

Courtesy of Hsu, Duke University

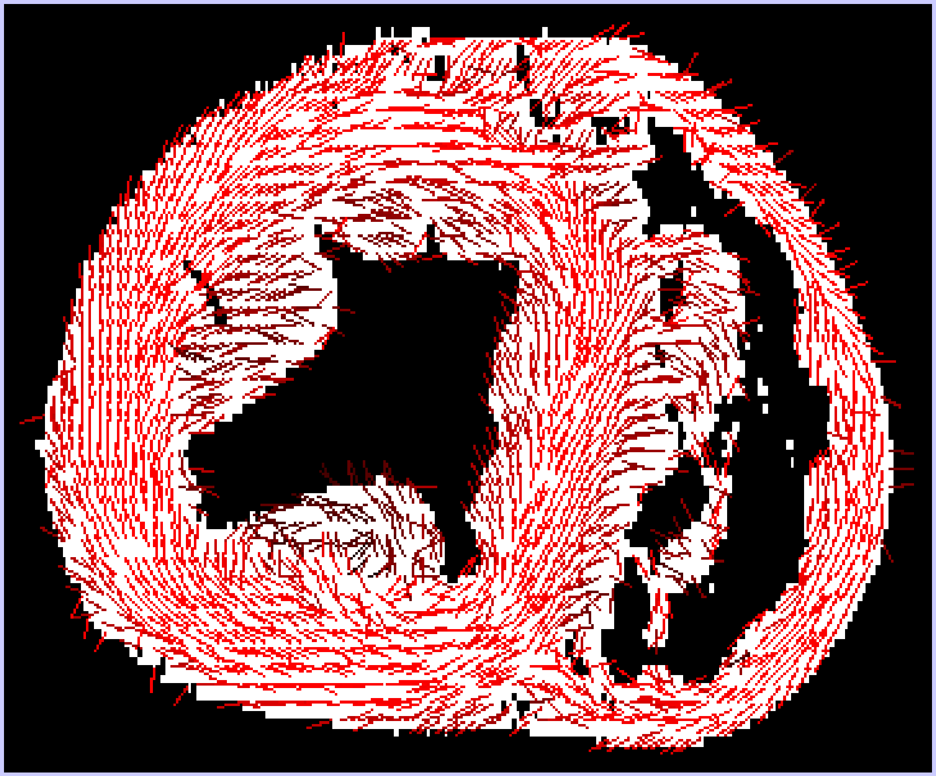


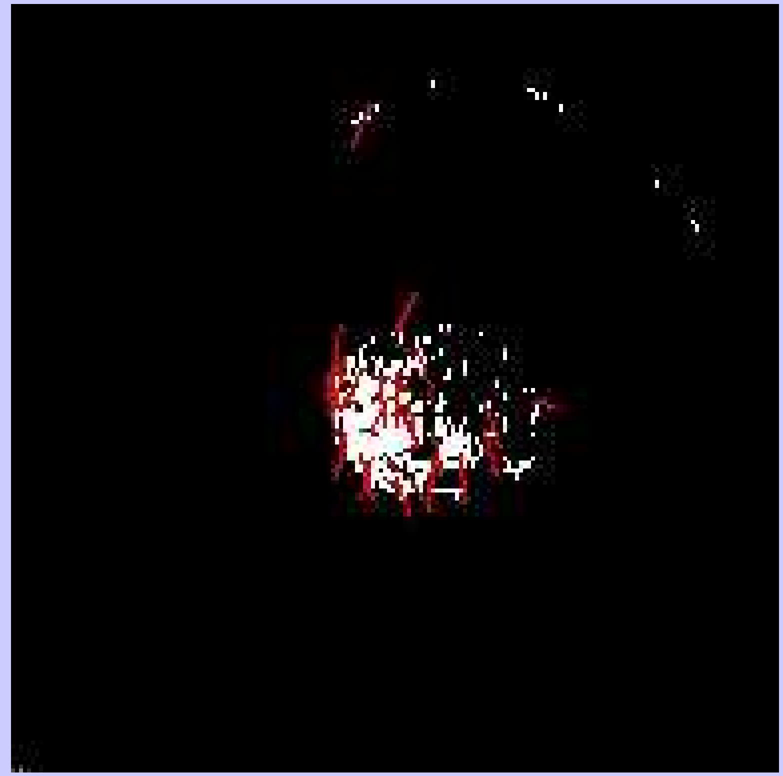


Other Data available, Courtesy of P. Hunter, Auckland University

Fiber Directions (canine Data)

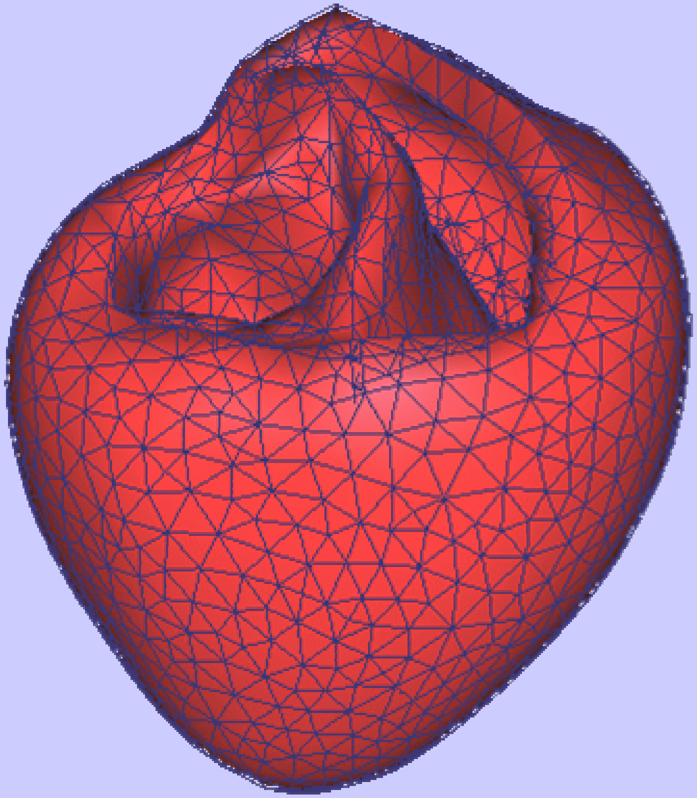
- From high resolution Diffusion Tensor MRI



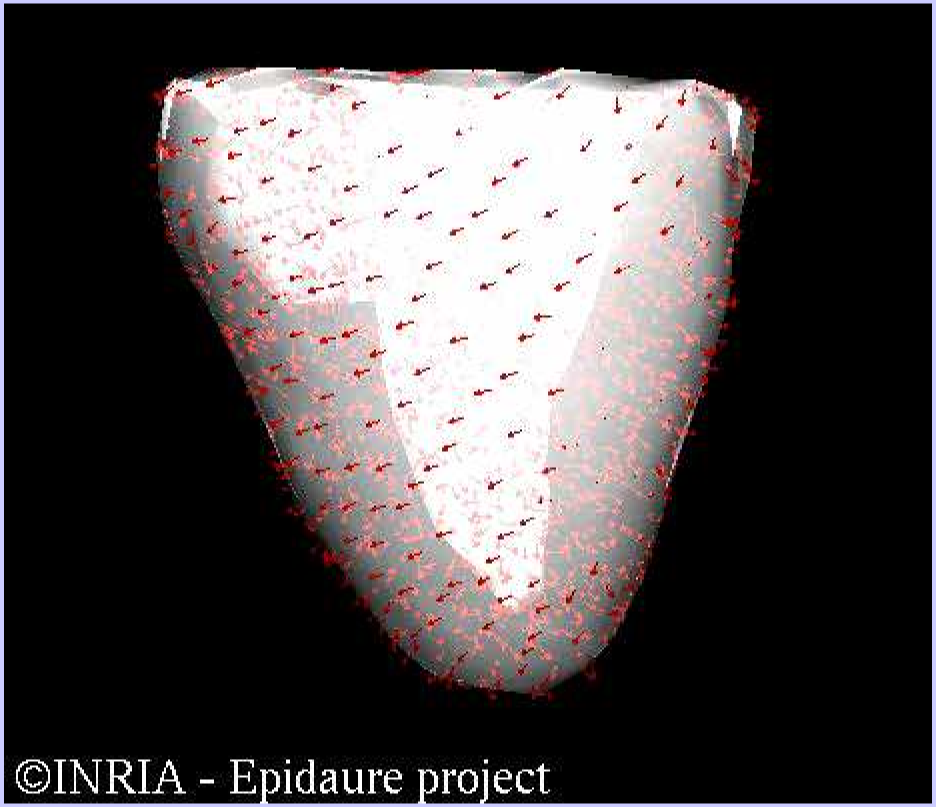


E.W. Hsu and C.S. Henriquez, *Myocardial fiber orientation mapping using reduced encoding diffusion tensor imaging*, Journal of Cardiovascular Magnetic Resonance, 2001.

