Soft Tissue Modeling

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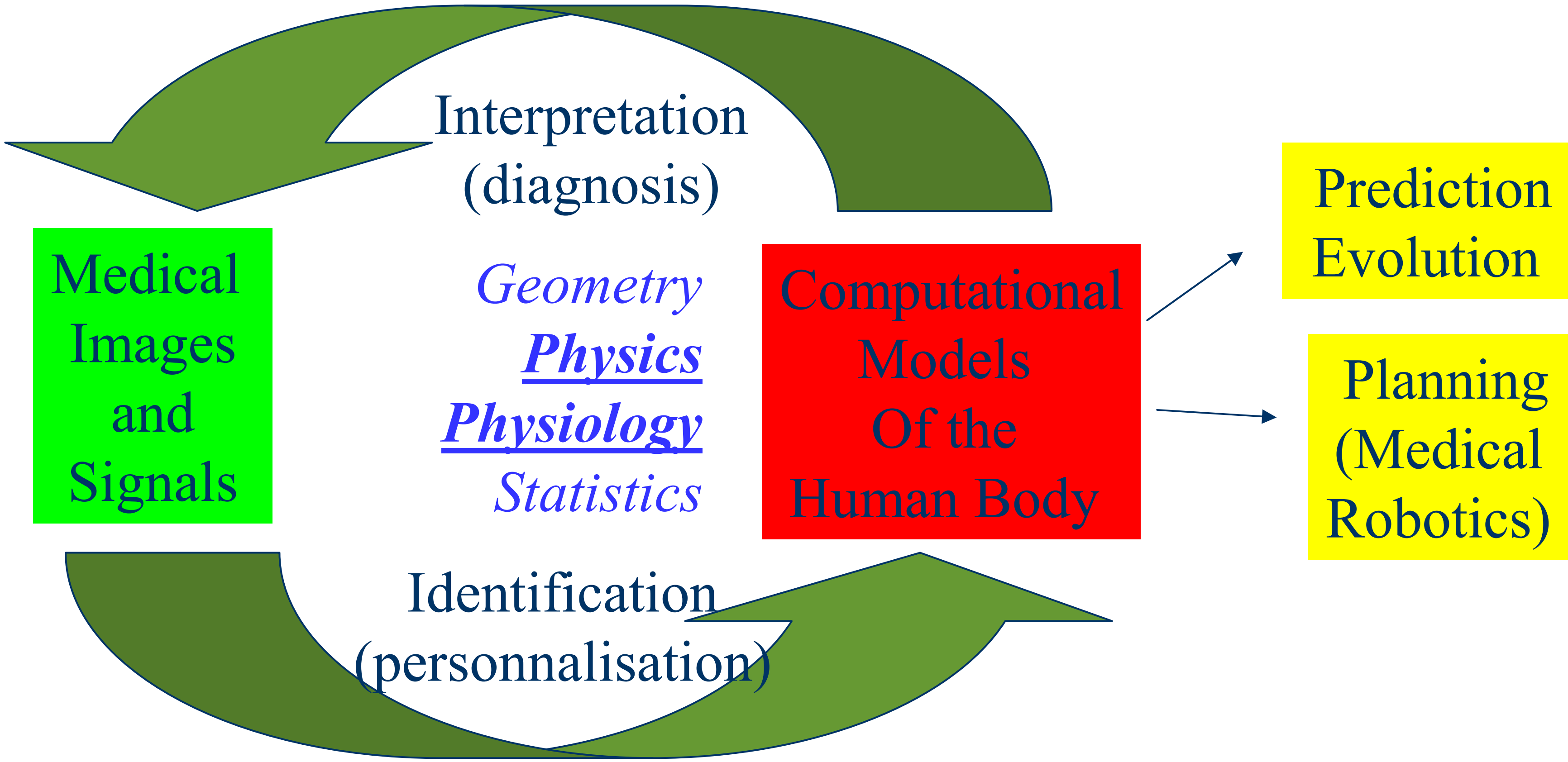
Overview

- Introduction
- Constitutive Modeling of Soft Tissue
  - Discrete Models (Particles + Spring Mass Models)
  - Continuum Models (Hyperelasticity)
- Measuring Soft Tissue Deformation
- Discretization Methods
- Example of Soft Tissue Models

Soft Tissue Modelling

- Important for :
  - Creating Physiological Models of the human body (understanding / teaching)
  - Image Registration (deformation prior)
  - Diagnosis : finding pathological behavior
  - Therapy planning
  - Procedure simulation

Computational Models of the Human Body



Soft Tissue Deformation

- Objectives of a soft tissue models :
  - Reproductive :
    - Able to produce simulation comparable to observations
  - Discriminative :
    - Associated parameters used to cluster a given population
  - Predictive :
    - Predict behavior under specific conditions

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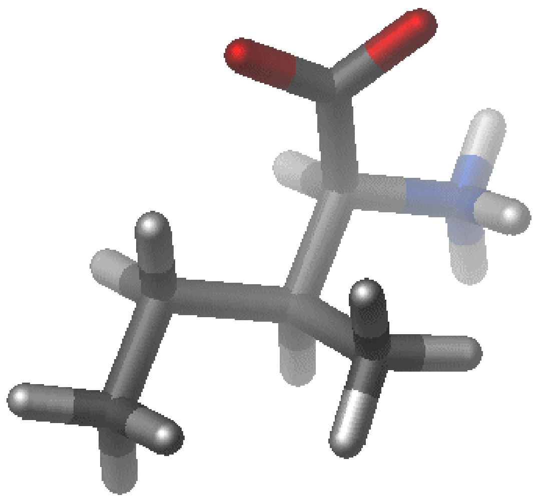
Discrete Constitutive Models

- Models defined on a set of points
  - Particles : material points in interaction
    - One particle interacts with all other particles (no fixed neighborhood)
    - Use of an interaction potential based on :
      - Distance between 2 particles
      - Angles between 3 particles
      - Angles between 4 particles
    - Link with molecular dynamics



Source http://www.ch.embnet.org/MD_tutorial/pages/MD.Part2.html

Discrete Constitutive Models

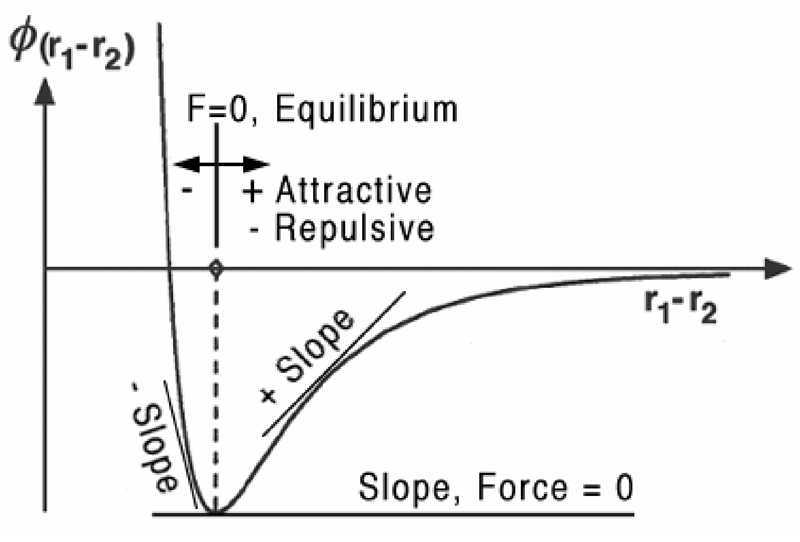
- Models defined on a set of points
  - Particles :
    - Example of potential : lennard-jones

Energy potential between particle I and j

$$\phi_{i \leftrightarrow j}$$

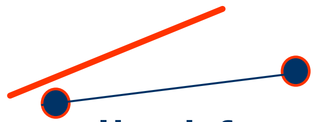
Force applied by particle j on particle i

$$F_{j \to i} = -\frac{\partial \phi_{i \leftrightarrow j}}{\partial P_i}$$

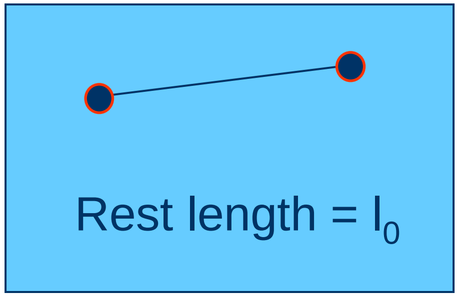


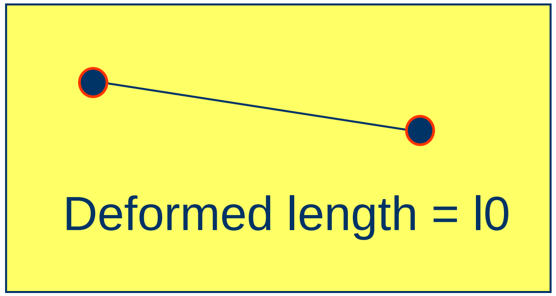
Discrete Constitutive Models

- Models defined on a set of points
  - Springs :
    - Based on a set of edges (fixed topology)



    - **Main idea** : applied force is proportional on the variation of edge length :

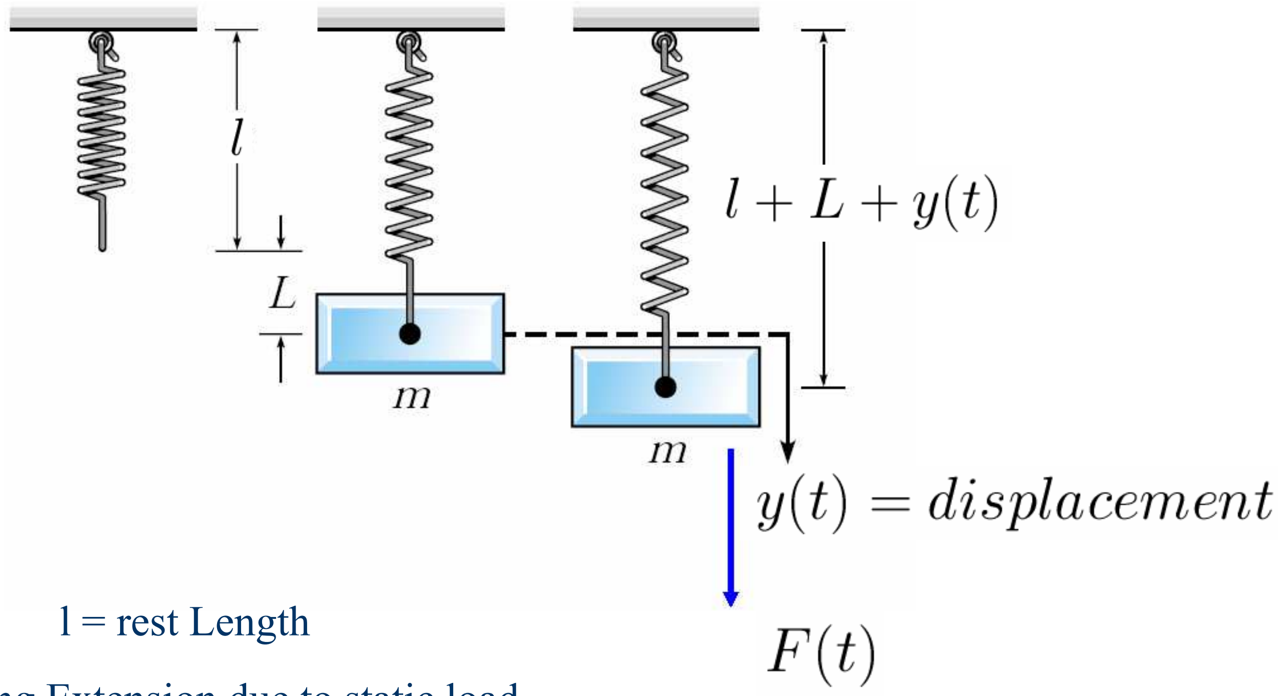




Force proportional to $l - l_0$ ⟺ Energy proportional to $(l - l_0)^2$

1D Spring-mass system

Dynamic problem: What is motion of the mass when acted by an external force or is initially displaced?



l = rest Length

L = Spring Extension due to static load

y(t) = Spring extension due to dynamic load

1D Spring-mass system

Newton's Second Law of Motion

the acceleration of an object due to an applied force is in the direction of the force and given by:

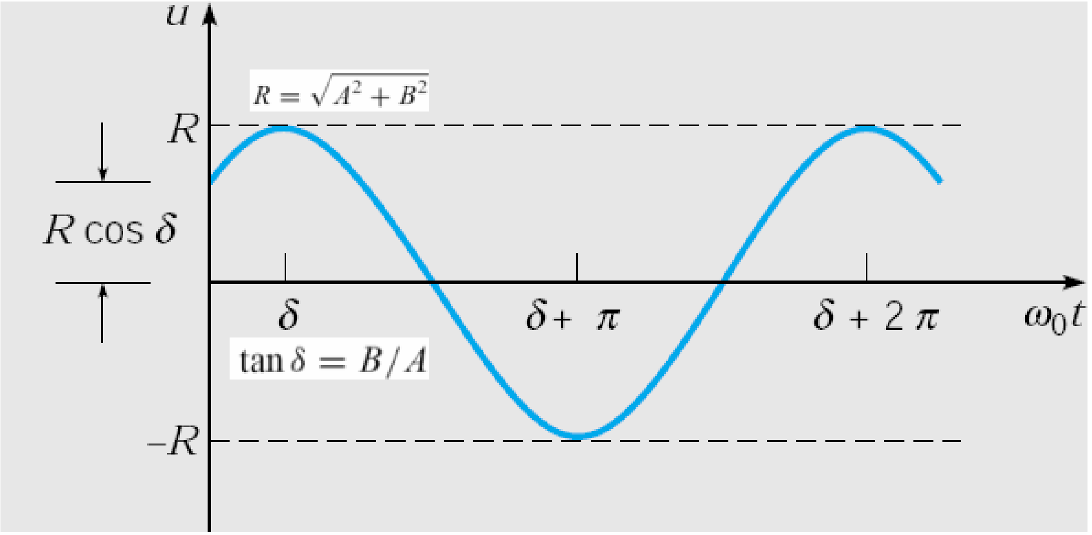
$$F_{net} = ma = my''(t)$$

For our spring-mass system

$$my''(t) = \underbrace{F(t) - cy'(t) - ky(t)}_{F_{net}}$$

$$\boxed{my''(t) + cy'(t) + ky(t) = F(t)}$$

Undamped Free Vibrations



$$T = \frac{2\pi}{\omega_0} = 2\pi \left(\frac{m}{k}\right)^{1/2}$$ **Period** of motion

$$\omega_0 = \sqrt{\frac{k}{m}}$$ **Natural frequency** of the vibration

$$R = \sqrt{A^2 + B^2}$$ **Amplitude** (constant in time)

Damped Free Vibrations

$$my''(t) + cy'(t) + ky(t) = \cancel{F(t)}$$

no external force

$$my''(t) + cy'(t) + ky(t) = 0$$

Assume an exponential solutio $y(t) = e^{rt}$

Then

$$y'(t) = re^{rt}, \quad y''(t) = r^2 e^{rt}$$

and substituting in equation above, we have

$$\boxed{mr^2 + cr + k = 0}$$

(characteristic equation)

Damped Free Vibrations

$$mr^2 + cr + k = 0$$

Solutions to characteristic equation:

$$r_1, r_2 = \frac{-c \pm \sqrt{c^2 - 4km}}{2m} = \frac{c}{2m}\left(-1 \pm \sqrt{1 - \frac{4km}{c^2}}\right)$$

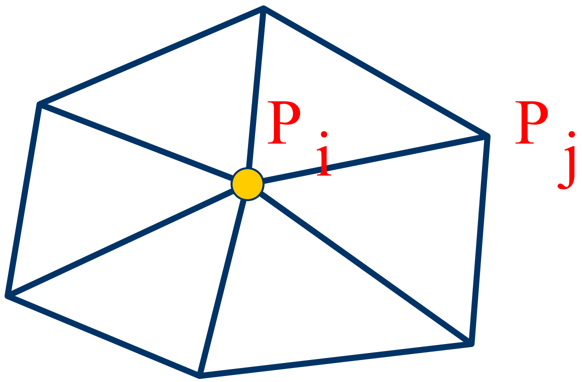
$c^2 - 4km > 0, \quad y = Ae^{r_1 t} + Be^{r_2 t}$ overdamped

$c^2 - 4km = 0, \quad y = (A + Bt)e^{-ct/2m}$ critically damped

$c^2 - 4km < 0, \quad y = e^{-ct/2m}(A\cos\mu t + B\sin\mu t)$ underdamped

3D Spring-Mass

- Point Masses linked with springs



$$W = \sum_{j \in N(\mathbf{P}_i)} k_{ij} \left(\left\| \mathbf{P}_i \mathbf{P}_j - l_{ij}^0 \right\| \right)^2$$

$$\mathbf{F}_i = -\frac{\partial W}{\partial \mathbf{P}_i} = \sum_{j \in N(\mathbf{P}_i)} k_{ij} \left(\left\| \mathbf{P}_i \mathbf{P}_j \right\| - l_{ij}^0 \right) \frac{\mathbf{P}_i \mathbf{P}_j}{\left\| \mathbf{P}_i \mathbf{P}_j \right\|}$$

Spring-Mass Advantages

- Use Point Mechanics instead of Continuum Mechanics

$$m_i \frac{d^2 \boldsymbol{P}_i}{dt^2} = \gamma_i \frac{d\boldsymbol{P}_i}{dt} + \boldsymbol{F}_i$$

- Easy to implement :
  - Only needs a graph
- Invariant to rigid body motion
- Widely used in computer graphics

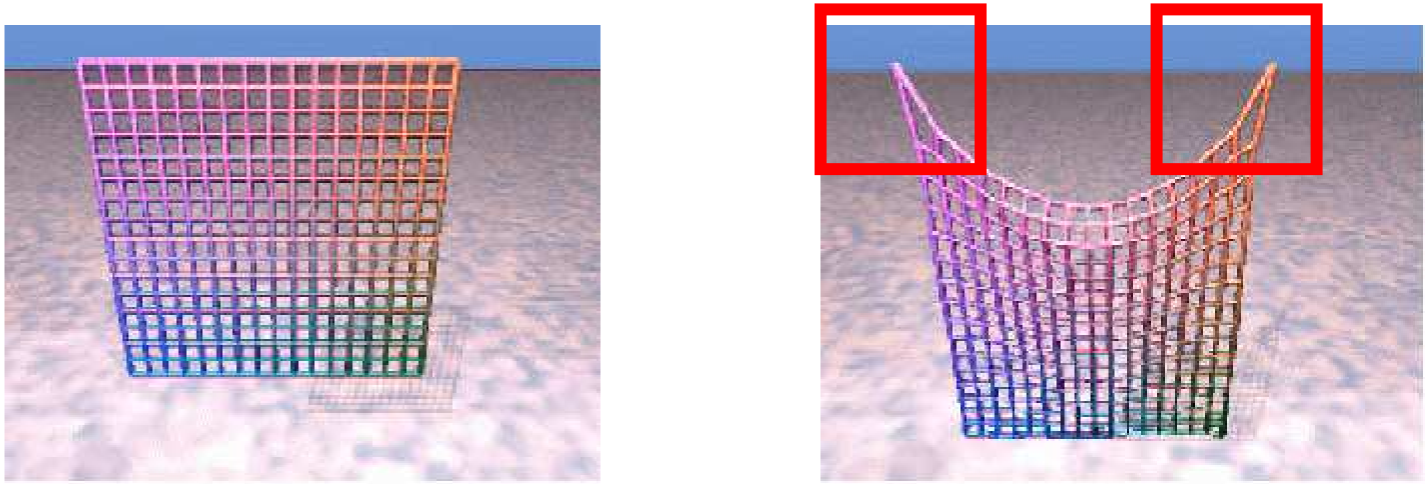
WinSpringies

Spring-Mass limitations

- Behavior depends on mesh topology
  - No formal link between topology and behavior (except for simplicial complexes)
  - Difficult to link spring stiffness to physical parameters (Young Modulus, Poisson Coefficient)
  - Use of Neural Network or genetic algorithms to identify stiffness parameters
  - Weak behavior for shear stress

Shear-Stress behavior

- Weak shear-stress
  - See X. Provot paper



(a) Initial position (b) After 200 iterations

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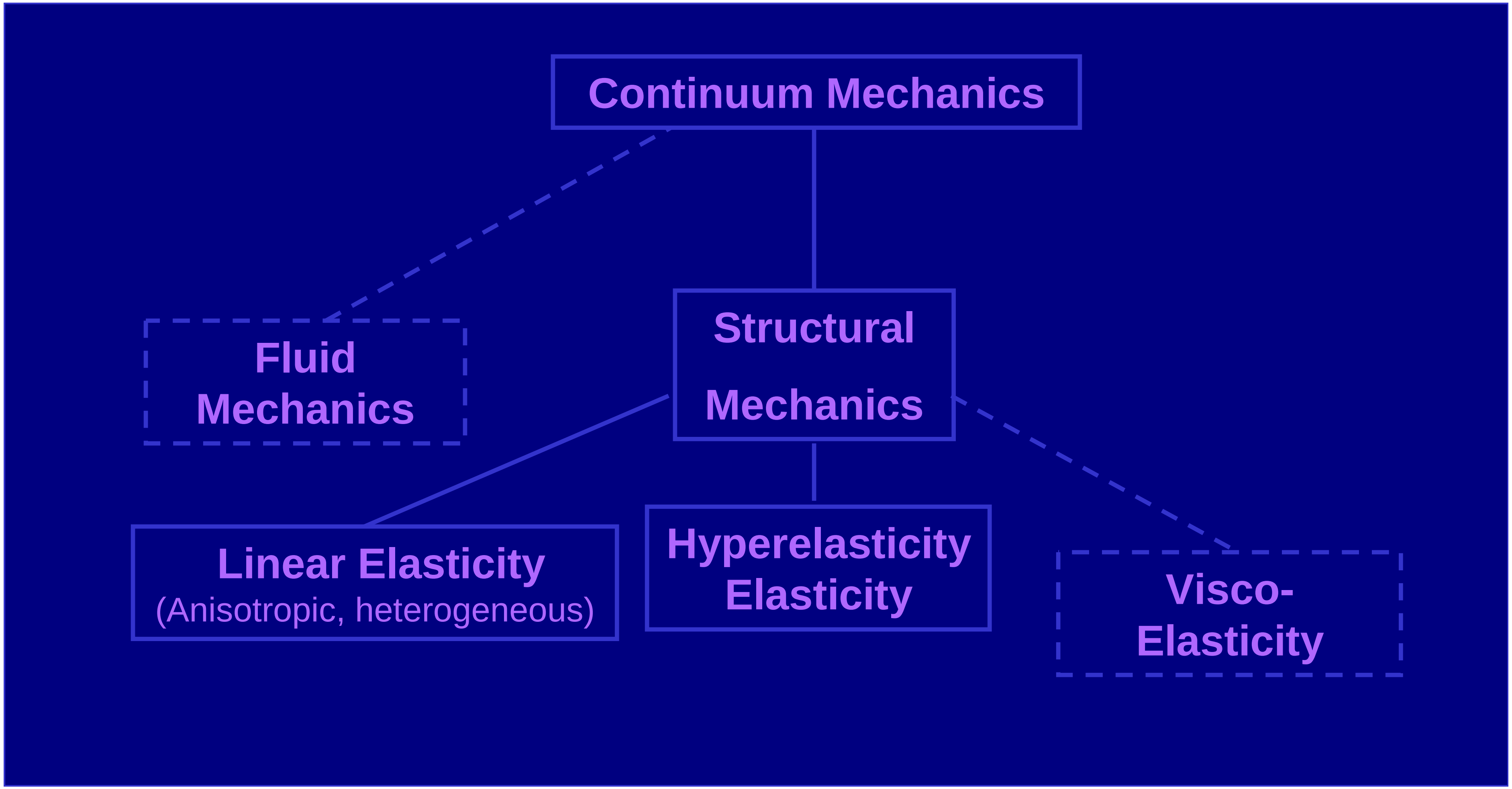
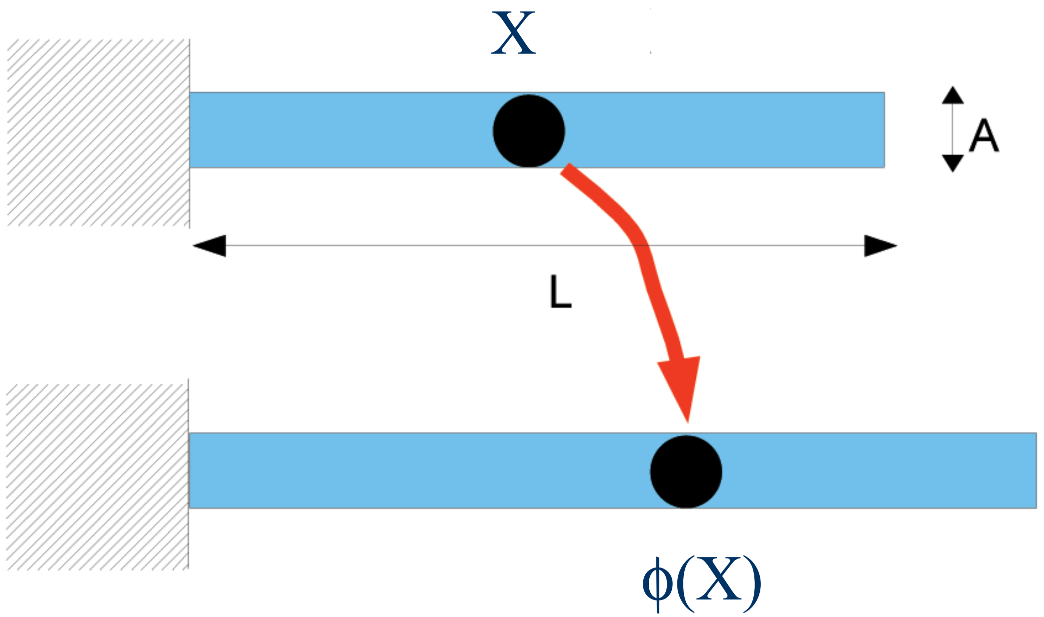
Continuum Mechanics



