Surgery Simulation

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Motivations of surgery simulation

- Increasing complexity of therapy and especially surgery

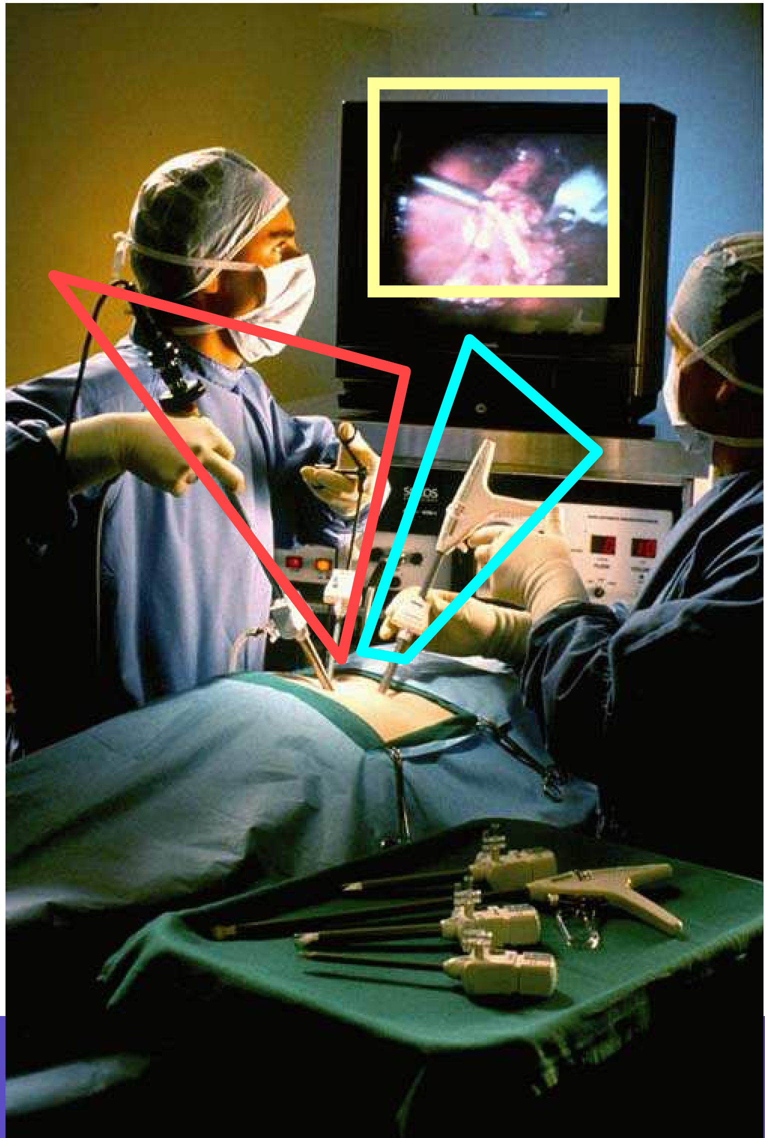
  ➡ Increasing need for training surgeons and residents

- Medical malpractice has become socially and economically unacceptable

  ➡ Increasing need for objective evaluation of surgeons (see Cordis Nitanol endovascular carotid stent)

- Natural extension of surgery planning

Need for Training



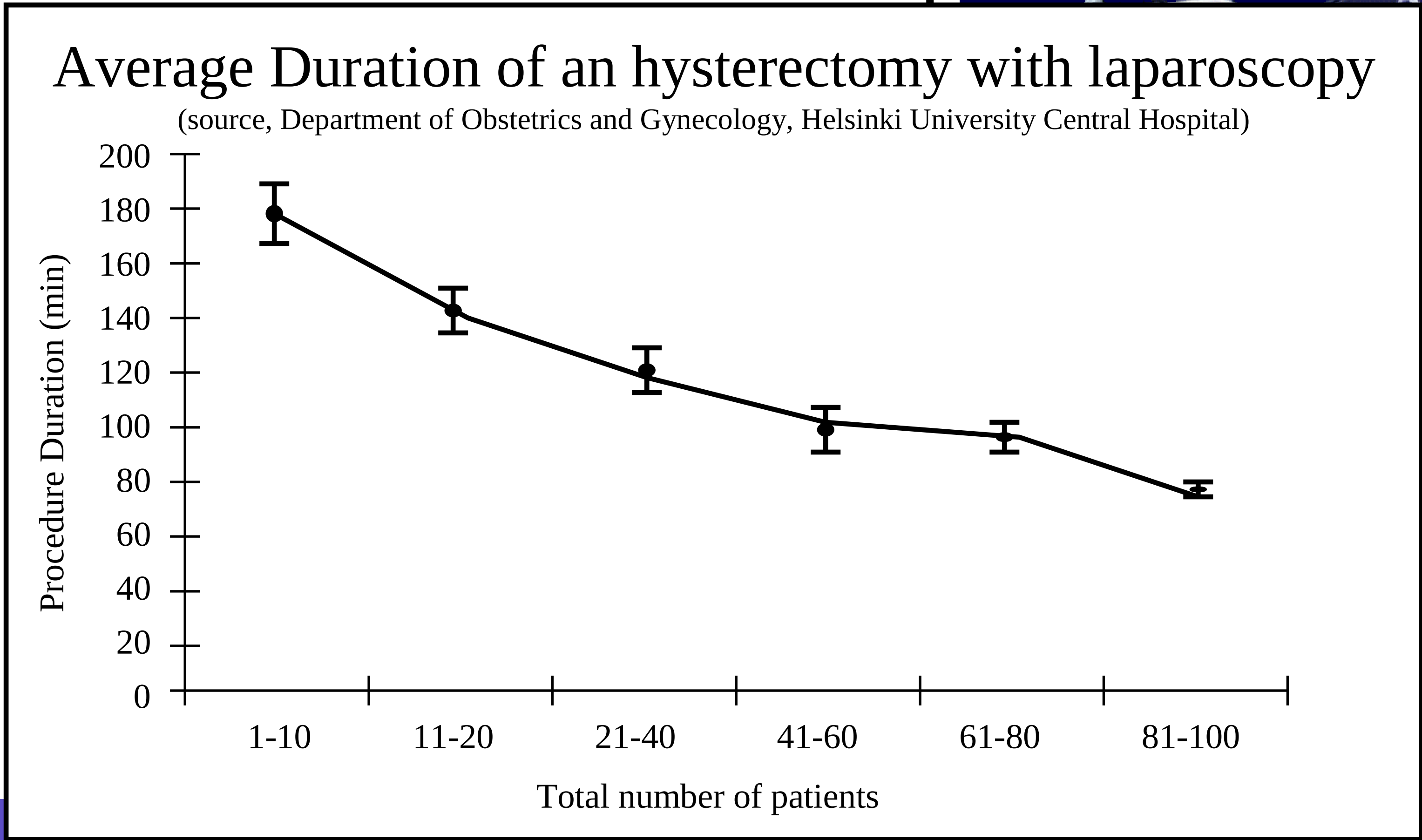
Hand-eye Synchronisation

Camera being manipulated by an assistant

Long instruments going through a fixed point in the abdomen

Current Training Techniques

- Mechanical Simulators



(source, Ayudamos)

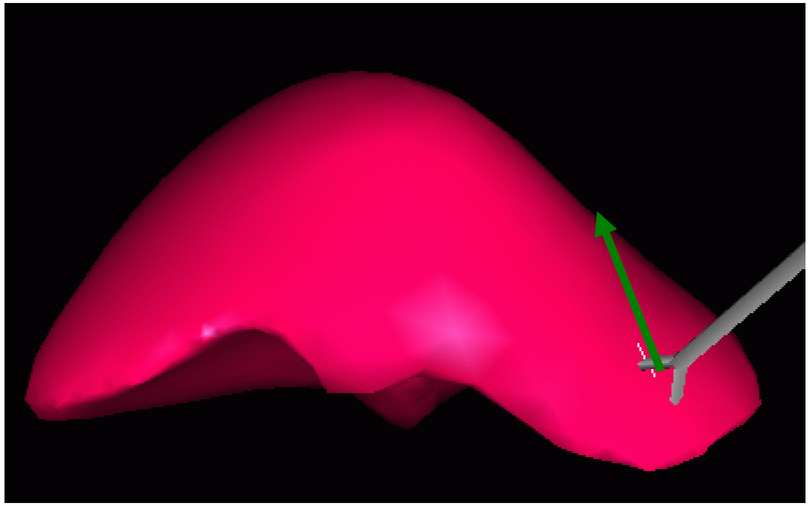
Goal for simulation : Training versus Rehearsal

- Training: Modelling a ***standard*** patient for teaching classical or rare situations

- Rehearsal: Modelling a ***specific*** patient to plan and rehearse a delicate intervention, and evaluate consequences beforehand

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

With real-time constraints

Different Technical Issues

- Mesh Reconstruction from Images
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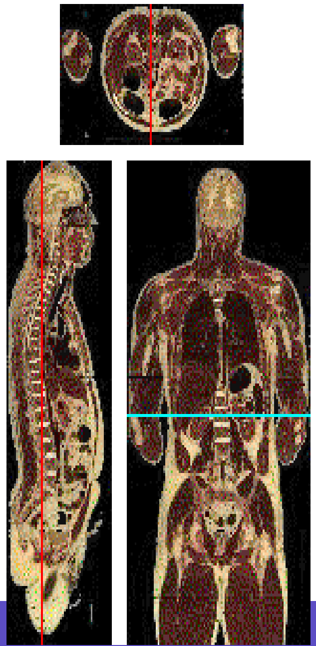
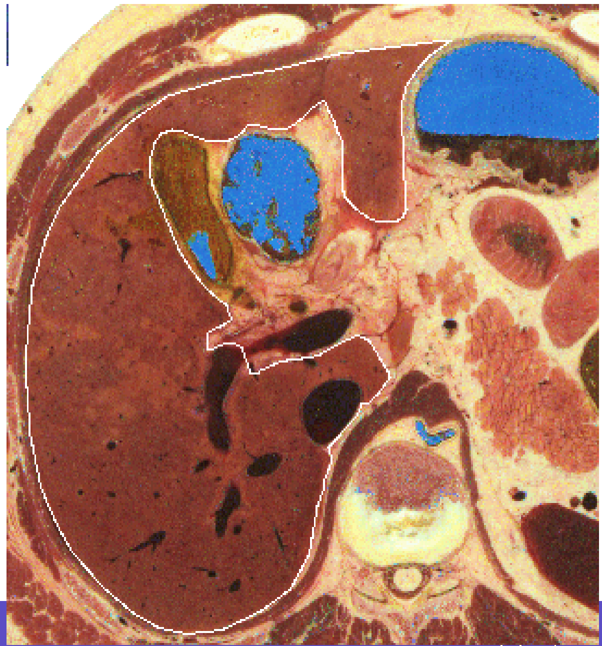
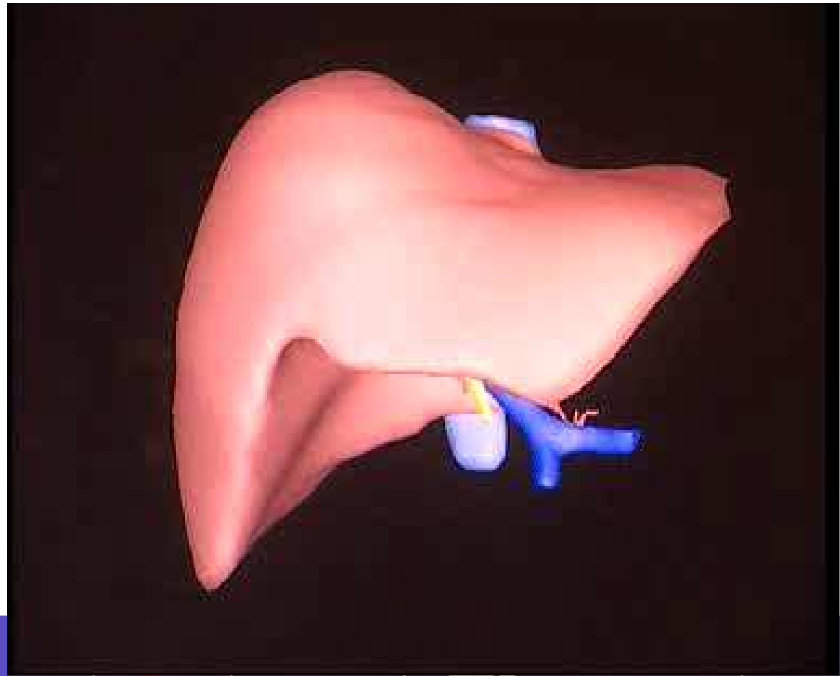
Liver Reconstruction

Deformation from a reference model reconstructed from the « *Visible Human Project* »







Different Technical Issues

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

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

Estimating material parameters

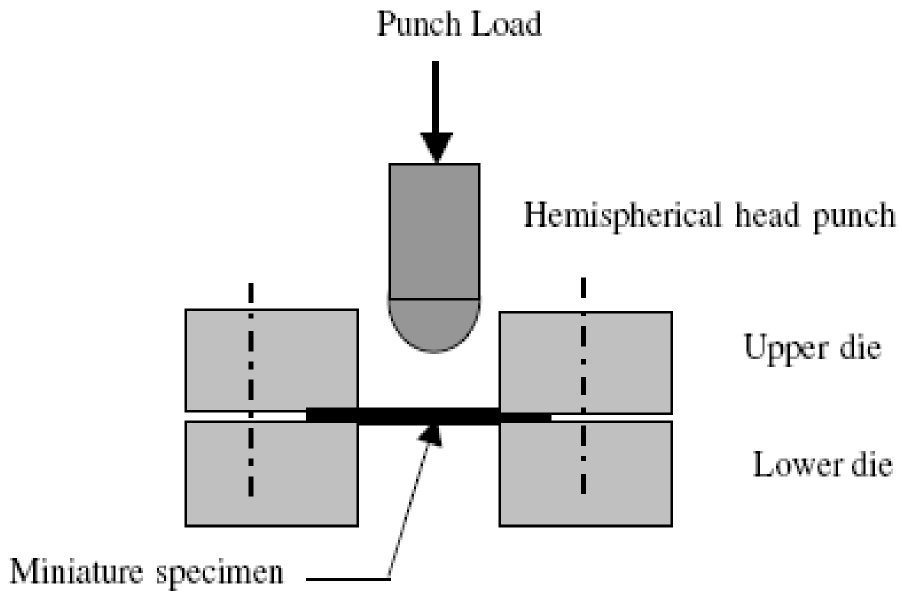
- Complex for biological tissue :
  - Heterogeneous and anisotropic materials
  - Tissue behavior changes between in-vivo and in-vitro
  - Effect of preconditioning
  - Potential large variability across population
  - Ethics clearance for performing experimental studies