Geometrical Model of Human Heart



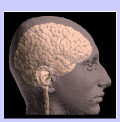
Finite Element Mesh



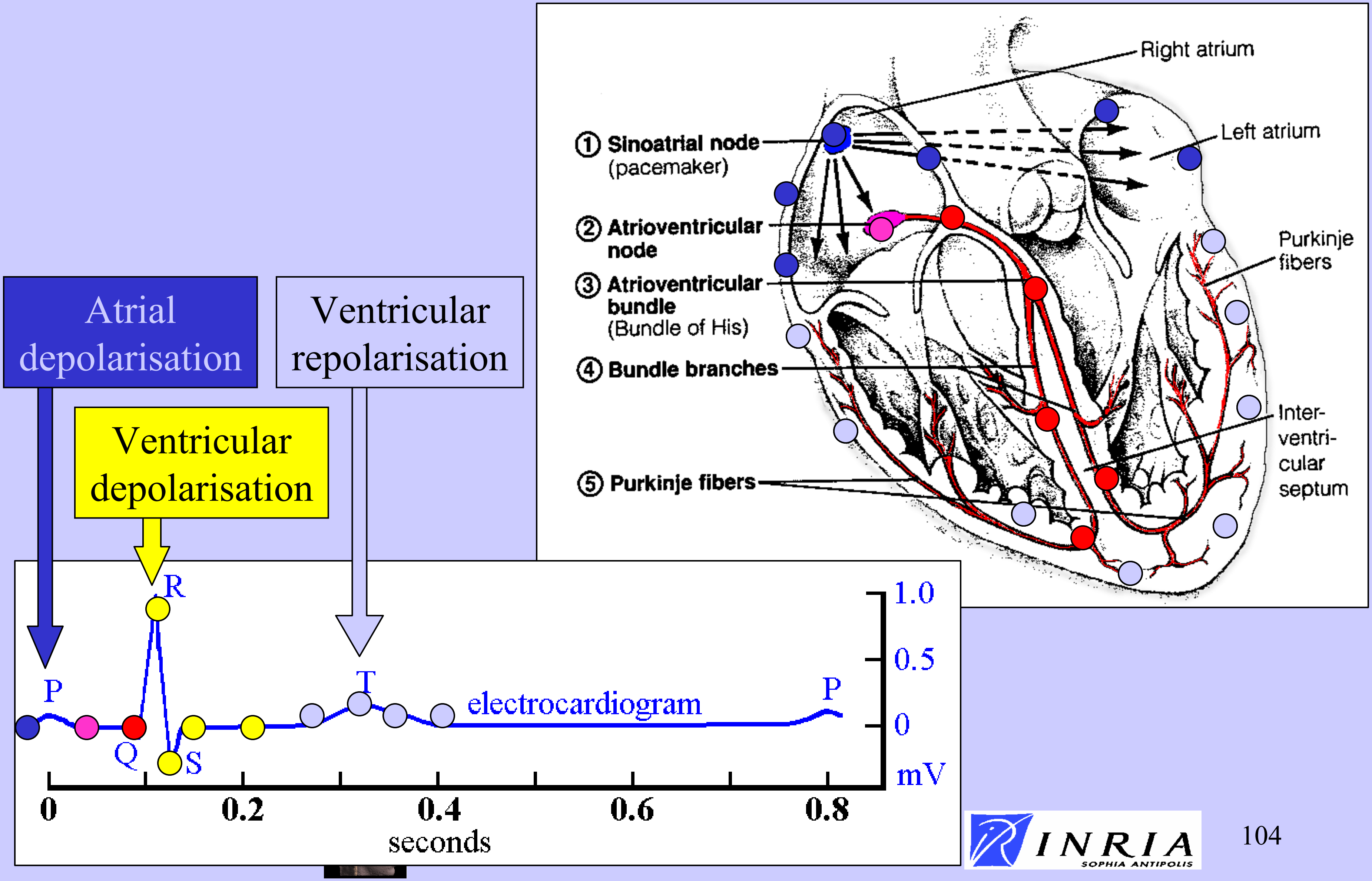
Fiber Directions

Different Levels of Modeling

- Geometrical Modeling
- Electrical Propagation
- Mechanical Coupling
- Parameter Identification



Heart Electrical Activity



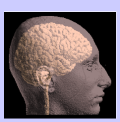
Electrical Model

- Action potential u computation: 2 variables
  FitzHugh-Nagumo *Reaction-Diffusion* system

$$\begin{cases} \dfrac{\partial u}{\partial t} = div(D\nabla u) + f(u) - z \\ \\ \dfrac{\partial z}{\partial t} = b(u - cz) \end{cases}$$

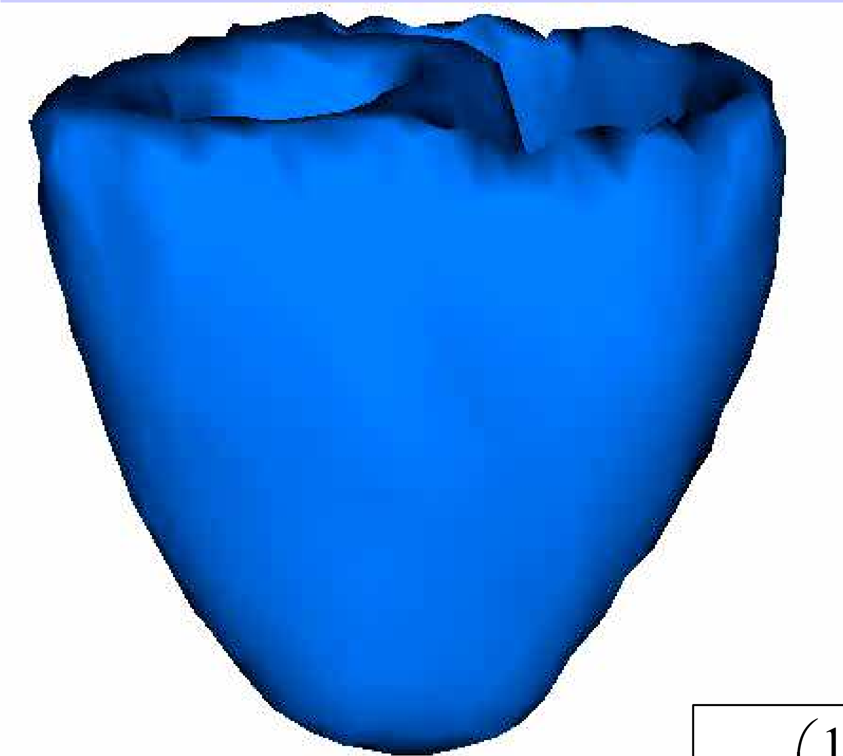
u action potential
D diffusion tensor
f ionic current
z repolarization variable
b repolarisation rate
c repolarisation decay

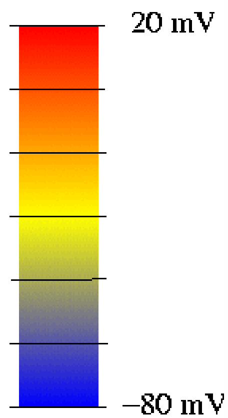
Or R. Aliev and A. Panfilov : A Simple Two-variable Model of Cardiac Excitation, *Chaos, Solitons & Fractals*, Vol 7, No 3, pp. 293-301,1996

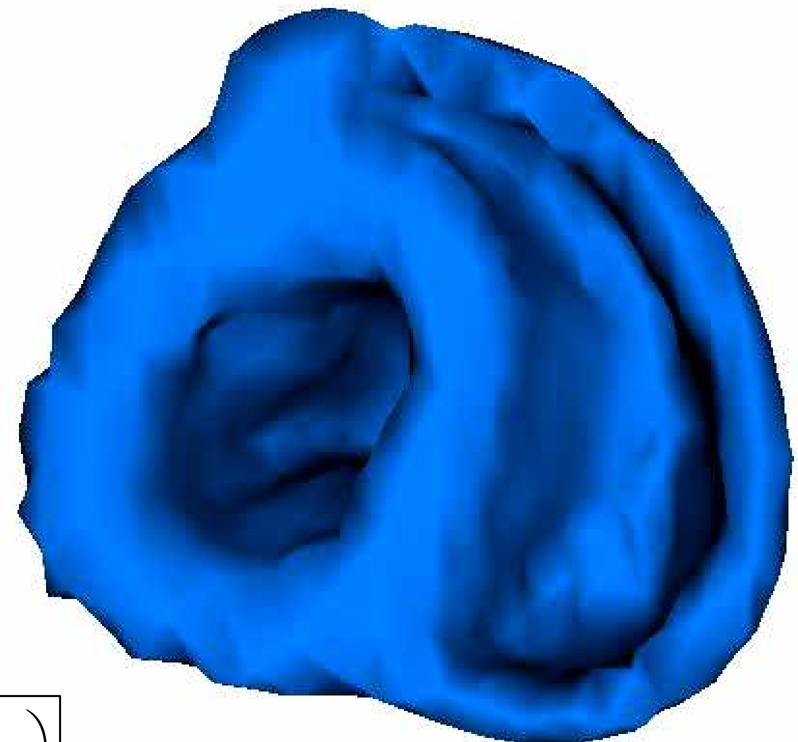


Electrical Simulation

Anisotropic model (fiber geometry + Purkinje network)



