POLITECNICO DI TORINO

Department of Mechanical and Aerospace Engineering

Master’s degree in Biomedical Engineering

Finite Element Analysis of femoral stem length in Endo-Model hinged knee prosthesis

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Abstract

Keywords: Revision Total Knee Arthroplasty, biomechanics, artificial implants, Finite Element Analysis

Introduction: Revision Total Knee Arthroplasty (TKA) substitutes a failed primary implant with a new prosthesis. To achieve joint stability such device usually has longer and thicker stems that will be fitted deeper inside the bone. Such procedure is more complicated than a primary implant as it requires extensive preoperative planning and specialized implants. Among the different possibilities of revision implant, when the collateral ligaments are compromised, a possible solution is the Endo-Model Hinged Knee Prosthesis: it consists in a femoral and tibial component, which are rigidly constrained, at the level of the flexion axis with a hinged constraint. Due to the presence of such joint, to guarantee a good transfer of the stress from the implant to the bone, femoral and tibial stems are usually present. According to patient anatomy and surgeon requirement, the stems can present different lengths, and can be placed in the bone cemented or uncemented. Even if different options are present, a proper guideline is still missing to help the surgeon in the right selection of the stem, from a biomechanical point of view, for a specific situation. Therefore, the goals of this study were to investigate, with a Finite Element Analysis, how the Endo-Model works during a squatting and the change in bone stress induced, in a hinged revision TKA, when short and long, both cemented and uncemented, femoral stems are used.

Materials and Methods: The model is based on a previously validated finite element model. The 3D geometries of the implant were recreated in Solid-Works starting from the industrial design. The bone geometries were obtained by 3D reconstruction from CT images.
Dynamic and static simulations were done to analyze a flexion-extension movement from 0 to 90° of the model (1) without the femoral stem, (2) within a cemented femoral stem (with a stem length of 50 and 120 *mm*), (3) within an uncemented femoral stem (with a stem length of 50 and 120 *mm*).
The models were analyzed in full extension with an axial force of 700N and at 90° of flexion, with an applied force of 2144.12N. These forces were obtained from previously experimental and numerical study.
In all the models the proximal part of the femur was constrained, and the forces were applied on the distal tibia.
The materials were chosen according to previously literature studies.

Results and Conclusion: Both in 0° and in 90° model, the presence of the cement helps to reduce the femoral bone stress in every region along the stem.
The 50*mm* stem configuration is more affected by the presence of cement, since the area of interaction between the femur and the stem is smaller than the 120*mm* stem configuration.
In the flexion configuration the lowest stresses appear when the cemented stem with a length of 50*mm* is used.
This study can be beneficial to help the surgeon within optimal implant choice.

1 Introduction

1.1 Knee Joint

The knee joint (**Fig.1**), together with the hip and ankle joints, supports the weight of the body during a series of activities, such as standing, walking and running. However, the knee joint must provide this support despite (1) exhibiting the greatest amplitude of movement (up to 160 degrees) compared to all the other joints of the lower limb, (2) lacking the large muscle mass that supports and strengthens the hip joint and (3) lacking the strong ligaments that support the ankle joint [1].
Even though the knee functions like a hinge joint is more complex than that of the elbow. The round femoral condyles flow on the upper surface of the tibia, causing a continuous change in the contact points. The knee is less stable than other hinge joints and, in addiction to flexion and extension movements, a certain degree of rotation is allowed. From a structural point of view, the knee is composed of two joints included within a complex joint capsule: one between the tibia and the femur (tibio-femoral articulation) and the other between the patella and the other patellar surface on the femur (patellofemoral joint).

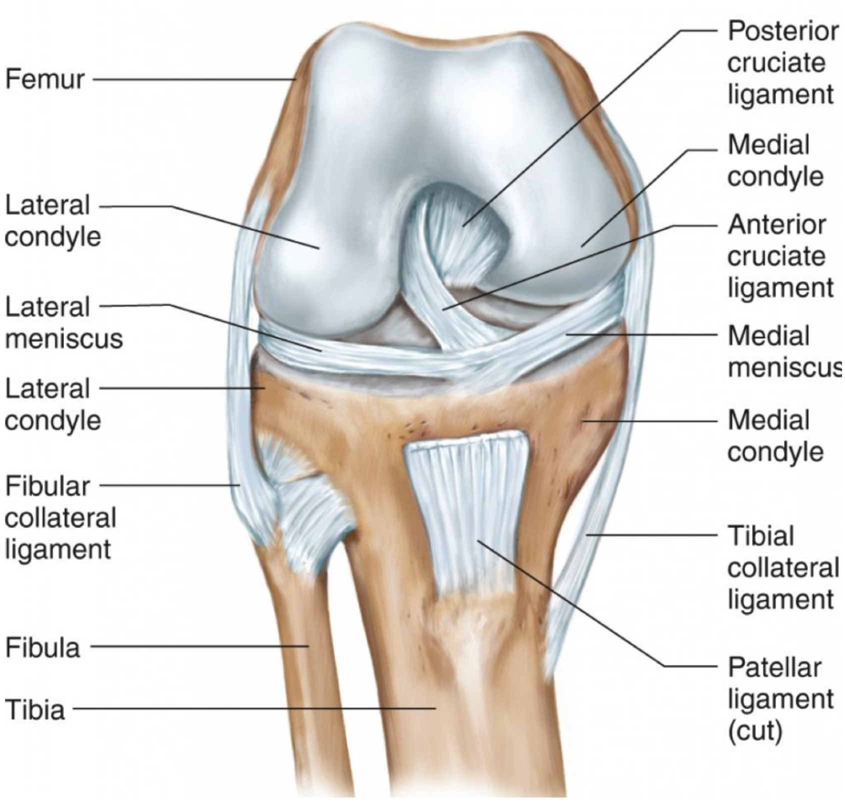


Fig.1 Knee Joint [2]

1.1.1 Bones

Fig.2 Bones of the Knee Joint [3]

Patella

The patella is a sesamoid bone that strengthens the quadriceps tendons and protects the anterior surface of the knee joint.

Femur

The femur is the longest bone in the body. Distally, it is articulated with the tibia in the knee joint. The body is massive and resistant, but has a curvature along the longitudinal axis, which facilitates weight tolerance.
Medial and lateral condyles, smooth and round, are distal to the epicondyles.
The smooth articular surfaces merge to determine the patellar surface, on to which the patella slips. On the back surface the two condyles are separated by a deep intercondylar fossa.

Tibia

The tibia is a voluminous medial leg bone. Medial and lateral condyles of the femur articulate with medial and lateral condyles of the proximal end of the tibia. The lateral condyle is more prominent and has a facet for the articulation with the fibula at the upper tibiofibular joint.
The anterior surface of the tibia presents the tibial tuberosity, which inserts the robust patellar ligament.

Fibula

The fibula is a thin bone parallel to the lateral margin of the tibia. The head of the fibula articulates with the lateral margin of the tibia, on the postero-inferior surface of the lateral condyle. The medial margin of the body is joined to the tibia by the interosseous membrane of the leg, extended between the interosseous margins of the two bones [1]. The fibula does not participate in the knee joint and does not transmit the weight to the ankle and to the foot, but it is nevertheless an important site for muscle insertion.