Soft Tissue Characterization

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

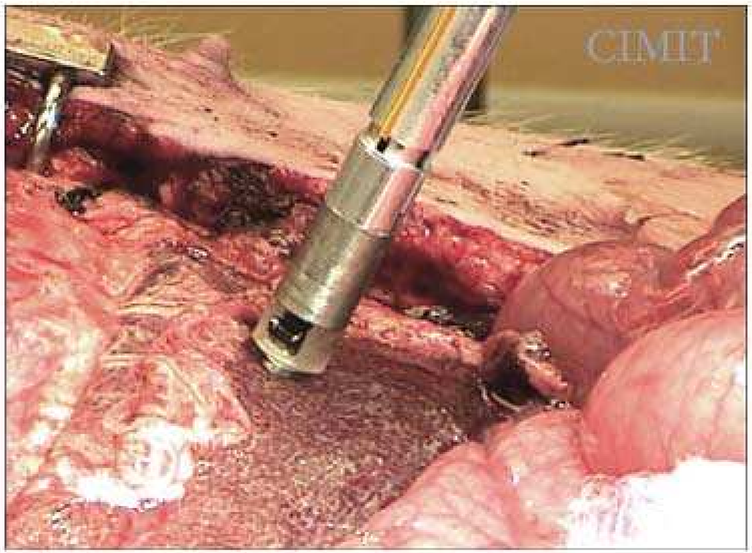
Soft Tissue Characterization

- 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

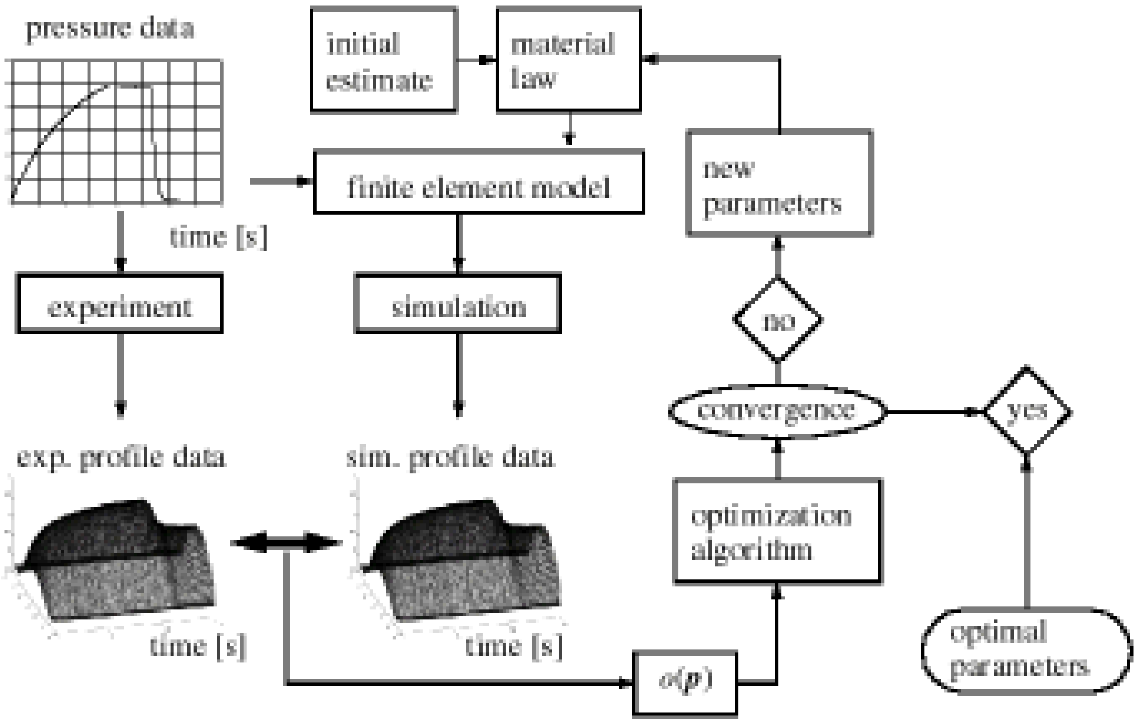
Soft Tissue Characterization

- Elastometry (MR, Ultrasound)
  - mesure property inside any organ non invasively
  - validation ? Only for linear elastic materials

Source Echosens, Paris

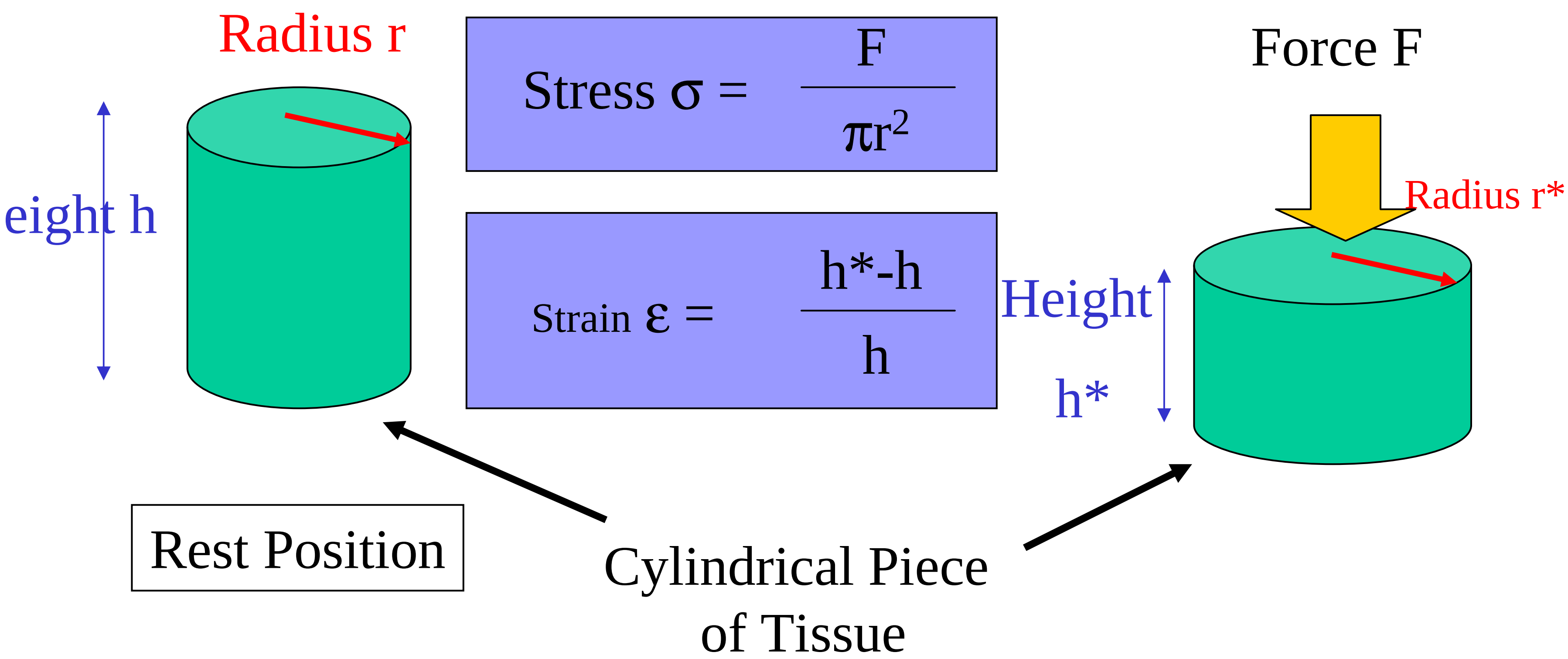
Soft Tissue Characterization

- Inverse Problems
  - well-suited for surgery simulation (computational approach)
  - requires geometry & BC before and after deformation



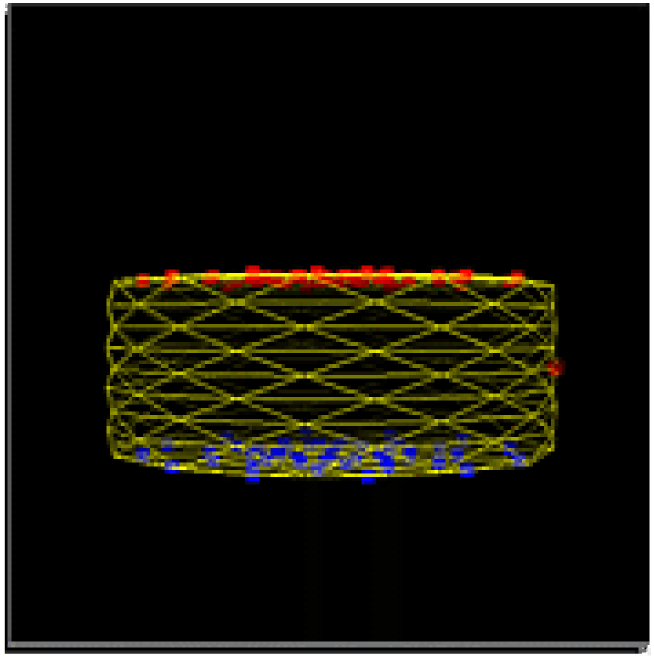
Soft Tissue Characterization

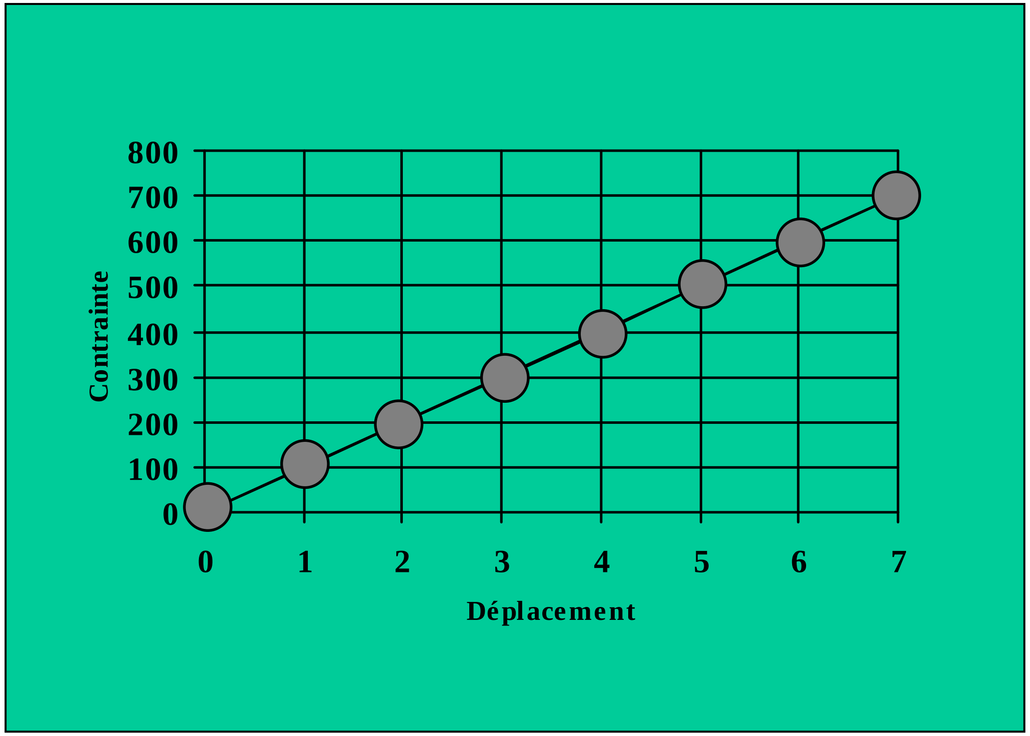
- To characterize a tissue, its stress-strain relationship is studied



Linear Elastic Material

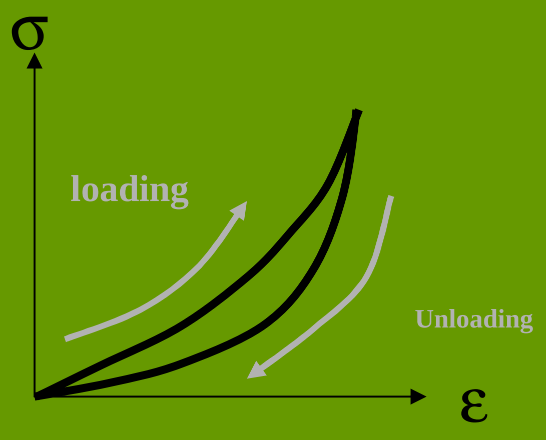
- Simplest Material behaviour
- Only valid for small deformations (less than 5%)