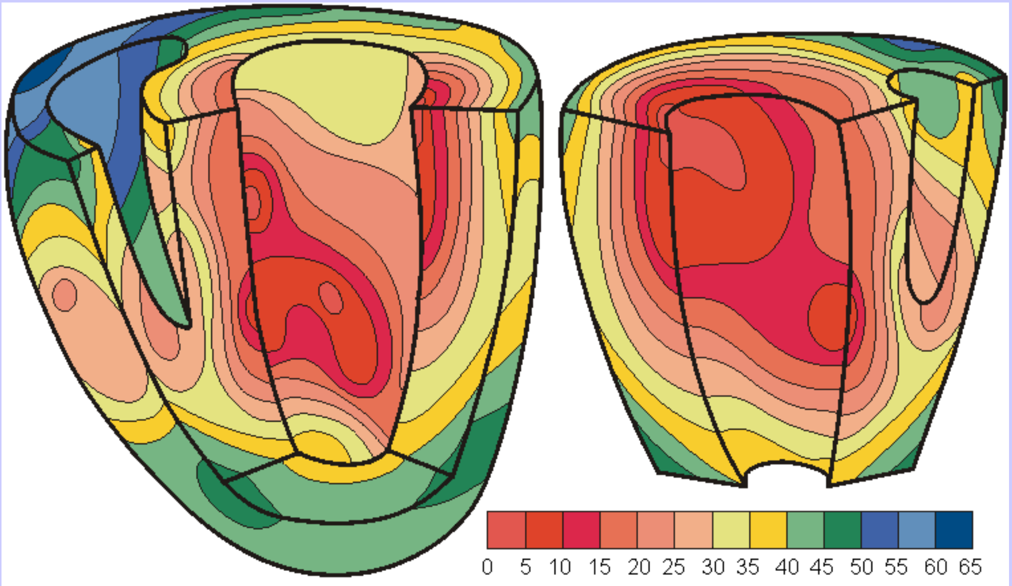




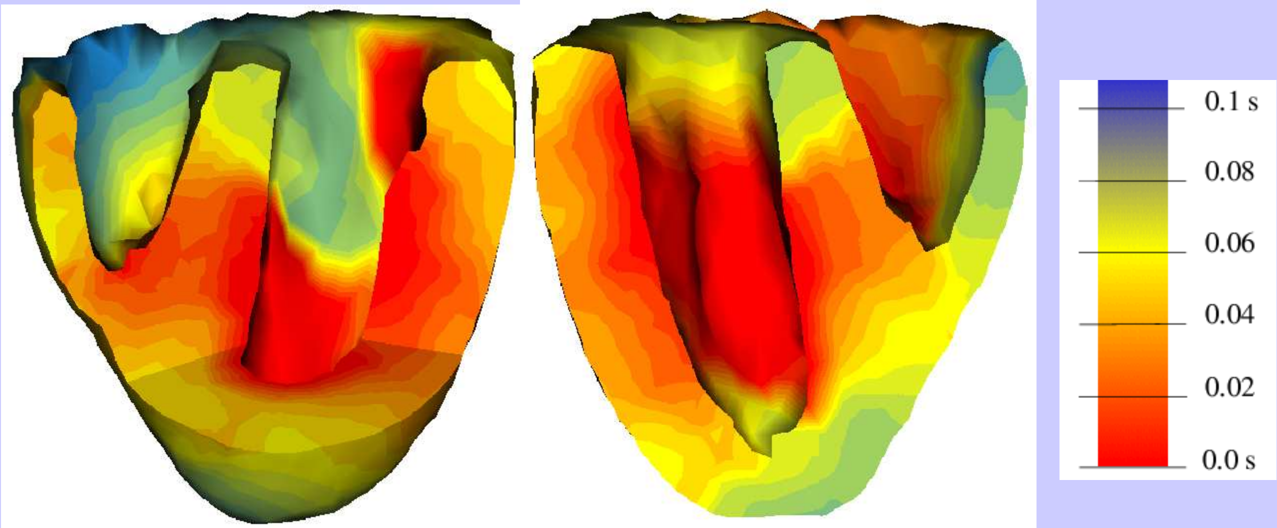
$$D = \begin{pmatrix} 1 & 0 & 0 \\ 0 & 0.25 & 0 \\ 0 & 0 & 0.25 \end{pmatrix}$$

Comparison of Isochrones

Durrer et al.



ICEMA at INRIA

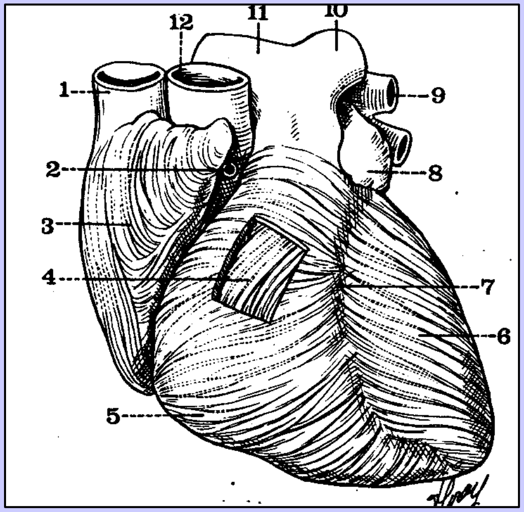


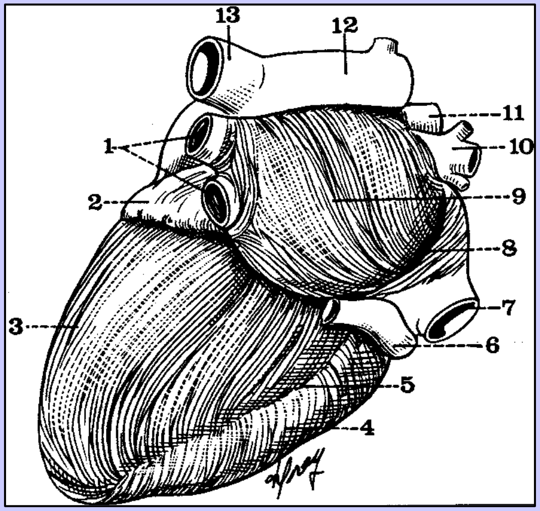
Different Levels of Modeling

- Geometrical Modeling
- Electrical Propagation
- Mechanical Coupling
- Parameter Identification

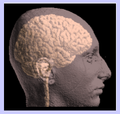
Biomechanical Properties

The myocardium is composed of muscle fibre bundles:



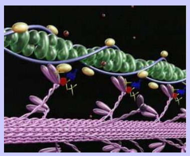
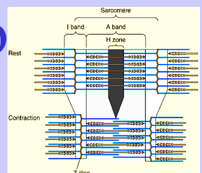
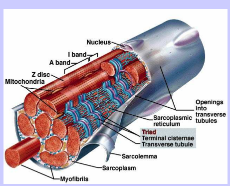
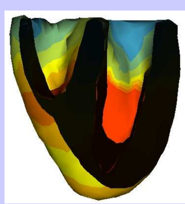


It is an active non-linear viscoelastic anisotropic incompressible material.