1.2 Knee diseases

The knee is one of the most complex and solicitated joint in the human body and it is for this reason that it can easily lead to complications.
The most common causes of knee replacement surgery are: osteoarthritis, rheumatoid arthritis and hemophilia.
The most important disease for the knee joint is the osteoarthritis, also known as "degenerative joint disease", that is a progressive form of arthritis characterized by the progressive damage to the cartilage in knee joint. In particular, the repair mechanisms are overwhelmed, and the thickness of the cartilage decreases until it disappears, causing non-negligible damages in the joint. Due to the absence of cartilage, bone excrescences, called Osteophytes, appear. They can be damaging to the joint's motion, leading to the inflammation of the soft tissues.
The risk factors for the osteoarthritis [4] are:

- *Age* (major risk factor): the risk of osteoarthritis increases with age;
- *Genetics*: an example is the Stickler syndrome, that interferes with the body's natural ability to produce the cartilage protein collagen XI;
- *Gender*: the osteoarthritis affects up to 12% of the population including 2/3 of women and 1/3 of men [5];
- *Weight*: the bodies of overweight people bare more stress and strain on their knee joint. In addition, fat tissue produces proteins that may cause harmful inflammation in and around the joint;
- *Congenital defects*: some people are more disposed to osteoarthritis symptoms at birth;
- *Joint injuries*: people who have suffered from joint injury or undergone joint surgery are predisposed to the development of osteoarthritis of those joints. This is often the cause when osteoarthritis occurs in younger people.

Rheumatoid arthritis, on the other hand, is an autoimmune disease, in which the immune system does not defend the body against infections. To pay the consequences are the joints that become rigid, painful and swollen.
Finally, hemophilia: the continuous bloody lesions weaken the joints which become stiff and become painful.

The most used solution to cope with these diseases is Total Knee Arthroplasty (TKA).

1.3 Primary Total Knee Replacement

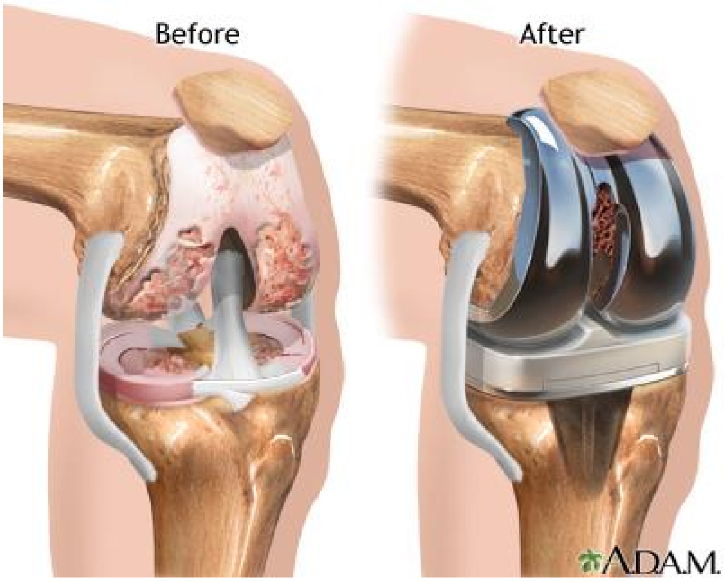


Fig.3 Before and after TKA [6]

Total Knee Replacement (TKR) or Total Knee Arthroplasty (TKA) is one of the most successful orthopedic procedures of the twentieth century: improvements in surgical materials and techniques have increased its effectiveness. In fact, according to the Agency for Healthcare Research and Quality, more than 600,000 knee replacements are performed each year in the United States. [7]

TKR is a technique in which a diseased knee joint is removed and is replaced by an artificial joint, to restore the function of the knee. In particular, the femoral condyles and the tibial plateau are replaced by metal components. A polyethylene part, positioned between the tibial and the femoral component, performs the role of the insert to absorb shocks and allow slipping. About the patella, according to the clinical case, it can be replaced or not.

The implant that is used by the surgeon depends on the condition of the patient (gravity of bone loss, allergies…) and on the economic possibilities of the hospital or the private clinic.

1.3.1 Implant

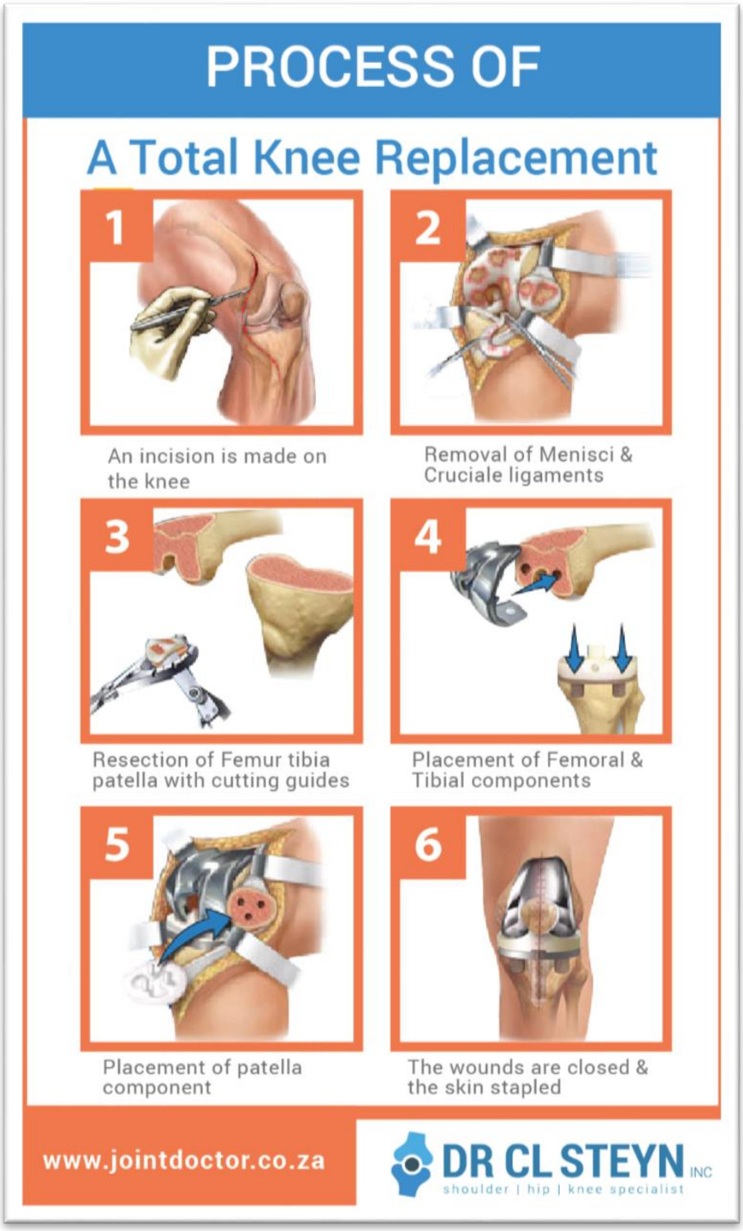


Fig.4 Process of a TKR [8]

The artificial knee joint helps in improving the mobility and reduces pain, although it will not be able to flex as much as a normal knee joint. This implant has an average life of 15-25 years, after which it can fail, causing pain or a decrease in knee function. Persistent pain and swelling can indicate loosening, wear, or infection, and the location of the pain can be all over the knee (generalized) or in one area (localized) [9]. The decline in knee function may result in a limp, stiffness, or instability.

