Ph.D. thesis defense

A Biomechanics-based Articulation Model for Medical Applications

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This thesis is about a 3D articulation model. So, why we need such a model? Well, the problem motivating this work is divided in two:

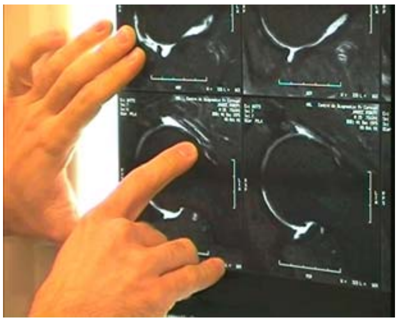
First, you know that, since long time, images are used to help clinicians on diagnosis. Medical imaging evolved along time, from x-rays, passing through fluoroscopy, computer tomography, MRI, and even 3D visualization is available. However, none of those techniques give functional information. Pressure on the tissues or muscular activation, for example, are not visible. Many diagnosis errors and consequently treatment errors are due to this lack of information.

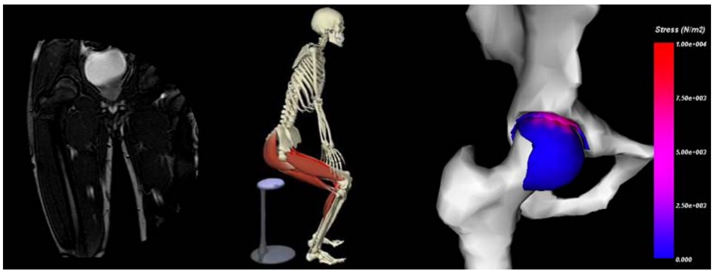
Second, would you go for a surgery if you were not sure it is safe and it will solve your problem? The answer is probably yes, because you have no choice. However, the AirBus engineers were sure the new A-380 would fly when they first put it on the runway for takeoff. The aircraft had been tested and simulated on computer using CAD systems. Wouldn't it be great if computer aided surgery design existed? A lot of pain would be avoided.

0. Intro 3

General objective

- To develop a 3D functional joint model for medical applications in Orthopedics
  - Computer-aided diagnosis
  - Computer-aided surgery design





We believe that a 3D functional joint model is necessary for both Computer-aided Diagnosis and Surgery Design. It could change the current diagnosis and treatment processes, replacing the image analysis by a functional analysis of the joint. The goal of this thesis was then to develop such model with sufficient realism for medical applications.

Approach

Ideal approach: mime physical world ≡ low-level modeling

↓

Inviable: lack of computer power

↓

Solution: simplification ≡ high-level modeling

↓

Usual approach: idealized joint based on robotics, axial rotations

↓

Our approach: combine bones kinematics with soft tissues simulation ⇨

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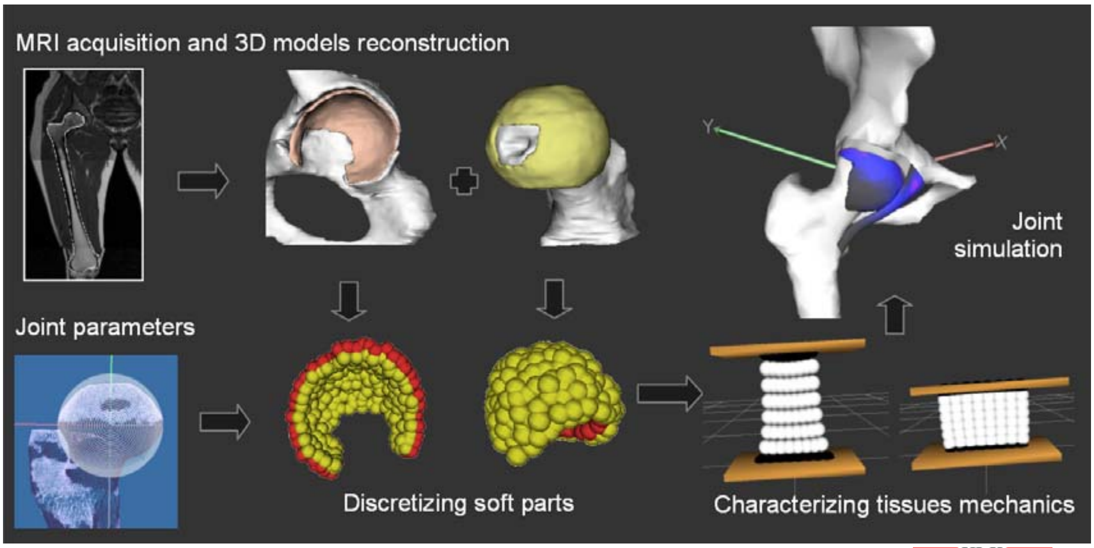
We approached the joint modeling problem in the following way. We realized that modeling the physical world as we know it in the atom level could be a good solution but the necessity of computer power would be prohibitive. Then we decided do model in a higher level. That's what most people do. But the existent approaches for joint modeling in CG are still too much based on the axial rotations of robotics and don't allow for medical applications. The compromise we propose is to combine the kinematics of the rigid parts with the physical simulation of the soft tissues.

Approach

Hybrid model = Kinematical skeleton + Deformable connective tissues



This schema illustrates the idea. Having static and dynamic MRI as input, both 3D shapes and kinematical information are extracted. The soft parts are then converted into a deformation model, material properties are configured for them, and everything is put together to be a 3D articulation that can be simulated.

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Outline

1. **Articulated motion**
2. Soft tissues deformation
3. Contact modeling

4. Our hybrid hip model
5. Medical applications
6. Conclusion

That was the introduction. The remaining of this presentation is divided like this:

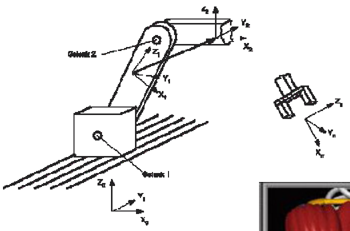
- 3 main parts to explain each of the 3 conceptual elements of our joint model

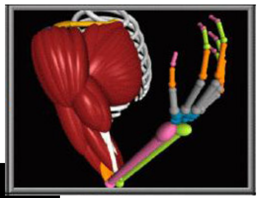
- Then 3 other parts to explain our case study, the derived applications, and to conclude.

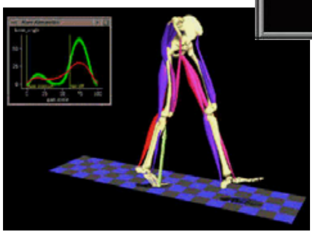
Let’s go into the first part: the articulated motion.

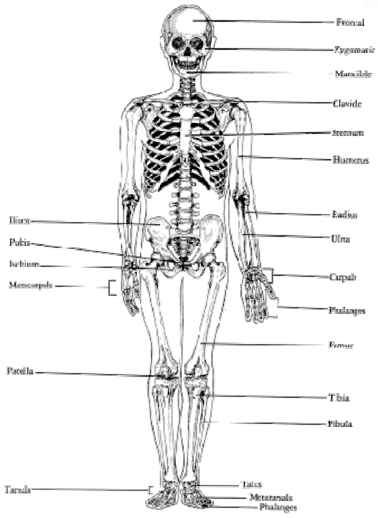
1. Articulated motion

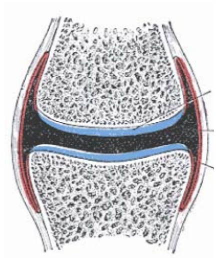
Simplified models **X** Complex anatomy

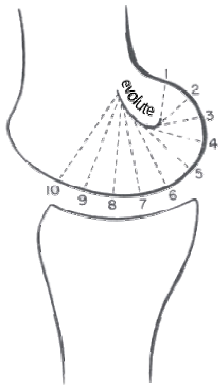












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Studying the existent models of joints in CG we understood that they are mostly based on former works in robotics. Consequently, they reflect idealized joints, represented by axial rotations. To achieve acceptable realism with those joints, animators must tune a large number of parameters.

On the other hand, the human anatomy is very complex. Joints are composed of surfaces sliding on each other, in a way that punctual rotation axes can only coarsely imitate. The axes in fact are subjected to momentary positions, as they slide while rotating. In addition, some motions are dependent of others, 2 motions of the same joint or even of 2 different joints. An example is the shoulder abduction, that can explore a greater range when the rotation puts the hand facing up.

To encapsulate the complexity, allowing simple motion specification at the same time as to achieve complex realistic motions, we proposed an anatomy based articulation model.

Joint types

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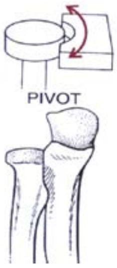
<tr>
<td>Hinges; pivots
Uniaxial – 1 DOF


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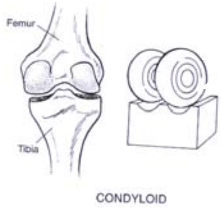
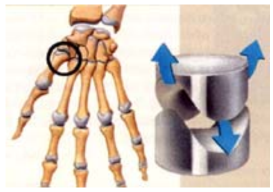
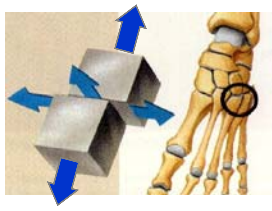
<td>Saddles; ellipsoids; condyloids
Biaxial – 2 DOF


</td>
</tr>
<tr>
<td>Enearthroses (Ball-and-socket)
Polyaxial – 3 DOF
</td>
<td>Planars
Planar – 6 DOF
</td>
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In addition to the internal functionality of a Joint, we also pay attention to the types of Joints in the human body.