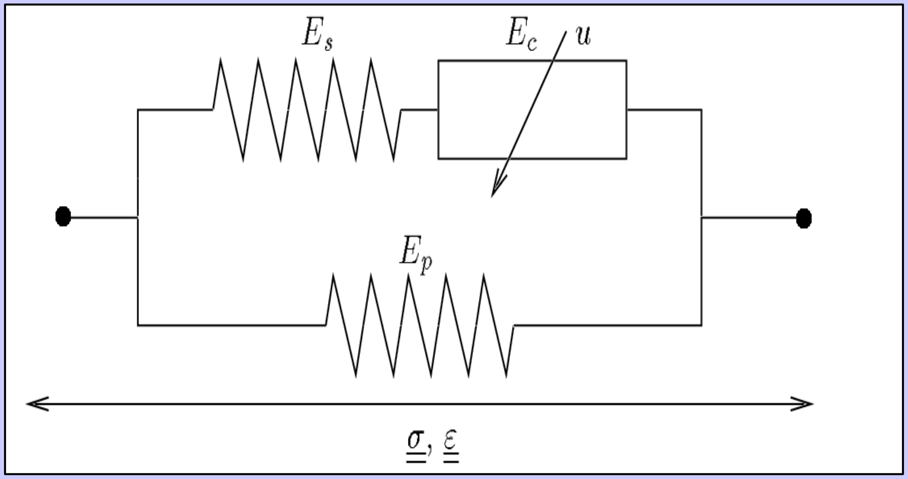
Excitation-Contraction Coupling

From Bestel-Clément-Sorine, MICCAI'01

Scale:		System:	Control:
Nano		molecular motors *Langevin equations* *(SDE) Ratchet or Power-stroke models*	Calcium ions *still to be designed…*
Micro		sarcomeres *Huxley-like models* *(PDE)*	ionic currents *Luo-Rudy-like models* *(ODE)*
Meso		myocytes *BCS* model *(ODE)*	action potential *FHN-like* models *(ODE)*
Macro		myocardium *dynamics equations* *(PDE with BCS Constitutive Law)*	action potential *FHN-like* models *(PDE)*

Electro-Mechanical Coupling

- Derived from nano to macroscopic scale
- Bestel-Clément-Sorine constitutive law in Hill-Maxwell type rheological model



E_S series element
E_p parallel element
E_c contractile element
u action potential
σ stress
ε strain

- E_S and E_p: elastic material laws,
- E_c contractile electrically-activated element.

J. Bestel, F. Clément, and M. Sorine. A Biomechanical Model of Muscle Contraction. In *Medical Image Computing and Computer-Assisted Intervention* (MICCAI'01), 2001.

Bestel-Clément-Sorine Myofiber Model

Electromechanical coupling system derived from *nanoscopic* to *mesoscopic* scale

$$\begin{cases} \dfrac{dk_c}{dt} = -\left(|u| + \left|\dfrac{d\varepsilon_c}{dt}\right|\right)k_c + k_0|u|_+ \\[2ex] \dfrac{d\sigma_c}{dt} = -\left(|u| + \left|\dfrac{d\varepsilon_c}{dt}\right|\right)\sigma_c + k_c\dfrac{d\varepsilon_c}{dt} + \sigma_0|u|_+ \end{cases}$$

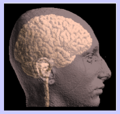
k_c contractile stiffness

u electrical action potential

ε_c contractile strain

σ_c contractile stress

J. Bestel, F. Clément, and M. Sorine. A Biomechanical Model of Muscle Contraction. In *Medical Image Computing and Computer-Assisted Intervention (MICCAI'01)*, 2001.



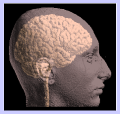
F_c: Simplified electromechanical model for image segmentation

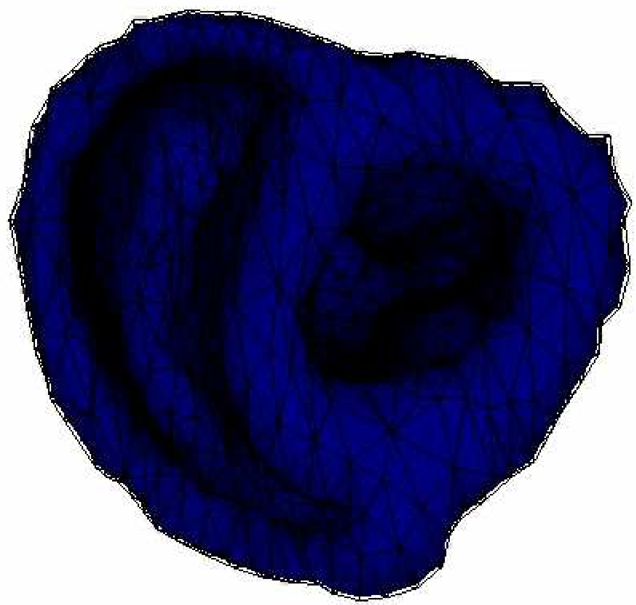
k piecewise constant

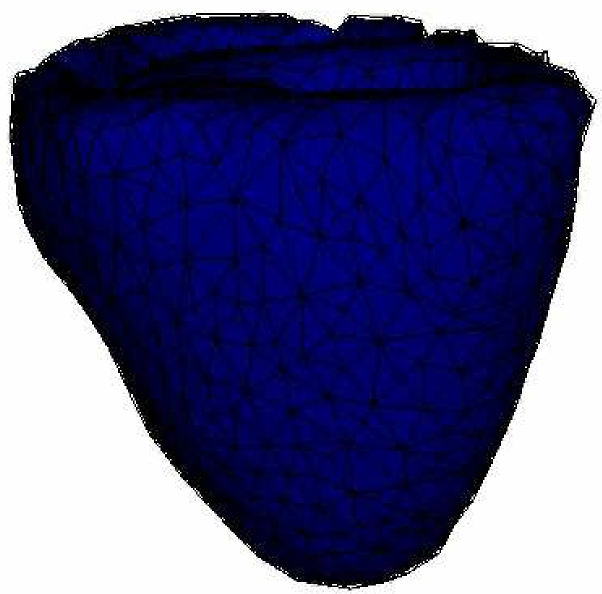
→ Piecewise linear viscoelastic anisotropic material

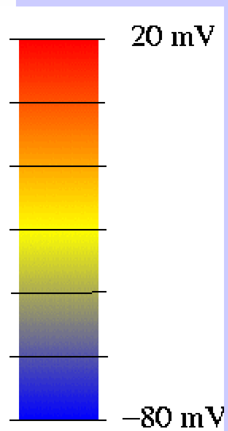
$$\frac{d\sigma_c}{dt} = -|u|\sigma_c + \sigma_0 |u|_+$$

→ Only electrical command on contraction stress σ_c









Electro-Mechanical coupling

- the "Action potential" u controls contractile element:

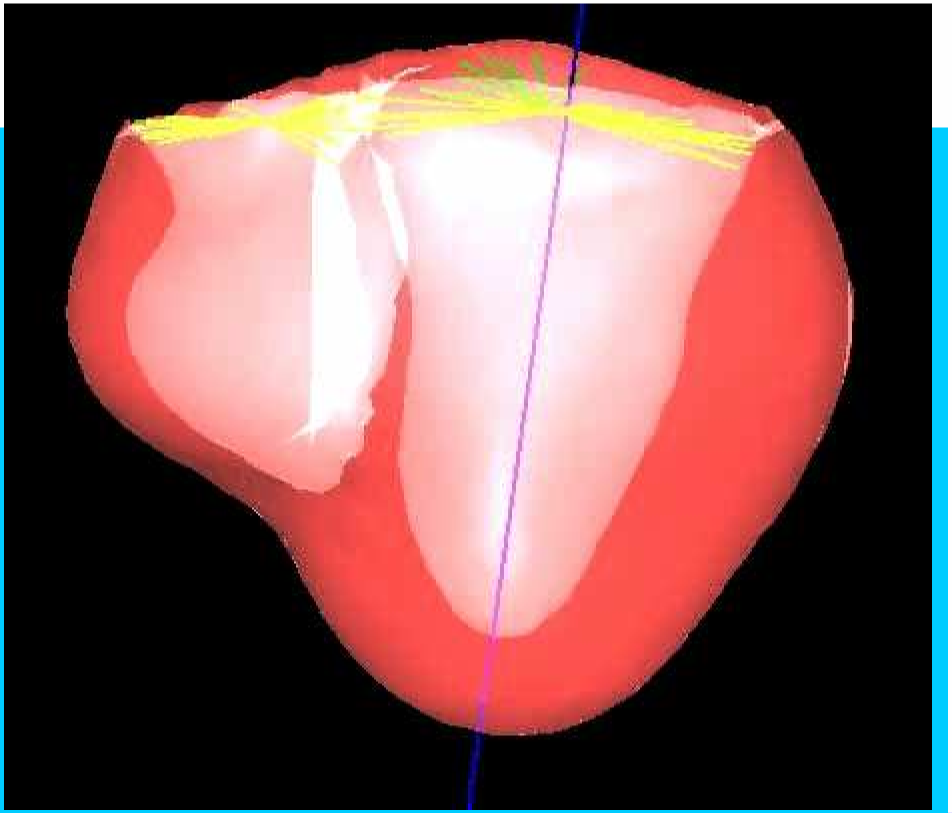
  $u > 0$: Contraction

  $u \leq 0$: Relaxation

- u also modifies stiffness k of the material.