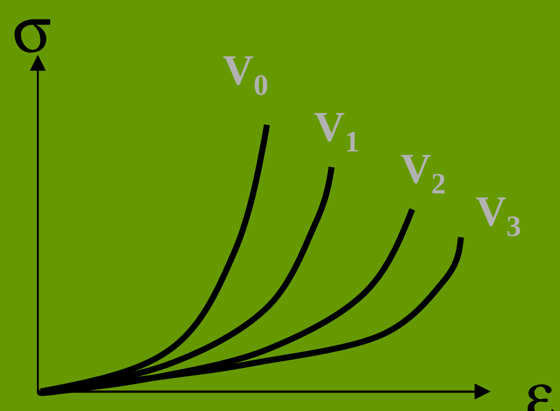


Biological Tissue

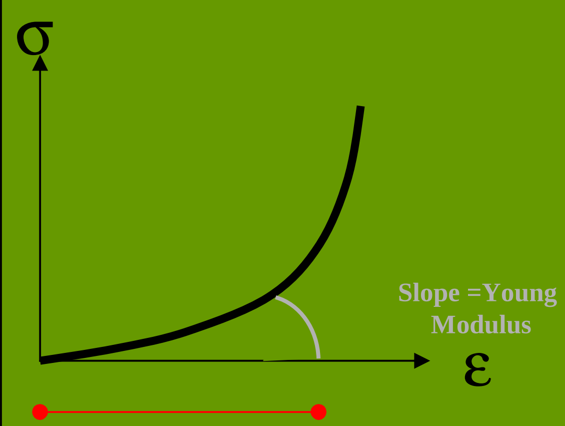
- Far more complex phenomena arises



Hysteresis

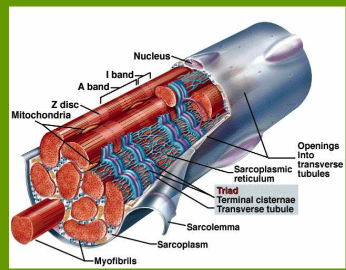


Visco-elasticity

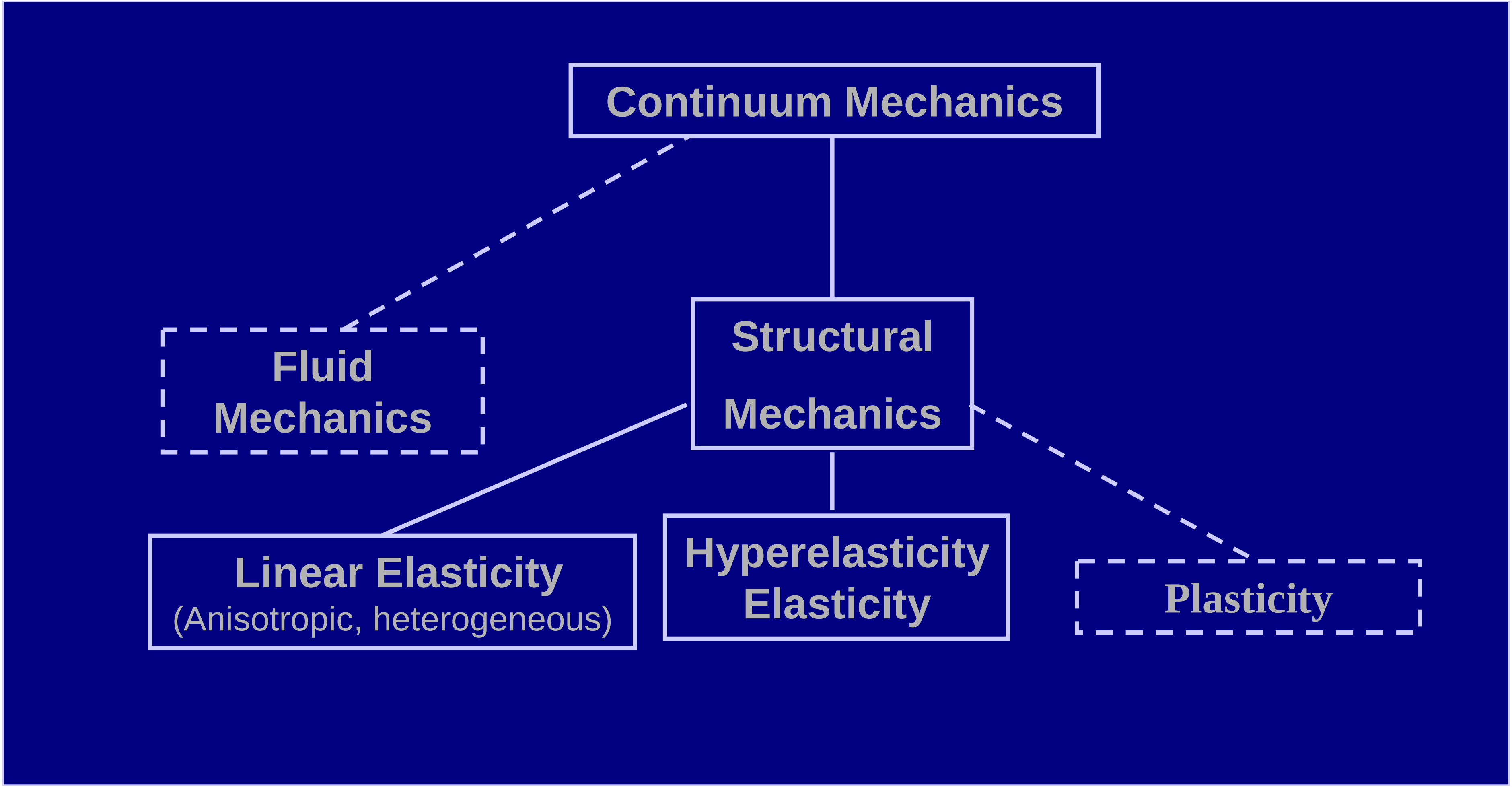


Non-Linearity

Anisotropy



Continuum Mechanics



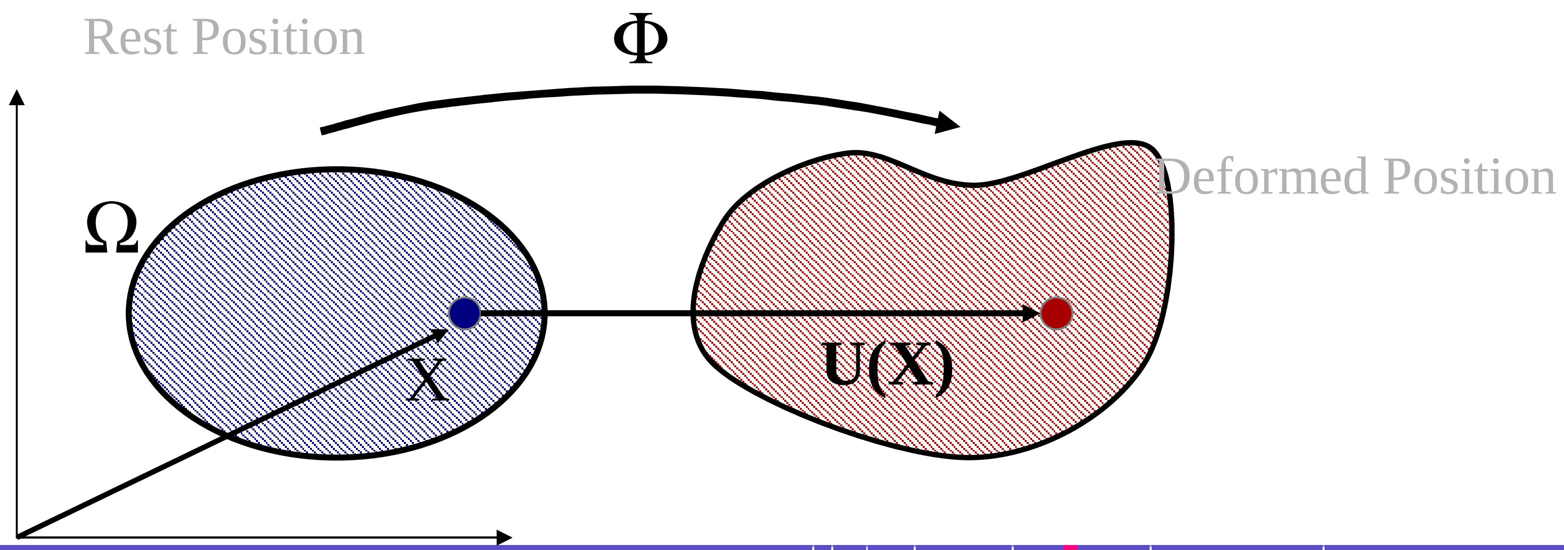
Material Modeling: Basics of Continuum Mechanics

- Deformation Function

$$X \in \Omega \mapsto \phi(X) \in \Re^3$$

- Displacement Function

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

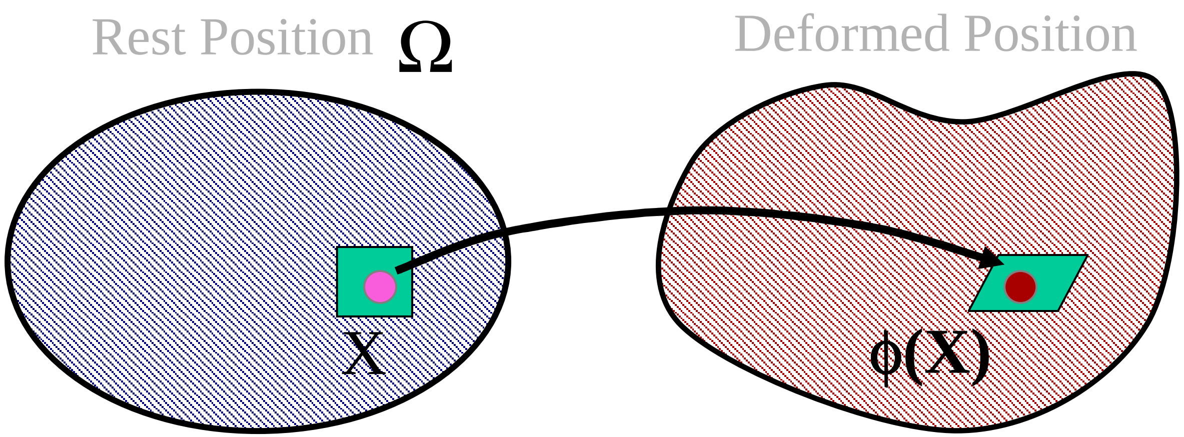


Basics of Continuum Mechanics

- The local deformation is captured by the deformation gradient :

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

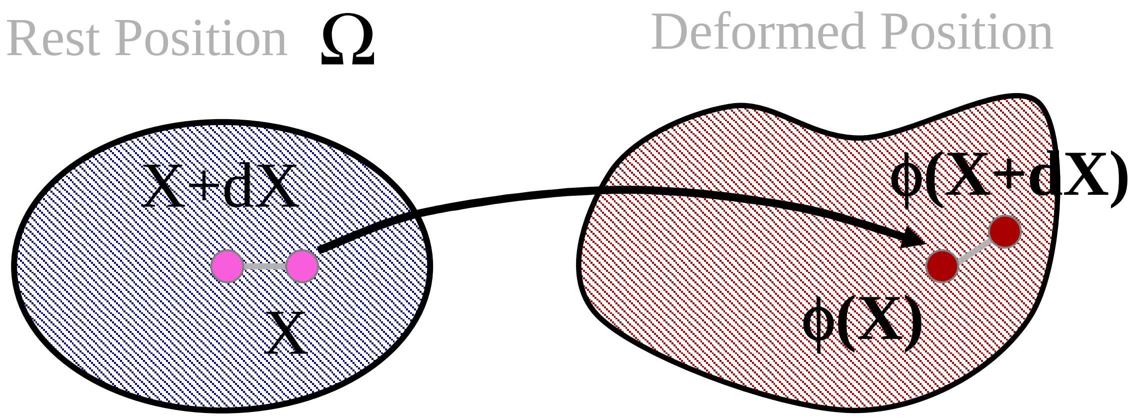
$$F_{ij} = \frac{\partial \phi_i}{\partial X_j} = \begin{bmatrix} \frac{\partial \phi_1}{\partial X_1} & \frac{\partial \phi_1}{\partial X_2} & \frac{\partial \phi_1}{\partial X_3} \\ \frac{\partial \phi_2}{\partial X_1} & \frac{\partial \phi_2}{\partial X_2} & \frac{\partial \phi_2}{\partial X_3} \\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



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

Basics of Continuum Mechanics

- Distance between point may not be preserved



- Distance between deformed points

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

- Cauchy-Green Deformation tensor

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

Measures the change of metric in the deformed body

Basics of Continuum Mechanics

- Example : Rigid Body motion entails no deformation

$$\phi(X) = RX + T$$

$$F(X) = \nabla\phi(X) = R$$

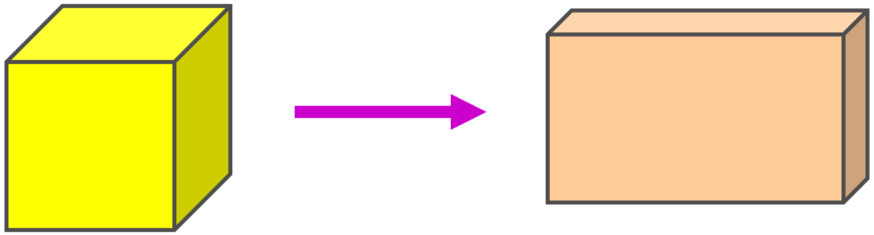
$$C = R^T R = Id$$

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

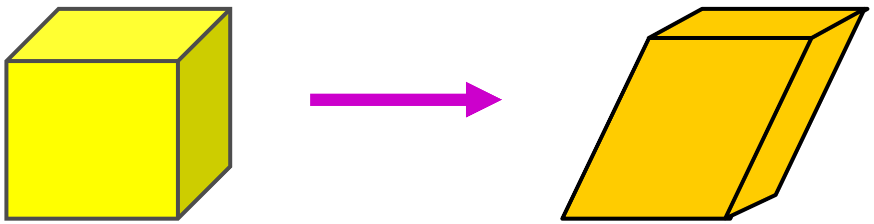
$$E = \frac{1}{2}\left(\nabla\phi^T\nabla\phi - Id\right) = \frac{1}{2}\left(C - Id\right)$$

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



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

Linearized Strain Tensor

- Use displacement rather than deformation

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

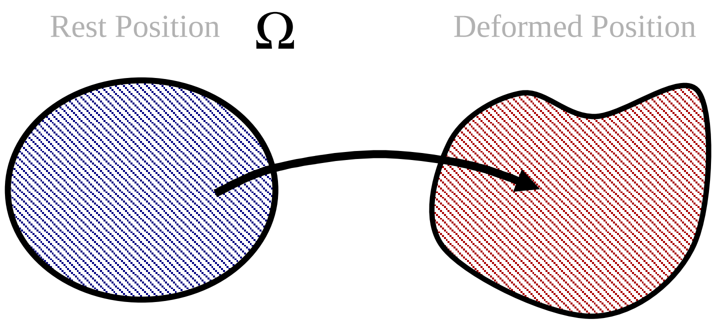
$$E = \frac{1}{2}\left(\nabla U + \nabla U^T + \nabla U^T \nabla U\right)$$

- 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

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

Hooke's Law

(λ, μ) **: Lamé coefficients**

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

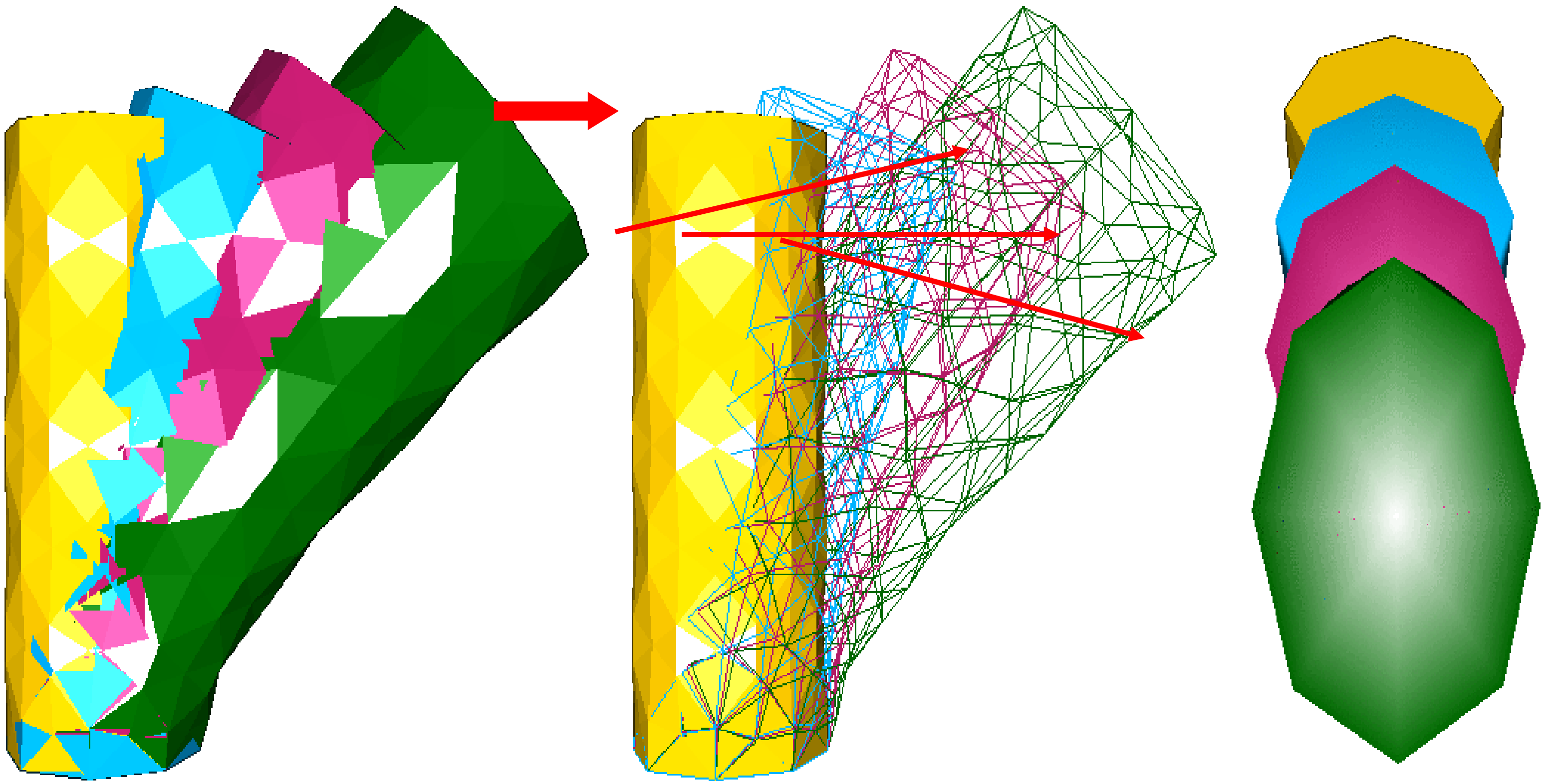
- Advantage :
  - Quadratic function of displacement