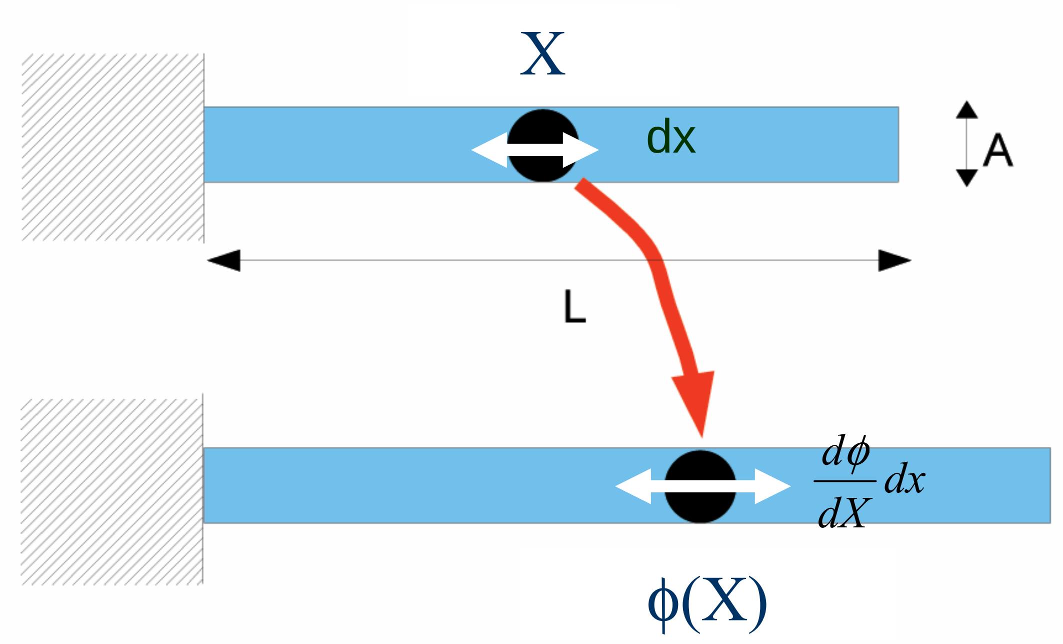
1D Elasticity



Point X is deformed into point $\phi(X)$

How much deformation around point X ?

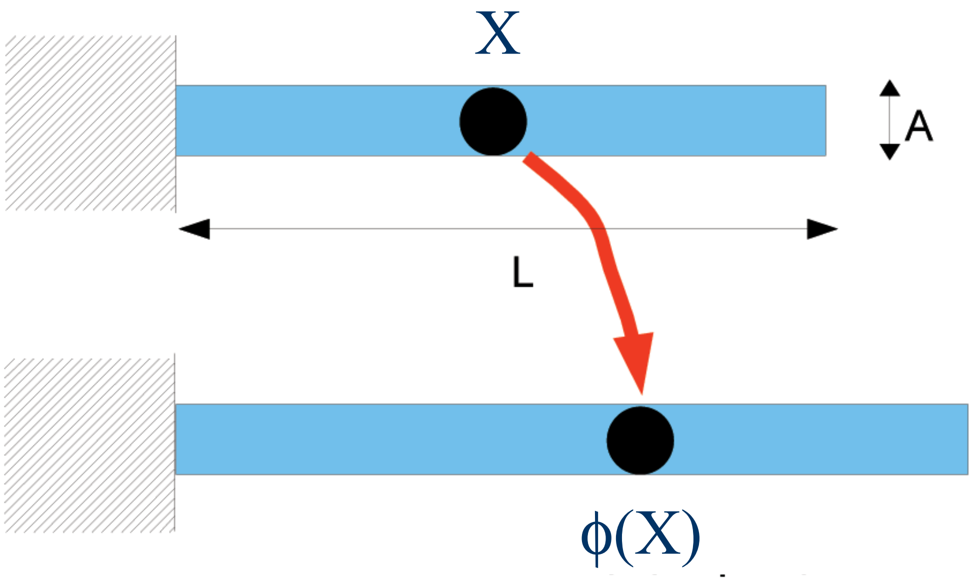
1D Elasticity : stretch ratio



Rest length : 2 dx   New length : ϕ(x+dx) – ϕ(x-dx)

Stretch ratio at X is $s(X) = \frac{d\phi}{dX}$

1D Elasticity : strain energy

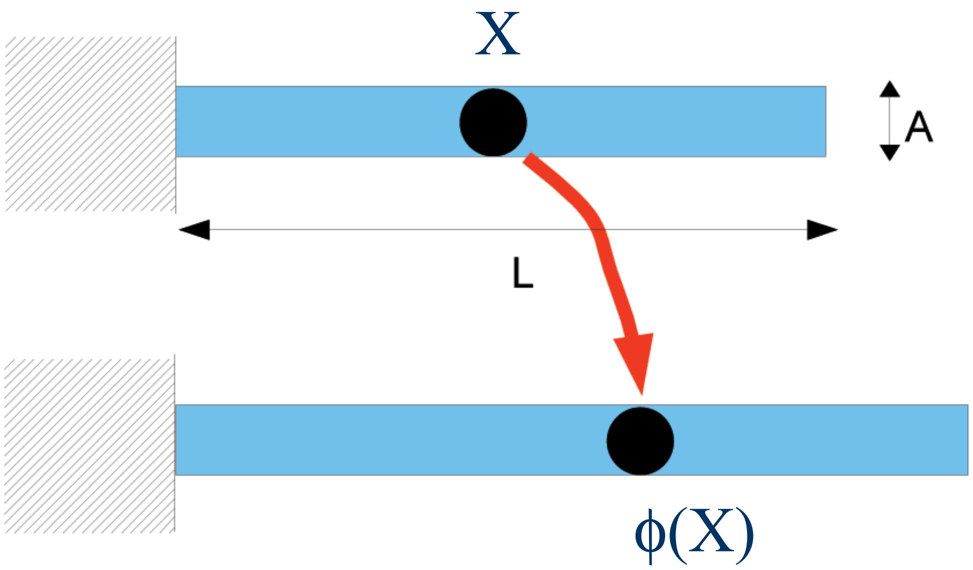


What is the energy necessary to deform the bar ?

Deformation energy W depends "how stretched" the bar is

W depends on strain ε = distance between s and 1

1D Elasticity : strain



Different choices of strain

$$\varepsilon(s) = \frac{1}{\alpha}\left(s^{\alpha} - 1\right) \quad \text{For } \alpha > 0$$

$$\varepsilon(s) = s - 1 \quad \text{For } \alpha = 1 \quad \text{Engineering strain}$$

$$\varepsilon(s) = \frac{1}{2}\left(s^{2} - 1\right) \quad \text{For } \alpha = 2 \quad \text{Green-Lagrange strain}$$

$$\varepsilon = \log s \quad \text{For } \alpha = 0 \quad \text{Henky strain}$$

1D Elasticity : stress

- Stress is the energy conjugate of strain

$$\sigma = \frac{\partial W}{\partial \varepsilon} \qquad \varepsilon = \frac{\partial W}{\partial \sigma}$$

Extensive Variable	Intensive Variable
Position	Force
Angle	Torque
Volume	Pressure
Strain	***Stress***

For $\alpha = 1$ (First Piola-Kirchhoff) nominal stress

For $\alpha = 2$ Second Piola-Kirchhoff stress

For $\alpha = 0$ Cauchy stress

St Venant Kirchhoff Material

- **Basic Material** :
  - W is a quadratic function of strain
  - Stress is proportional to strain

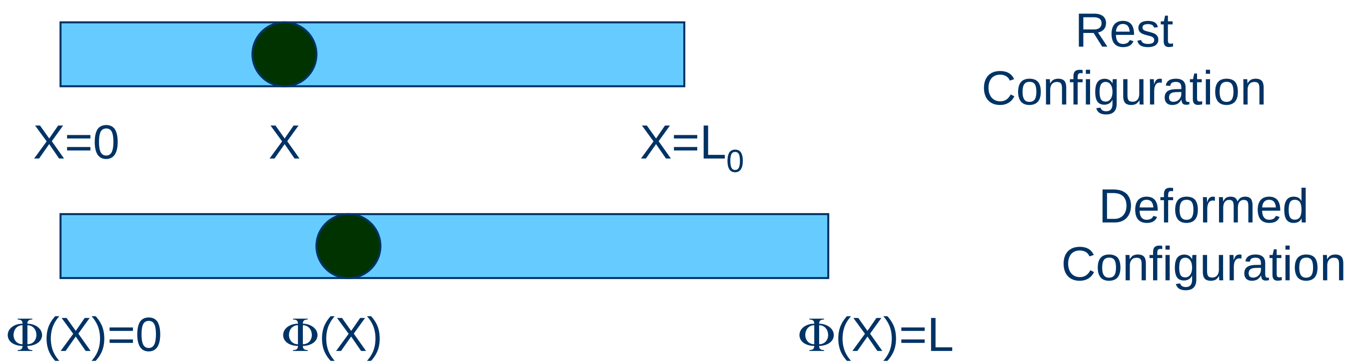
$$\sigma = \frac{\partial W}{\partial \varepsilon}$$

- **1D case** : λ is the material stiffness

$$W = \int_{\Omega} \frac{1}{2}\sigma\varepsilon = \int_{\Omega} \frac{\lambda A}{2\alpha^2}\left(\left(\frac{d\phi}{dX}\right)^{\alpha} - 1\right)^2 dX$$

1D Elasticity : discretization

- Represent the bar with a single segment



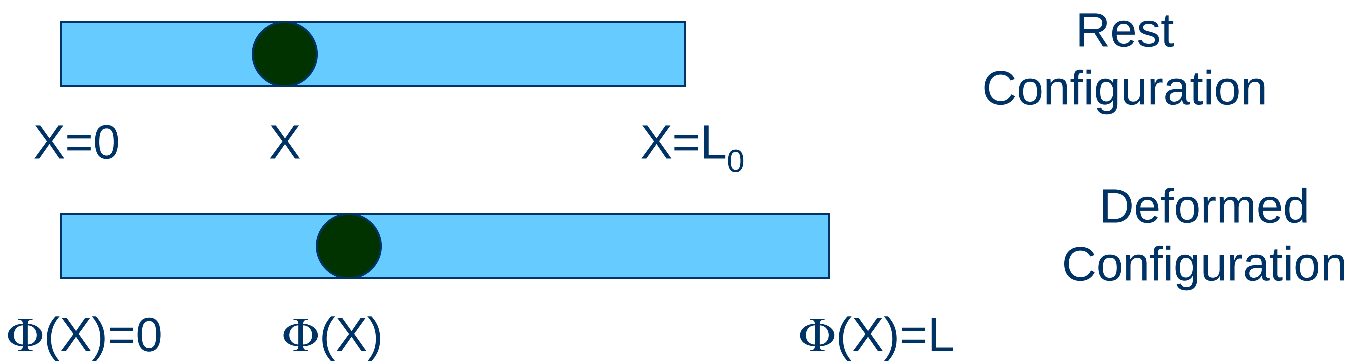
Stretch Ratio $s = \frac{L}{L_0}$

Strain $\varepsilon = \frac{1}{\alpha}\left(\frac{L^\alpha}{L_0^\alpha} - 1\right)$

Strain Energy $W = \frac{\lambda A L_0^{1-2\alpha}}{2\alpha^2}\left(L^\alpha - L_0^\alpha\right)^2$

1D Elasticity : discretization

- Represent the bar with a single segment



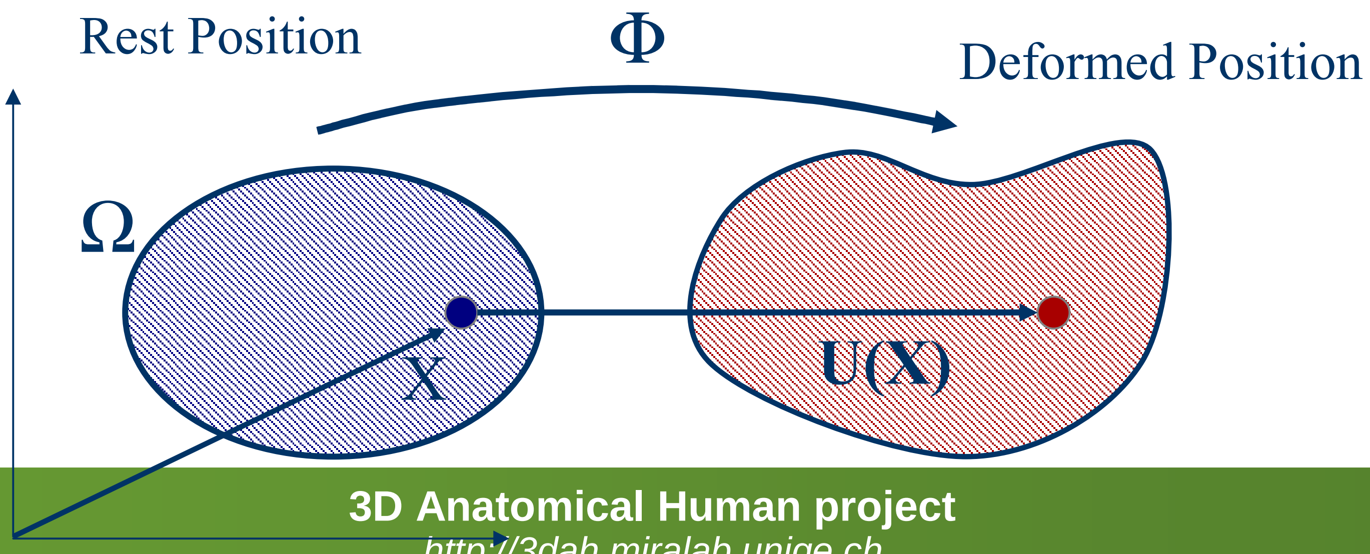
For $\alpha = 1$ $\quad W = \frac{\lambda A}{2L_0}\left(L - L_0\right)^2$ → (Quadratic) Spring Energy