- Ayache-Chapelle-Clément-Coudière-Delingette-Sermesant-Sorine-FIMH'01
- Sermesant-Coudière-Delingette- et al., MICCAI'02

Cardiac Cycle Simulation



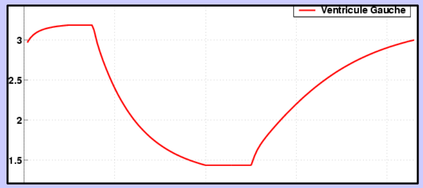
- 4 phases :
  - filling
  - isovolumetric contraction
  - ejection
  - isovolumetric relaxation
- 2 Boundary Conditions :
  - Pressure constraint on the endocardium
  - Isovolumetric constraint on the endocardium

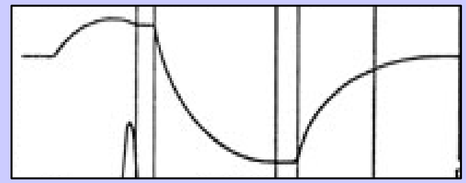
Cardiac Cycle Parameters

Simulated

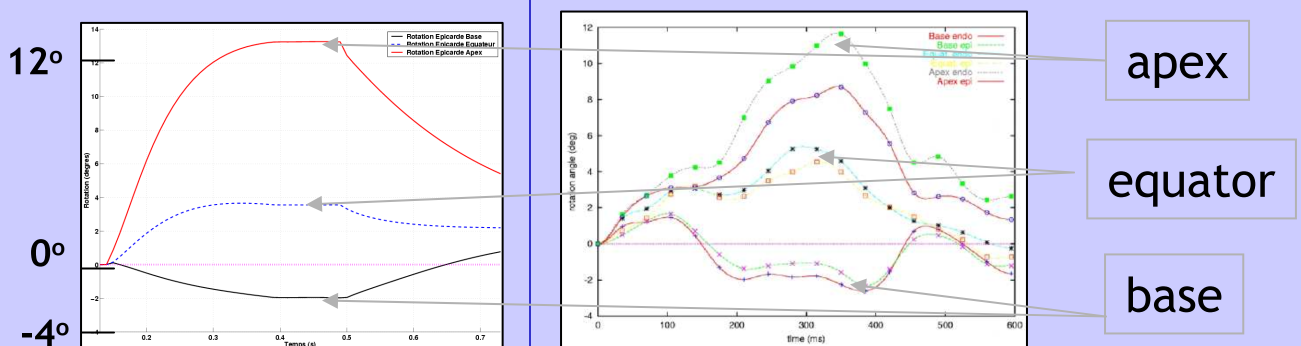
Measured par Philips
(tagged MRI)