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

- 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

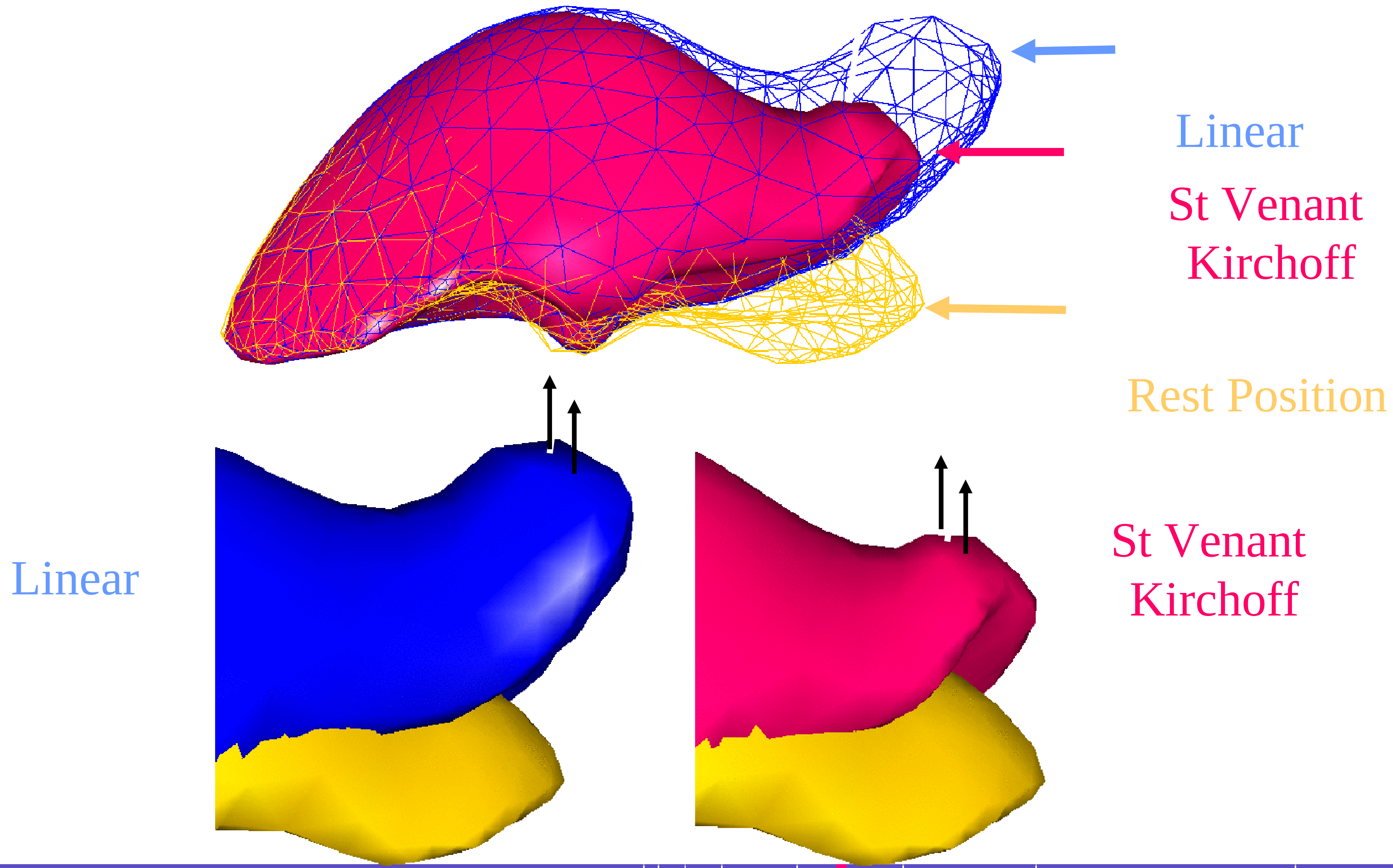
- Isotropic Energy

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

(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

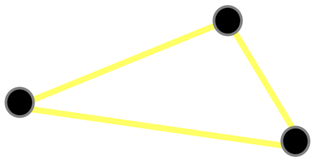
- Neo-Hookean Model $$w(X) = \frac{\mu}{2} trE + f\left(I_3\right)$$
- Fung Isotropic Model $$w(X) = \frac{\mu}{2} e^{trE} + f\left(I_3\right)$$
- Fung Anisotropic Model $$w(X) = \frac{\mu}{2} e^{trE} + \frac{k_1}{k_2}\left(e^{k_2\left(I_4 - 1\right)} - 1\right) + f\left(I_3\right)$$
- Veronda-Westman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

Discretisation techniques

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

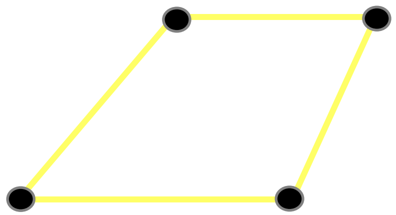
Different types of meshes

- Surface Elements :



Triangle

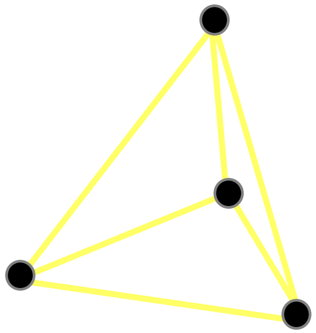
3, 12 nodes and more



Quad

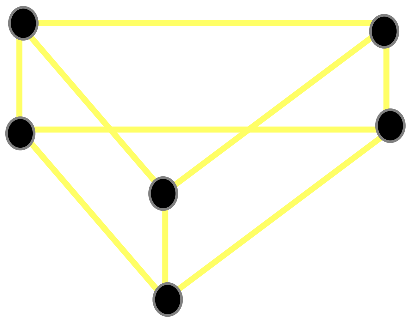
4, 8 nodes and more

- Volume Elements



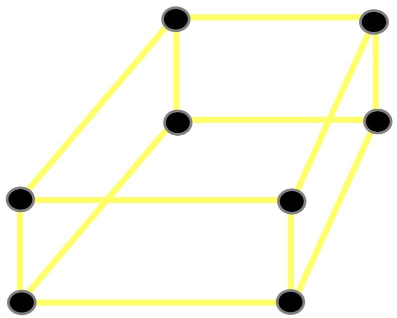
Tetrahedron

4, 10 nodes



Prismatic

6, 15 nodes and more



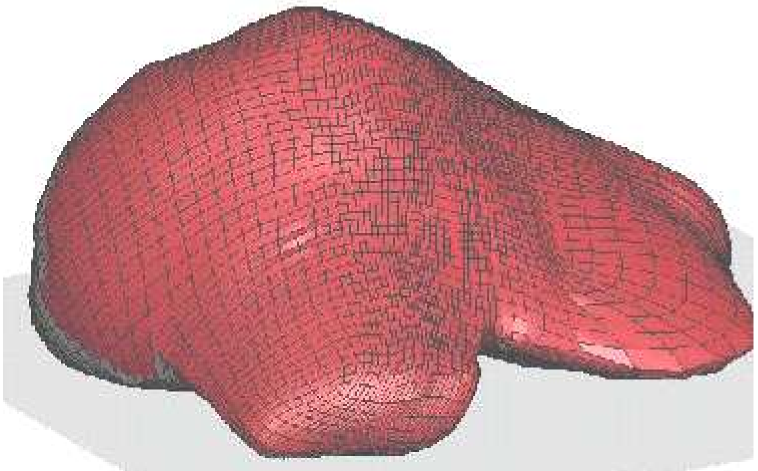
Hexahedron

8, 20 nodes and more

Structured vs Unstructured meshes

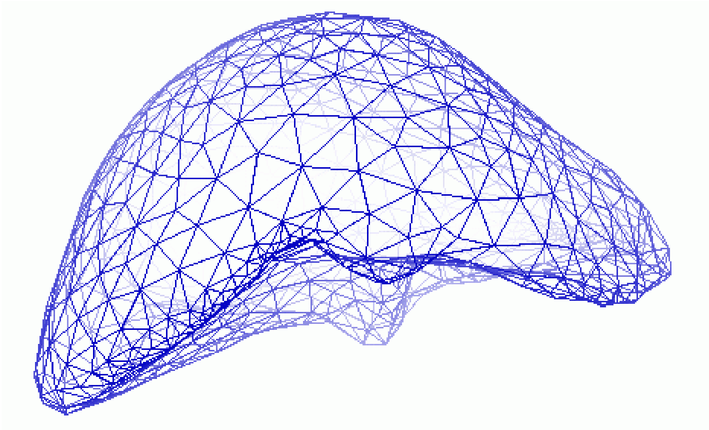
- Example 1 : Liver meshed with hexahedra

3 months work
(courtesy of ESI)



- Example 2: Liver meshed with tetrahedra

Automatically
generated (10s)

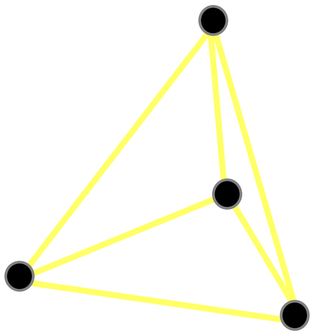


Volumetric Mesh Discretization

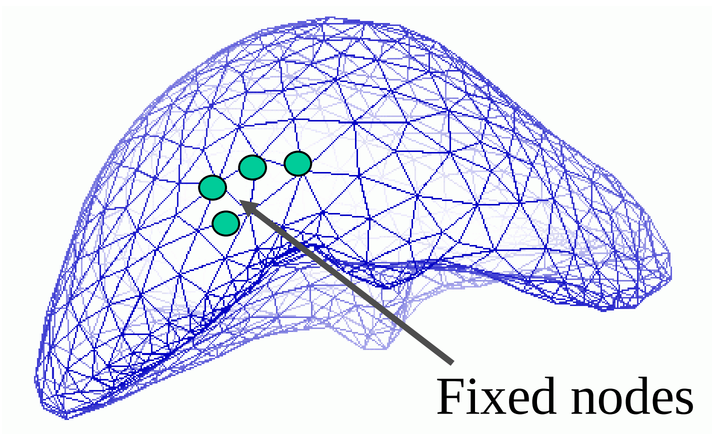
- Classical Approaches :
  - Finite Element Method (weak form)
  - Rayleigh Ritz Method (variational form)
  - Finite Volume Method (conservation eq.)
  - Finite Differences Method (strong form)
- FEM, RRM, FVM are equivalent when using linear elements

Rayleigh-Ritz Method

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



Tetrahedron
4 nodes



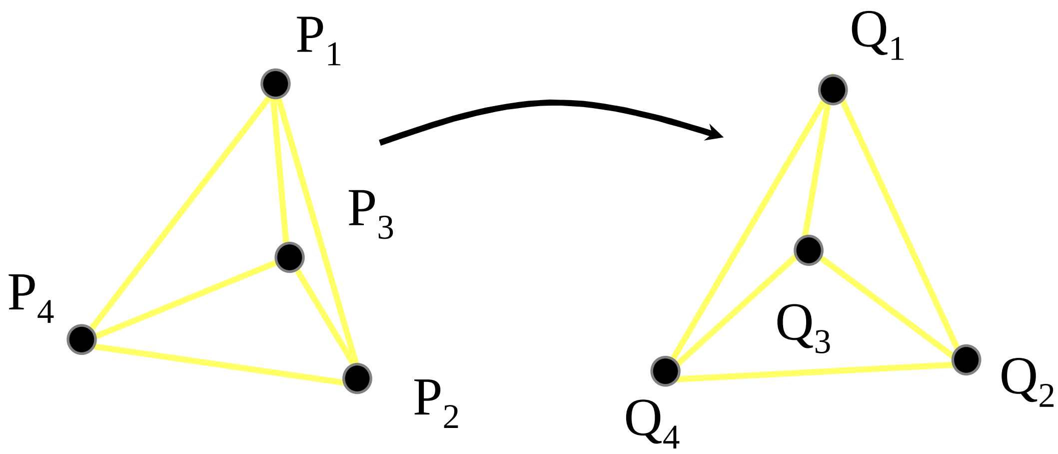
$$w(X) = \frac{\lambda}{2}\left(tr\,E\right)^2 + \mu\, tr\, E^2$$

Young Modulus

Poisson Coefficient

Rayleigh-Ritz Method

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



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

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

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

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

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

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

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

$$KU = R \qquad \text{Static case}$$

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

HyperElasticity=NonLinear Elasticity

$$K(U) = R \qquad \text{Static case}$$

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

Surface-Based Methods

- Only consider the mesh surface under some hypothesis :
  - Linear Elastic Material (sometimes homogeneous)
  - Only interact with organ surface
- Pros :
  - No need to produce volumetric meshes
  - Much faster than volumetric computation
- Cons :
  - Only linear material
  - No cutting

Evolution

- Dynamic evolution
  - Discrete models = lumped mass particles submitted to forces
  - Newtonian evolution (1st order differential system):

$$\begin{cases} \delta P = V.dt \\ \delta V = M^{-1}F(P,V).dt \end{cases}$$

  - Explicit schemes:
    - Euler: $\begin{cases} \delta P = V_t .dt \\ \delta V = M^{-1}F(P_t , V_t).dt \end{cases}$
    - Runge-Kutta: several evaluations to better extrapolate the new state [press92]

    → Unstable for large time-step !!

  - Semi-Implicit schemes:
    - Euler: $\begin{cases} \delta P = V_{t+dt} .dt \\ \delta V = M^{-1}F(P_t , V_t).dt \end{cases} \rightarrow \begin{cases} P_{t+dt} = 2P_t - P_{t-dt} + M^{-1}F(P_t , V_t).dt^2 \\ V_{t+dt} = (P_{t+dt} - P_t)dt^{-1} \end{cases}$

Evolution

– Implicit schemes [terzopoulos87], [baraff98], [desbrun99], [volino01], [hauth01]

- First-order expansion of the force:

$$F(P_{t+dt}, V_{t+dt}) \approx F(P_t, V_t) + \partial F/\partial P\ \delta P + \partial F/\partial V\ \delta V$$

- Euler implicit

$$\rightarrow \begin{cases} \delta P = V_{t+dt}.dt \\ \delta V = H^{-1} Y \end{cases} \quad \text{with} \quad \begin{aligned} H &= I - M^{-1}\ \partial F/\partial V\ dt - M^{-1}\ \partial F/\partial P\ dt^2 \\ Y &= M^{-1}\ F(P_t, V_t) + M^{-1}\ \partial F/\partial P\ V_t\ dt^2 \end{aligned}$$

- Backward differential formulas (BDF) : Use of previous states

→ Unconditionally stable for any time-step

… But requires the inversion of a large sparse system

– Choleski decomposition + relaxation

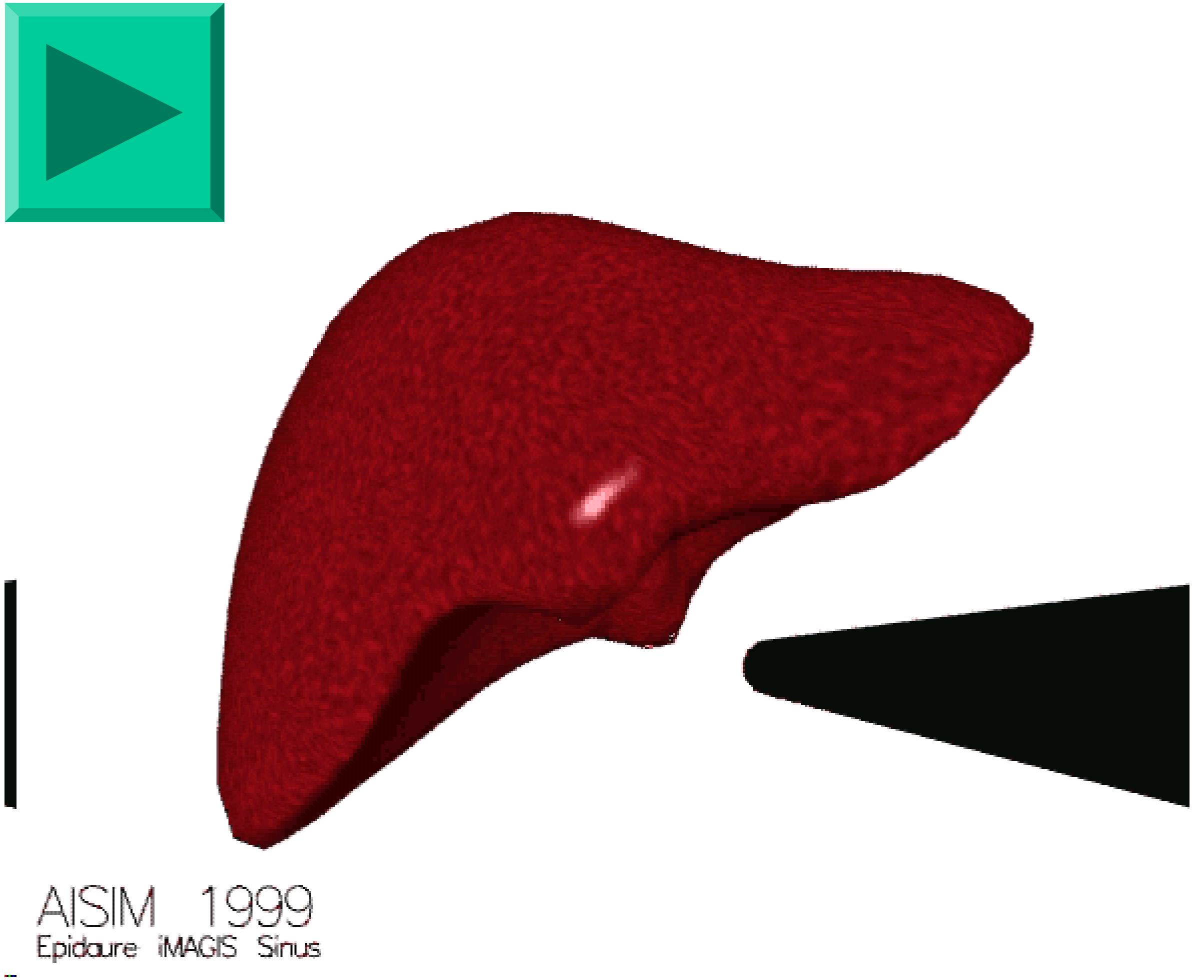
– Conjugate gradient

– Speed and accuracy can be improve through preconditioning (alteration of **H**)