We concluded that 4 types are necessary and enough to cover all possibilities of motion. They are:

Dofs

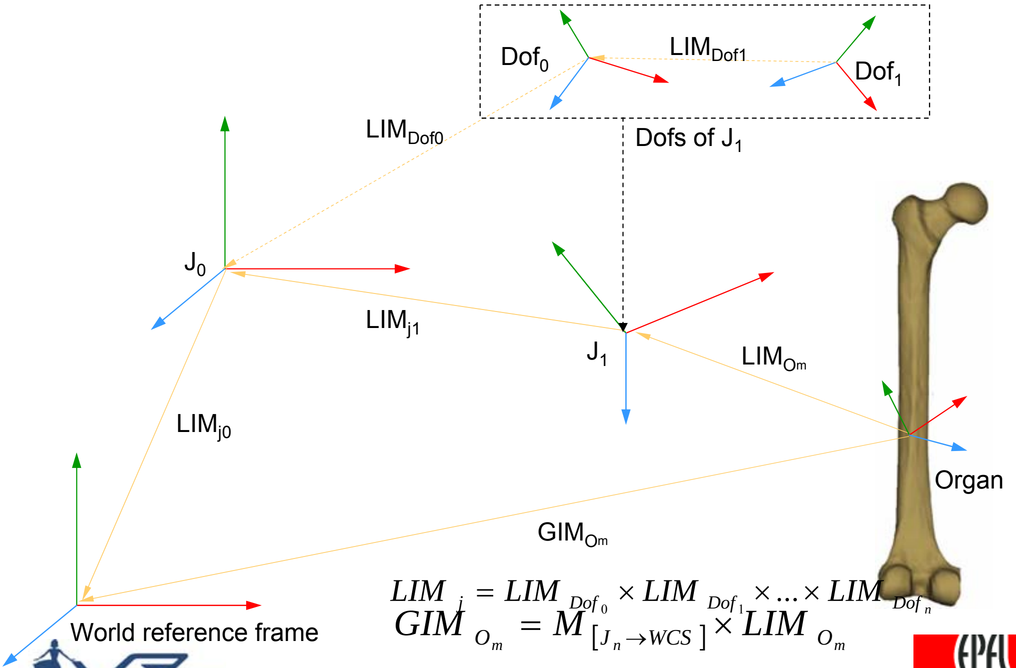
- DOF → *Degree of Freedom*
- *Dof* → An element of the model
  - Joints are a composition of Dofs
- A Dof defines:
  - The motion axis
  - The angular limits
  - The rest position
  - The current position
  - Etc.

One key structure in the model is what we call Dof.

The acronym DOF, for Degrees of Freedom, is widely known. However, when we say Dof, in small caps, we mean more than a motion capability. A Dof is an element of the conceptual model which defines many parameters of a motion possibility, like a motion axis, its angular limits, and standard positions.

1. Joints 10

Dofs as components of joints



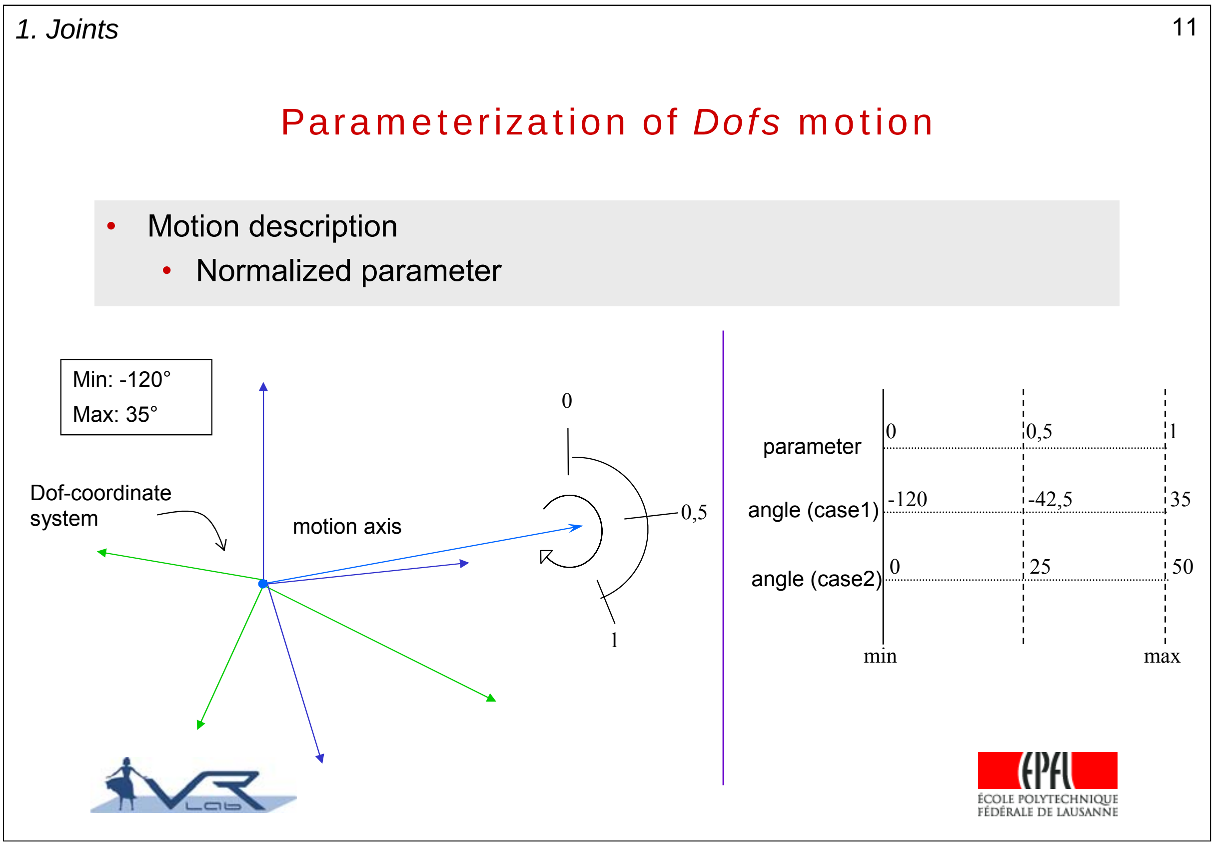
$$LIM_j = LIM_{Dof_0} \times LIM_{Dof_1} \times \ldots \times LIM_{Dof_n}$$

$$GIM_{O_m} = M_{[J_n \rightarrow WCS]} \times LIM_{O_m}$$

Conventionally, joints are grouped hierarchically to compose a body. In our model, the hierarchical relation is represented by transformation matrices we call LIM, standing for Local Instance Matrix. Every Joint has a set of one or more Dofs, and the LIM of a joint is, in fact, the combination of all its Dofs.

The Dofs 1 and 2 for joint J1 are combined to compose the LIM of J1, like in the equation.

Finally, when a shape, let's say a bone, is instantiated to a joint, the composition of the LIMs of all Joints in the branch can be used to calculate the object's GIM, that can be used later to visualize the object, for example.

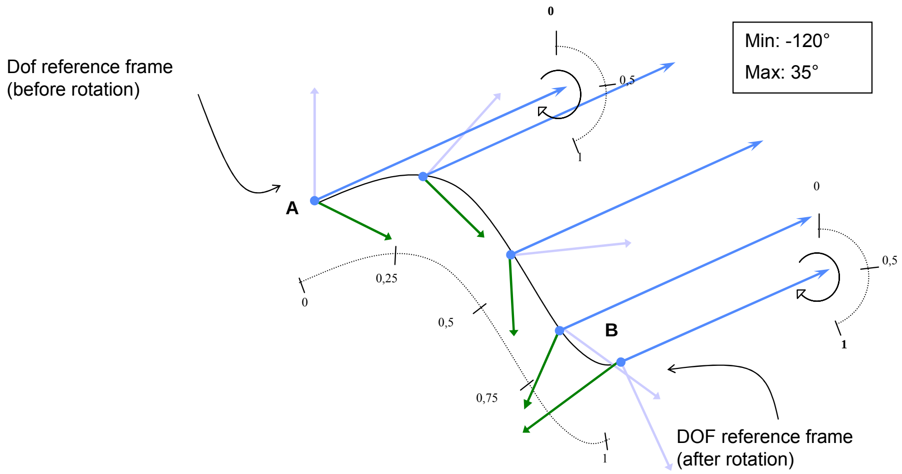


Let's now see how a Dof move. The rotation motion is done around an axis, and is constrained to be between the minimum and maximum angles using a normalized Dof parameter. In the example, when the parameter is 0 the angle is -120°, when it is 1 the angle is 35°. All intermediary values are obtained by interpolation, 0.5 is -42.5 for instance. Supposing a second example, when min is 0° and max is 50°, 0.5 would correspond to 25.

Other two axes complete the Dof-coordinate system, which moves like this. Up to now it looks much like any other articulated system in Graphics.

Parameterization of *Dofs* motion

- Sliding curve

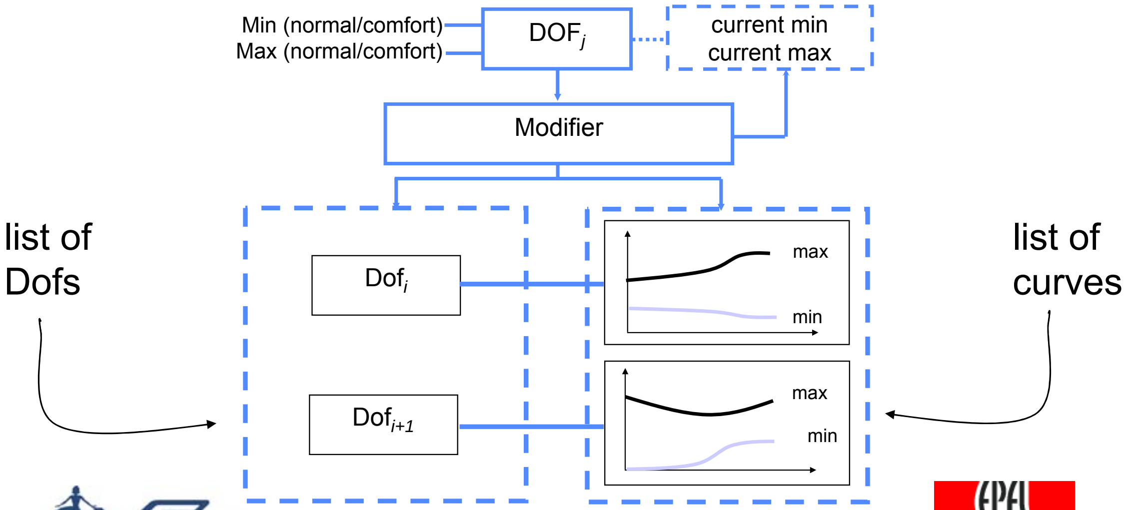


One novelty is that we use a parametric curve to represent the rotational axis displacement.