global
volume

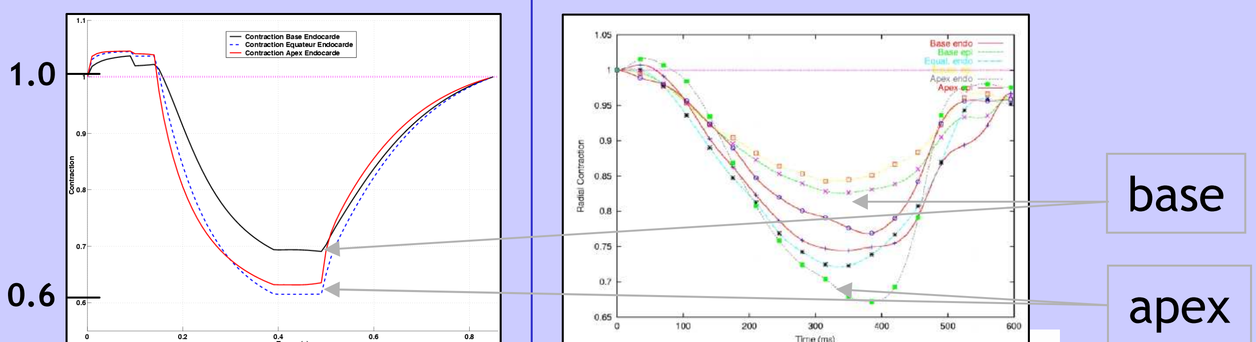


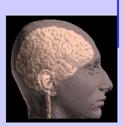


local
twist

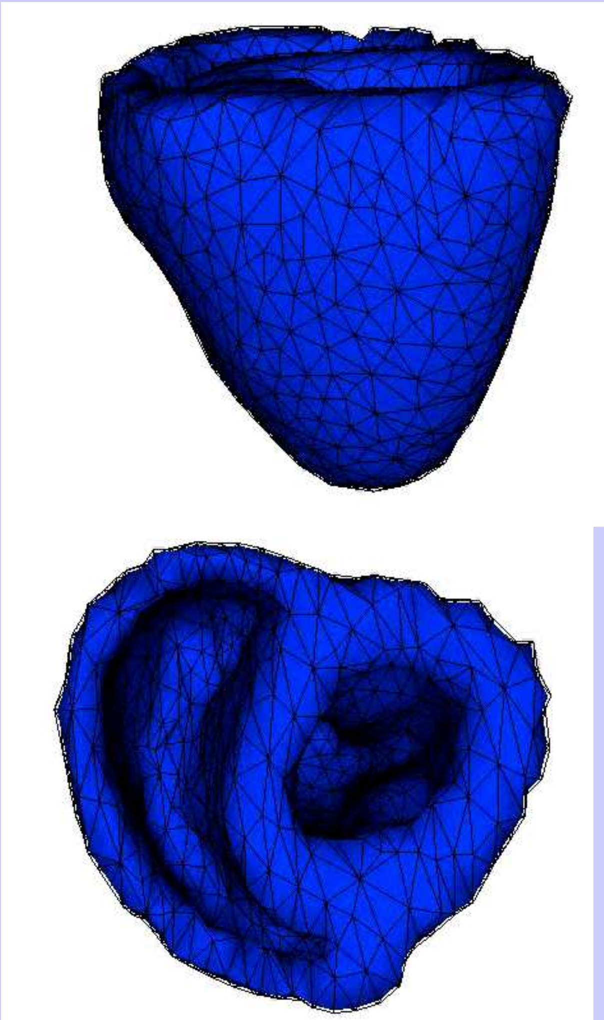


radial
contraction

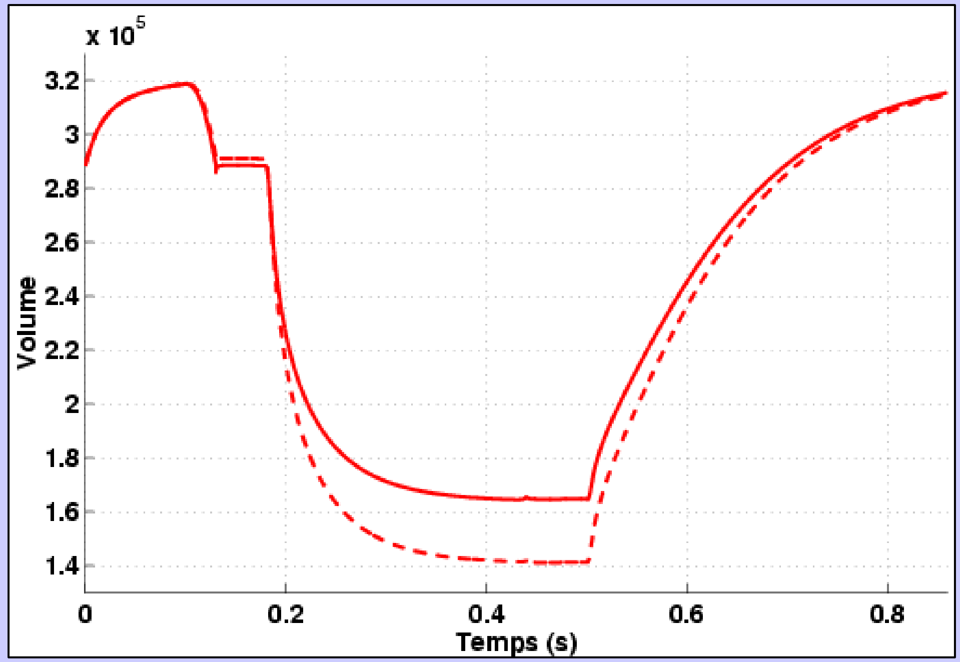




Simulating an Infarct



Infarcted zone simulation

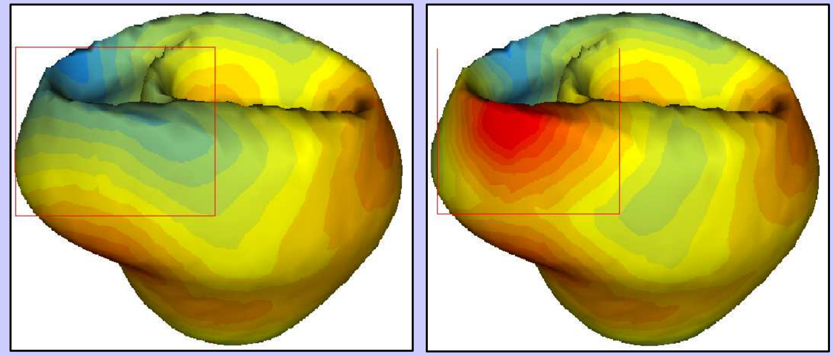


volume

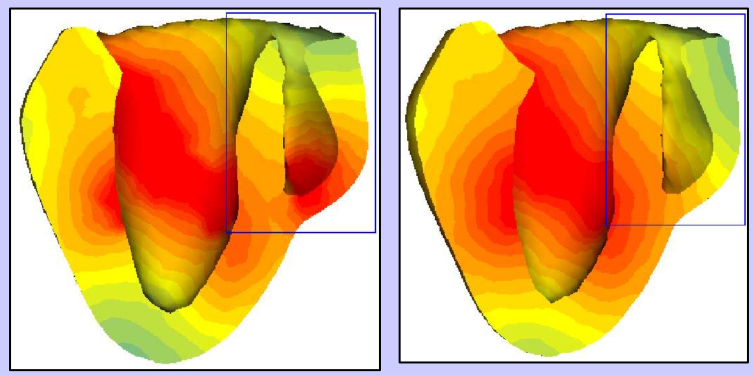
ejection fraction:
56 % → 48 %

Simulating Electrical Pathologies

- Ectopic Focus in RV
  - Wolff-Parkinson-White

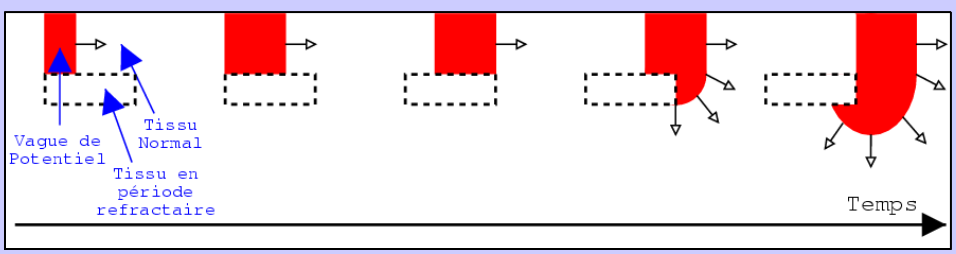


- Bundle of His defect in RV
  - Right Branch Block

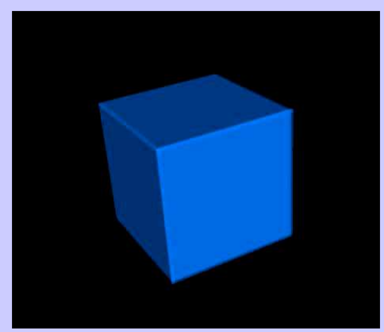


pathological

- Fibrillation



Wave-break



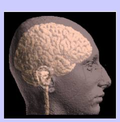
Brain Soft Tissue Modeling

Acknowledgments:

- Olivier Clatz, Nicholas Ayache
- Hopital La Pitié Salpétrière (D. Dormont)
- Centre Antoine Lacassagne (P.-Y. Bondiau)
- Brigham and Women's Hospital (S. Warfield)

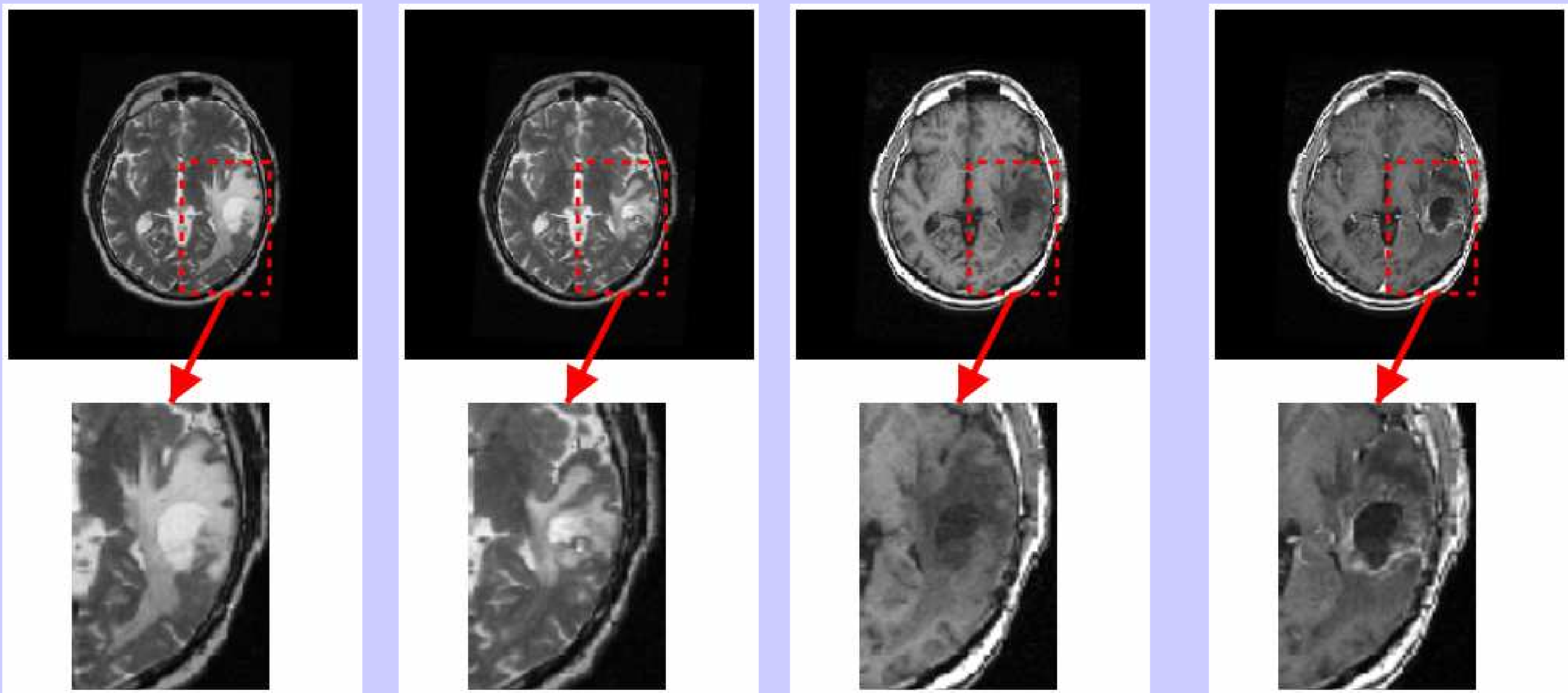
Applications

- Glioblastoma growth modeling
  - Objectives :
    - Understanding tumor growth
    - Improve radiotherapy planning
- Brain-shift modeling
- Image registration based on biomechanical modeling



Input Data (patient)

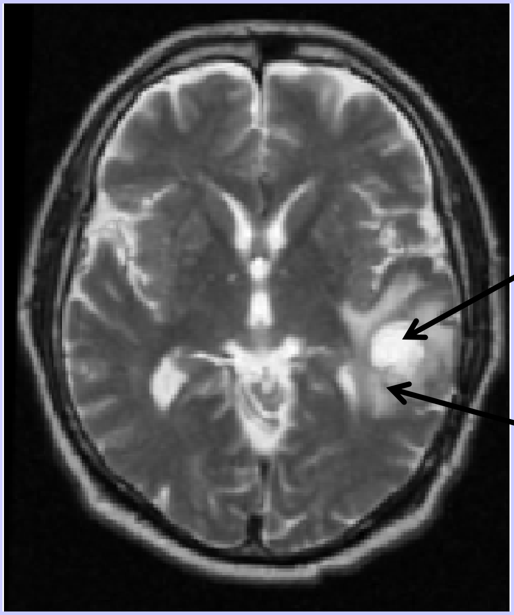
- 2x3 images acquired 6 months apart



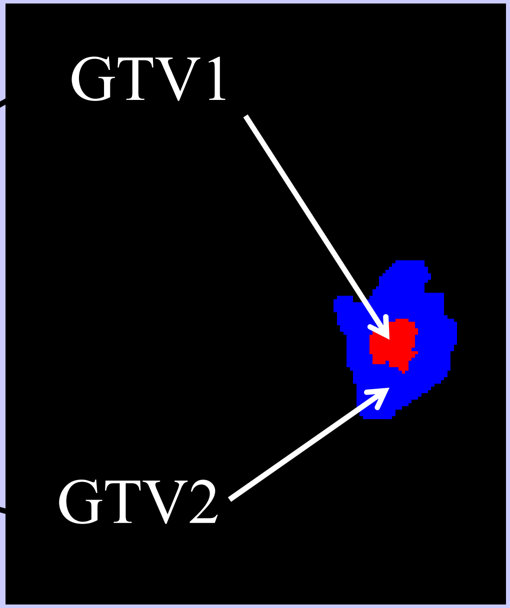
T2 september	T2 march	T1 march	T1 + gad march