Example of Soft Tissue Models

	Pre-computed Elastic Model	Tensor-Mass and Relaxation-based Model	Non-Linear Tensor-Mass Model
Computational Efficiency	+ + +	+	-
Cutting Simulation	-	++	++
Large Displacements	-	-	+

Precomputed linear elastic model



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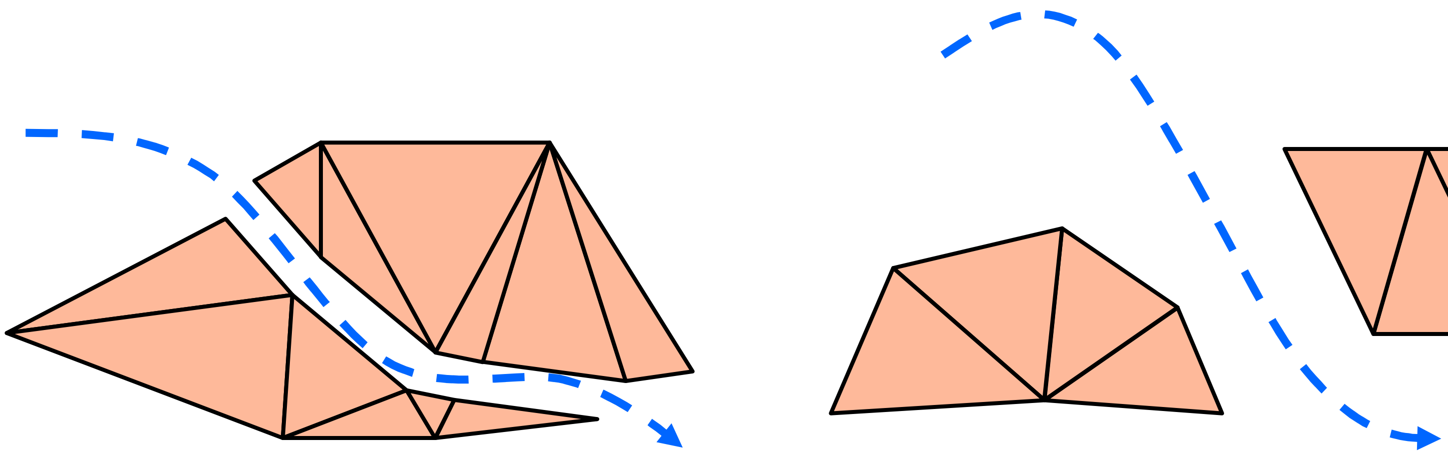
Tetrahedra

Different Technical Issues

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

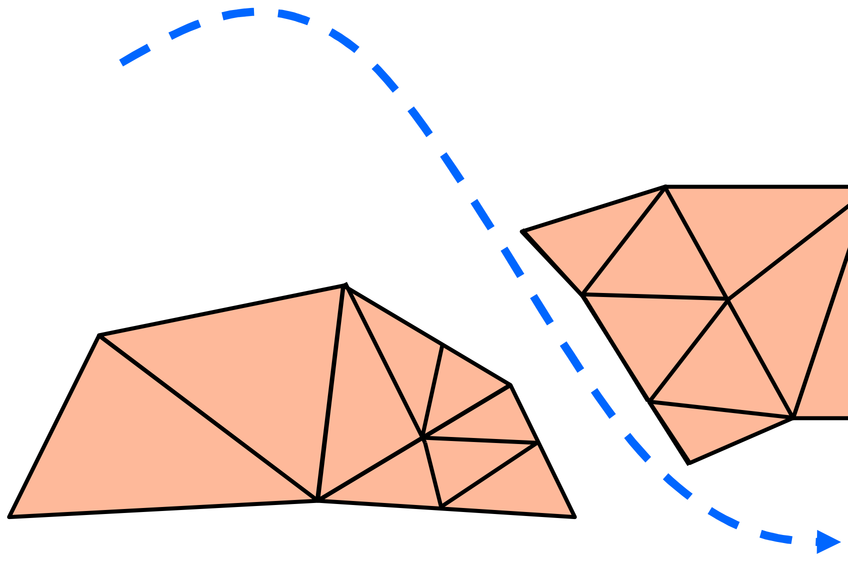
Different algorithms for cutting tetrahedral meshes

- Split of tetrahedra
  [Bielser, 2000] [Mohr, 2000] [Nienhuys, 2001]
  - **+** Accurate, realistic
  - **-** Decrease of Mesh Quality
- Removing Tetrahedra
  [Forest, 2002]
  - **+** Keeps a good mesh quality
  - **-** Gross cut

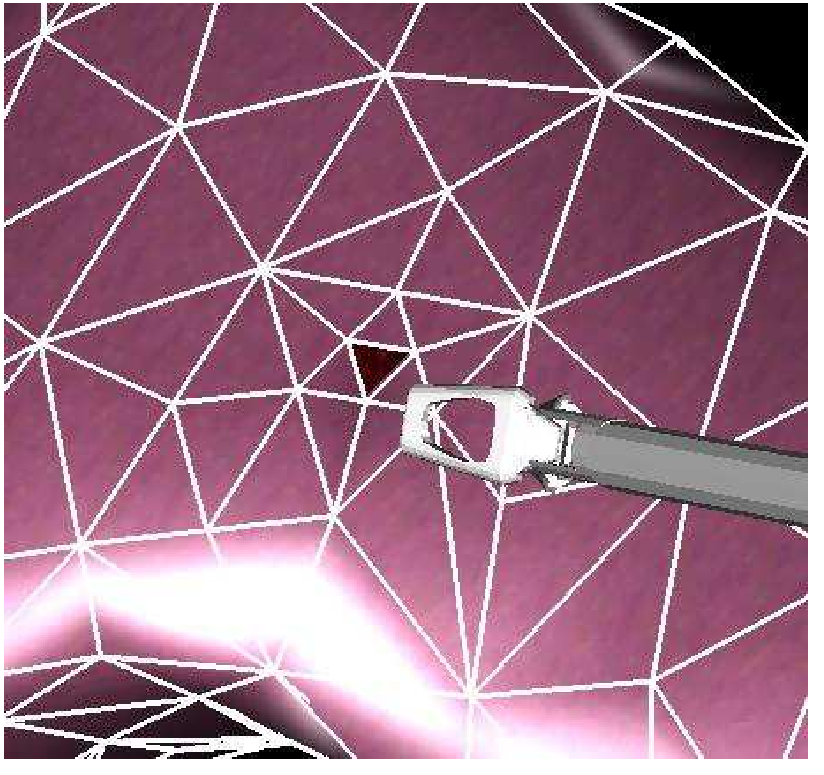


Proposed Technique

- Remove Tetrahedra
- Refine Mesh before removing material

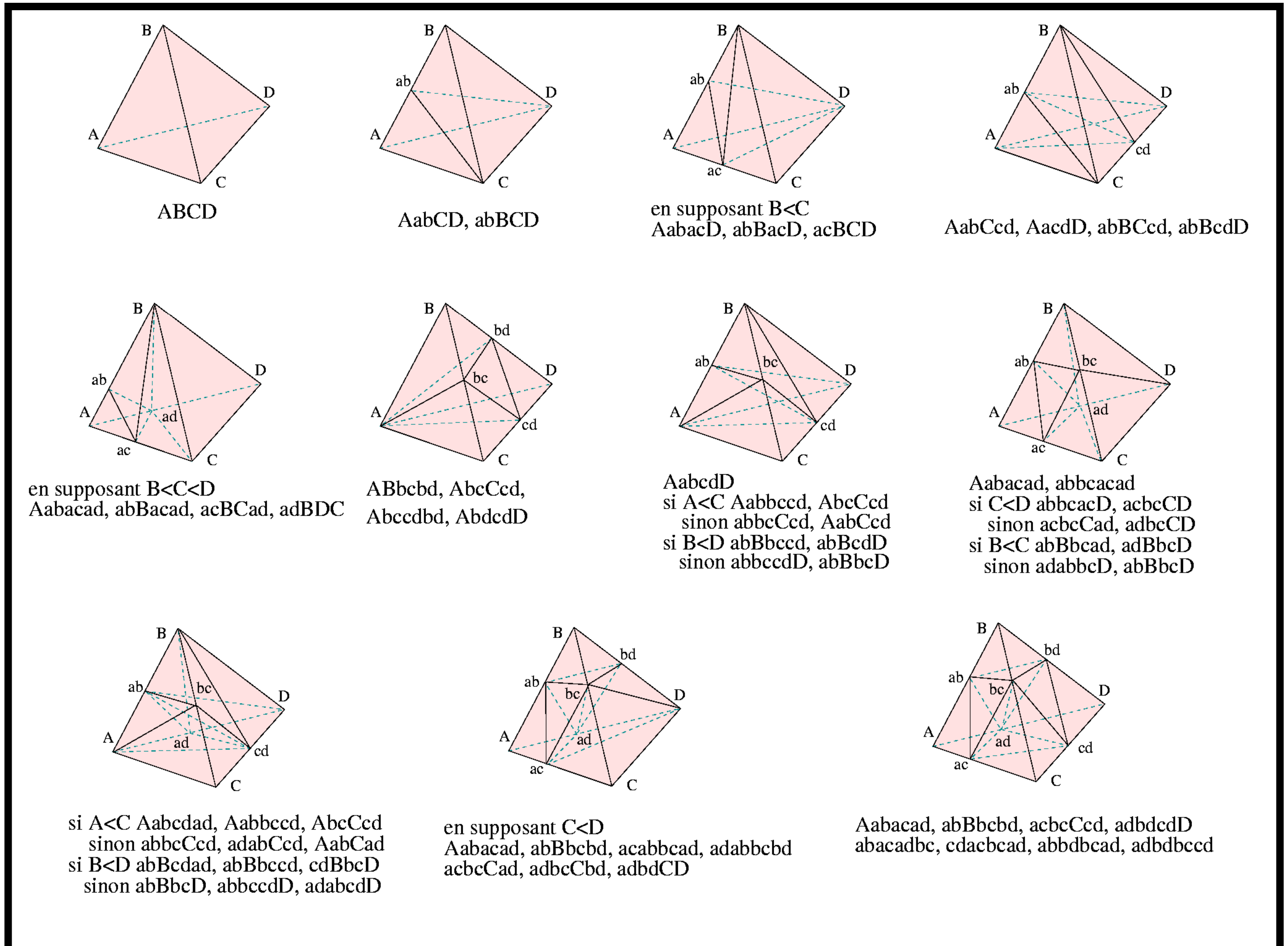


Dynamic Refinement



Example of refine-before-cutting

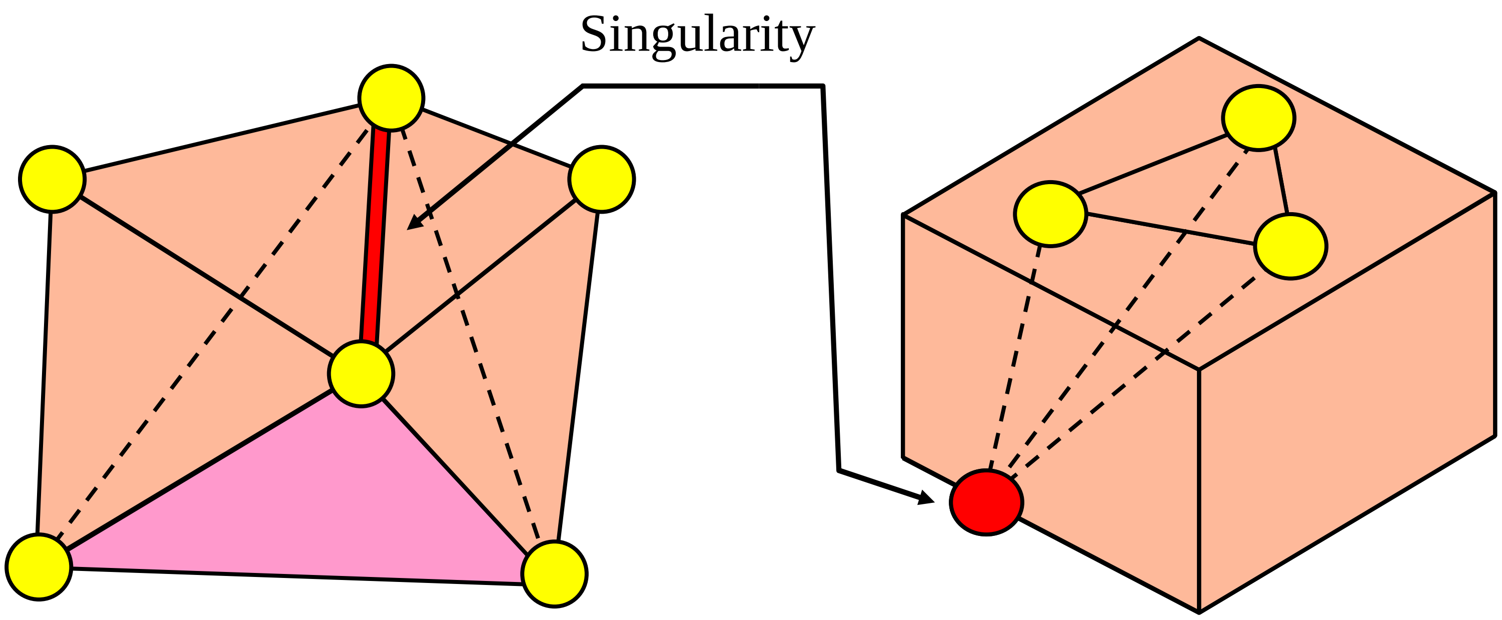
Refinement by Edge Split



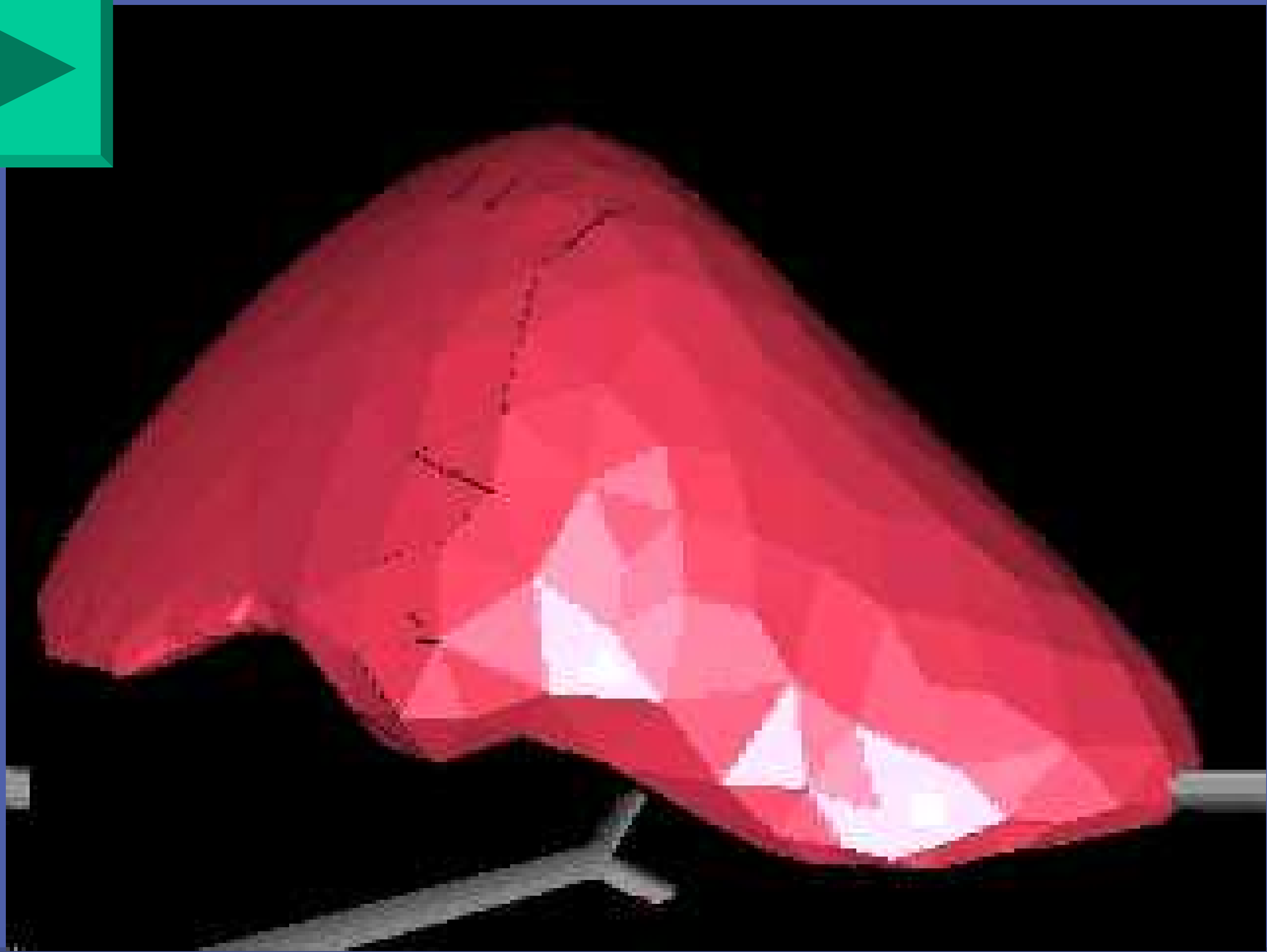
Decomposition of tetrahedra by edge split