For $\alpha = 2$ $\quad W = \frac{\lambda A}{8L_0^3}\left(L^2 - L_0^2\right)^2$ → Biquadratic Spring Energy

3D Elasticity

- Deformation Function
- $$X \in \Omega \mapsto \phi(X) \in \Re^3$$
- Displacement Function

$$U(X) = \phi(X) - X$$

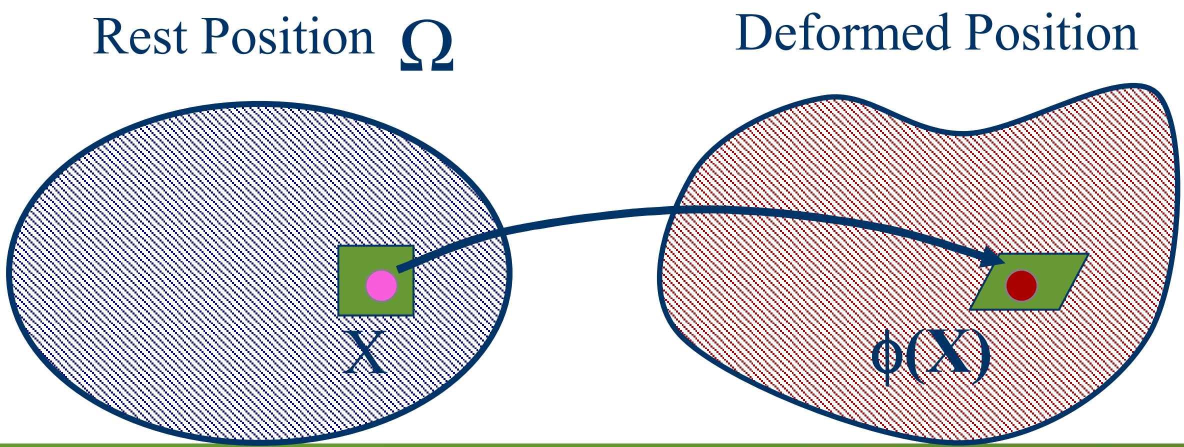


Deformation Gradient

- The local deformation is captured by the deformation gradient :

$$F = \frac{\partial \phi}{\partial X}$$

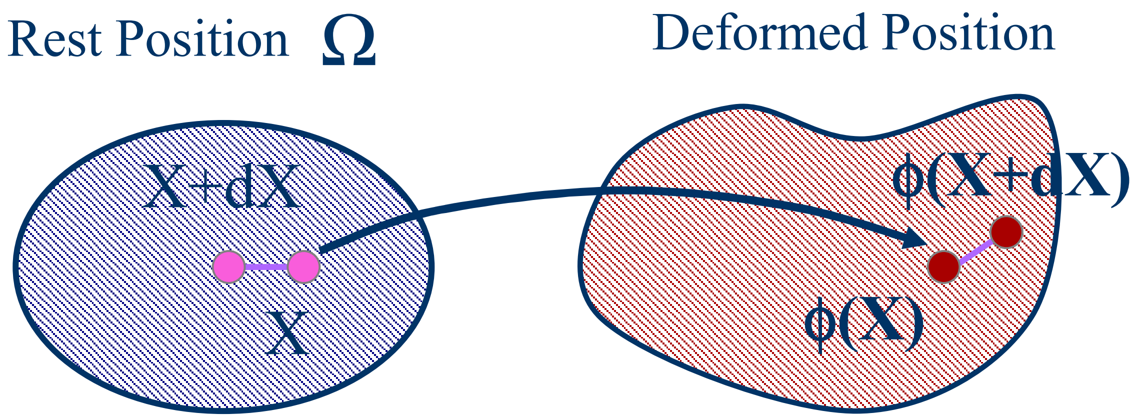
$$F_{ij} = \frac{\partial \phi_i}{\partial X_j} = \begin{bmatrix} \frac{\partial \phi_1}{\partial X_1} & \frac{\partial \phi_1}{\partial X_2} & \frac{\partial \phi_1}{\partial X_3} \\ \frac{\partial \phi_2}{\partial X_1} & \frac{\partial \phi_2}{\partial X_2} & \frac{\partial \phi_2}{\partial X_3} \\ \frac{\partial \phi_3}{\partial X_1} & \frac{\partial \phi_3}{\partial X_2} & \frac{\partial \phi_3}{\partial X_3} \end{bmatrix}$$



F(X) is the local affine transformation that maps the neighborhood of X into the neighborhood of ϕ(X)

Stretch Tensor

- Distance between point may not be preserved



- Distance between deformed points

$$(ds)^2 = \|\phi(X + dX) - \phi(X)\|^2 \approx dX^T \left(\nabla\phi^T \nabla\phi\right) dX$$

- Right Cauchy-Green Deformation tensor

$$C = \nabla\phi^T \nabla\phi$$

Measures the change of metric in the deformed body

Strain Tensor

- Example : Rigid Body motion entails no deformation $\phi(X) = RX + T$

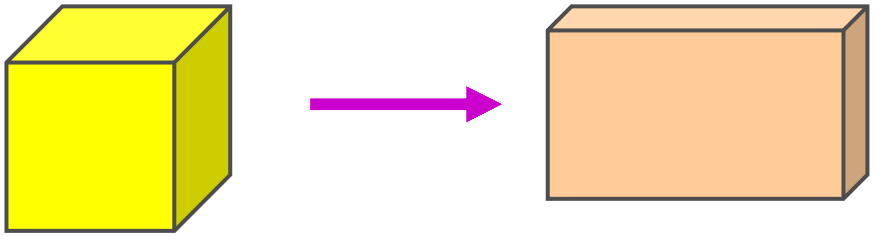
$F(X) = \nabla\phi(X) = R$ $\qquad C = R^T R = Id$

- Strain tensor captures the amount of deformation
  - It is defined as the "distance between C and the Identity matrix"

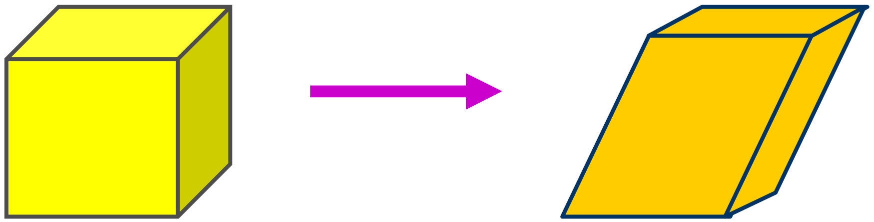
$$E = \frac{1}{2}\left(\nabla\phi^T\nabla\phi - Id\right) = \frac{1}{2}\left(C - Id\right)$$

Strain Tensor

- Diagonal Terms : ε_i
  - Capture the length variation along the 3 axis



- Off-Diagonal Terms : γ_i
  - Capture the shear effect along the 3 axis



$$E = \begin{bmatrix} \varepsilon_x & \frac{1}{2}\gamma_{xy} & \frac{1}{2}\gamma_{xz} \\ \frac{1}{2}\gamma_{xy} & \varepsilon_y & \frac{1}{2}\gamma_{yz} \\ \frac{1}{2}\gamma_{xz} & \frac{1}{2}\gamma_{yz} & \varepsilon_z \end{bmatrix}$$

Analogy 1D-3D Elasticity

1D Elasticity		3D Elasticity	
Deformation Gradient	$\frac{d\phi}{dX}$	Deformation Gradient	$\nabla\phi(X)$
Square Stretch Ratio	$s^2 = \left(\frac{d\phi}{dX}\right)^2$	RCG-Deformation Tensor	$C = \nabla\phi^T \nabla\phi$
Green Strain	$\varepsilon(s) = \frac{1}{2}\left(s^2 - 1\right)$	Green Strain Tensor	$E = \frac{1}{2}\left(\nabla\phi^T \nabla\phi - Id\right)$
SVK Strain Energy	$w(X) = \frac{\lambda A(\varepsilon(s))^2}{4}$	SVK Strain Energy	$w(X) = \frac{\lambda}{2}(tr\, E)^2 + \mu\, tr\, E^2$

Linearized Strain Tensor

- Use displacement rather than deformation

$$\nabla \phi(X) = Id + \nabla U(X)$$

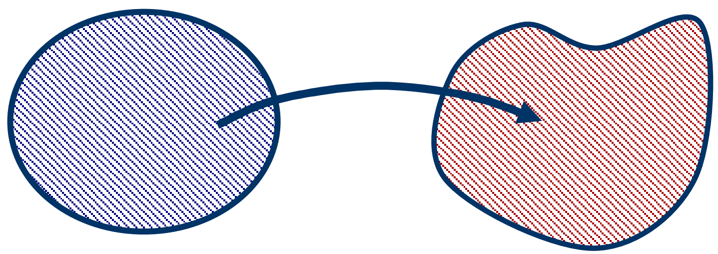
$$E = \frac{1}{2}\left(\nabla U + \nabla U^T + \nabla U^T \nabla U\right)$$

- Assume small displacements

$$E_{Lin} = \frac{1}{2}\left(\nabla U + \nabla U^T\right)$$

Hyperelastic Energy

- The energy required to deform a body is a function of the invariants of strain tensor E :
  - Trace E = I_1
  - Trace E*E= I_2
  - Determinant of E = I_3



$$W(\phi) = \int_{\Omega} w(I_1, I_2, I_3)\, dX$$

Total Elastic Energy

Linear Elasticity

- Isotropic Energy

$$w(X) = \frac{\lambda}{2}(tr\,E_{Lin})^2 + \mu\, tr\, E_{Lin}^2$$

Hooke's Law

(λ, μ) **: Lamé coefficients**

$w(X)$ **: density of elastic energy**

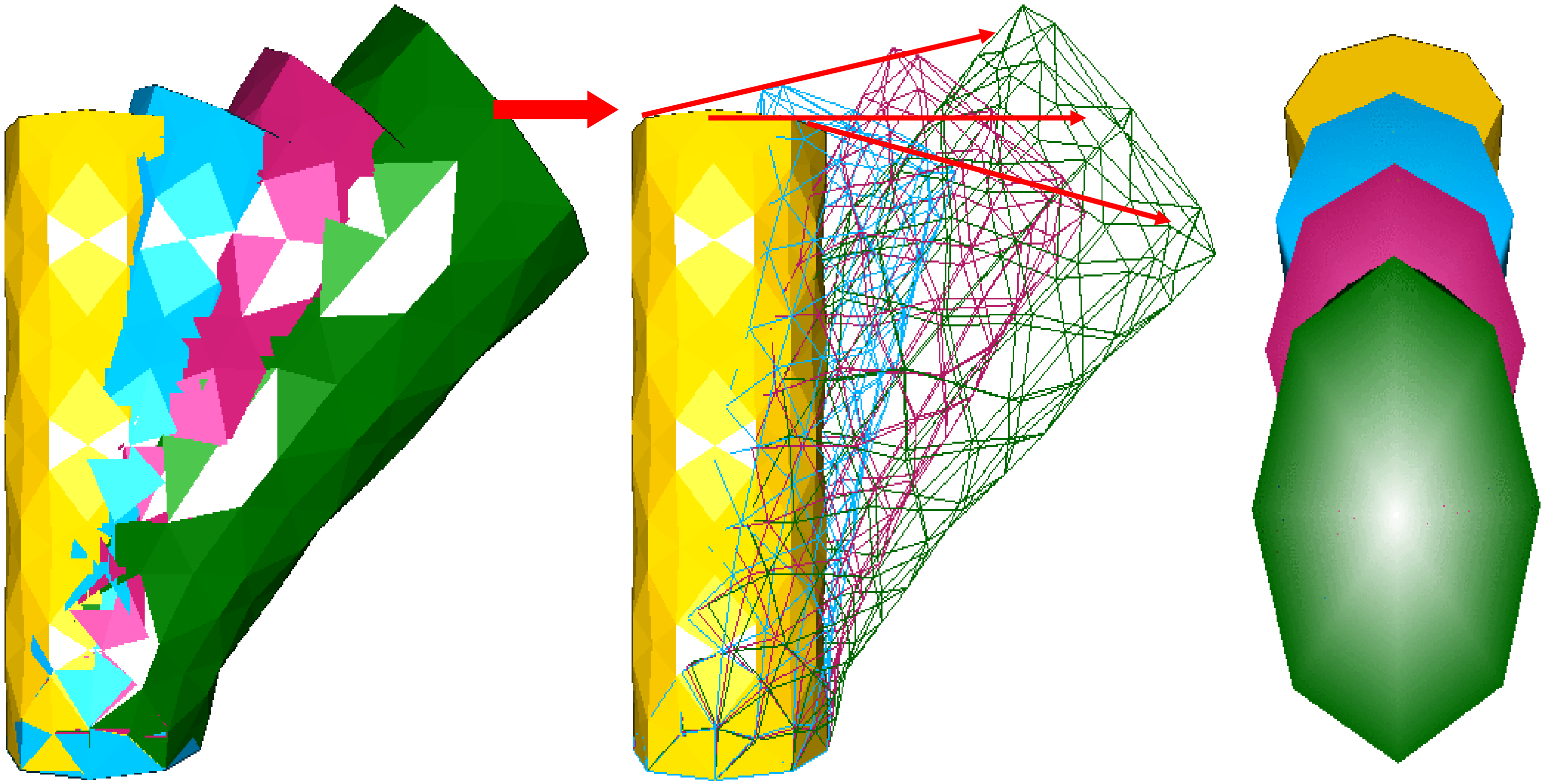
- Advantage :
  - Quadratic function of displacement

$$w = \frac{\lambda}{2}(div\,U)^2 + \mu\|\nabla U\|^2 - \frac{\mu}{2}\|rot\,U\|^2$$