Besides the rotation movement, the Dof can also slide on the space while rotating. So, we define a sliding curve to the Dof and the same normalized value is used to determine the position of the Dof on the curve.

Dofs coupling

- Interferences between joints



After sliding curves, another anatomical particularity present in our model are the modifiers of dof's ranges of motion.

It is used to constrain the original limits according to the status of another Joint or Dof.

A list of Dofs with their corresponding relations is associated to the modifier.

Now, let's zoom in this part to see how curves are used to obtain current min/max values.

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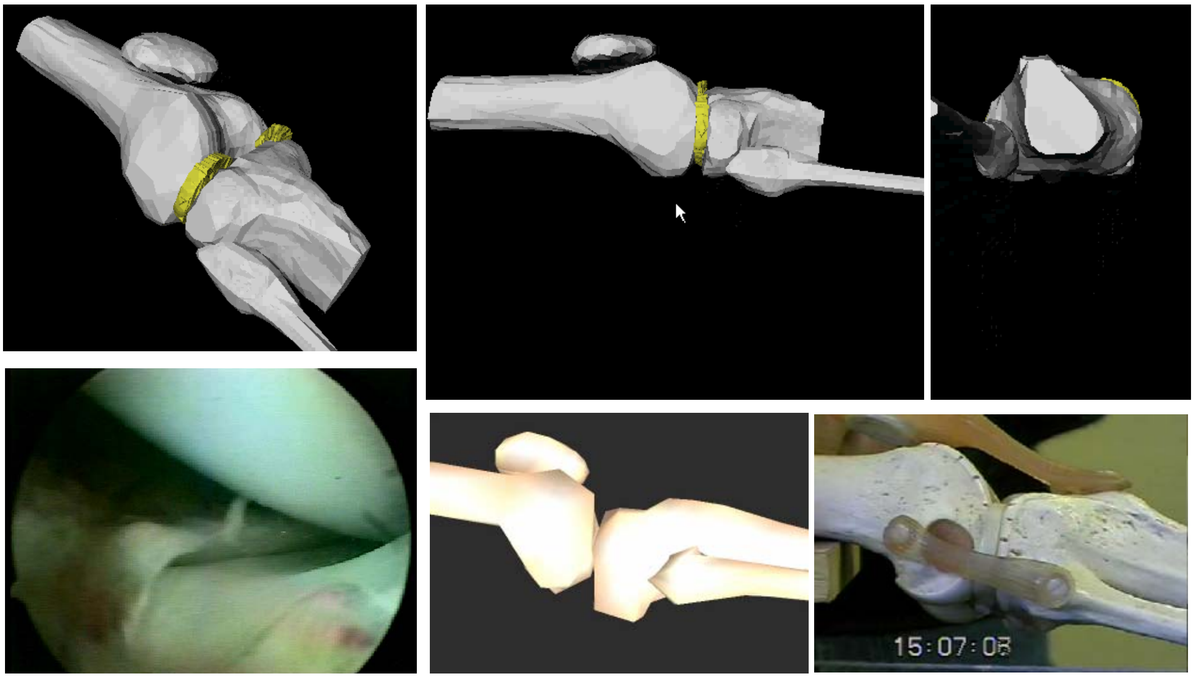
Knee case study



As a case study we built a model of the knee. On the top row you have 3 different views of a knee flexion cycle. Observe that the patella motion is coupled with the tibia's, that the flexion axis slides during motion; a terminal twist is also present for the knee lock at standing position but is difficult to observe.

The bottom row show sample motions we compared to ours for evaluation. From left to right: a real knee during arthroscopy, the simulation system SIMM by the Stanford group, and a plastic knee model handled by an orthopedist.

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2. **Soft tissues deformation**
3. Contact modeling

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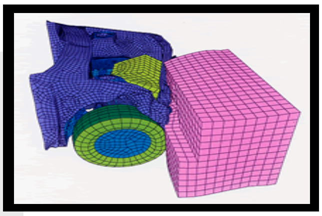
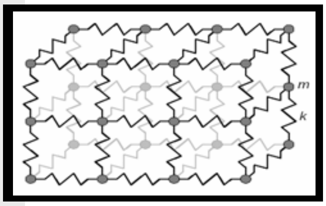
That was all about the articulated motion of the skeleton. Let's now pass to the second conceptual model of our biomechanical joint: the deformation of soft tissues.

2. Soft tissues deformation





- Soft tissues modeling
  - Finite Elements Method (FEM)
  - Mass-spring(-damper) systems (MSD)
- How to represent the behavior of complex tissues?
- Using a mass-spring like system:
  - Defining appropriate Hooke's constant *k* for every spring
  - Depends of material properties (Young's modulus) and springs structure (topology)

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A number of techniques has been applied to model deformations in CG. However, two of them are of greatest success: FEM and MSD.

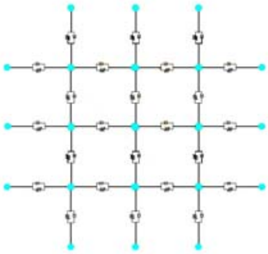
FEM model the space as a continuum, their advantages are that they give very precise results and are relatively easy to configure from real materials properties. The main drawbacks are the low performance and the very complex implementation. It may take hours or days to simulate depending on the model complexity and constraints. Optimizations exist exploring especially levels of details, that allow for interactive applications, but the model must be very simplified, softening the advantages of the method.

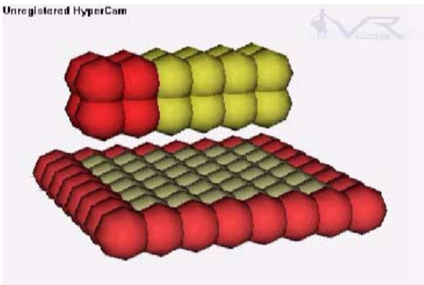
MSD discretize the space into mass-points conected by springs. It is simple to implement and usually performs in real-time, except for very stiff materials. The main drawback is the difficulty to characterize the tissues properties.

Finding appropriate values for all springs k's from a give material elasticity is extremelly complex, there is no definitive solution available in the literature. We address that problem in this work.

Basics of our deformation model

- Molecular model based on a generalized mass-spring model where mass points are spherical mass regions.





$$E=\{e_1,e_2,\cdots e_n\}\quad C=\{C_{e_1},C_{e_2},\cdots C_{e_n}\}\quad C_e=\{C_1,C_2,\cdots C_n\}$$

$$\vec{F}_e=\vec{F}_G+\vec{F}_L+\vec{F}_C+\vec{F}_{collision}$$

$$\vec{F}_G=m_e\vec{g}$$

$$\vec{F}_C=\vec{F}_b+\vec{F}_d+\vec{F}_f$$

$$\vec{F}_L=-\Pi r_e^{2}\rho\left|\vec{V}_e\right|^2\frac{\vec{V}_e}{\left|\vec{V}_e\right|}$$

$$\begin{cases}\vec{F}_b=\sum_{i=0}^{|C_e|}-k_c\left(\left|\vec{P}_e-\vec{P}_p\right|-l_c\right)\frac{\vec{P}_e-\vec{P}_p}{\left|\vec{P}_e-\vec{P}_p\right|}\\ \vec{F}_d=\sum_{i=0}^{C_e}-b_c\left(\vec{V}_\parallel\right)\\ \vec{F}_f=\sum_{i=0}^{C_e}-\left|\mu_e\vec{F}_N\right|\frac{\vec{V}\perp}{\left|\vec{V}\perp\right|}\end{cases}$$

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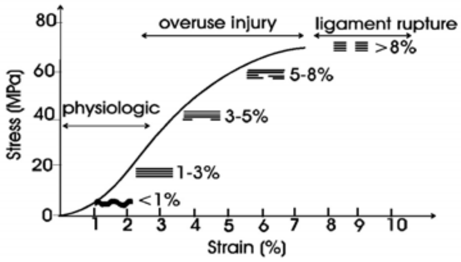
Our model of soft tissues is a kind of generalized mass-spring-damper system. It is composed by a set of mass-elements connected by damped springs. The forces involved are these. At any given moment, the force on an element is the composition of gravity, ambient viscous friction, connections forces and in some cases collision forces. The connections forces, in turn are composed by elastic forces, internal damping (heat), and sliding friction.

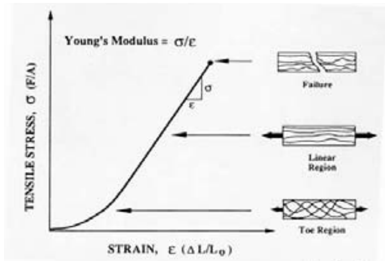
Up to here it looks like a conventional mass-spring. The spherical regions are new. They represent the volume of influence of one mass point, analogously of what happens with atoms and molecules. This volume is used first to determine the lattice connectivity, then to configure the material properties, to detect collisions, to simulate fluid flowing in and out of the tissue and so on.

Bio-tissues behavior

- **Ligament, cartilage, tendon, muscle...**
- Viscoelastic
- Anisotropic
- Non-linear
- Heterogeneous
- Sensitive to: age, gender, activity…

Young's modulus ≡ Elasticity modulus





Let's now check how the tissues we want to model behave, and remember some concepts.

Biological tissue can be very viscoelastic, anisotropic, non-linear and heterogeneous. Moreover, those features vary according to many factors like age, gender and activity.

One property of materials that's especially interesting here is the Elasticity modulus, or Young's modulus. It is derived from the stress strain relationship, represented by the two graphics here.

Usually, the graphic, has 4 different phases: one non-linear in the beginning, then a linear one, another non-linear in the end, and rupture.

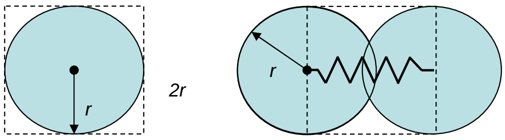
The slope of the linear part determines the Young's modulus. Most of the physiologic motion is in this part. So, the Young's modulus should be taken into consideration when characterizing the tissue.

Configuring springs: trivial approach

Input:
- Young's modulus of material (E)
- Spheres distribution
  - r = radius
  - l_0 = nominal distance between centers

Output: k = Hooke's constant

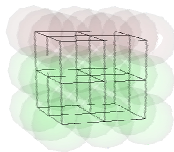
$$k = \frac{E(2r)^2}{l_0}$$



cross-sectional area = (2r)²

l_0

This approach works straight forward when applied to objects which springs have only right angles.