1.3.2 Causes of an implant failure

The causes of failure of the implants are [7]:

- Wear and Loosening;

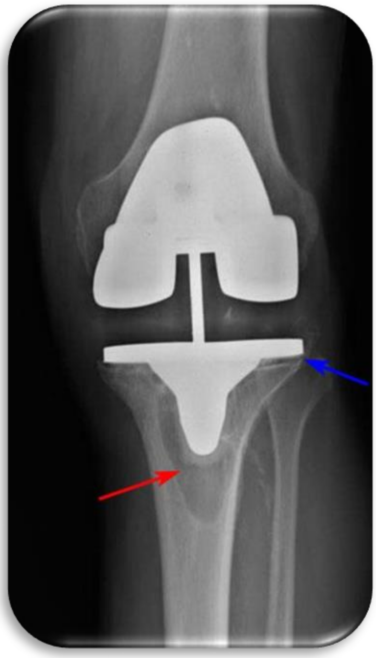


Fig.5 Osteolysis (red arrow) has occurred around the tibial component, causing it to become loosened from the bone (blue arrow)

- Infection;

Fig.6 An antibiotic spacer placed in the knee during the first stage of treatment for joint replacement infection

- Fractures;

Fig.7 The femur (thighbone) has broken in several places around the implant [10]

- Instability;

Fig.8 Injured ligaments can make the knee unstable [11]

- Patient-Related Factors.

The solution used to cope with this causes of failure of a primary TKR is the revision TKR.

1.4 Revision Total Knee Replacement

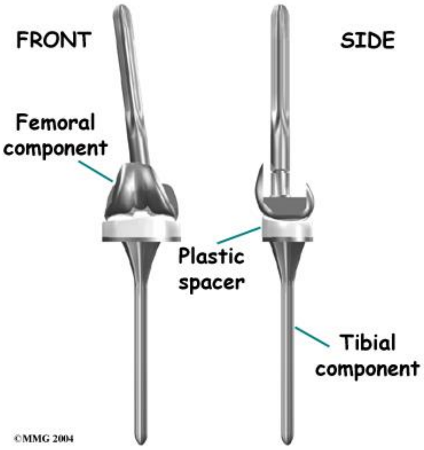


Fig.9 Revision prosthesis [12]

Revision Total Knee Replacement (rTKR) or Revision Total Knee Arthroplasty (rTKA) provides for the replacement of the failed primary implant prosthesis with a new prosthesis, called a revision prosthesis, that usually has longer and thicker stems that will be fitted deeper inside the bone for extra support. It is more complicated than a primary total knee replacement and requires extensive preoperative planning, specialized implants and tools, prolonged operating times and the mastery of difficult surgical techniques to achieve good results.
Nevertheless, it is estimated that more than 22.000 knee revision operations are performed in the United States each year [13].
Revision TKR studies included variable quantities of bone loss and soft tissue ineffectiveness that necessitated improved fixation and increased constraint, respectively, to maintain joint stability. The use of the stems can help distributing the increased stress of a constrained articular fixation and reducing the risk of implant loosening. On the other hand, the use of long stems provides less loosening, but a greater stress on the bone. Furthermore, since a revision prosthesis has a life equal to half the life of a primary prosthesis and since people are physically more active and they live longer than patients several decades ago, if a long stem is used at the first revision, a longer stem must be used in a second revision.
In fact, the number of people candidates for the knee prosthesis rises proportionally to the progressive increase in average life expectancy, as can be seen in this graph:

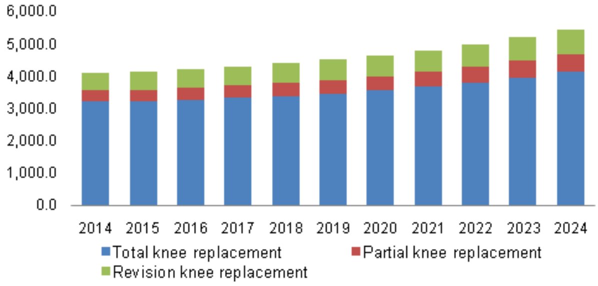


Fig.10 U.S. knee implants market revenue, by procedure type, 2014-2024 (USD Million) [14]

Stem lengths are variable but are decisive to the overall success of the implant. Stem length is principally determined by the stem fixation technique. The major stem fixation techniques in modern rTKR are fully cemented and uncemented fixation. Both techniques have advantages and disadvantages that influence their use. [15]
Relevant factors in the decision to use cemented or uncemented stems include stress shielding, osteolysis, stem-end pain, implant wear, periprosthetic fracture and aseptic as well as septic loosening. However, no evidence-based guidelines are available to standardize stem length and stem type.
Removal of the revision implant should be easier, lethal micro embolism and the main reactions to cement should be avoided when uncemented stems are used. Moreover, some studies report that the cemented interface may break down gradually with time, because of cyclic loading, producing component failure [16].
However, cement consents an increase of the contact area between the stem and the bone, a better centralization of the stem in the femoral canal, the possibility of adding topical antibiotics and an initial rigid bond between the prosthesis and the bone, which is a fundamental requirement for the long-term success of a rTKR.
Thus, the main advantage of cemented stems is a fast and good fixation of the stem with the bone and good results in long-term studies.
The use of cemented stems is commonly recommended in patients with low bone quality or altered anatomy.

1.4.1 Things to do before the revision implant

Standard assessments are performed, including x-rays, laboratory tests, and possibly other imaging modalities like bone scans, CT scans, or Magnetic Resonance Imaging (MRI) studies. X-rays may demonstrate a change in the position or condition of the components. MRI helps to determine the cause, location, and the amount of bone loss before surgery [9].
In case of infection it may be necessary to aspirate the joint fluid with a needle, which will then be sent to a laboratory to analyze it and identify the specific type of infection.

1.4.2 Implant

Most revision total knee replacements take longer to perform than primary procedures (about two or three hours). The first step is removal of the primary implant.
In the presence of a significant bone loss, a bone graft is necessary, which can be either autograft or an allograft. In some cases, metal wedges, wires or screws may be used to strengthen the bone.
After removing the primary implant and checking the amount of bone loss, a hole is made in the femoral canal and in the tibial canal where respectively the femoral and tibial stem will then be inserted, following the longitudinal direction of the bone.
The femoral component is then inserted with the femoral stem and the tibial component with the tibial stem.
In the case of an uncemented stem, cement will be placed on the femoral component, while in the case of cemented stem cement will also be placed in the femoral canal. The same thing is done for the tibia.
Finally, the polyethylene insert is introduced between the femoral and the tibial component.

Fig.11 (Left) In this x-ray, the primary knee replacement implant is unstable due to weakened bone. (Right) In revision surgery, components with longer stems fit more securely into the bones and provide stability [17].