- Drawback :
  - Not invariant with respect to global rotation
- Extension for anisotropic materials

Shortcomings of linear elasticity

- Non valid for «large rotations and displacements»



St-Venant Kirchoff Elasticity

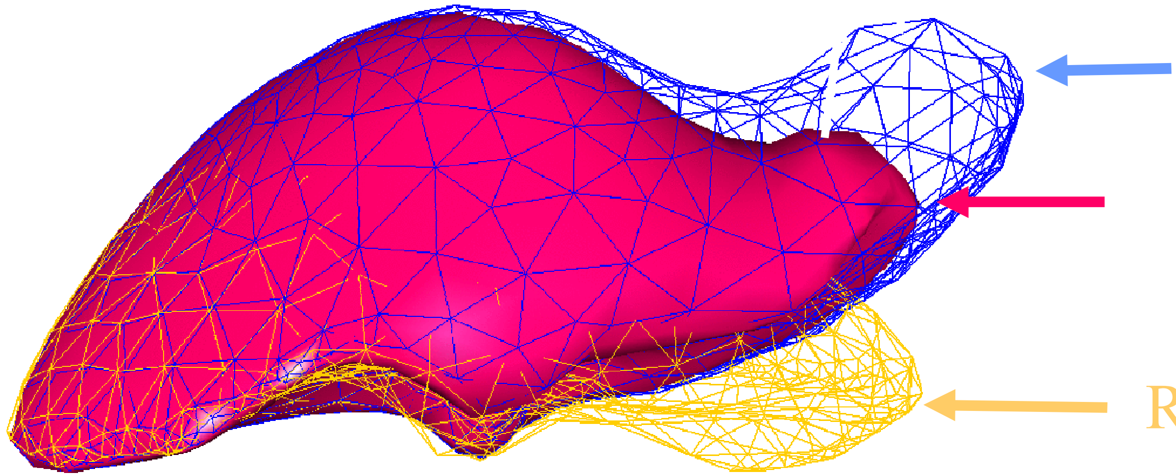
- Isotropic Energy

$$w(X) = \frac{\lambda}{2}(tr\,E)^2 + \mu\, tr\,E^2$$

(λ, μ) **: Lamé coefficients**

- Advantage :
  - Generalize linear elasticity
  - Invariant to global rotations
- Drawback :
  - Poor behavior in compression
  - Quartic function of displacement
- Extension for anisotropic materials

St Venant Kirchoff vs Linear Elasticity



Other Hyperelastic Material

- Neo-Hookean Model $$w(X) = \frac{\mu}{2} trE + f(I_3)$$
- Fung Isotropic Model $$w(X) = \frac{\mu}{2} e^{trE} + f(I_3)$$
- Fung Anisotropic Model $$w(X) = \frac{\mu}{2} e^{trE} + \frac{k_1}{k_2}\left(e^{k_2(I_4 - 1)} - 1\right) + f(I_3)$$
- Veronda-Westman $$w(X) = c_1\left(e^{\gamma trE}\right) + c_2\, trE^2 + f(I_3)$$
- Mooney-Rivlin : $$w(X) = c_{10} trE + c_{01} trE^2 + f(I_3)$$

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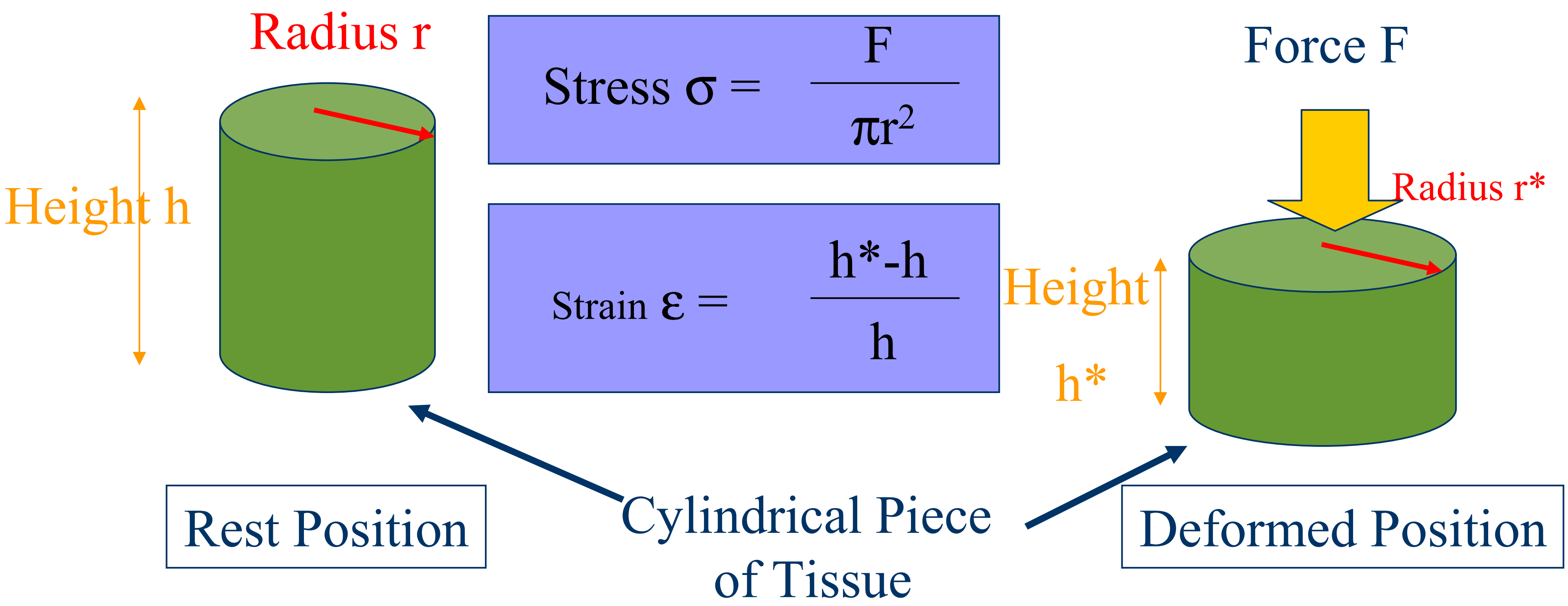
Soft Tissue Characterization

- Biomechanical behavior of biological tissue is very complex
- Most biological tissue is composed of several components :
  - Fluids : water or blood
  - Fibrous materials : muscle fiber, neuronal fibers, …
  - Membranes : interstitial tissue, Glisson capsule
  - Parenchyma : liver or brain

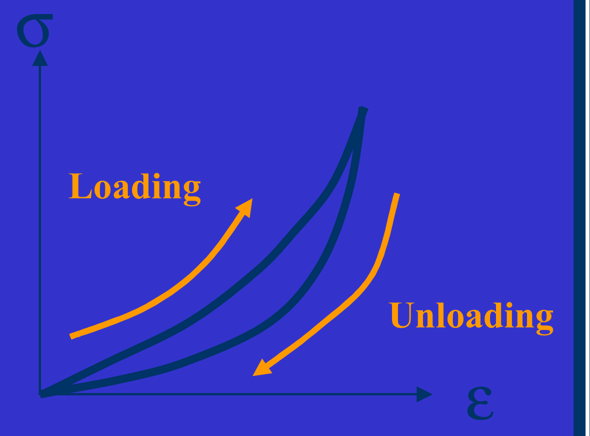
Soft Tissue Characterization

- To characterize a tissue, its stress-strain relationship is studied

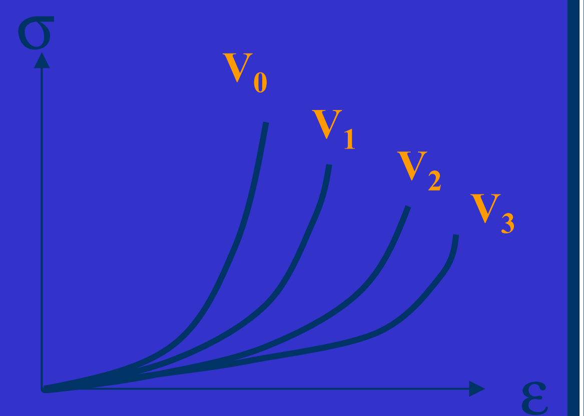


Soft Tissue Characterization

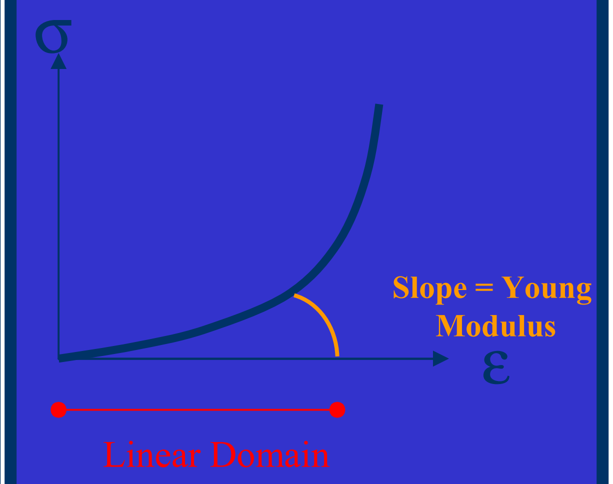
- In stress-strain relationships there are :



Hysteresis phenomenon

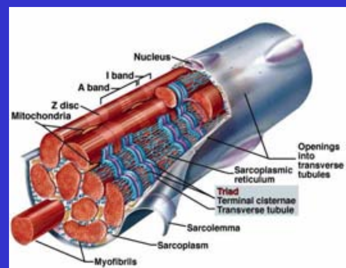


Visco-elasticity phenomemon



Non-linearity

Anisotropy



Parameter estimation

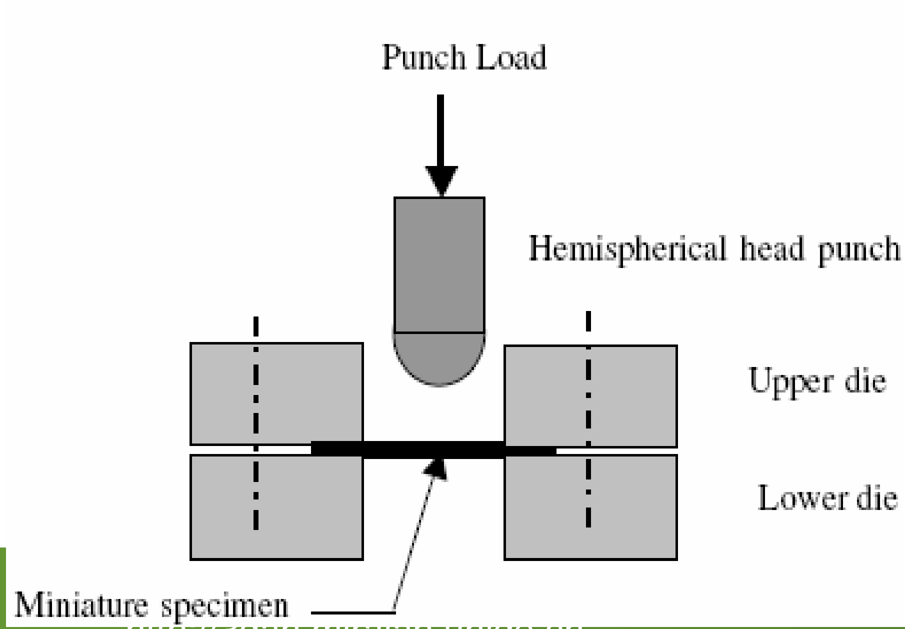
- Complex for biological tissue :
  - Heterogeneous and anisotropic materials
  - Tissue behavior changes between in-vivo and in-vitro
  - Ethics clearance for performing experimental studies
  - Effect of preconditioning
  - Potential large variability across population