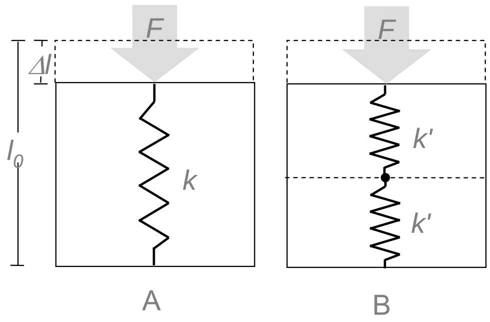
So, as I mentioned, one of our goals is to configure tissues elasticity from real materials. If we suppose an object represented by only one spring, there's a straight forward solution using this equation. In fact, we have E and a spring connecting two mass-spheres as input, and we want to calculate the k constant for it.

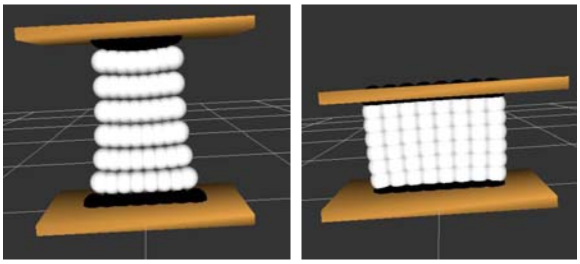
The principle say that any solid object can be seen as a spring of stiffness k, where k is the E of the material of which the object is made, multiplied by the cross-section area of the object and divided by its length.

Unfortunately this approach only works for objects in which all springs are aligned, and objects of this type are not even stable, they collapse because of topological ambiguities.

Iterative approach

$$E = \frac{F \cdot l_0}{\Delta l \cdot A}$$





- Pre-processing phase
  - Iteratively approximate value of spring constants

- Estimate effective *E* at each time step
  - A given force
  - Rest elongation
  - Current elongation variation
  - Cross sectional area
- Adapt *k* values
  - Minimize difference between effective and target *E*

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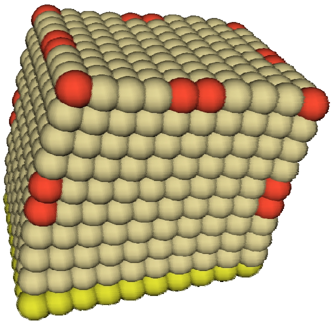
We finally propose a solution that works in the general case.

We initialize k's according to the straight forward solution seen before, and then iteratively modify the k's and assess the E to approximate the correct value.

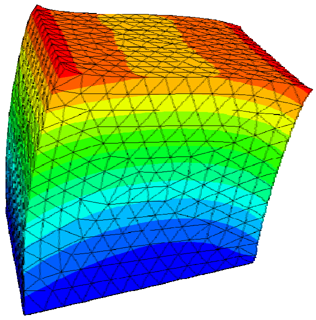
Given a known force applied onto a know area of an object which the lenght is also known, we used this equation to estimate E at each step. Then, if the E estimated is grater than we wish, all k's are reduced to make the object softer. If it is lesser, we increase the k's.

Evaluation

- Elastic behavior corresponding to elasticity of real materials



Result of a simulation with our model predicting behavior of a cube after force application



FEM static analysis by IMES - Center of Mechanics/ETHZ using the same physical parameters and applying the same forces

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To evaluate the outcome of this method, we compared our results with the results obtained with FEM analysis.

We used the same parameters as our colleagues in Zurich and obtained very similar results, not only qualitatively, but also quantitatively. We tracked key-points and compared their displacements. We show here only one example for illustration, but we tested a number of different setups, always obtaining good results.

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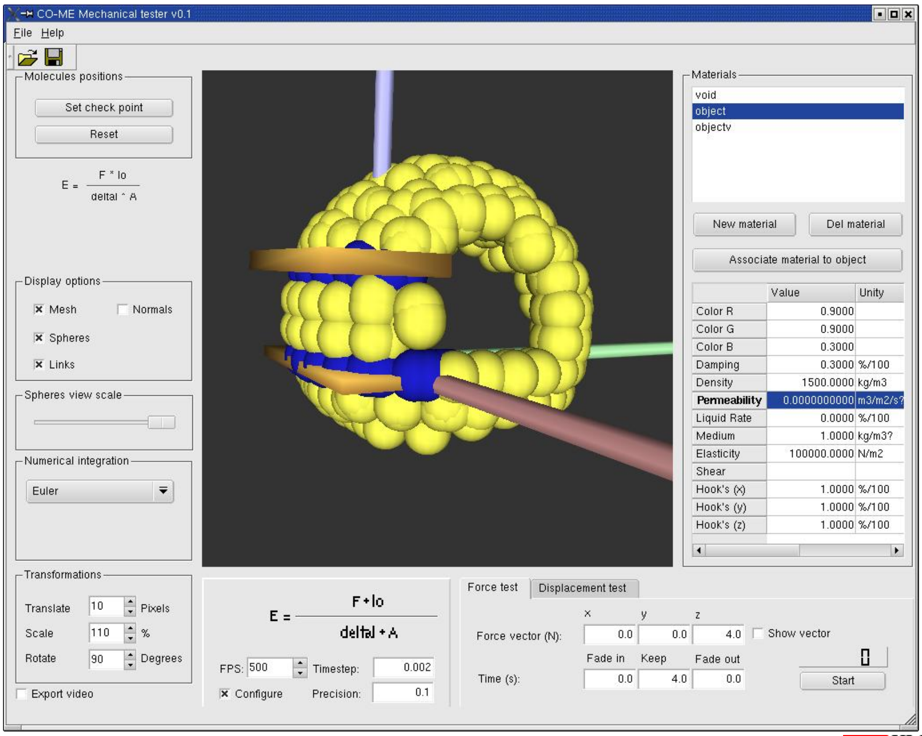
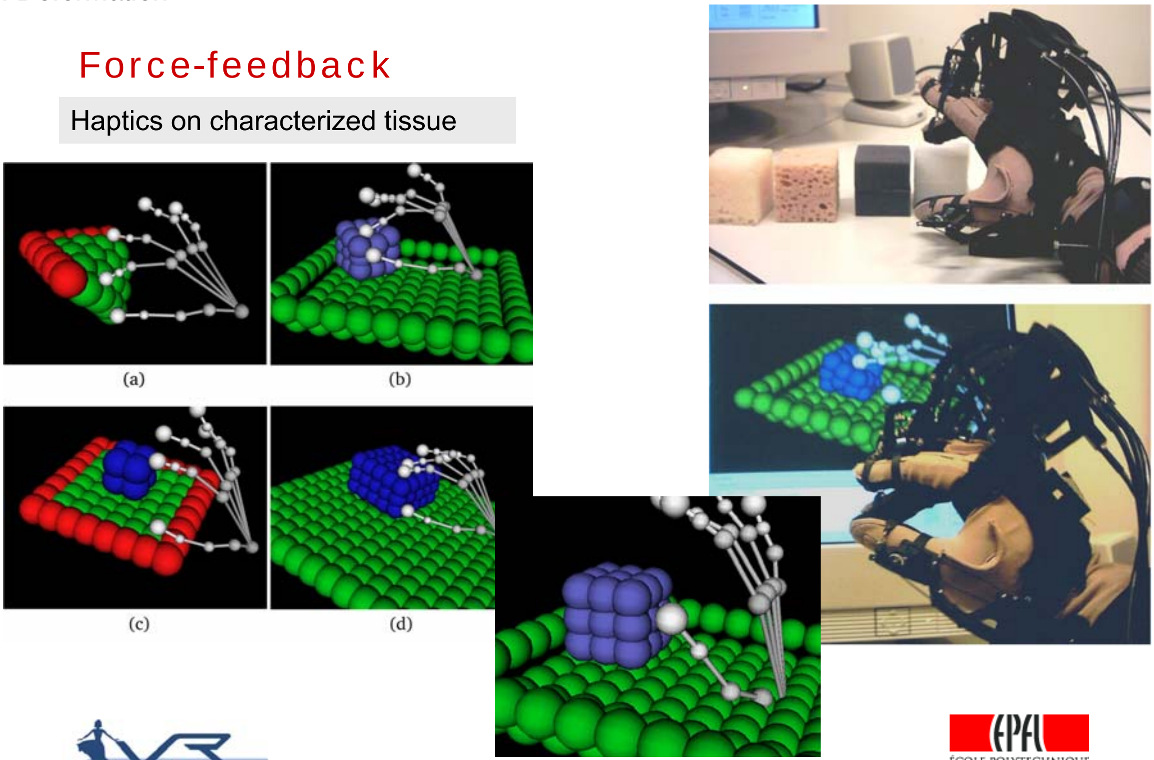


Concerning the joints parts, they are individually configured like this. Virtual clamps are used to fix the object and forces are applied.