Possible risks and complications [7]:

- Bleeding;
- Infection;
- Damage to nerves or blood vessels;
- Intra-operative fractures;
- DVT (Deep Venous Thrombosis) or blood clots;
- PE (Pulmonary Embolism)

1.4.3 After the surgery

Postoperative care after knee revision surgery is very similar to the treatment of a primary knee replacement. This includes a combination of physical therapy (**Fig.12**) that will be started within 24 hours of surgery and pain medication if necessary. Therapy will usually continue for up to three months following the surgery.
Assistive devices, such as a walker or crutches (**Fig.13**), will be used early in the convalescent period, and patients will progress to a cane or walking without any assistance as their condition improves [9].
Advanced techniques and materials for revision knee surgery usually allow substantial pain relief and improved functionality; however, this may not always be possible.

Fig.12 Physical Therapy [18]

Fig.13 Crutches [19]

1.5 Endo-Model Hinged Knee Prosthesis

One of the prosthesis for revision TKA is the LINK's Endo-Model Hinged Knee Prosthesis:

Fig.14 Prosthesis with Femur and Tibia

Fig.15 Prosthesis without bones

When there is case of large bone loss or gross ligamentous instability, a Hinged Knee Prosthesis is a good solution: it allows flexion-extension and intra-extra rotation movements, as in a physiological knee. These movements are achieved by means of a cross joint (**Fig.16**).

Fig.16 Hinged Knee Prosthesis Movements

This prosthesis consists of femoral and tibial components, which are simply pushed together, and a polyethylene (UHMWPE) tibial plateau. The two components are coupled by the plateau to prevent luxation without reducing the range of motion or rotation. There are also two stems that are inserted into the femur and tibia. Depending on the patient and the damage, the stems can be of different lengths, cemented or uncemented.

The cemented stems do not have a structuring, but instead the uncemented ones have longitudinal ribs and a microporous surface.
To achieve a central position within the medullary cavity, the tips of the cemented stems are fitted with star shaped UHMWPE centralizers. Direct contact of the metal stems with the inner wall of the bone is thereby prevented, and a gradual stress translation between metal and bone is achieved.

The stems are supplied in lengths of 50mm up to 280mm and the designs are showed here below:

Fig.17 Cemented Stems

Fig.18 Uncemented Stems

The modular stems are secured by a tapered connection. To ensure rotational stability, the stem has two opposing flanges, which are inserted into the medial and lateral grooves on the femoral/tibial components.

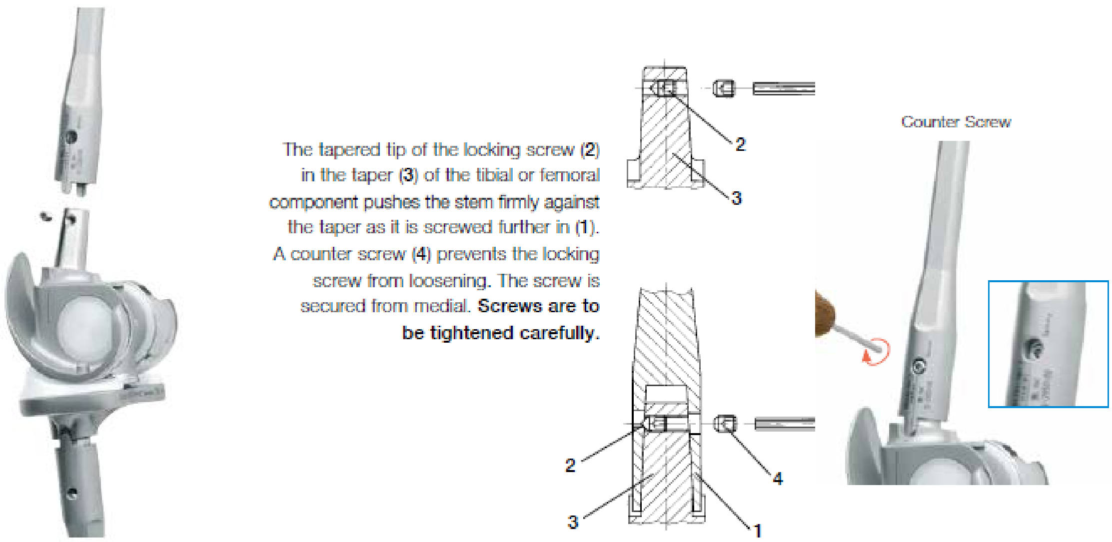
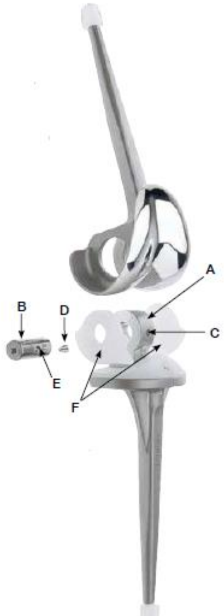


Fig.19 Details of the Endo Model Hinged Knee Prosthesis [20]

Furthermore, due to its design and dimensions, this prosthesis is considered one of the knee joints requiring very little bone resection and therefore it offers the highest possible sparing of bone substance in cases of revision.
Here below is reported a figure and a description of the components of the prosthesis to comprehend it better [20]:



The connecting piece **A**, which is fixed to the hinge knee prosthesis tibial component and links it to the femoral component, has a through-hole to accommodate the joint axis **B**. The ventral hole **C** is provided for the grub screw **D**, whose tip fits into the recess **E** on the axis and locks the latter once the upper and lower components have been joined.
From inside the intra-condylar box of the femoral component, polyethylene bearings **F** for the prosthesis axis are pressed into medial and lateral holes. The femoral and tibial prosthesis components are joined by introducing the tibial coupling into the intra-condylar box of the femoral component, such that the prosthesis axis can be inserted using the threaded rod. Articulation takes place between the prosthesis axis and the two bearings.

1.5.1 Problems of a Revision Hinged Knee Prosthesis

One of the problems of the Revision Hinged Knee Prosthesis is that it is ultra-constrained: it completely replaces the original joint, but the new artificial joint allows only a flexion-extension movement and a small movement of intra-extra rotation and this, according to some surgeons is a problem.
When cases of high bone loss occur, to ensure that there will be no loosening of the prosthesis, stems of considerable length are inserted in the bone. This, however, could be a problem in the case of a second revision, since the stem inserted in the first revision must be replaced with a longer stem, which will stress more the bone.
Furthermore, the use of a revision implant with long stems could be a problem if the patient suffers from osteoporosis, since it could happen that the stem pierces the bone epiphysis (**Fig.20**), with the consequent necessity of a re-intervention.

Fig.20 X-Ray image of prosthetic failure caused by tibia perforation with tibial stem [21]

Other problems that can occur are: the wear of the polyethylene insert or the use of cement, which can cause trouble of allergy to the patient, also due to failure of the implant.
Since there are no precise guidelines, it is therefore important that the surgeon carefully studies the clinical case before proceeding with the operation, so that he makes the most appropriate choices on the type of implant to be used in terms of materials, stems length, use of cement or not, to try to have a low failure rate. In addition, the surgeon must be careful not to infect the area affected by the implant and not risk complications during the operation, since these could be additional causes of revision prosthesis failure.

1.6 Aim of the Work

The rTKR is a success story of modern medicine and is expected to be increasingly used by surgeons in the coming years, but not enough is yet known about patient's outcomes or the effectiveness of different implants and there is no agreement among specialists on precise indications for the implant procedure. For these reasons, the intent of this thesis is to understand how the LINK's revision hinged knee prosthesis works when a patient does a squat and to produce data that could assist the clinical decision on the length and type of femoral stems (cemented or uncemented) to use. Specifically, investigate the change in the femur stress induced, when cemented and uncemented stems of different lengths (50 mm, 120 mm) are used. To accomplish this, a Finite Element Analysis (FEA) will be performed that enables detailed biomechanical investigations with the potential to detect effects of the different configurations on the bone that cannot be investigated in vivo.