Soft Tissue Characterization

- Different possible methods
  - In vitro rheology
  - In vivo rheology
  - Elastometry
  - Solving Inverse problems

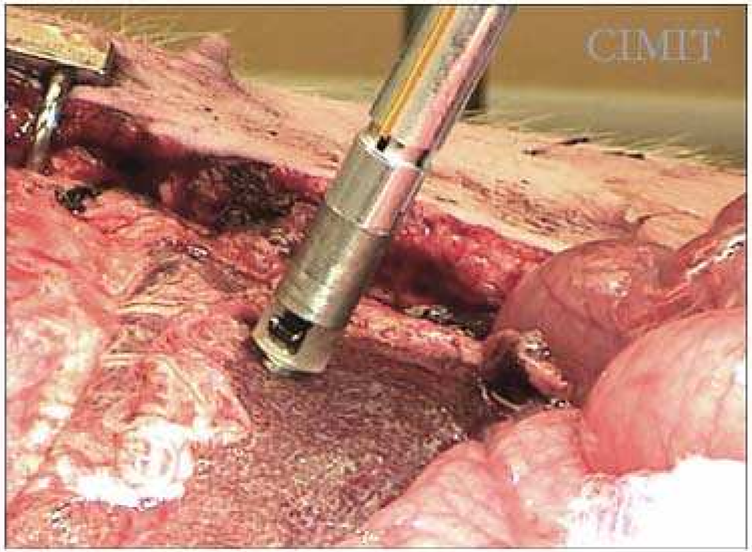
Soft Tissue Characterization

- In vitro rheology
  - can be performed in a laboratory. Technique is mature
  - Not realistic for soft tissue (perfusion, …)



Soft Tissue Characterization

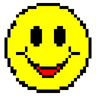
- In vivo rheology
  - can provide stress/strain relationships at several locations
  - Influence of boundary conditions not well understood



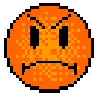
Source: Cimit, Boston USA

Soft Tissue Characterization

- Elastometry (MR, Ultrasound)



  - mesure property inside any organ non invasively

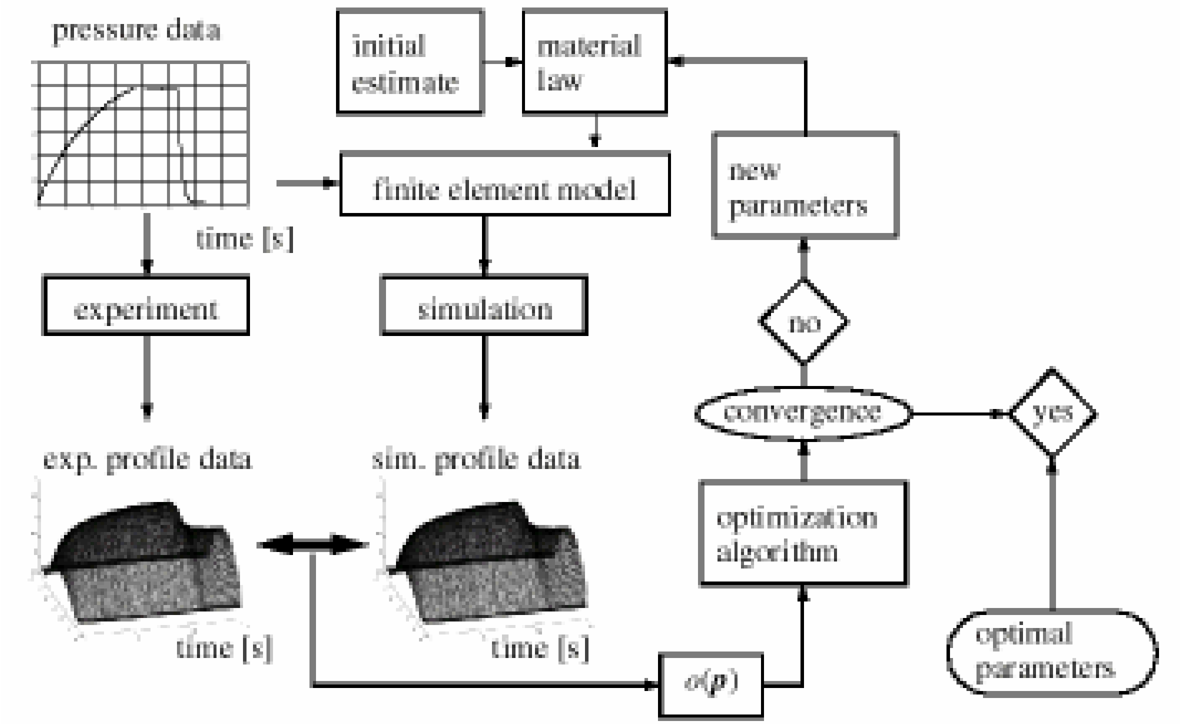


  - validation ? Only for linear elastic materials

Source Echosens, Paris

Soft Tissue Characterization

- Inverse Problems
  - well-suited for surgery simulation (computational approach)
  - require the geometry before and after deformation



Soft Tissue Characterization

- Still difficult to find “reliable” soft tissue material parameters
- Example : Liver soft tissue characterization

First Author	Experimental Technique	Liver Origin	Young Modulus (kPa)
Yamashita [111]	Image-Based	Human	Not Available
Brown [15]	*in-vivo*	Porcine Liver	≈ 80
Carter [17]	*in-vivo*	Human Liver	≈ 170
Dan [27]	*ex-vivo*	Porcine Liver	≈ 10
Liu [62, 61]	*ex-vivo*	Bovine Liver	Not Available
Nava [76]	*in-vivo*	Porcine Liver	≈ 90
Miller [74]	*in-vivo*	Porcine Liver	Not Available
Sakuma [92]	*ex-vivo*	Bovine Liver	Not Available

Table 2: List of published articles providing some quantitative data about the biomechanical properties of the liver.

Overview

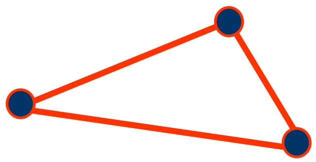
- Introduction
- Constitutive Modeling of Soft Tissue
  - Discrete Models (Particles + Spring Mass Models)
  - Continuum Models (Hyperelasticity)
- Measuring Soft Tissue Deformation
- Discretization Methods
- Example of Soft Tissue Models

Discretisation techniques

- Four main approaches :
  - Volumetric Mesh Based
  - Surface Mesh Based
  - Meshless
  - Particles

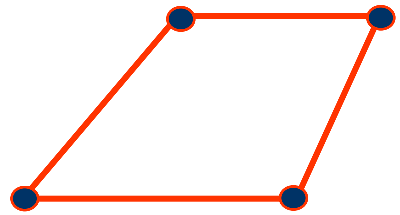
Different types of meshes

- Surface Elements :



Triangle

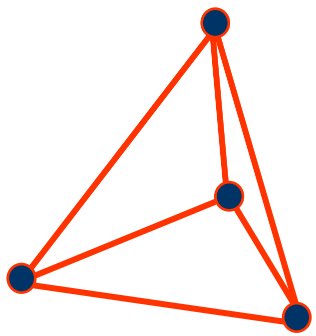
3, 12 nodes and more



Quad

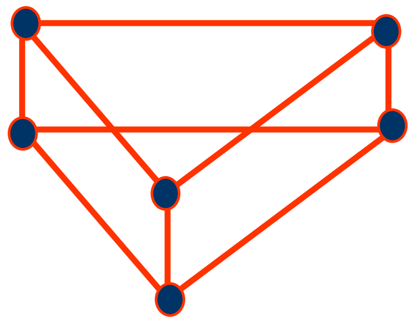
4, 8 nodes and more

- Volume Elements



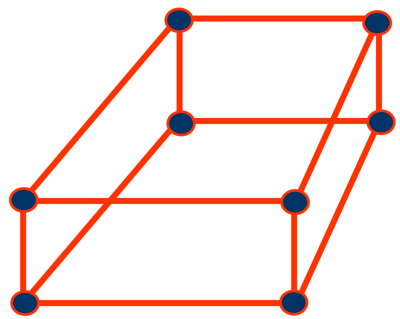
Tetrahedron

4, 10 nodes



Prismatic

6, 15 nodes and more



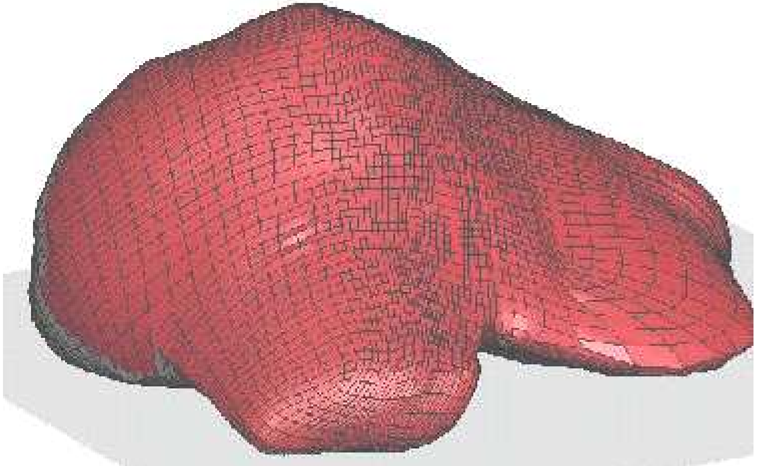
Hexahedron

8, 20 nodes and more

Structured vs Unstructured meshes

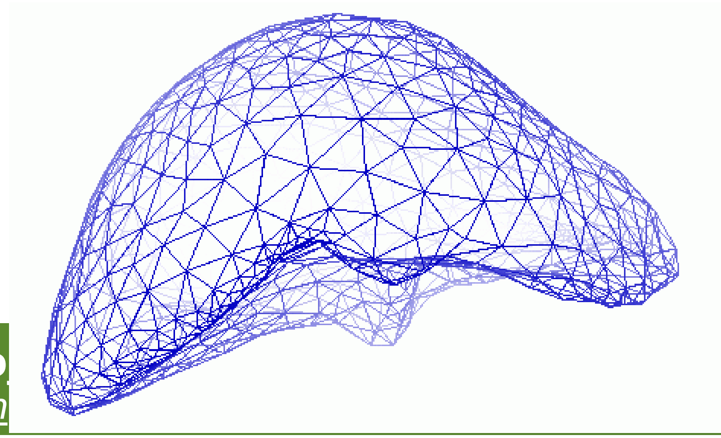
- Example 1 : Liver meshed with hexahedra

3 months work
(courtesy of ESI)



- Example 2: Liver meshed with tetrahedra

Automatically
generated (1s)

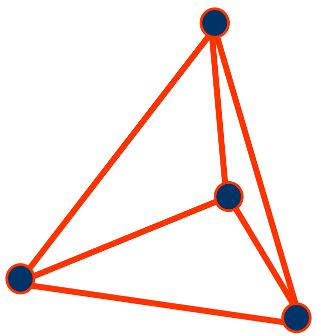


Volumetric Mesh Discretization

- Classical Approaches :
  - Finite Element Method (weak form)
  - Rayleigh Ritz Method (variational form)
  - Finite Volume Method (conservation eq.)
  - Finite Differences Method (strong form)
- FEM, RRM, FVM are equivalent when using linear elements

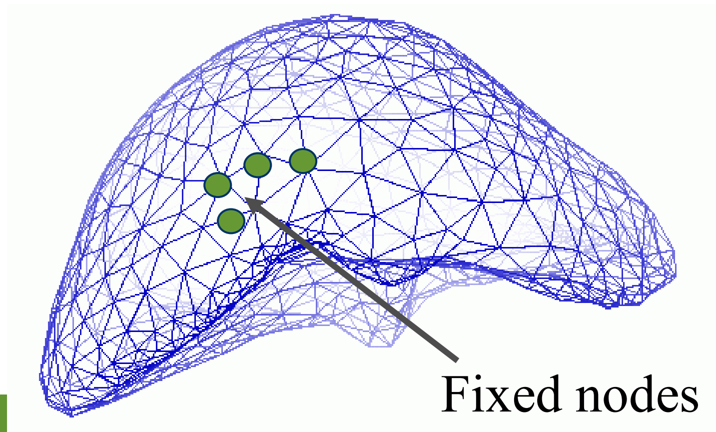
Rayleigh-Ritz Method

- Step1 : Choose
  - Finite Element (e.g. linear tetrahedron)
  - Mesh discretizing the domain of computation
  - Hyperelastic Material with its parameters
  - Boundary Conditions