Input Data (patient)

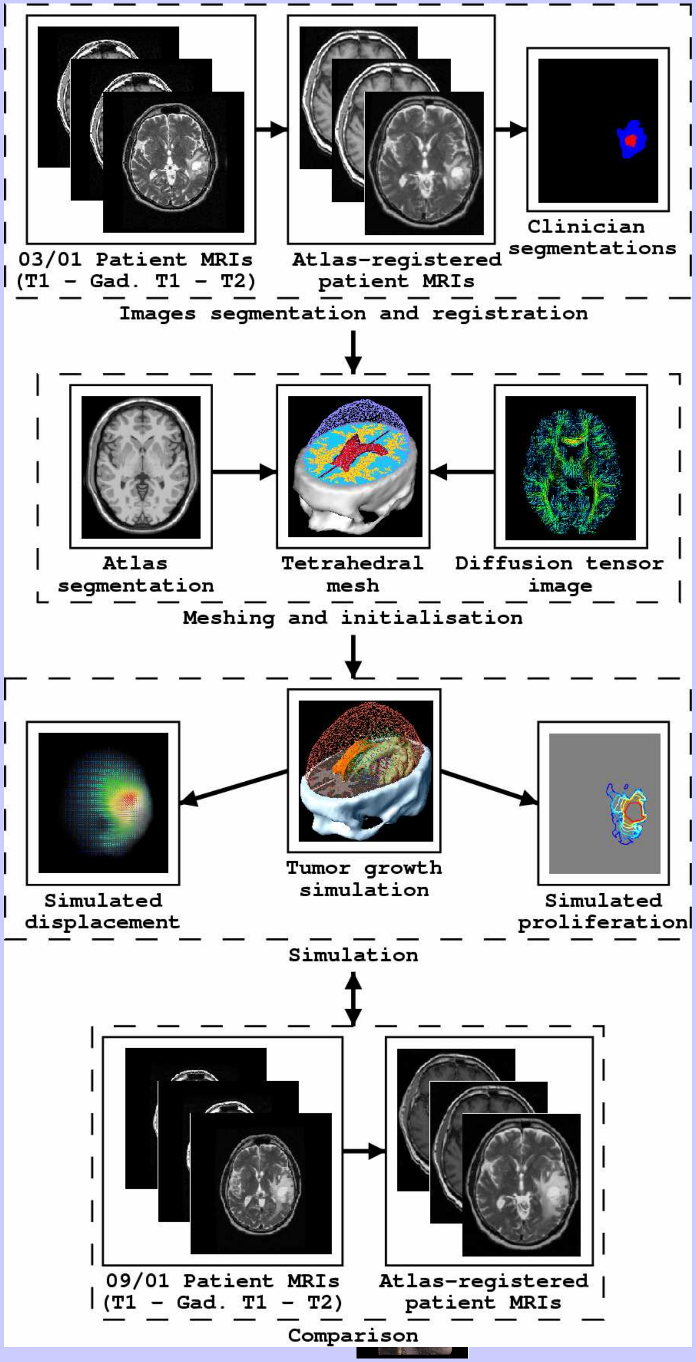
- Segmentation of target tissues



T2 MRI



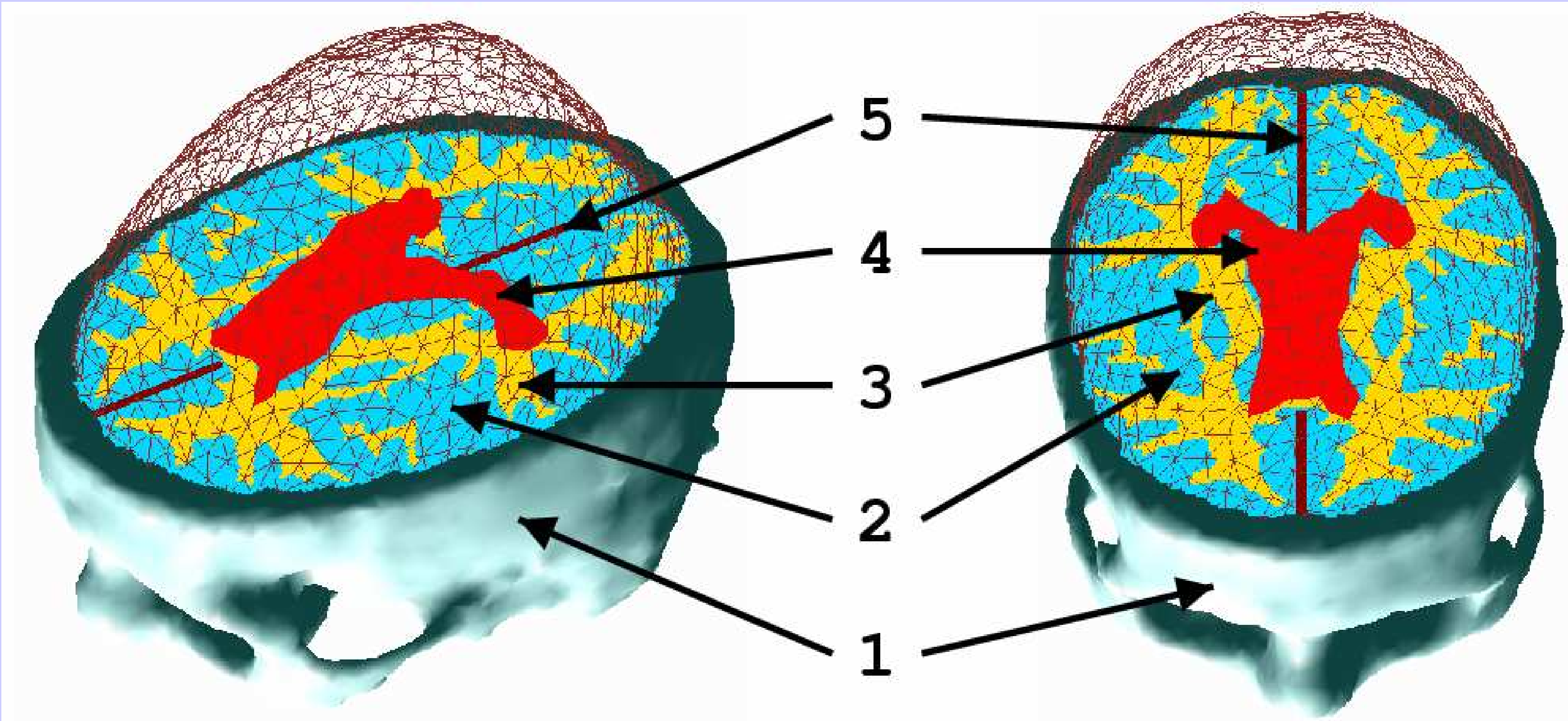
Segmentation GTVs



Methods

- Images + segmentation
- Initialization with atlas
- FE Method
    - Diffusion
    - Mechanical
- Comparaison with patient