Topological Singularities

- Removing a tetrahedron may create a singularity (zero thickness at edge and vertices) (see [forest])



Non-linear Tensor-Mass Models



Different Technical Issues

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

Previous Work

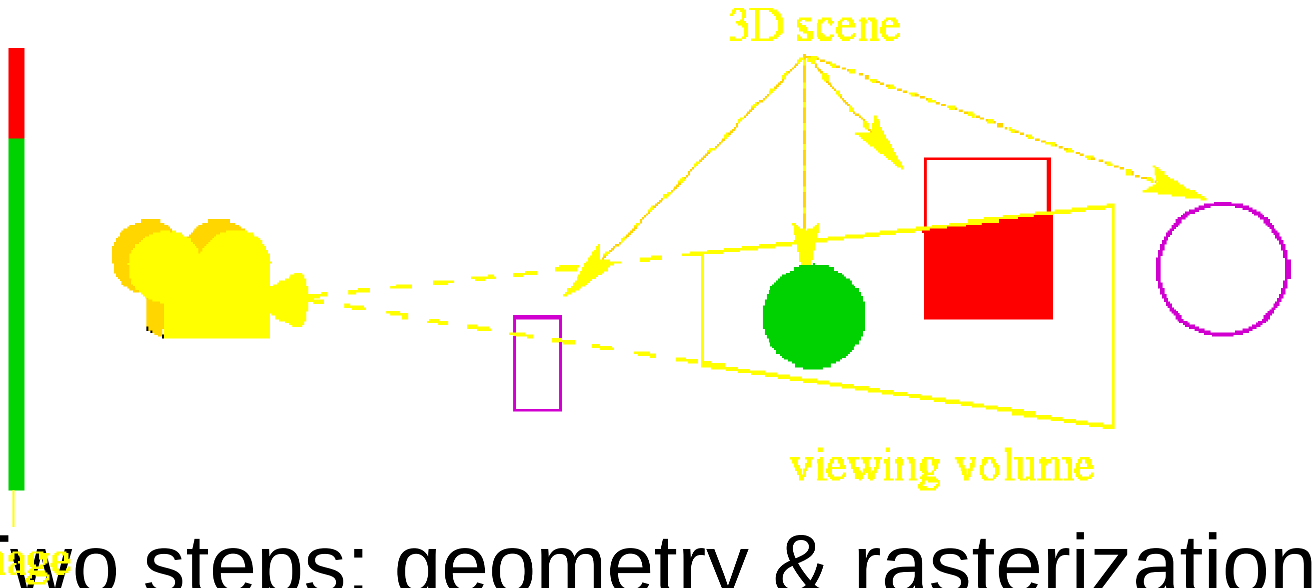
- A lot of research on Collision Detection

- Hierarchy of oriented bounding boxes:

  Gottshalk & al. - *Obb-tree: A hierarchical structure for rapid interference detection* - SIGGRAPH'96
- public domain package *RAPID*
- Very efficient,
  but needs pre-computation

The Rendering Process

- Camera = viewing volume + projection



- Two steps: geometry & rasterization

Collision Detection and Rendering analogy

a tool collides the organ

⇔

a part of the organ is inside the tool

⇔

if we define a camera with a viewing volume that matches the tool geometry, the organ will be in the picture.

Different Technical Issues

- Mesh Reconstruction from Images
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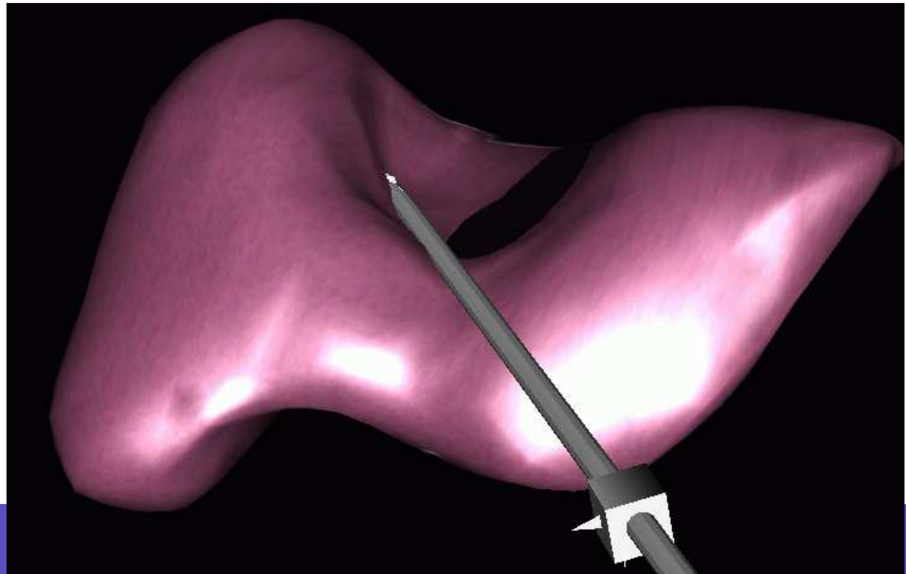
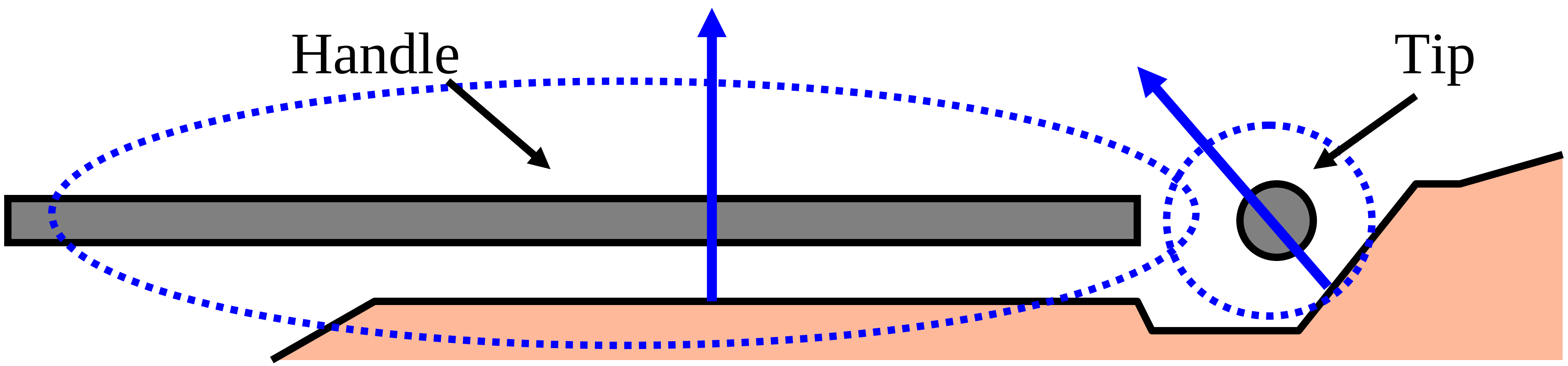
Tool-Soft Tissue Interaction

- Prevent penetration of tool inside the soft tissue
  - Detect intersections
  - Push explicitly mesh vertices outside the tool



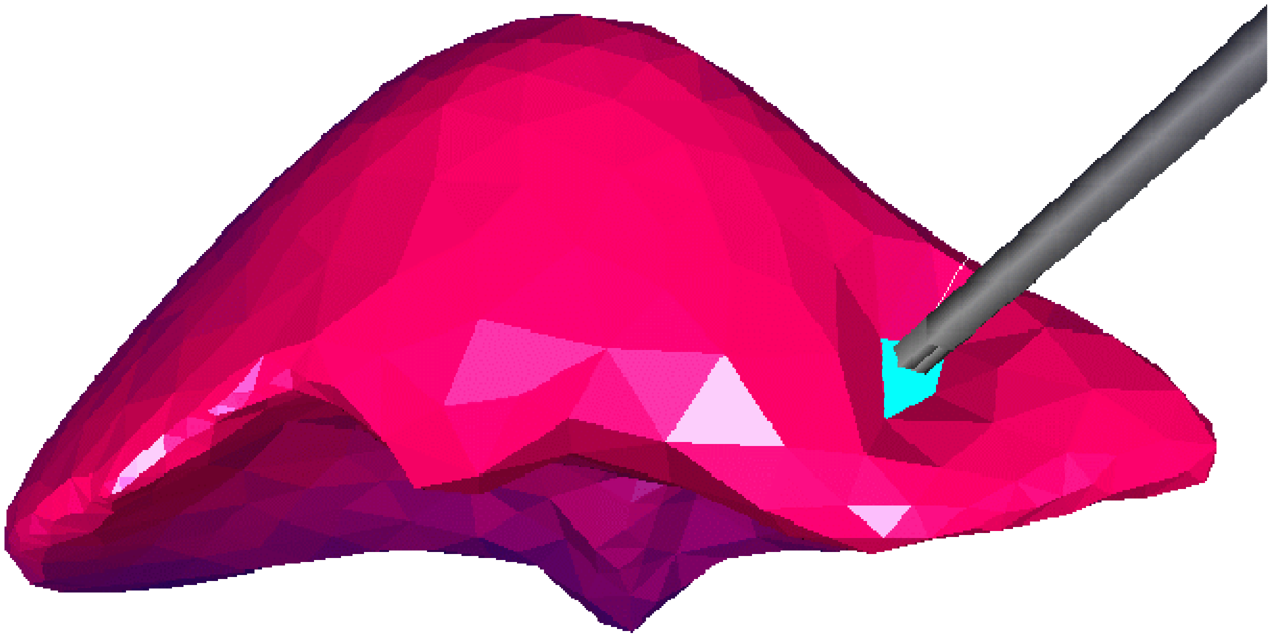
First Approach *[Picinbono, 2001]*

- 2 different tools : tip and handle
- Compute average normal in the neighborhood of the contact
- Projection of vertices in this plane



Collision Processing

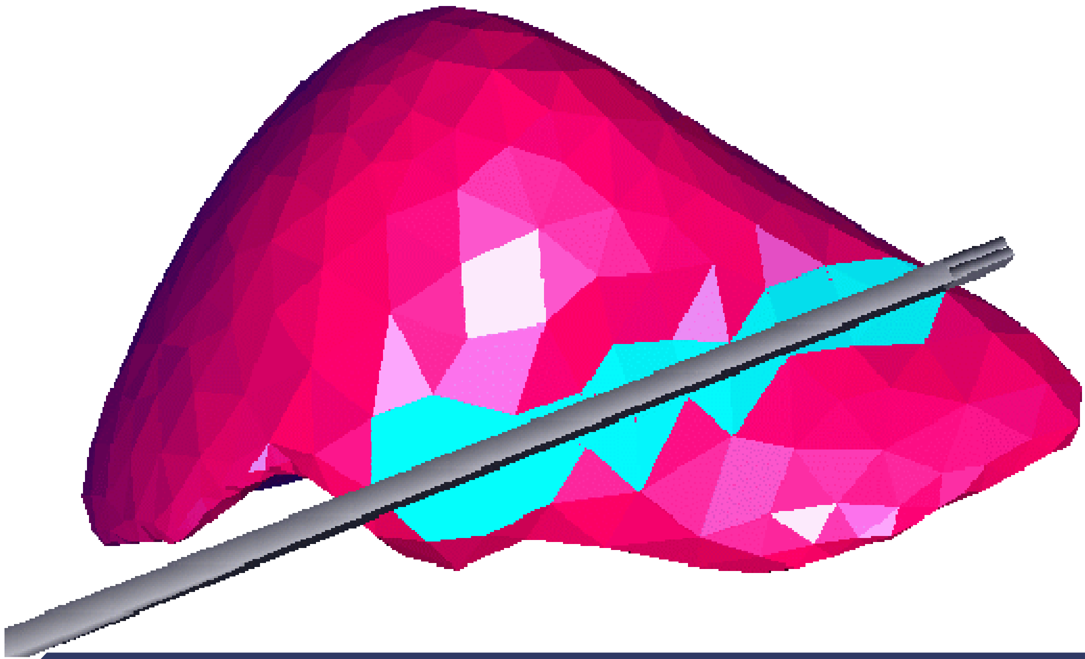
- Contact with the tip of the instrument



➡ **Projection on the plane defined by the tip of the instrument and the average normal of intersected triangles**

Collision Processing

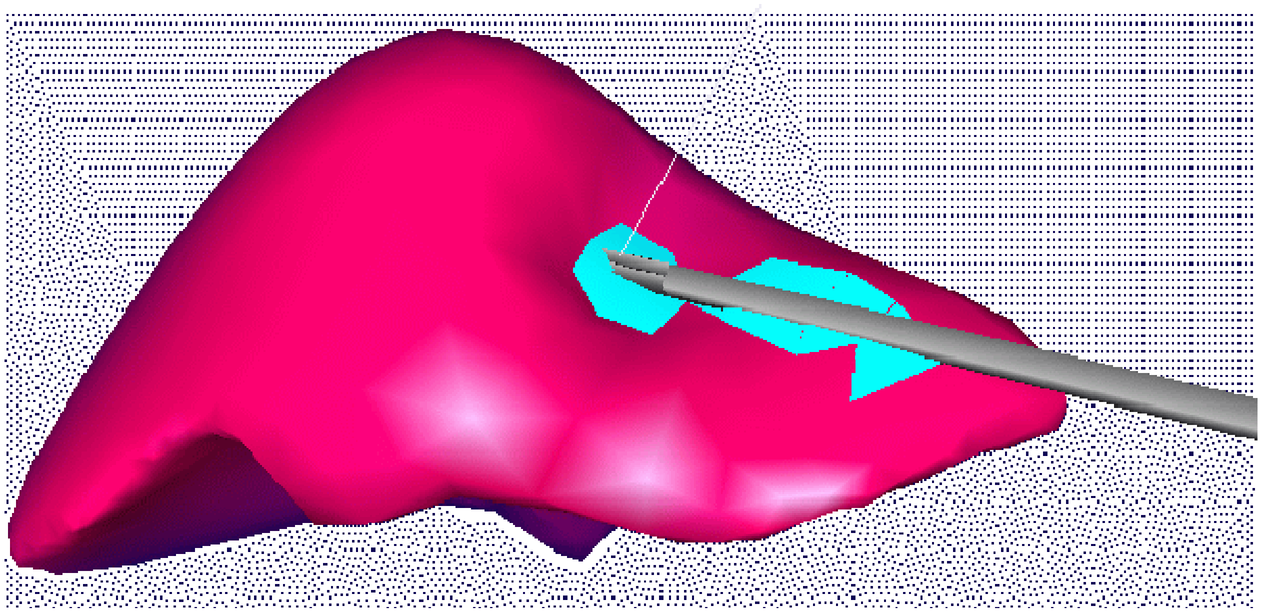
- Contact with the handle of the tool



Projection on the plane defined by the tool direction and the averaged normal direction