Tetrahedron

4 nodes



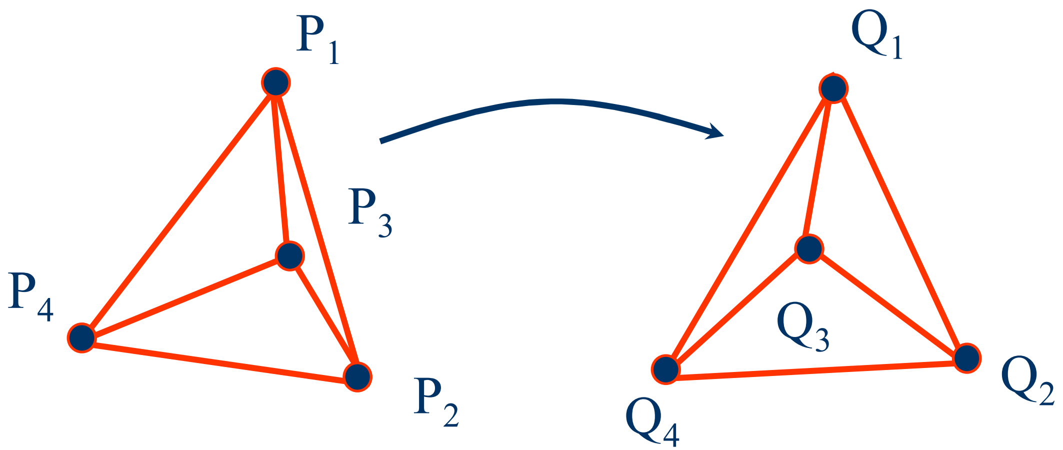
$$w(X) = \frac{\lambda}{2}(tr\,E)^2 + \mu\,tr\,E^2$$

Young Modulus

Poisson Coefficient

Rayleigh-Ritz Method

- Step2
  - Write the elastic energy required to deform a single element



$$u(P_i) = Q_i - P_i = U_i$$

$$u(X) = \sum_{i=1}^{4} \lambda_i(X) u(P_i)$$

$$\nabla \lambda_i(X) = -\frac{M_i}{6V(T)}$$

$$trE = -\sum_i \frac{M_i \cdot U_i}{6V(T)}$$

$$W_{T_i} = \sum_{jk} U_j^t \left[K_{jk}^{T_i} \right] U_k$$

$$[\mathbf{K}_{jk}^{T_i}] = \frac{1}{36.\mathrm{V}(T_i)} \left(\lambda_i \mathbf{M}_k \mathbf{M}_j^T + \mu_i \mathbf{M}_j \mathbf{M}_k^T + \mu_i (\mathbf{M}_j \cdot \mathbf{M}_k)[\mathbf{Id}_{3x3}] \right)$$

Rayleigh-Ritz Method

- Step3
  - Sum to get the total elastic energy

$$W(U) = \int_{\Omega_h} w(I_1, I_2, I_3)\, dX = \sum_{T_i} W_{T_i} = U^T K U$$

  - Write the conservation of energy

$$\underbrace{W(U)}_{\text{Internal Energy}} = \underbrace{F^T U}_{\text{Nodal Forces}} + \underbrace{\int_{\Omega} \rho(X)(X \cdot g)\, dX}_{\text{Gravity Potential Energy}}$$

Rayleigh-Ritz Method

- Step3
  - Write first variation of the energy :

Linear Elasticity

$$KU = R$$ Static case

$$M\ddot{U} + C\dot{U} + KU = R(t)$$ Dynamic case

HyperElasticity=NonLinear Elasticity

$$K(U) = R$$ Static case

$$M\ddot{U} + C\dot{U} + K(U) = R(t)$$ Dynamic case

Surface-Based Methods

- Possible approaches :
  - Boundary Element Models (BEM)
    - Based on the Green Function of the linear elastic operator
    - Requires homogeneous material
  - Matrix Condensation
    - Full Matrix inversion
  - Iterative Precomputed Generation
    - Solve 3*Ns equations F=KU

Other Methods

- Meshless Methods
  - Use only points inside and specific shape functions
  - Can better optimize location of DOFs
  - Can cope with large deformations
  - Deformation accuracy unknown
- Particles
  - Smooth Particles Hydrodynamics that interact based on a state equation

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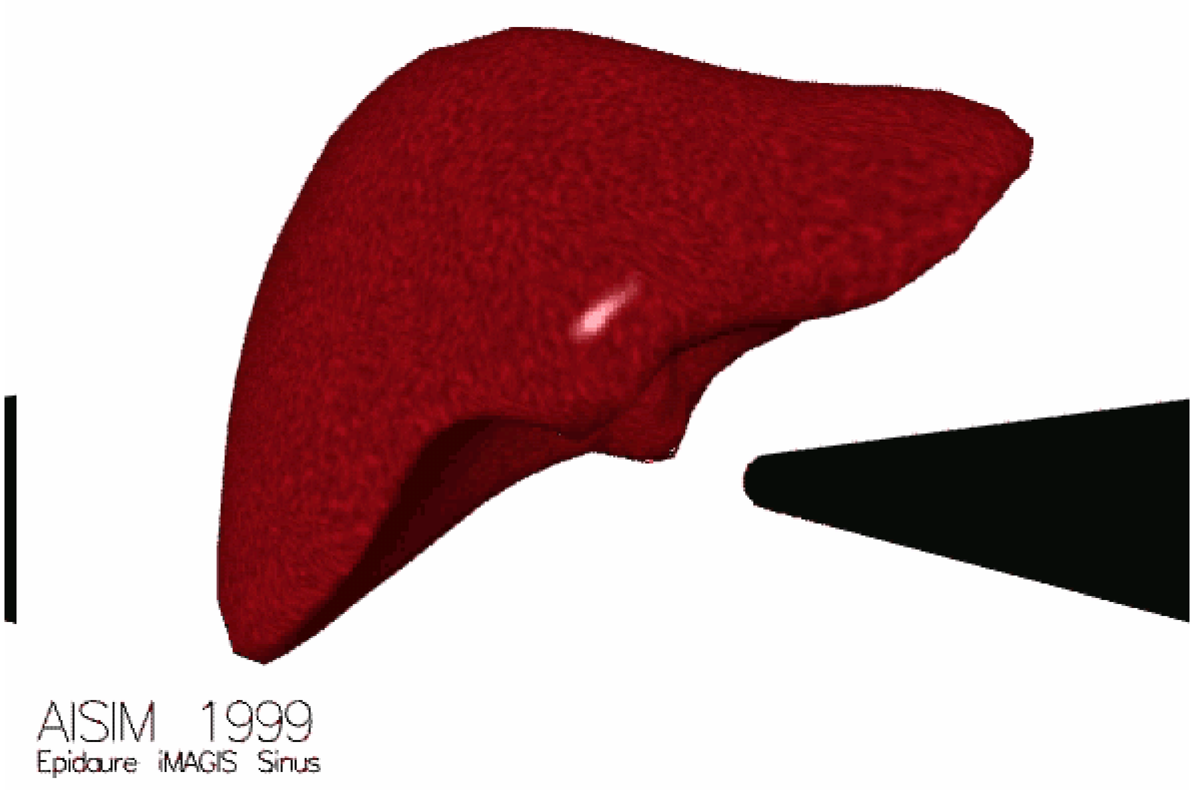
Surgery Simulation

- Surgery Simulation must cope with several difficult technical issues :
  - Soft Tissue Deformation
  - Collision Detection
  - Collision Response
  - Haptics Rendering
- Real-time Constraints :
  - 25Hz for visual rendering
  - 300-1000 Hz for haptic rendering

Precomputed linear elastic model

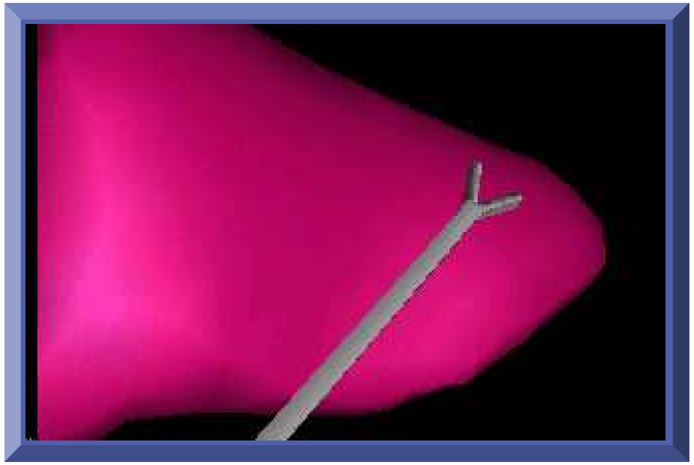


9517
Tetrahedra

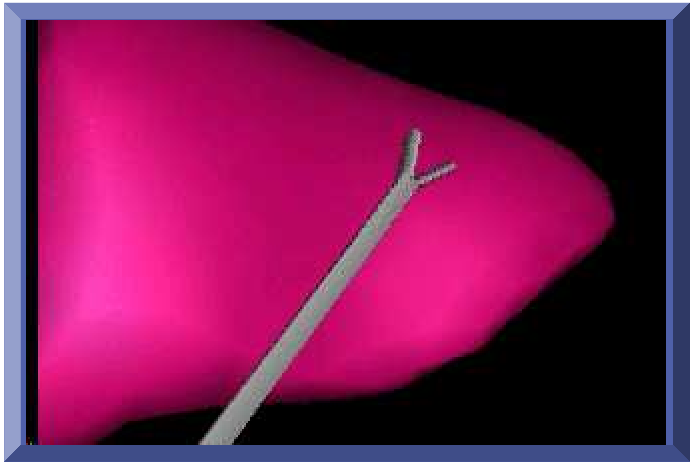
Tensor-Mass Models (low resolution)

N = 1394 (6342 Tétraèdres)

Simulation of surgical gestures

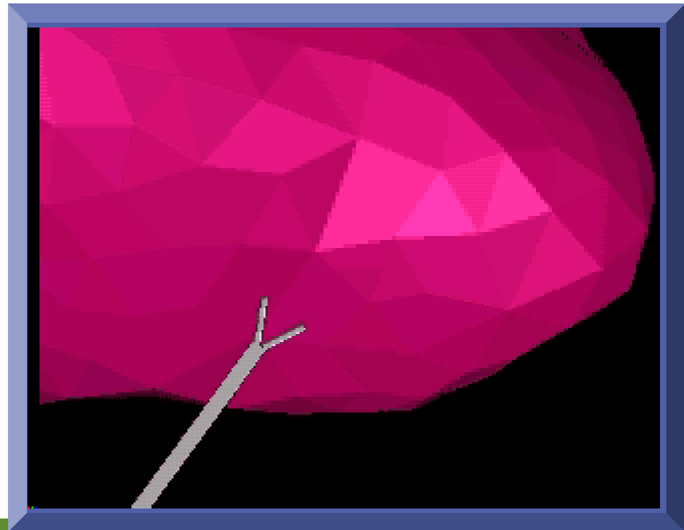


Gliding



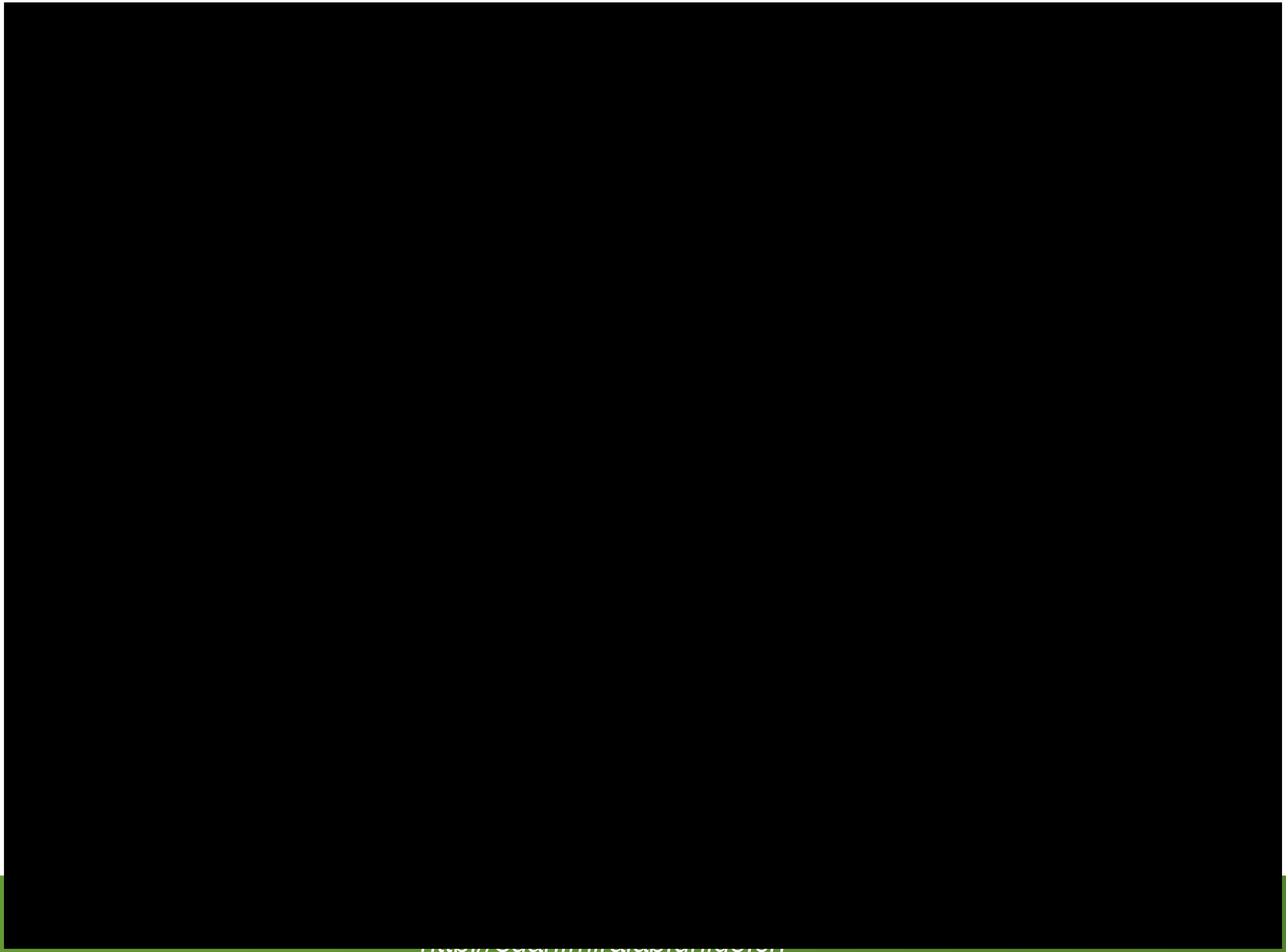
Gripping

Cutting (pliers)