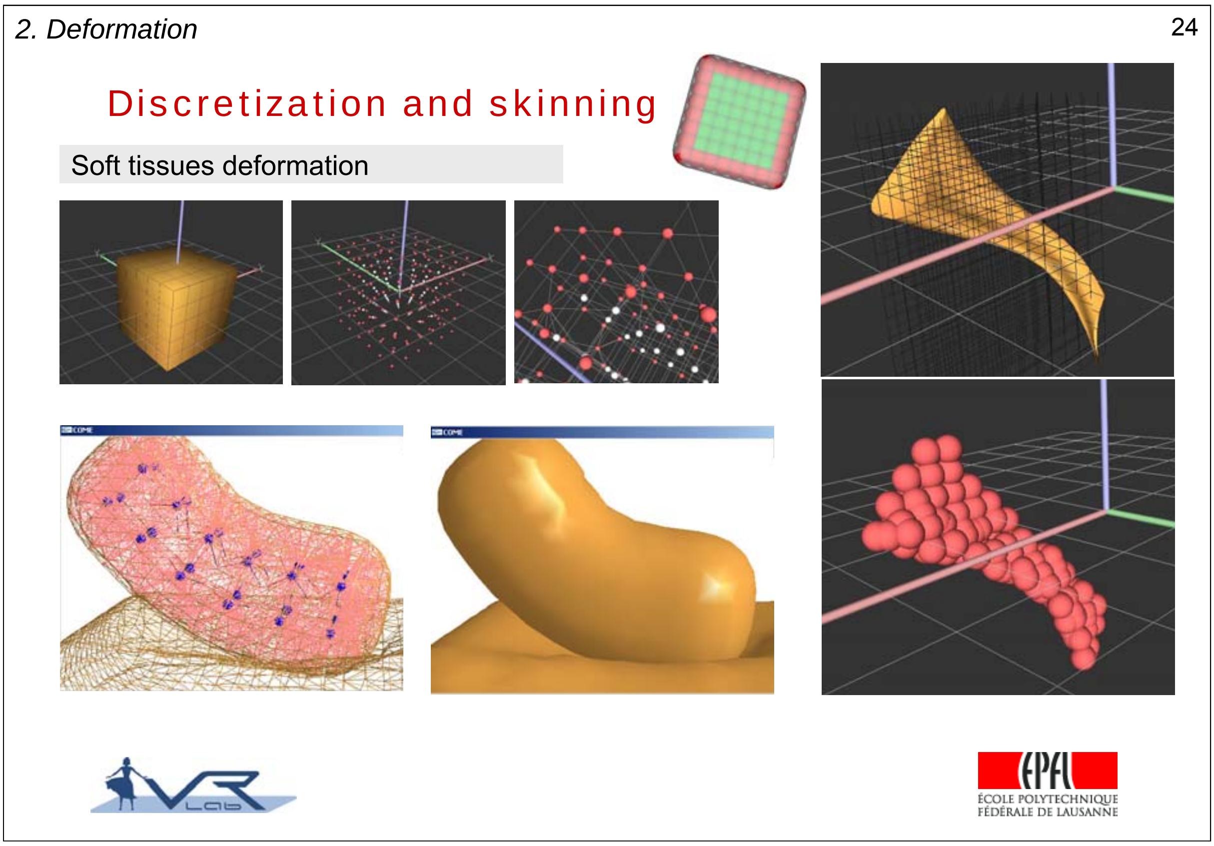
Force-feedback

Haptics on characterized tissue



We also evaluated the different elasticities with a force-feedback interface. We developed simple scenes to be able to run in real-time (it is required by the force-feedback device) and then asked some people to wear a cyber grasp haptic glove and to try to recognize different objects by their elasticities.

Most of them succeeded on ordering the objects from the softest to the stiffest, despite the limitations of the device.



Finally, for rendering and collision detection, the spheres surfaces are not smooth enough. The solution we proposed is to cover them by a 3D mesh like the skin covers the human body. Mesh vertices are anchored to the molecules underneath and follow their deformation.

In practice, we start from 3D meshes of the different organs and discretize their volume into sets of spheres.

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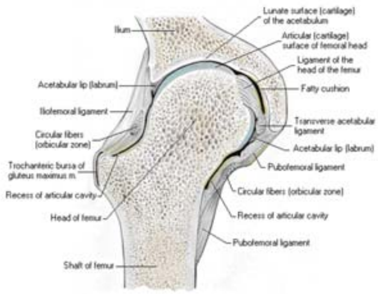
That was all about our soft tissues conceptual model. Let's go now into the contact management model, the third of the 3 main components of our 3D joint model.

3. Contact modeling

- Collision detection and avoidance
  - Required for realism in VE
- Inside a joint
  - Deformation
  - Permanent contact
- Inside a 3D joint
  - Equilibrium between opposite forces
  - Numerical instability

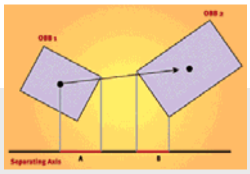


Collision detection is a classical requirement in CG to improve realism in Virtual Environments. Real objects do not penetrate and this constraint must be present in realistic virtual objects interaction.

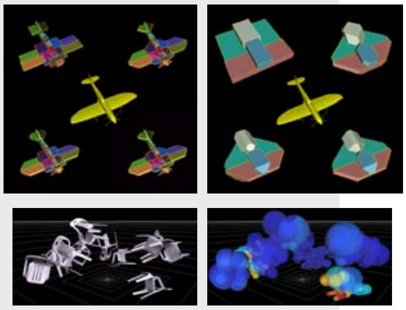
However, inside a joint the problem is more complex for two main reasons: objects can be deformable; and they are in quasi-permanent contact. (two very tricky situations to deal with)

In addition, depending on how we model a 3D joint, we have to deal with strong opposite forces which can cause numerical instability.

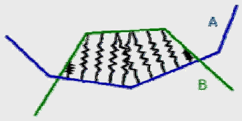
Previous works



- Intersection determination
  - Intersection test: separating axis
  - Distance calculation



- Collision detection: optimization methods
  - Bounding volume hierarchies
  - Spatial partitioning
  - Probabilistic methods
  - Distance fields
  - Image-based



- Collision response
  - Penalty method
  - Constraint-based
  - Impulse-based

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Reviewing previous works we detected three main problems involving contact management.

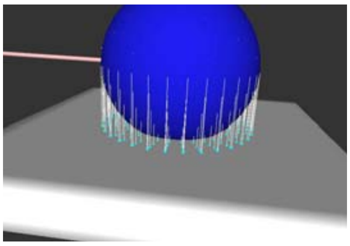
First is to identify if two static objects intersect. Many geometric and algebraic methods exist. For intersection determination, distance calculation methods have also been presented.

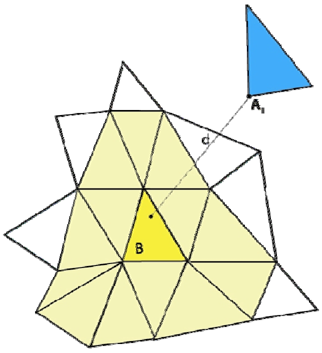
Second is to that quickly along time, when the objects can move or deform. This is the real goal of collision detection. And many techniques exist to avoid testing every little element for intersection. Different techniques are best adapted for a specific situation. Everything depends on the type and amount of objects.

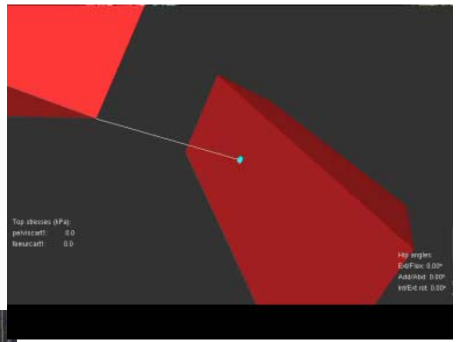
Third, once a penetration is detected, collision response methods are responsible to take them to a non-intersection position. Here also, the choice of the method will depend on the objects nature.

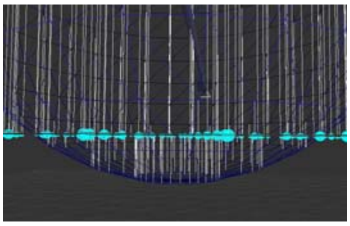
Collision detection: proximities

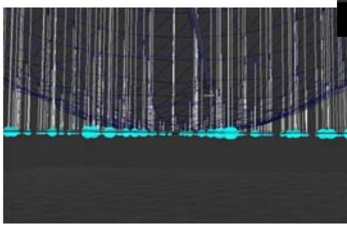
- Spatial coherence
- Fast update











VR Lab

EPFL ÉCOLE POLYTECHNIQUE FÉDÉRALE DE LAUSANNE

To deal with a small number of deformable objects in quasi-permanent contact, we proposed two methods.

The first, after an exhaustive initialization procedure, keeps updated for each pair of objects, a structure containing for each point of one object the closest triangle on the other, as well as the closest point on the triangle.

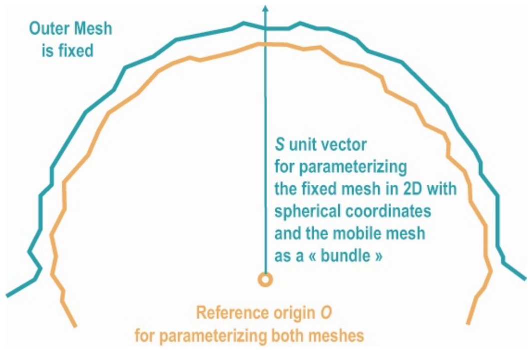
The advantage is that, relying on spatial coherence, we quickly update the proximities structure. At each step, only the neighboring triangles to the closest must be tested.

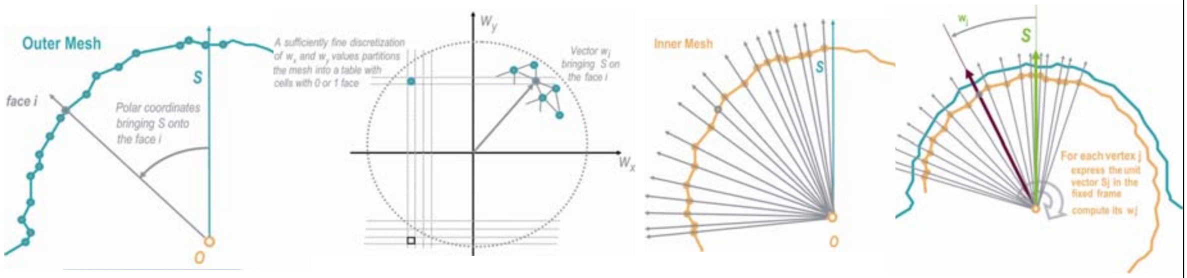
Collision detection: spherical sliding

- Spatial coherence
- No update

- Vector through the face center (fixed)
- Polar coordinates define hash key
- Store face into matrix

- Vector through vertex (mobile)
- Polar coordinates define hash key
- Query on table and calculate distance





The second method we propose, also exploits spatial coherence, but it requires a particular topological relationship. The two surfaces to be tested must be placed facing each other in a more or less spherical topology. Despite this constraint, the method adapts well for collisions inside a joint, and is very efficient as no update is required.