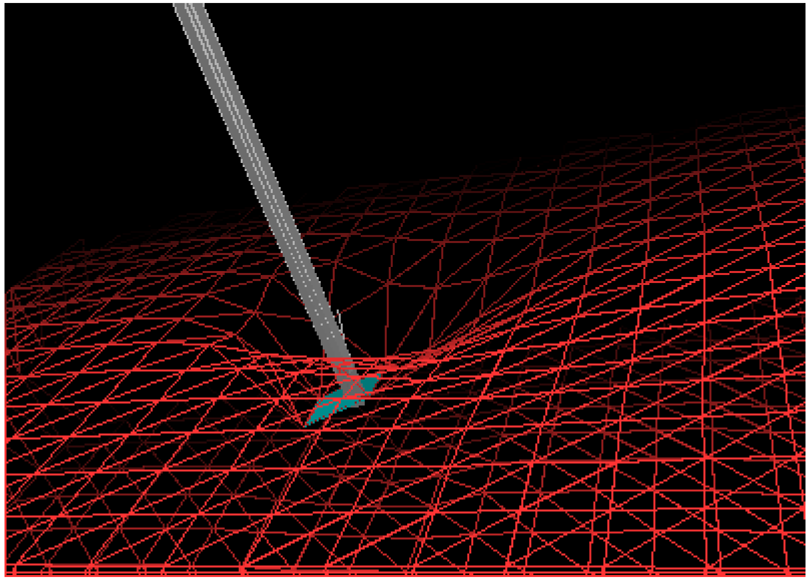
Collision Processing

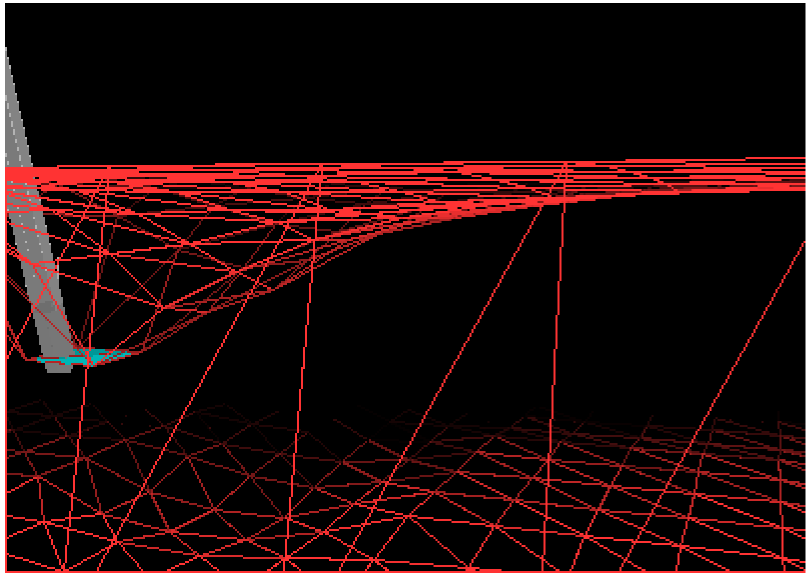
- Perform 2 detections simultaneously



Possible interactions

- Slip on the surface



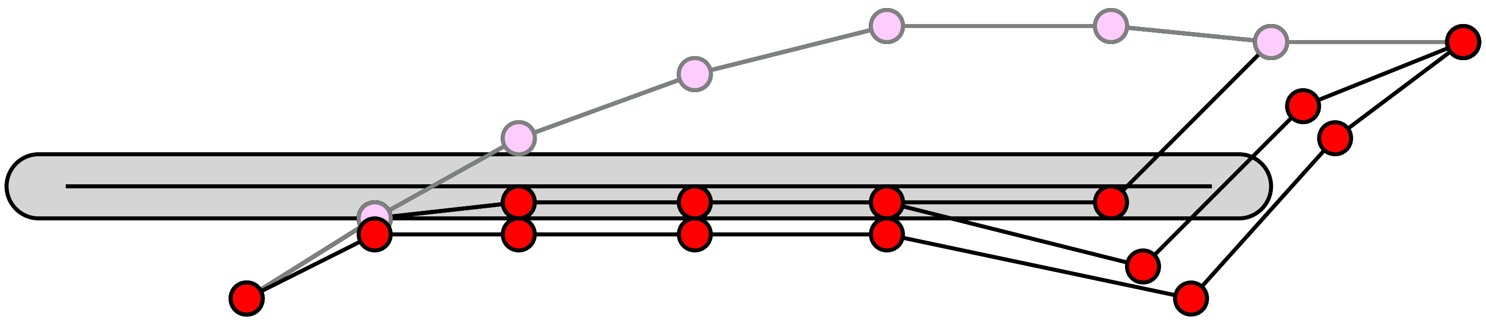


Limitations of this approach

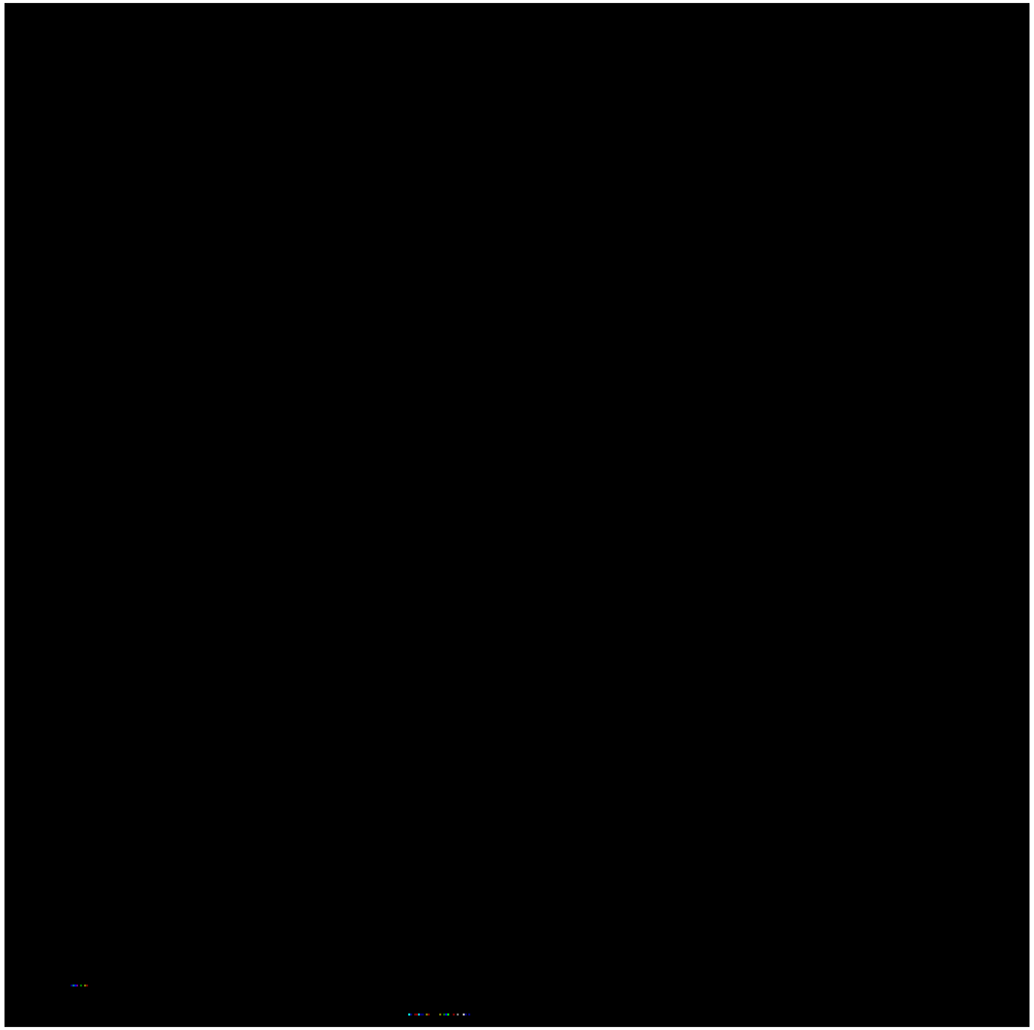
– Same normal vector for all triangles in the same neighborhood
– Leads to instabilities when handling a complex geometry

New approach

- Three steps
  - Prevent vertices to collide with the tool axis
  - Move vertices near the tip of the tool
  - Move vertices outside the volume of the tool



Example



Different Technical Issues

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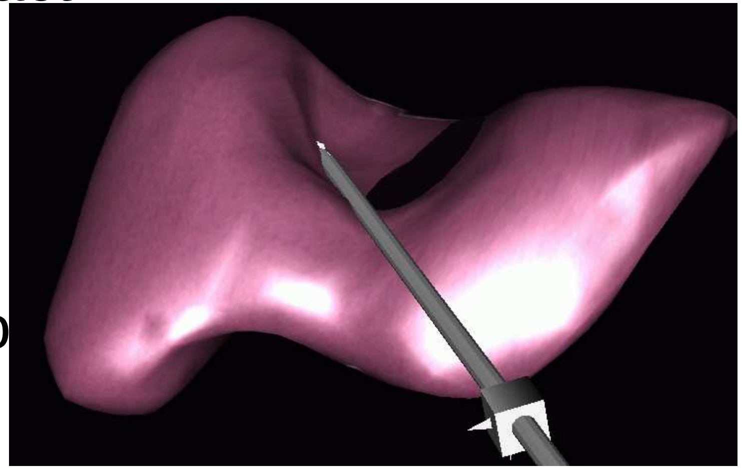
- Surface Rendering
- Haptic Feedback

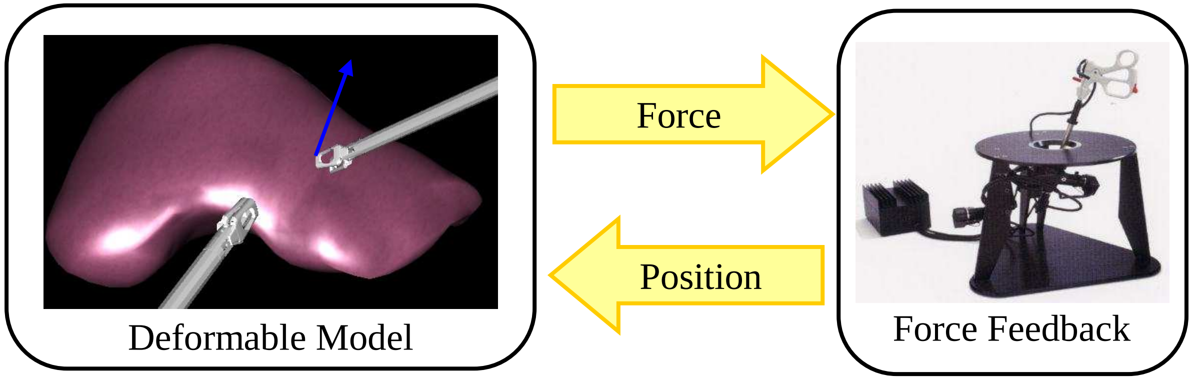
Haptic Feedback

- Principle
  - Give a realistic sense of contact with the soft tissue
- Motivation
  - Increase realism
  - Naturally limit the amplitude o hand motion
- Pitfalls
  - Frequency update of haptics > 500 Hz
  - Frequency update of deformable models ≈ 30 Hz



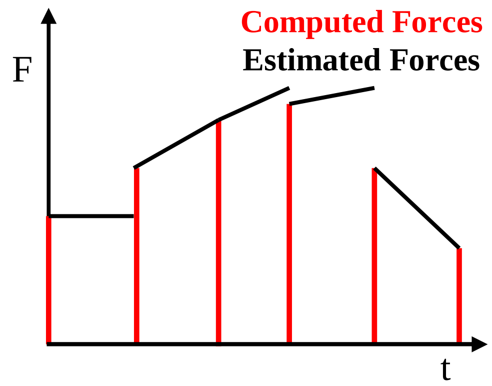
Mouvement non contraint par le retour d'effort

First approach *[Picinbono, 2001]*



Frequency Update 20 Hz
Force Computation

Frequency Update 300 Hz
Force Extrapolation

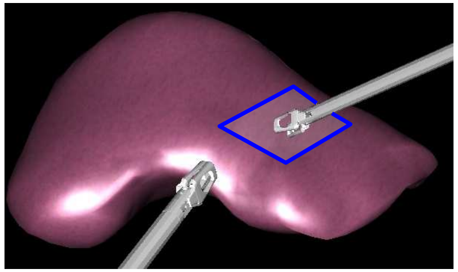


Example of extrapolation

- Unstable if complex geometry
- Difficult extrapolation for large hand motions

$$\boldsymbol{F}^{p}(t) = \boldsymbol{F}_n + \frac{\left\|\boldsymbol{P}' - \boldsymbol{P}_n\right\|}{\left\|\boldsymbol{P}_n - \boldsymbol{P}_{n-1}\right\|}\left(\boldsymbol{F}_n - \boldsymbol{F}_{n-1}\right) \qquad t_n \leq t < t_{n+1}$$

Local Model
[Mendoza, 2001] [Balaniuk, 1999] [Mark, 1996]

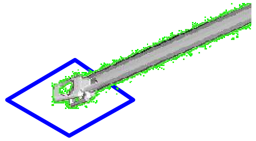


Deformable Model

Modèle local

Position

Retour d'effort



Update Frequency 20 Hz
Computation of a local model

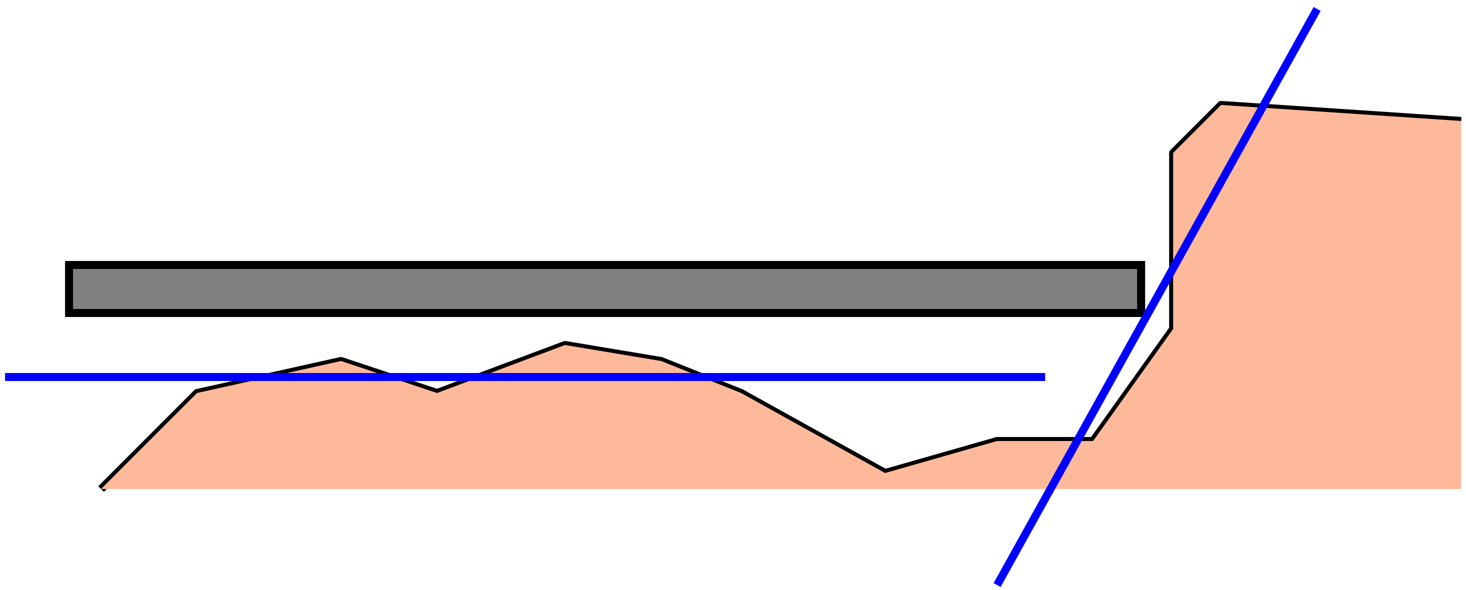
Update Frequency 300 Hz

Force Computation from a local model

- Smooth Transition from one local model to the next

Computing the local model

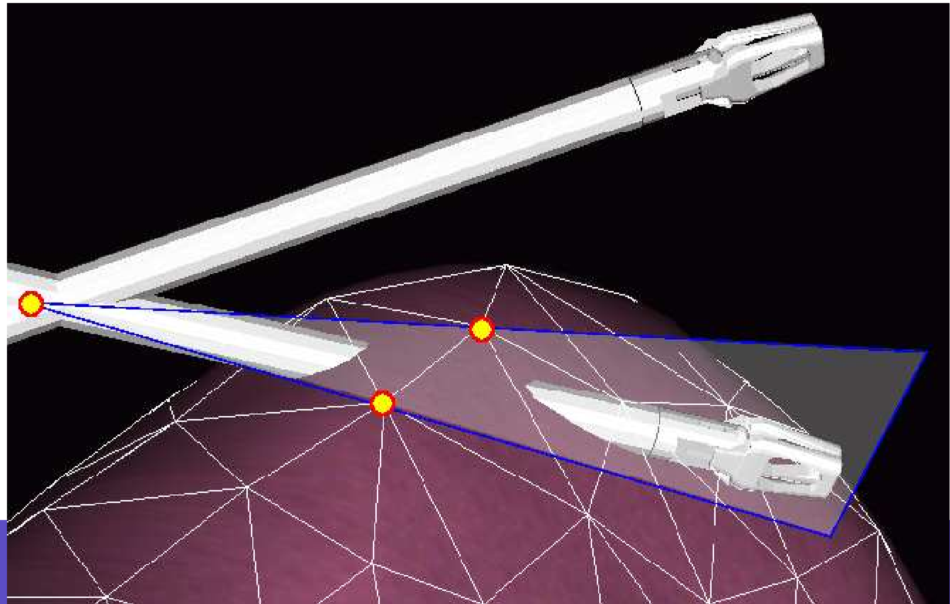
- Described as a set of planes
- One model for the tip
- One model for the handle