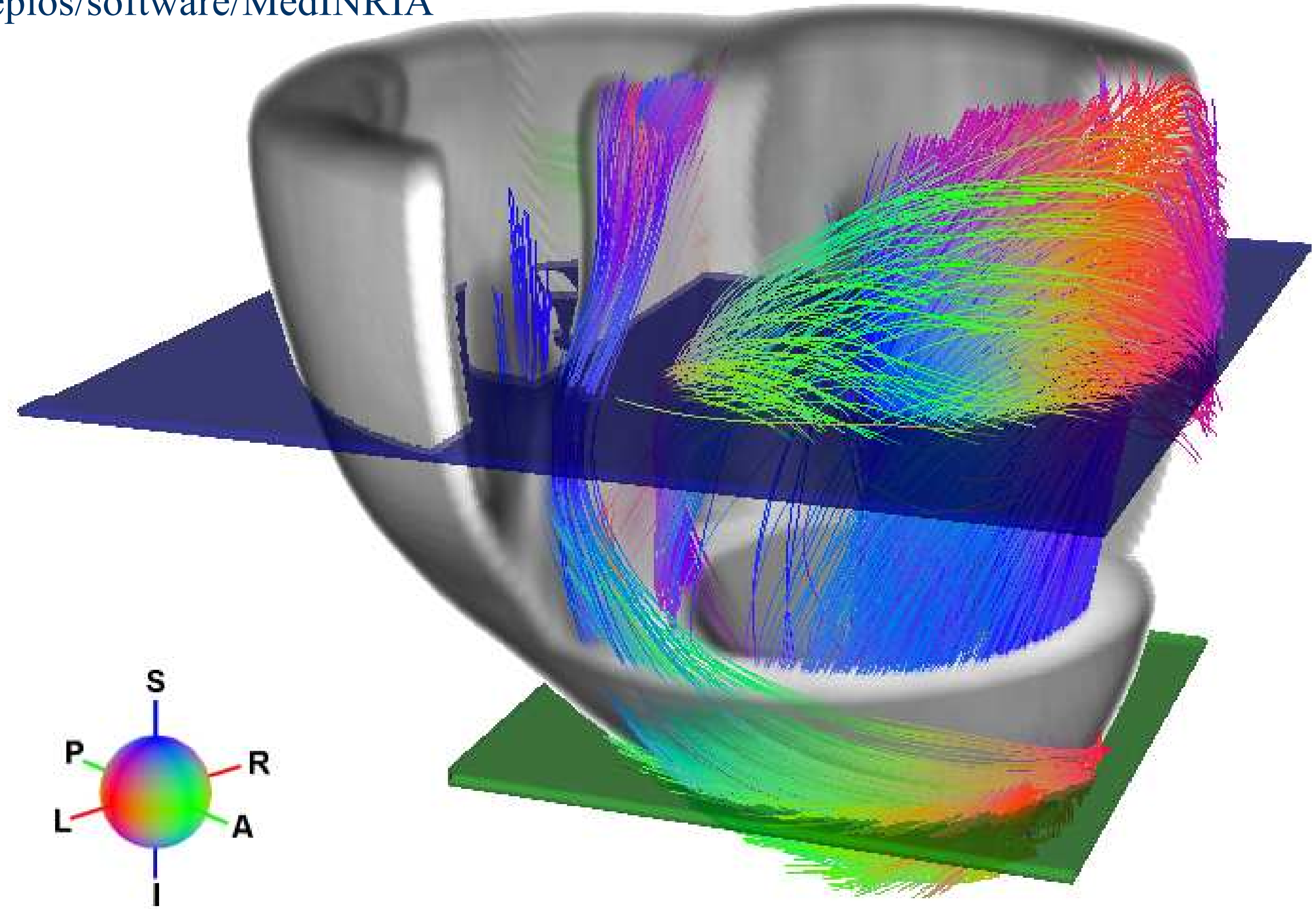
Cutting (US)

Hepatic Surgery Simulation

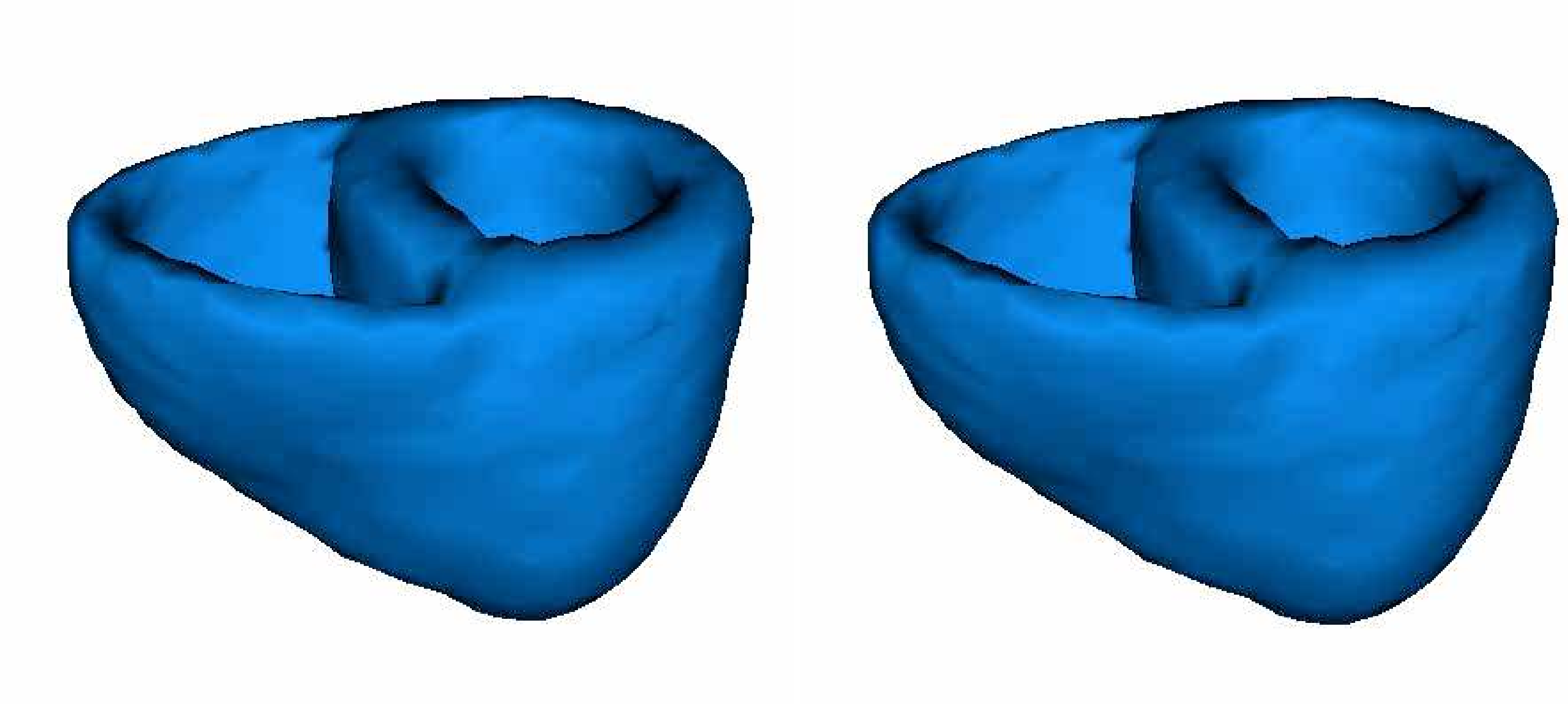


Fiber Tracking on the Average Cardiac DTI

http://www.inria.fr/asclepios/software/MedINRIA



Electrophysiology Simulation

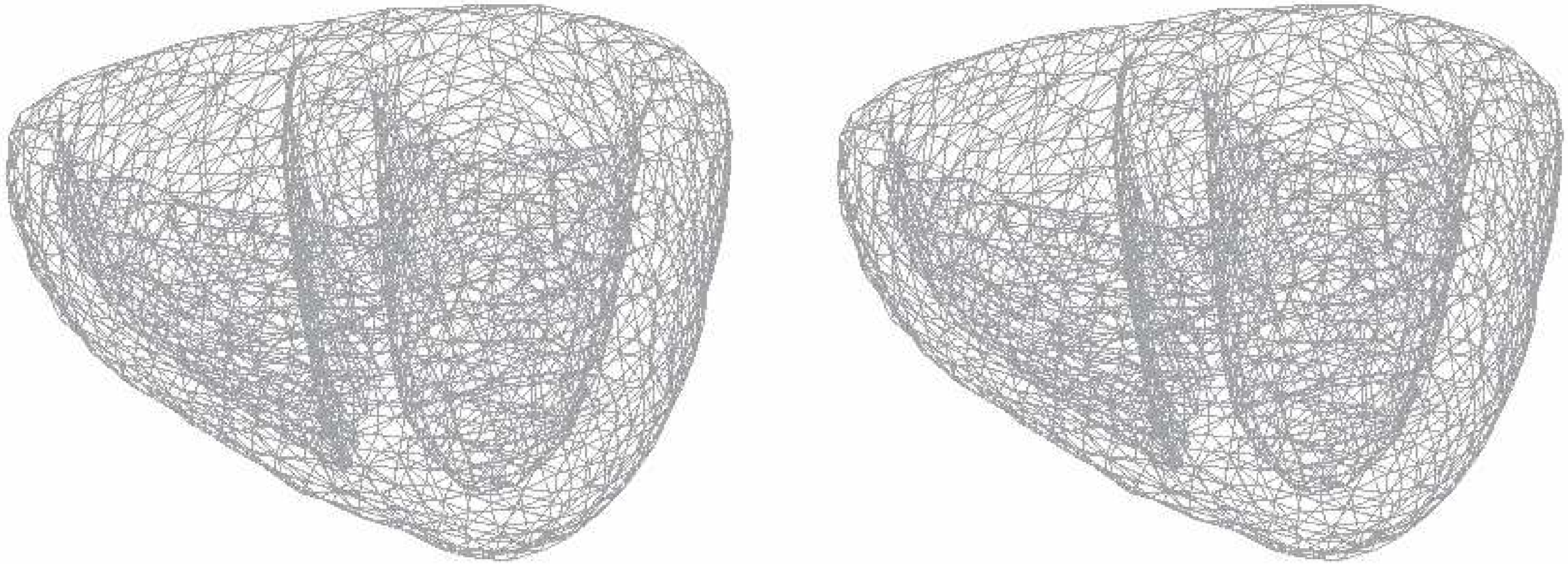


Normal heart
Ectopic Pacing

Pseudo-potential
Blue: excitable
Red: depolarised
Yellow: refractory

Infarcted Area
10 times less conductive
→Ventricular tachycardia
→Ventricular fibrillation?

Electrophysiology Simulation



Normal heart
Ectopic Pacing

Depolarisation Front
Blue: depolarised side
Red: excitable side
*Black: **Repolarisation Front***

Infarcted Area
10 times less conductive
→Ventricular tachycardia
→Ventricular fibrillation?

Cardiac Simulation

- 4 Cardiac Phases:
  - Filling
  - Isovolumetric Contraction
  - Ejection
  - Isovolumetric Relaxation
- 2 Volumetric Conditions:
  - Pressure Field in the endocardium
  - Isovolumetric Constraint of myocardium

Slowed

6 times

74

Thank you for your attention