Let’s see how it works:

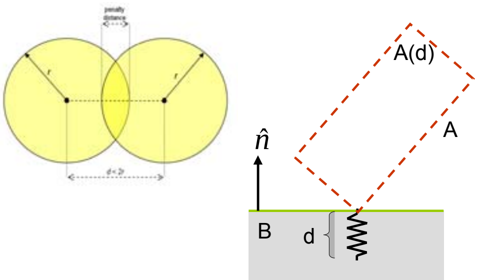
One mesh is considered to be fixed and the other mobile, undergoing a rigid transformation. The method is based on the geometrical hashing of the surfaces.

At initialization we build the hash table. Vectors are built going from the center to every triangle on the fixed mesh. Spherical coordinates of the vectors are used to calculate hash indices, and then, the triangle is associated with the cell of the table corresponding to the indices.

For collision detection, similar vectors are build passing through vertices of the mobile mesh. Indices are calculated from their spherical coordinates and the triangle found on the respective cell of the table is used for distance calculation with the vertex. The distance determines if yes or no a collision exists, and gives information for collision response.

Collision response

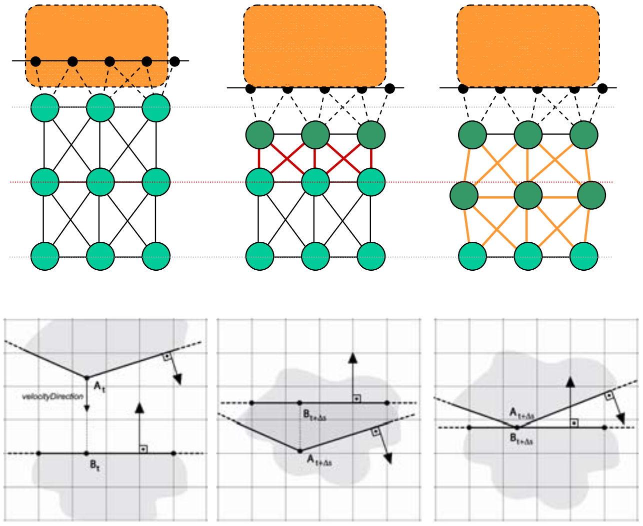
- Spheres: penalty method



$$\vec{f}_p = \begin{cases} -kd\hat{n} & if\ d > 0 \\ 0 & if\ d \le 0 \end{cases}$$

k penalty coefficient
d penetration distance
n force direction

- Mesh: position and velocity corrections



I said collision response, so, how we did it?

First approach was to use the spheres themselves for collision detection and response. A penalty method was applied using as penalty distance the spheres penetration (difference between the distance of the two centers and the sum of the two radii), and the average of k’s of the springs associated to each sphere were used as penalty coefficient. Thus an instantaneous spring placed between the spheres avoid them to penetrate.

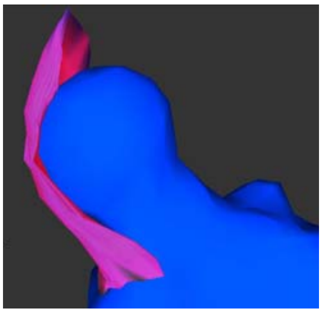
When we put a skin around the objects, the collisions must be taken using this skin. In that case, the penalty parameters have shown to be difficult to obtain in a general case. That’s why we searched for another solution.

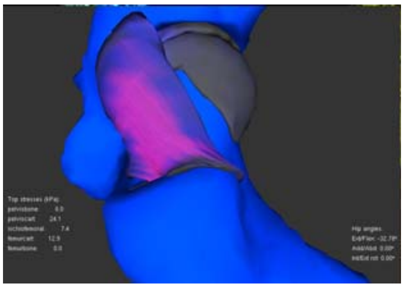
The second approach is then a geometric one. Suppose the situation of penetration here. If the object in the top is rigid, we must move the vertices of the other such that they lie on the surface of the first. But, because of our skinning system, it means moving the underlying molecules instead. Doing so, we stress their springs. The stressed springs try to release energy and that way, the geometric modification becomes physical, propagating the forces through the whole object till the equilibrium is reached.

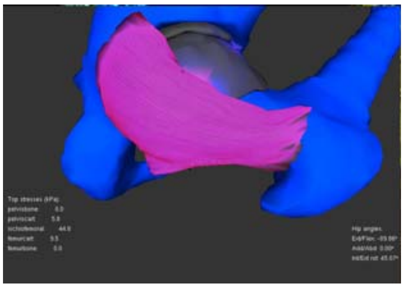
In practice, the method has shown to be a bit unstable for large and abrupt penetrations. To solve that problem, we also act on the velocities. When a penetration in time t does not exist, we calculate vertices positions in time t+1 (just use 2t when calculating positions from current velocity). If a vertex penetrates at that time, we tune its velocity in time t such that in time t+1 its position is on the surface of the other object.

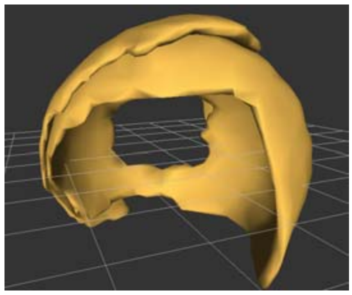
Results within a joint

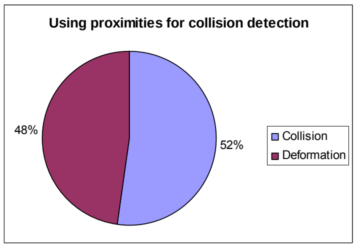
- Hip bones, cartilages and ligaments in contact.
- Collision strain causes stress on tissues.

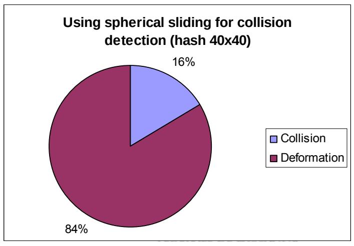












So, let’s see now some results.

Here you have some typical situations of a hip joint with bones, cartilages and ligaments in contact. Even with large stressed regions the penetration is avoided.

In terms of performance, we tested our two methods on the case of the two hip cartilages in multi-axial motion.

The proximities approach takes 52% of the total simulation time, more than the deformation itself.

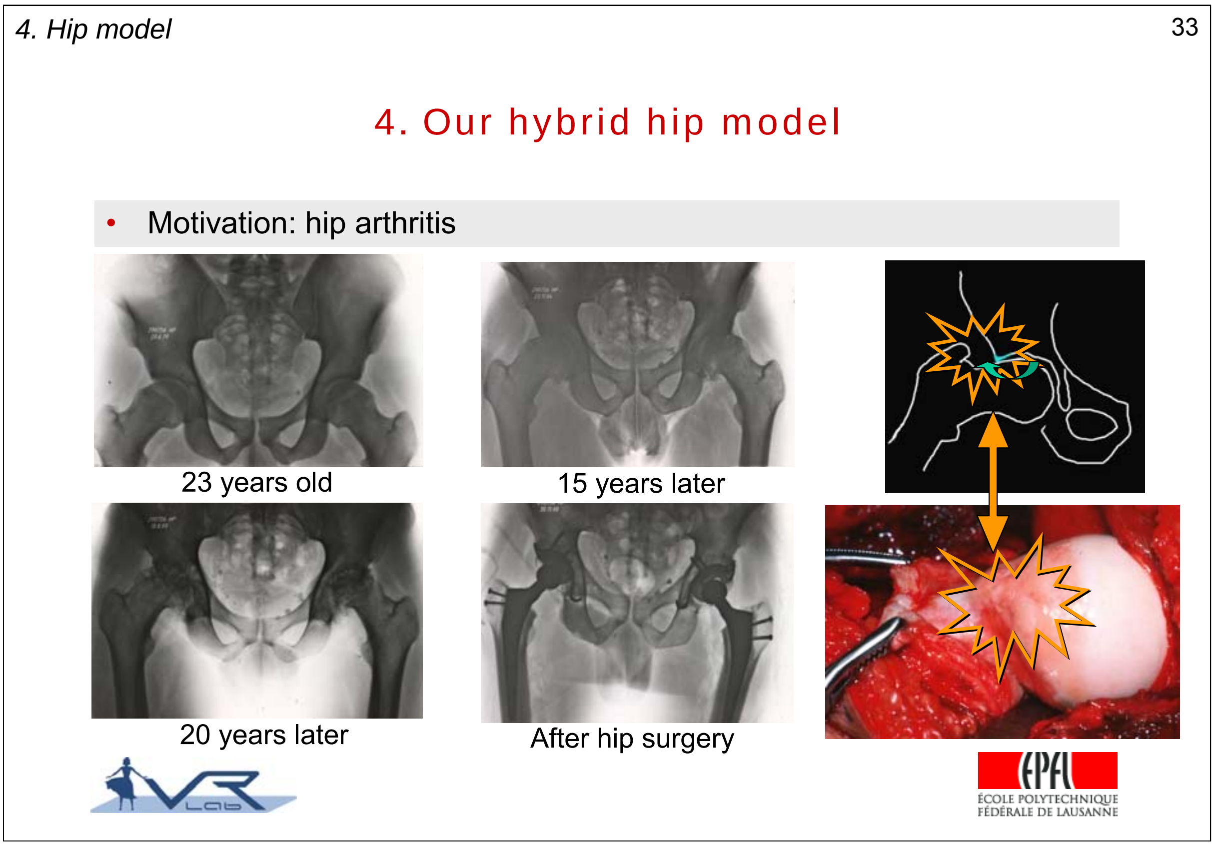
The spherical sliding, in turn, takes only 16% for a hash table of 1600 elements, being much faster.

Outline

1. Articulated motion
2. Soft tissues deformation
3. Contact modeling

4. **Our hybrid hip model**
5. Medical applications
6. Conclusion

Ok, the conceptual part is done. Let's see now a case study. First, I will show how we built a joint model for the analysis.



First question is: why the hip?

We were motivated by a degenerative pathology called hip arthritis. This pathology affects every human being after a certain age and in certain cases, like extreme sportive activity, can affect young people.