The thesis is divided into two parts:

1- Dynamic Analysis to study how the Endo-Model prosthesis works during a squat movement;

2- Static Analysis, both in an extension and in a flexion configuration, to study the stresses in the femur when two types of stem (cemented and uncemented), with two different lengths ($50mm$, $120mm$).

2 Materials and Methods

2.1 Geometry

From LINK's industrial designs of some parts of a left Endo-Model Hinged Knee Prosthesis, it was possible to view the measures of the parts and their characteristics (rugosity, chamfers, fittings, type of screws).

2.1.1 3D Components

Since some 3D parts had not been made before, it was possible to made 3D parts by recreating important industrial designs within the Software SolidWorks, with an extrusion in revolution as a result.
Shown below are the 3D parts that have been made:

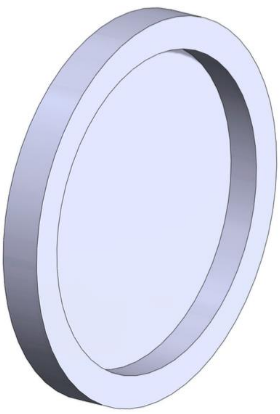


Fig.21 Cover of the Medial Bearing

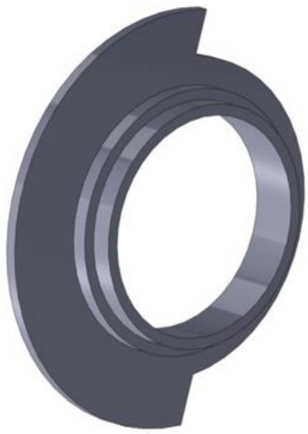


Fig.22 Medial Bearing

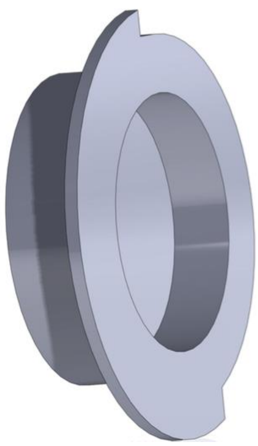


Fig.23 Lateal Bearing

Table 1 *Tibial Component*

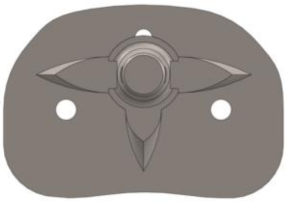
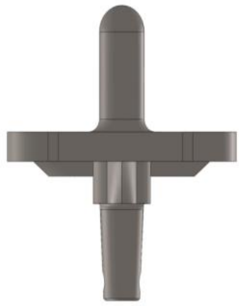
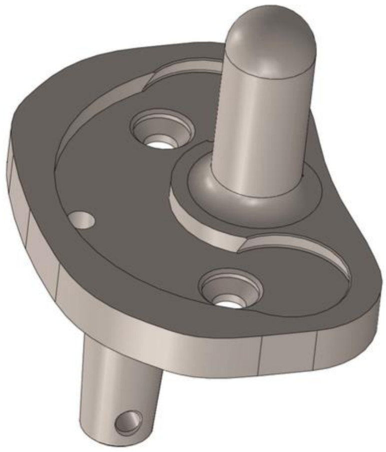
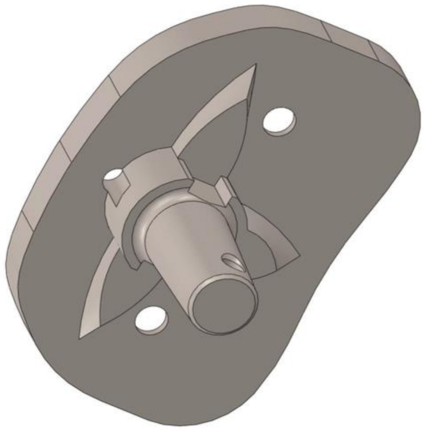
<table>
<tr><th colspan="4">Tibial Component</th></tr>
<tr><th>Frontal view</th><th>Bottom view</th><th>Lateral view</th><th>Behind view</th></tr>
<tr><td></td><td></td><td></td><td></td></tr>
<tr><th colspan="4">General view</th></tr>
<tr><td colspan="2"></td><td colspan="2"></td></tr>
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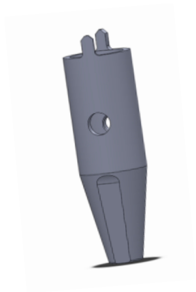
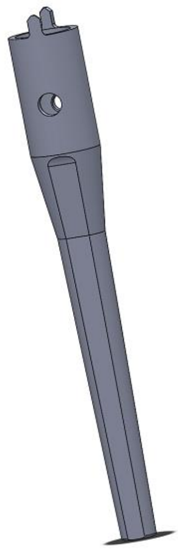
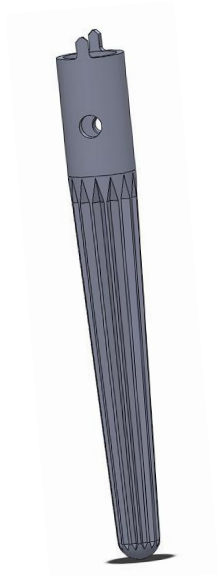
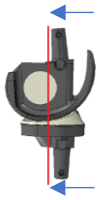
Table 2 *Stems*

Cemented Stems	
50 mm	**120 mm**
	
Uncemented Stems	
50 mm	**120mm**
	

2.2 Part I – Dynamic study

Assumed to have a linear, elastic and isotropic model for the Cobalt-Chrome (CoCr) and a linear, elastoplastic and isotropic model for the Polyethylene (PE), the materials have been assigned to components in this way (CoCr in grey and PE in white):



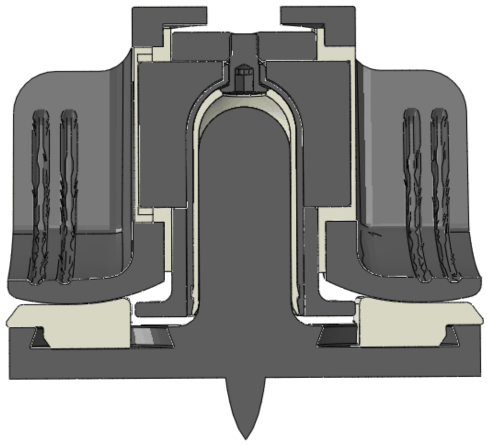


Fig.24 Section of the prosthesis

Table 3 *Properties of the materials [16]*

Cobalt-Chrome	Mass density: 8270 Kg/m^3 Young's Modulus: 240 GPa Poisson's Ratio: 0.3
Polyethylene	Mass density: 960 Kg/m^3 Young's Modulus: 685 MPa Poisson's Ratio: 0.4