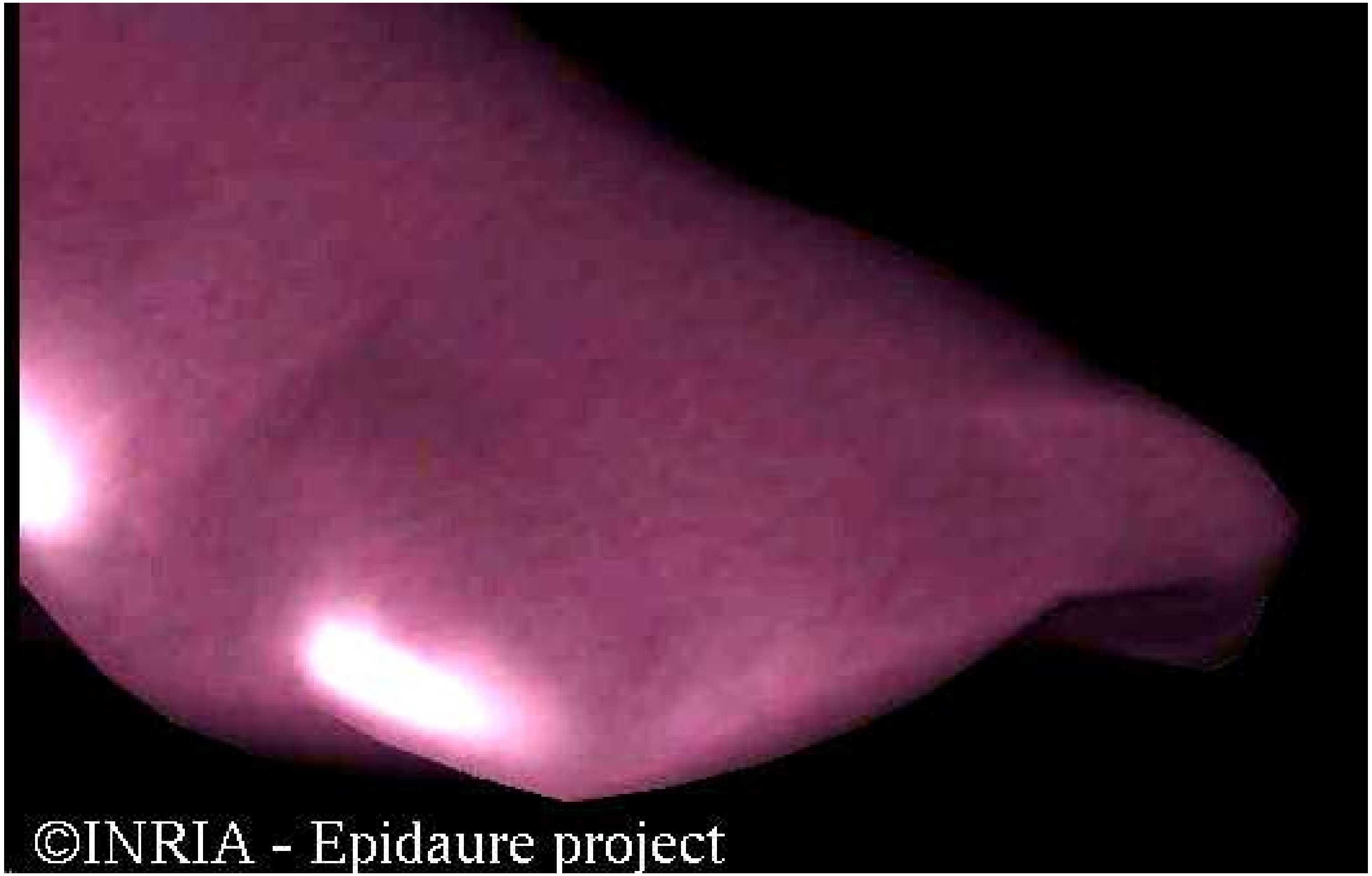
Force Computation

- Proportional to the penetration of the tool tip in the planes described by the local model

$$F = k.(EndP - O_P).\vec{n}_P$$

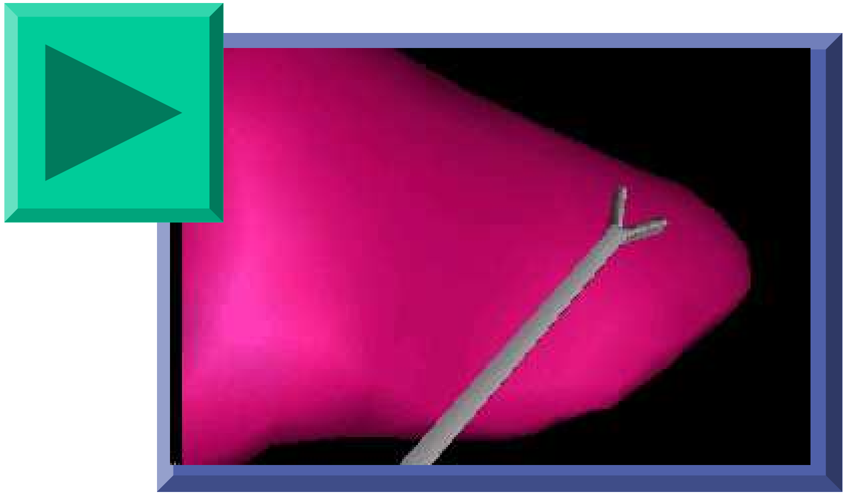


Tensor-Mass Models (low resolution)

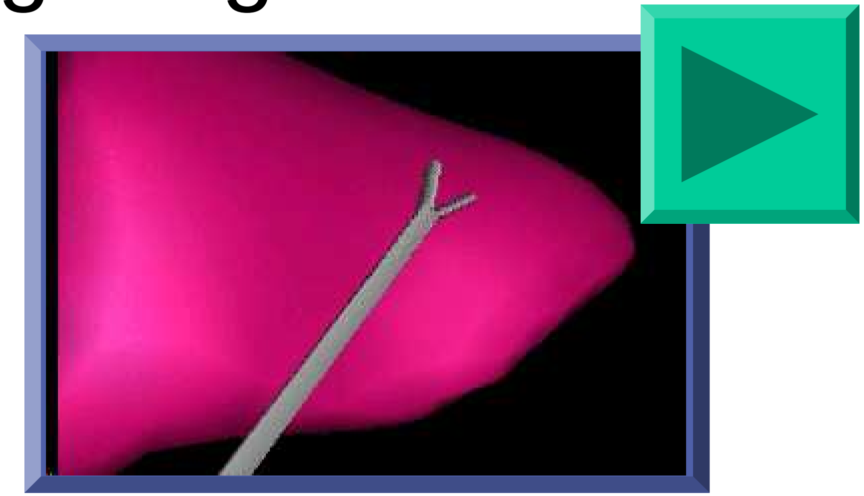


N = 1394   (6342 Tétraèdres)

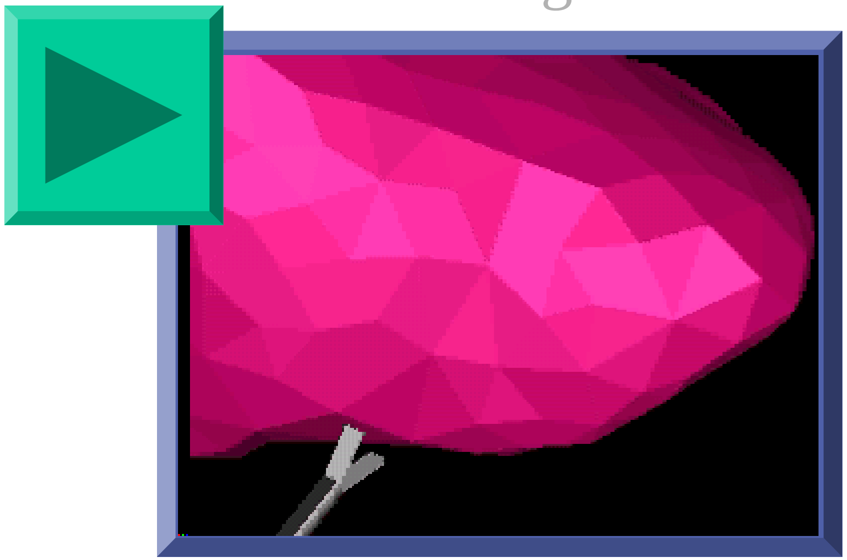
Simulation of surgical gestures



Gliding



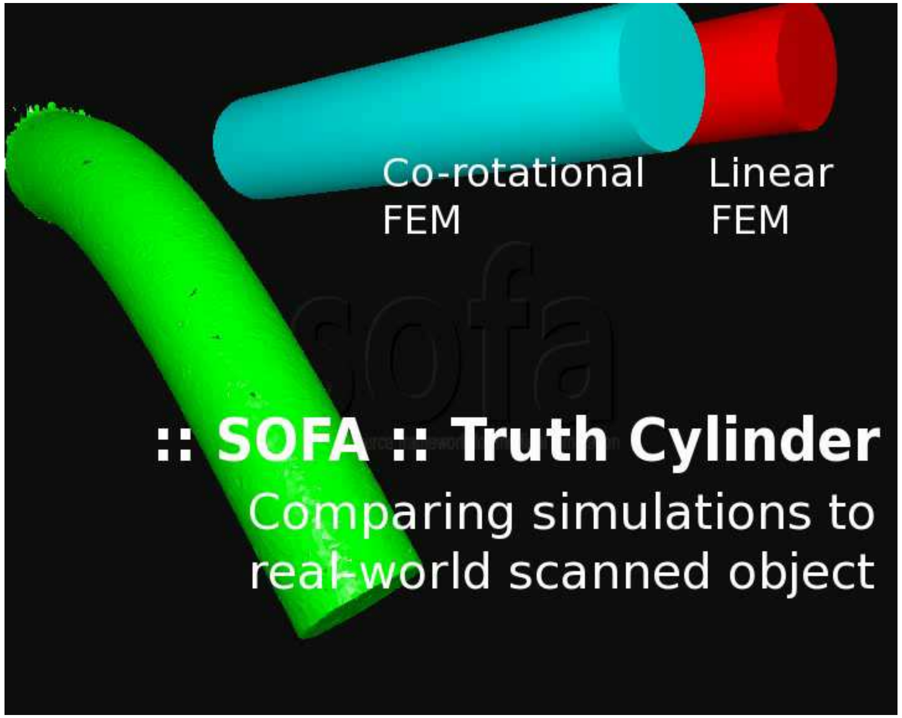
Gripping



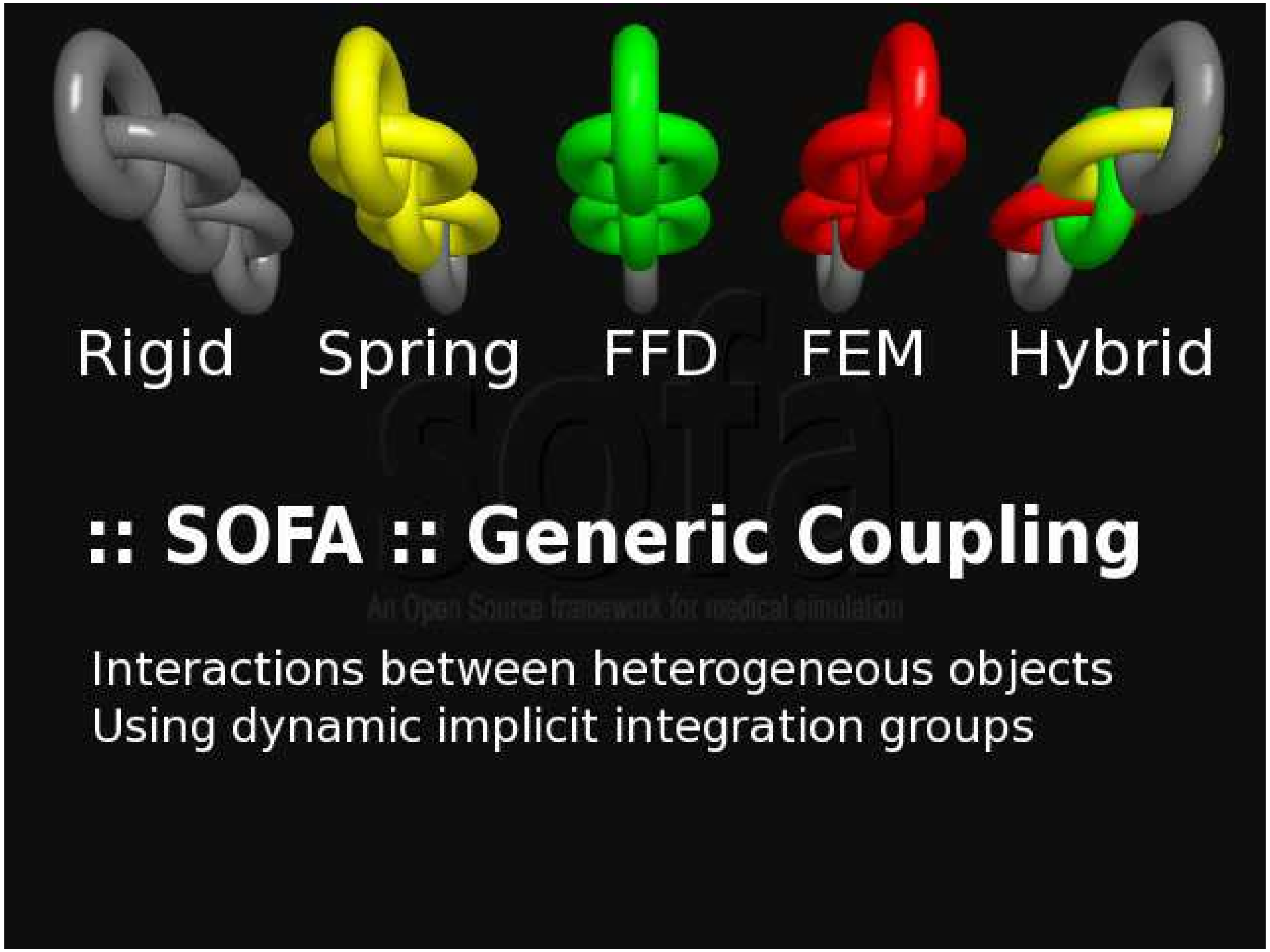
Cutting (pliers)

Cutting (US)

SOFA : www.sofa-framework.org



SOFA : www.sofa-framework.org



Thank you