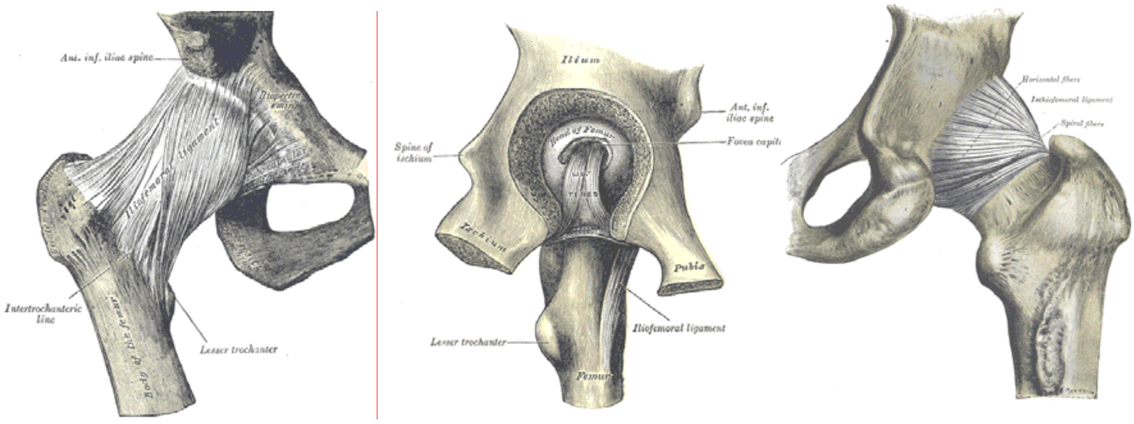
What happens is that the femoral neck bounces continuously on the pelvis making the acetabular labrum, a soft structure, become rigid. This hard labrum scratch the cartilage until it completely wears out. The usual treatment is the replacement of the hip by a prosthesis.

With the extension of the life expectance these last decades, the usual treatment became ineffective. The prosthesis lifetime is shorter than the patient life expectance.

Early detection of the problem can avoid the hip replacement and we believe a functional joint model can help in early diagnosis.

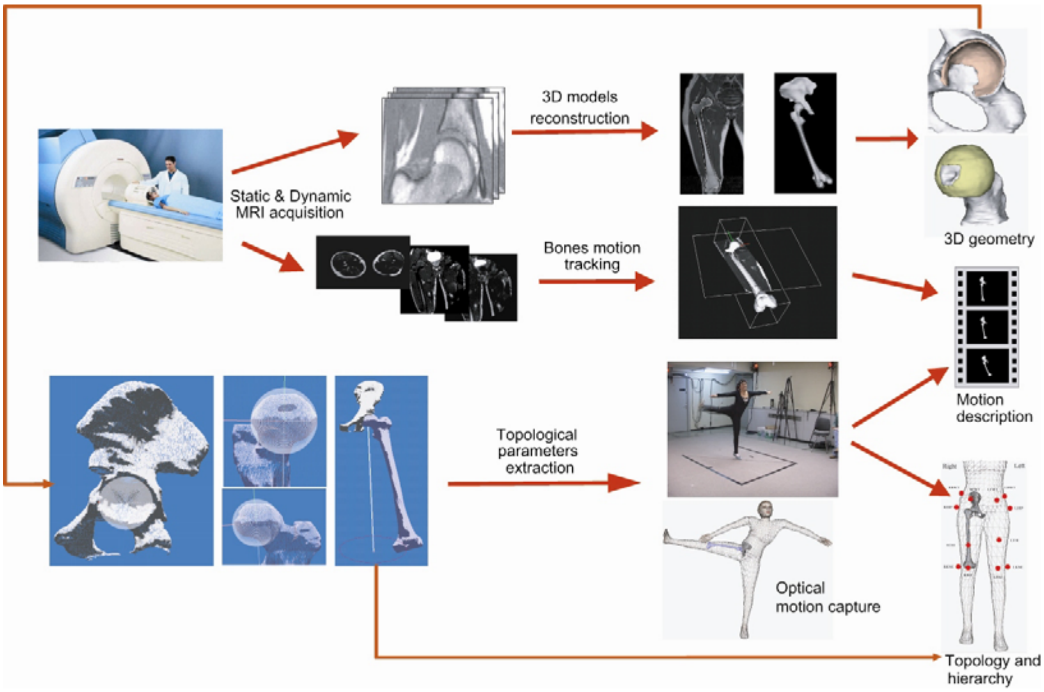
The human hip joint

- Ball-and-socket joint: 3 DOF



The hip is a ball-ahd-socket joint between the pelvic and femoral bones. Two cartilage caps provide excellent joint lubrication, and a set of 7 ligaments compose its tight articular capsule.

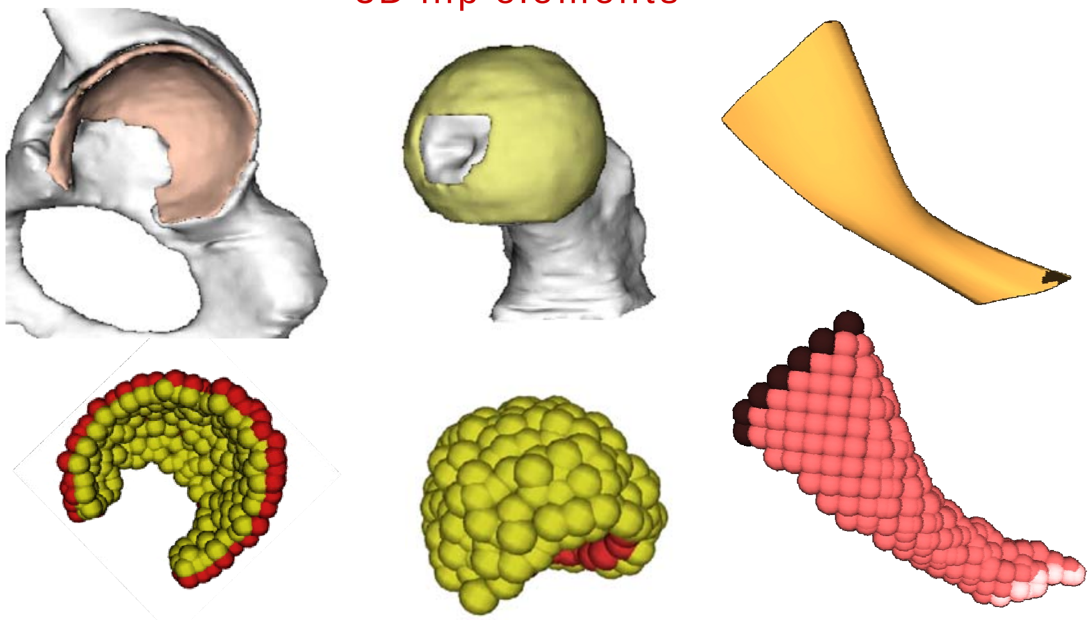
Data model



The conceptual model must be fed with data. Our data come from MRI analysis and motion capture, like in this schema.

After acquisition 3D models are reconstructed and motion is tracked. Shapes and motion are analyzed to extract joint information. Material properties come from the literature. At this point we have all the data we need to configure our joint.

3D hip elements



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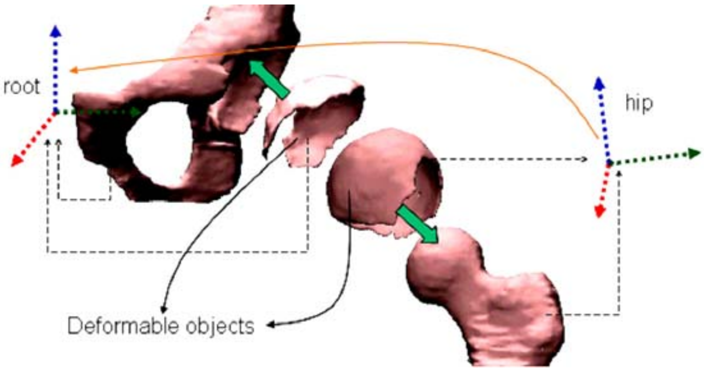
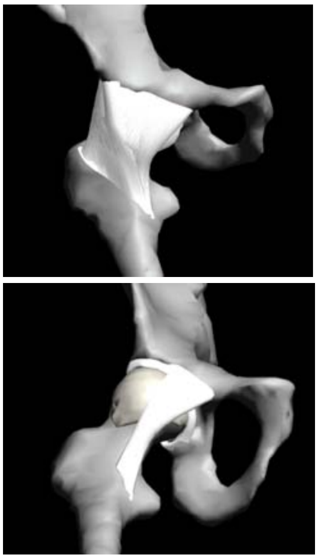
We modeled bones as rigid structures governed by uor articulation model presented before. And cartilages and ligaments as soft objects using our deformation model also presented before.

3D hip topology





The many elements are put together to compose a 3D hip, more or less like this.

Here the articulated structure, and here the complete hip with the complete capsule and only with the pubofemoral ligament.

Outline

Now we have our hip model. Let's see how we used it to analyze medical problems.

5. Medical applications

- Case study on the 3D hip joint
- Computer-aided:
  - Joint analysis
  - Medical diagnosis
  - Surgery planning
- Assess the hybrid joint model composed by:
  - The anatomically based skeleton kinematics
  - The biomechanically characterized soft tissues
  - The quasi-permanent contact model

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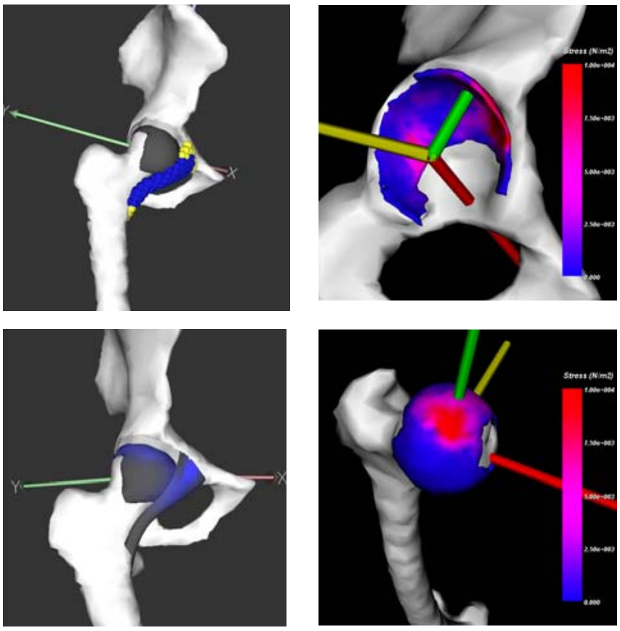
I will show 4 applications aiming at analyzing the joint, aid on medical diagnosis, and allow for surgery planning.