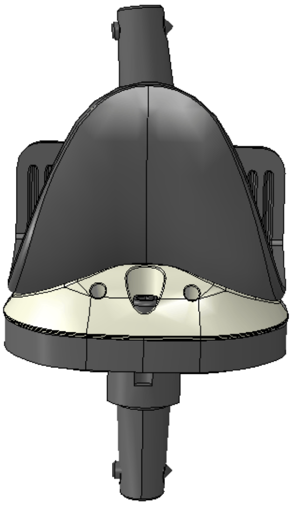
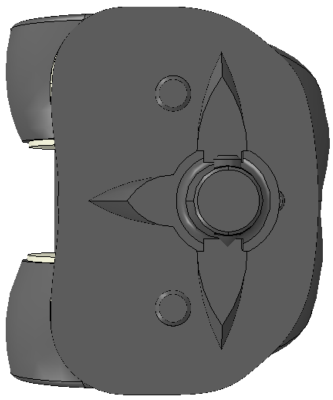
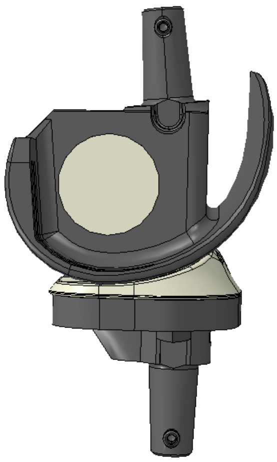
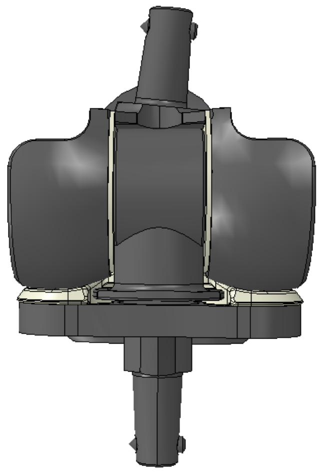
Yield Stress (MPa)	Plastic Strain
22.7	0.00
22.9	0.05
25.3	0.10
36.3	0.20
54.7	0.30

2.2.1 Assembly and Contacts

With the use of the Software Abaqus, it was possible to show all the parts created in SolidWorks and then to assemble them, through translations and rotations of the individual components.

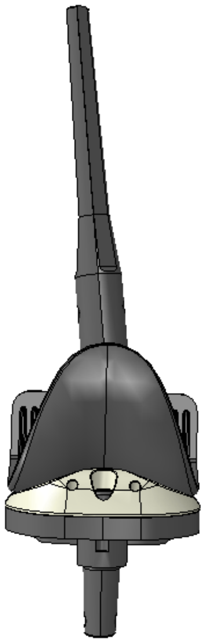
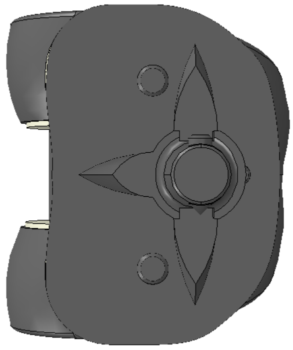
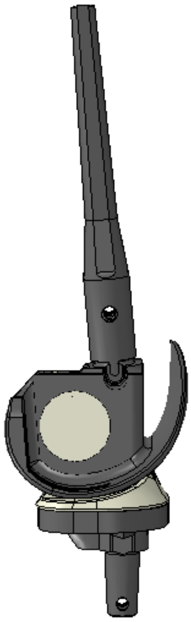
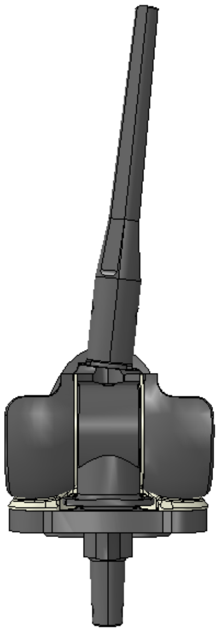
Shown below are the principle views of the primary assembly without the femoral stem:

Table 4 *Principal Views of the Prosthesis*

Frontal view	Bottom view
	
Lateral view	**Behind view**
	

Then a cemented stem of $120mm$ was assembled at the model:

Table 5 *Principal Views of the Prosthesis with the Cemented Stem of 120mm*

Frontal view	**Bottom view**
	
Lateral view	**Behind view**
	

Once the assembly was finished, the contacts between the parts were created.
To start, a general contact was made along with tie contacts between screws and CoCr components, to avoid any consequential displacement during the explicit simulation. No tie contact was made between screws and PE components because micromotions will be studied there.

The tie contacts that have been created between screws and CoCr components are as follows:

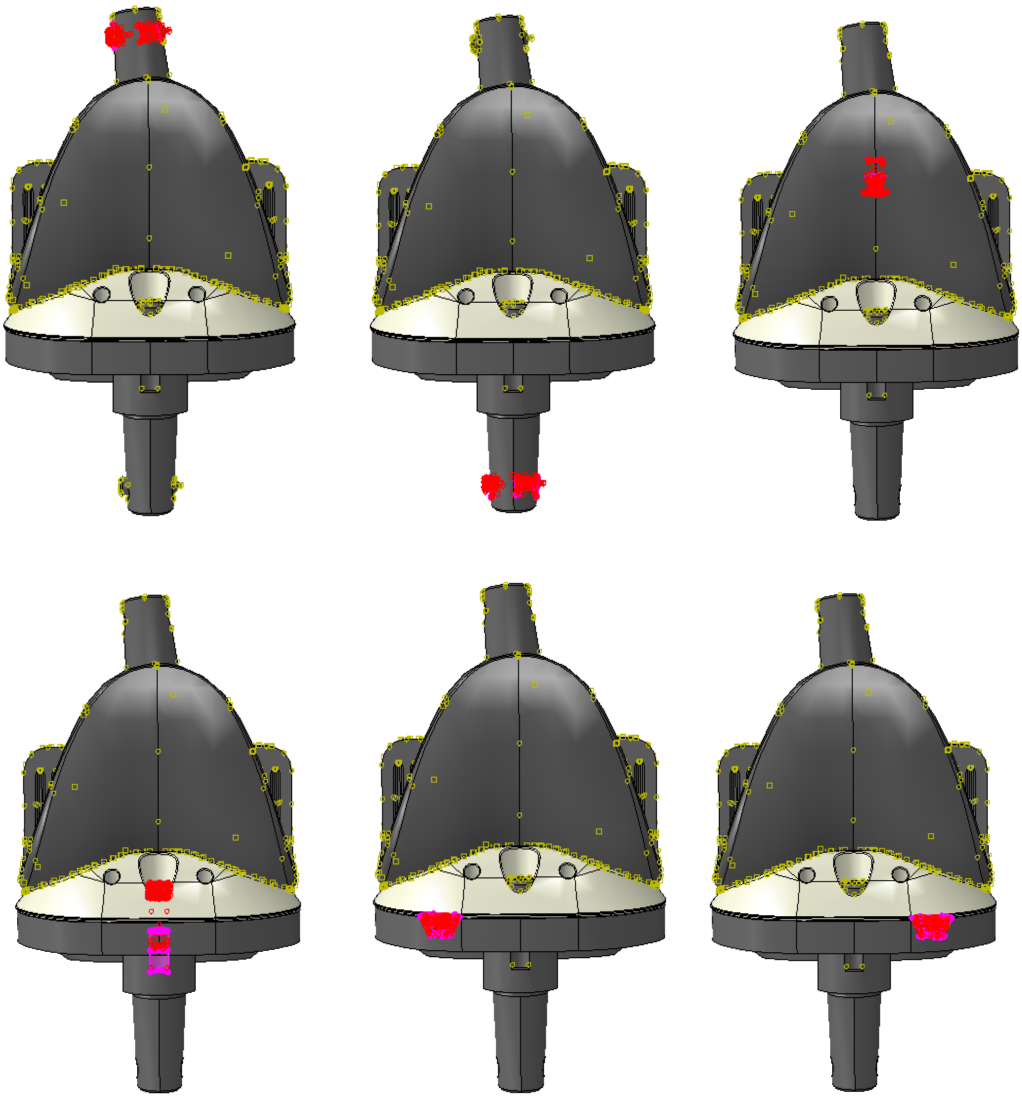


Fig.25 Tie Contacts

The pink surfaces in the tie contacts figure like ‘slave surfaces’, while the red parts figure like ‘master surfaces’, which are generally more rigid than the slave surfaces.