Mesh Construction



1 Skull. 2 Grey Matter. 3 White Matter
4 Ventricles. 5 Falx

Model Equations

Diffusion + source terms

$$\frac{\partial u}{\partial t} = div(D\nabla u) + \rho u$$

u = density of tumor cells

D = Diffusion Tensor

ρ = Source Term

Mechanical Constitutive Law

Linear Elasticity : $\sigma = \mathrm{k}\varepsilon$

Équilibre : $\mathrm{div}\ \sigma + F_{ext} = 0$

σ = stress

ε = strain : $\varepsilon = \frac{1}{2}(\nabla x + \nabla x^{t})$

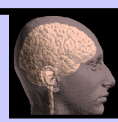
k = stiffness

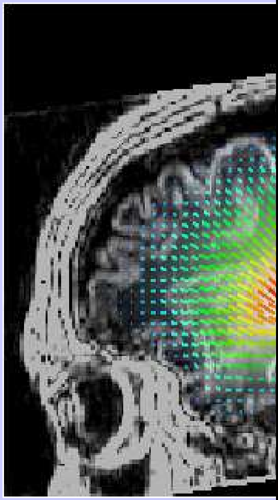
F_{ext} = External force

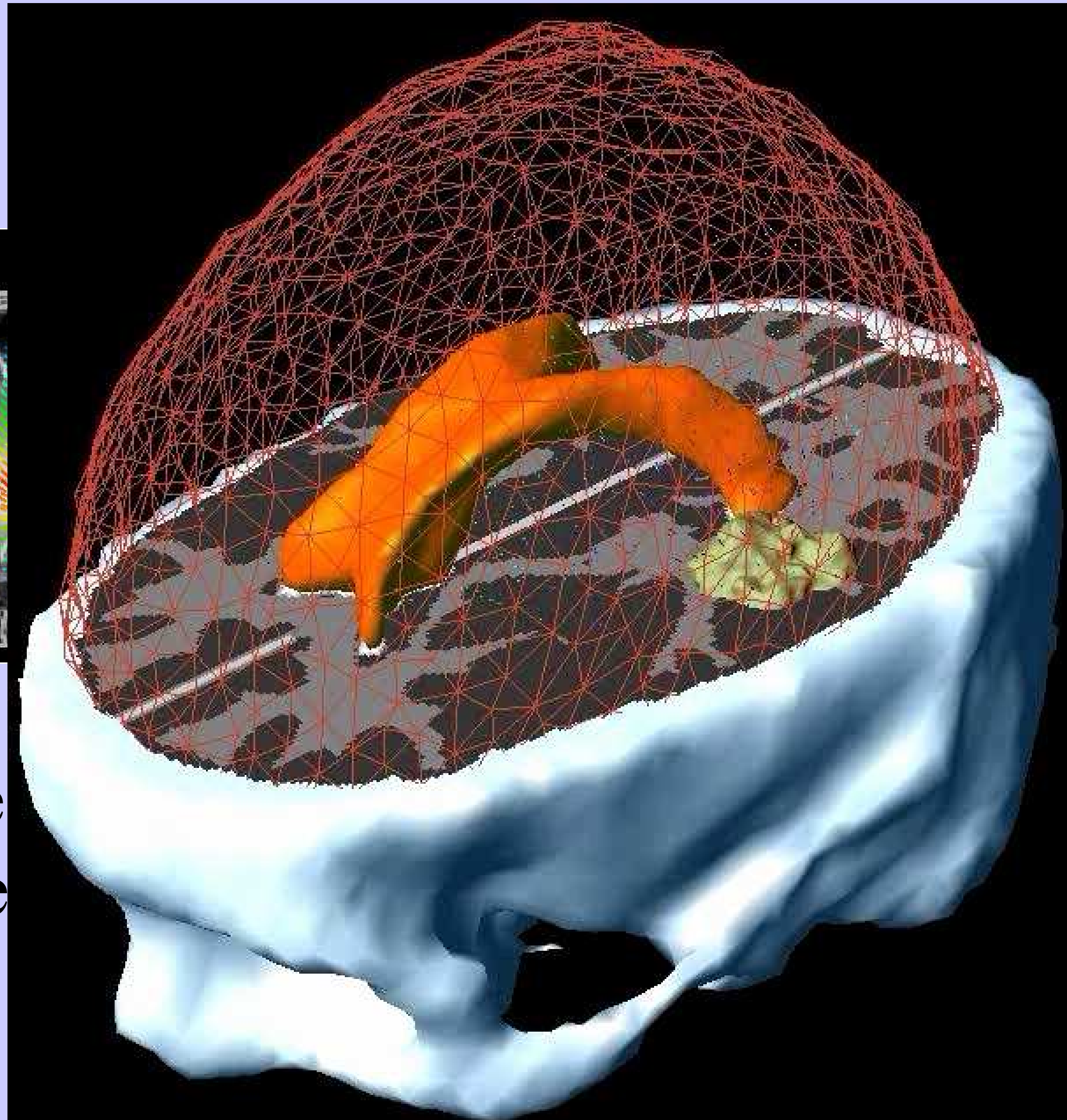
Coupling

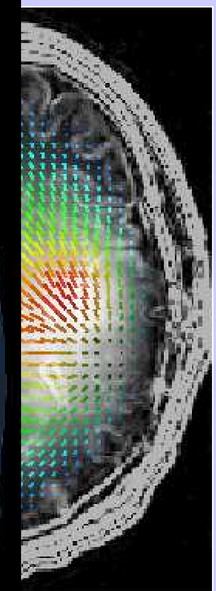
$$F_{ext} = \lambda \nabla u$$

λ = coupling factor

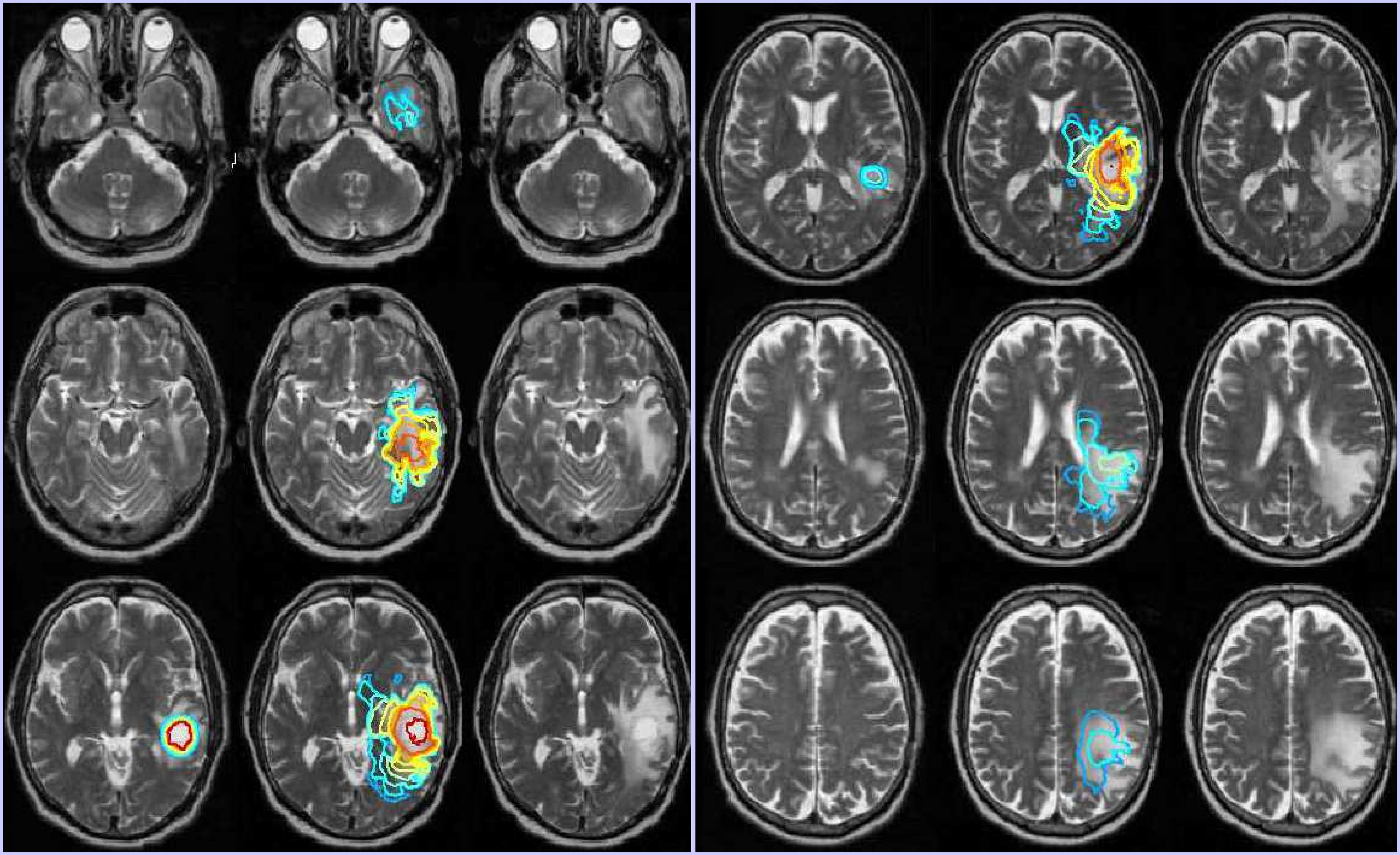








- Me n
  ave



March | September simulation | September | March | September simulation | September