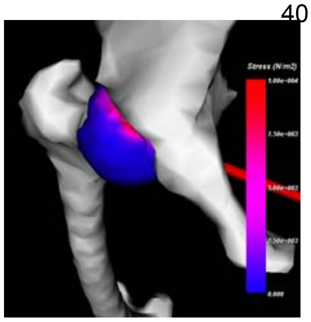
More than answer medical questions, our goal with these applications is to assess the applicability of our hybrid joint model in medical matters.

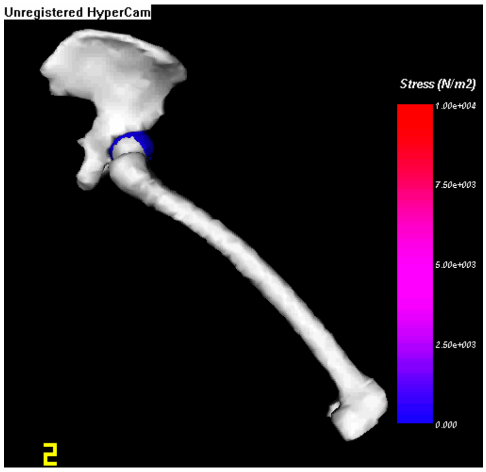
Application #1

Stress distribution assessment



$$\sigma = \frac{F}{A}$$





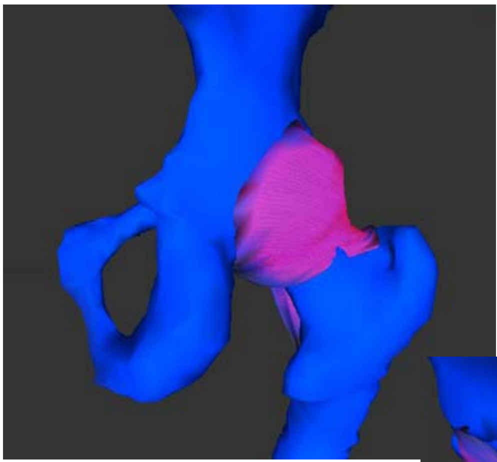
The first application allows analyzing the joint to evaluate its congruity, or efficiency on moving.

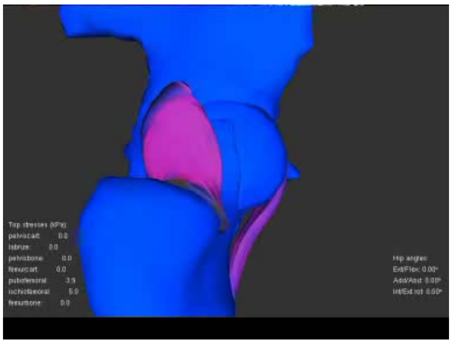
We used this equation to calculate the stress all over the deformable volumes, and used color mapping to visualize stress distribution in the form of a color scale rendered on the surfaces.

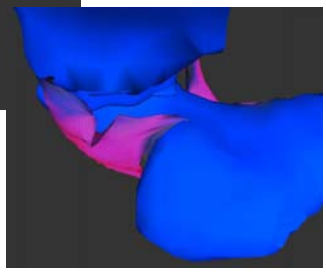
Hidden parts can be inspected interactively. Differently of the real joint, here all the operations are non invasive.

Application #2

ROM estimation from ligaments







- Ischiofemoral ligament
- Total internal rotation at 90° of flexion
- Failure stress

Many elements define the range of motions a joint can achieve. Bones impingement, fat, strong connective tissues… For the internal rotation of the hip at 90° of flexion, it is the tension on the ischiofemoral ligament.

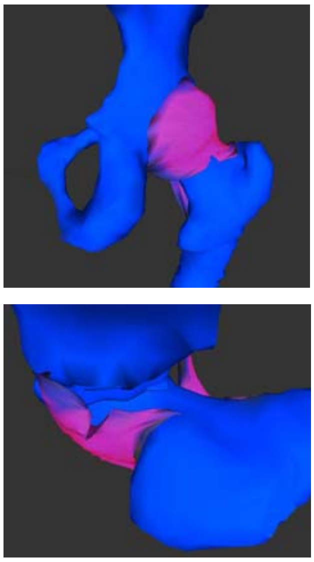
So we tried to estimate with our hip model what's that range. We found the E of that ligament and also its failure stress, the maximum stress it can bare before rupture, in the literature. Then, we setup the model and simulate that motion. At each step, we measured the maximum stress on the tissue. We stop simulating when that stress reached the failure stress. The position where we stopped defines the maximum angle the joint accepts.

Application #3

Elasticity estimation from ROM

Goniometers

$$\sigma = \frac{F}{A} \qquad E = \frac{\sigma \cdot l_0}{\Delta l}$$



Increase stiffness to reach failure stress

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In this application we did the inverse approach. Knowing that the E can vary a lot from one person to another and it is difficult to measure, we try here to estimate that E.

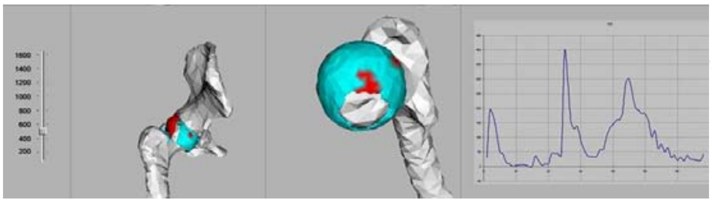
We used range of motion data from clinical measurement as input. With a goniometer, the clinician measured the range of motion for the same situation of the previous application. Then, we configured the E of our virtual ischiofemoral with a very soft material and drove the joint to the extreme posture. At that posture, we continued the simulation, but we started to increase the stiffness of the tissue at every step. Also at every step, we compared the maximum stress with the failure stress. When we reach it we stop to increase the stiffness, and from the current stress and length we estimate the E of the tissues using the equation.

Application #4

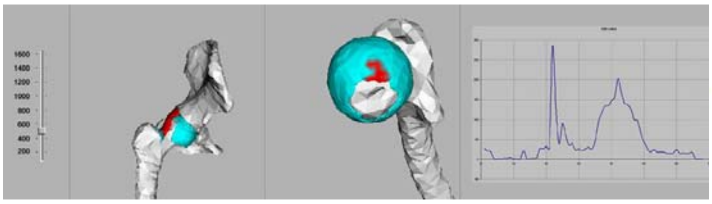
Pre- and post-operative comparison

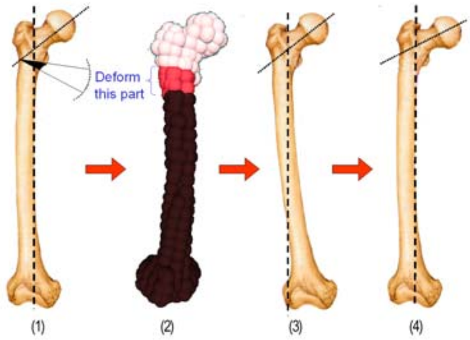
- Femur-head reoriented by osteotomy of Imhäuser
- Difference of stress distribution before and after the operation

Before



After





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For the last application, we analyzed the joint congruity (stress distribution) before and after a virtual surgery.

We first simulated a multi-axial movement. The chart shows the maximum stress along time.

Then we deformed the femur close to the neck to simulate an osteotomy of Imhäuser. This obliged the virtual patient to change their rest angles to keep the good angle for the knee.

Finally, we applied the same motion as previously and again plotted the chart of max stress along time. We observe that it is different now, the profile is a bit shifted on time, what was expected, and the max stress has been in average decreased.

So, this is an example on how we can evaluate a surgical procedure before its effective application on a patient.

Outline

1. Articulated motion
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4. Our hybrid hip model
5. Medical applications
6. **Conclusion**

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Now we approach the end and go into the conclusions.

6. Conclusion

- Goal:
  - To develop a 3D functional joint model allowing for computer-aided diagnosis and computer-aided surgery design
- Choices:
  - Hybrid joint
  - Discrete physically based model for deformation
  - Anatomy-based kinematical model
  - Permanent contact treatment

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First of all, let's recall the goal of this work. It was to develop a functional joint model for medical applications on Graphics.

Our scientific choices were to create a hybrid joint model, to use a discrete model for physically based deformation, to consider the skeleton as rigid and take anatomical features into account when moving it, and develop specific methods for contact management.

Summary of contributions

- Anatomy-based articulation model
- Biomechanics-based deformation model
- Quasi-permanent contact model
- Biomechanics-based articulation model
- Force-feedback from deformable objects
- Medical applications

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Our main contributions were:

-To include anatomical features in the articulated skeleton at the same time that the motion control remains simple.

-To simulate soft tissues with a discrete model at the same time that biomechanical properties are taken into account.

-To provide an optimized collision detection and response method for the type of contact we have inside a joint.

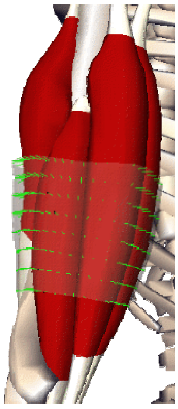
-To group the conceptual models above into a model for human joints based on biomechanics.

-To test the suitability of a force-feedback device with deformable objects.

-To test applications allowing to make medical conclusions using our joint model.

Future perspectives

- Clinical validation
- New image acquisition hardware and techniques
- Material properties: shear modulus
- Performance: deformation on GPU (parallelism)
- Biomechanical joint enhanced virtual characters



Muscles: A. Aubel Ph.D. Thesis, VRlab, 2002.

Coming backwards and seeing the work done from a wider point o view, we see that there’s a lot of work still to be done before we have functional models of the body being used on day-by-day clinical procedures.

Among them we can mention:

- a full clinical validation

- improved technology for image acquisition and measurement of material properties

- specifically in our model, shear modulus should be considered to complete the material characterization

- performance can be improved exploiting the parallelism of the new programmable graphics cards

- and last, but not least, apply such biomechanical joint on virtual characters animation to see how it can improve the visual realism.

End

Acknowledgements

- This work was supported by the Swiss National Science Foundation in the frame of NCCR/CO-ME.