2.2.2 Loads

To simulate a flexion-extension movement, an axial force was applied in a caudal-cranial direction to the lower surface of the tibial plate (**Fig.26 and Fig.27**):

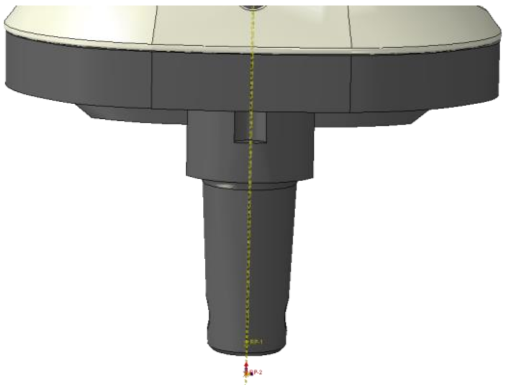


Fig.26 Direction Axial Force

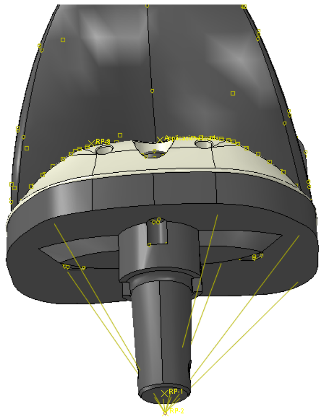


Fig.27 Coupling

The amplitude trend of the axial load is represented as a function of the time in the graph below:

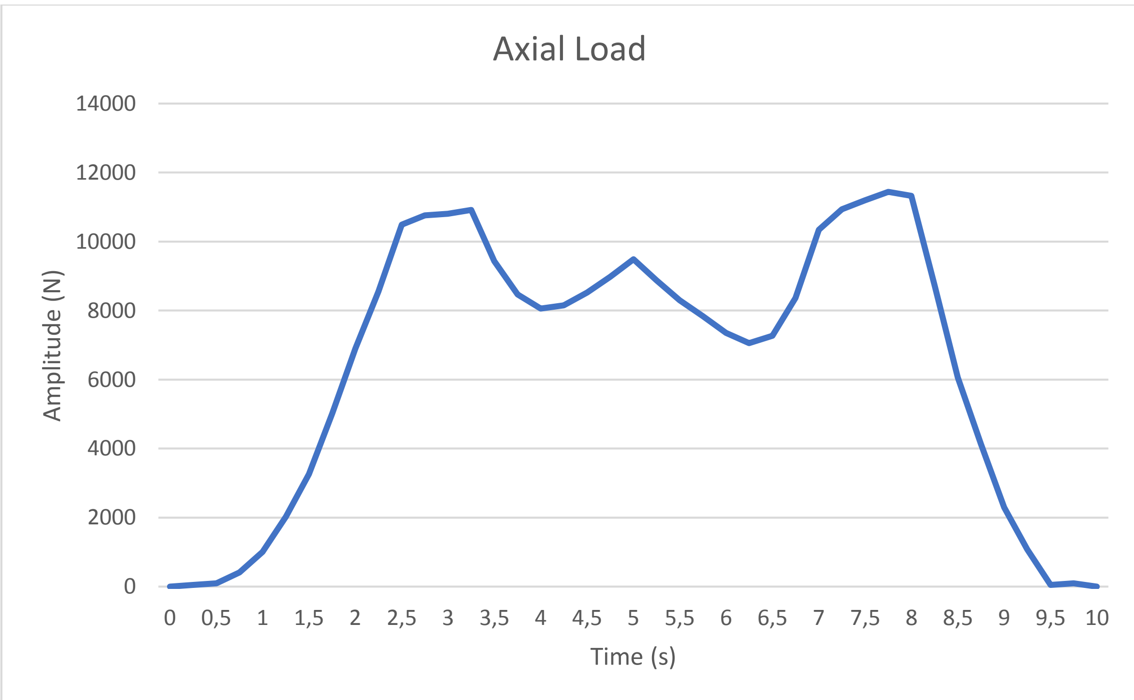


Fig.28 Amplitude Trend of the Axial Load

2.2.3 Constraints

A constraint has been placed on the femoral component to allow it only a flexion-extension movement: rotation was allowed only around the middle-lateral axis.

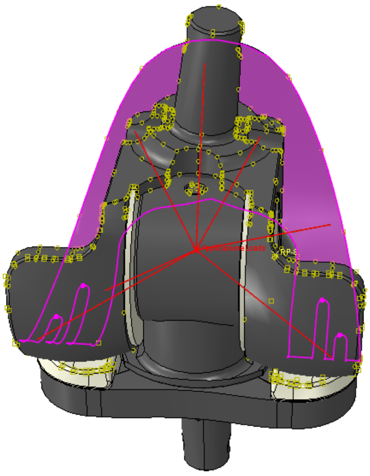


Fig.29 Coupling with the internal surface of the femoral component

A flexion-extension movement with a range of 0°-90° was given to the femoral component, with a trend amplitude that is as follows:

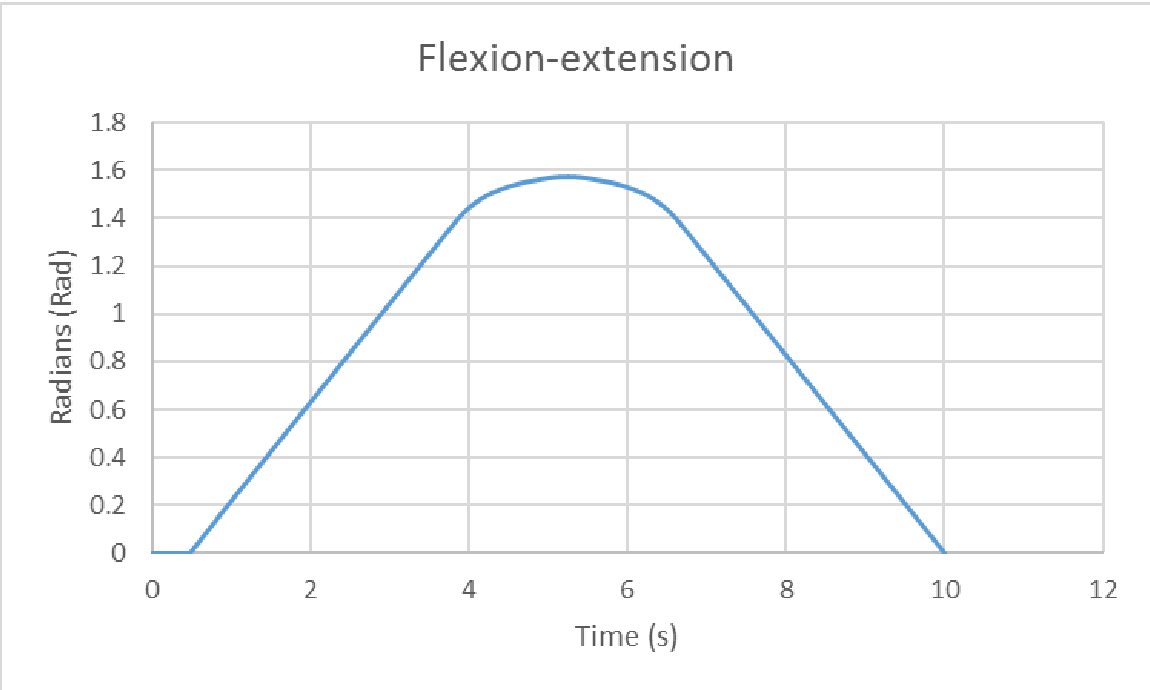


Fig.30 Amplitude Trend of Flexion-Extension

2.2.4 Mesh

In order to simulate the flexion-extension movement, the model was meshed using tetragonal elements with an approximate element size of $2mm$.
Element size was chosen based on a convergence test to make sure that the selected mesh did not influence the result.
The result is shown below:

Fig.31 Full Assembly meshed

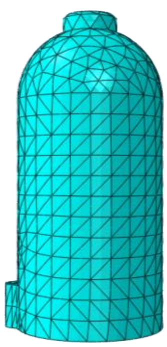


Fig.32 Metallic Cylinder

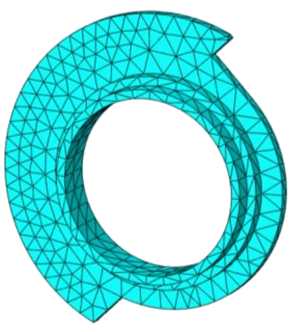


Fig.33 Medial Bearing

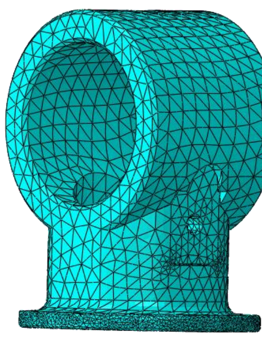


Fig.34 T-Component

2.3 Part II – Static Study

Implant materials (UHMWPE, Cement, CoCr) and cancellous bone were assumed to be homogeneous, isotropic and linearly elastic, while the cortical bone was considered as transversely isotropic, with properties varying along the mechanical axis of the femur.

In the static simulations, for all the models, the cement (PMMA) was placed between the femoral component and the distal part of the femur.

A shell of $3mm$ was done on the femoral component, to mimic what happens during the implant of the prosthesis:

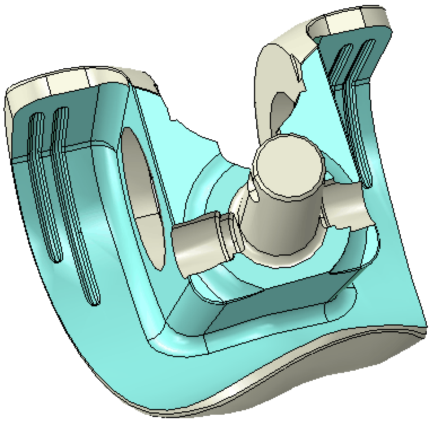


Fig.35 PMMA on the femoral component

However, cement has also been positioned in the femoral canal in the models with cemented stems:

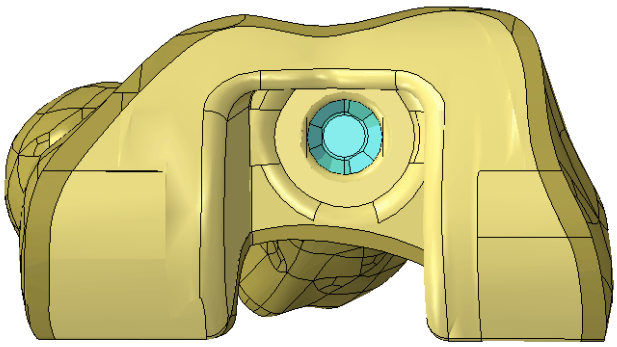


Fig.36 Femoral canal with PMMA

Fig.37 Detail: PMMA

Table 6 *Properties of the materials [16]*

PMMA	Elastic isotropic Young's Modulus: 3000 MPa Poisson's Ratio: 0.3
Cancellous Bone	Elastic isotropic Mass density: 130 Kg/m^3 Young's Modulus: 2130 MPa Poisson's Ratio: 0.3
Cortical Bone	Transversely isotropic Mass density: 1800 Kg/m^3 Young's Modulus: E_1=11500 MPa E_2=11500 MPa E_3=17000 MPa Poisson's Ratio: ν_{12} =0.58 ν_{13} =0.31 ν_{23} =0.31

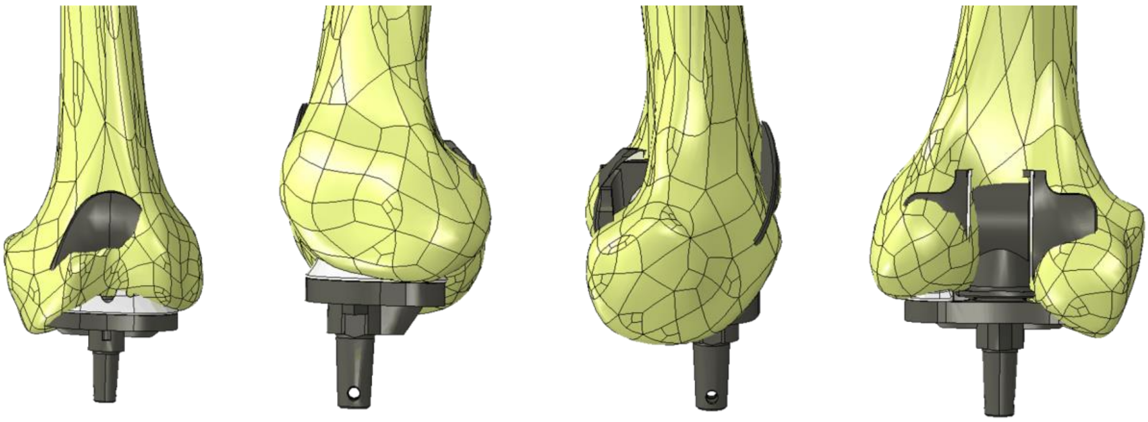
2.3.1 Assembly and Contacts

Here below are represented the four configurations of the prosthesis with the cemented femoral stem (**Fig.38**) and uncemented femoral stem (**Fig.39**) with a length of 50mm and 120mm:

Fig.38 Medial view of the prosthesis with cemented stems with a length of 50mm (left) and 120mm (right)

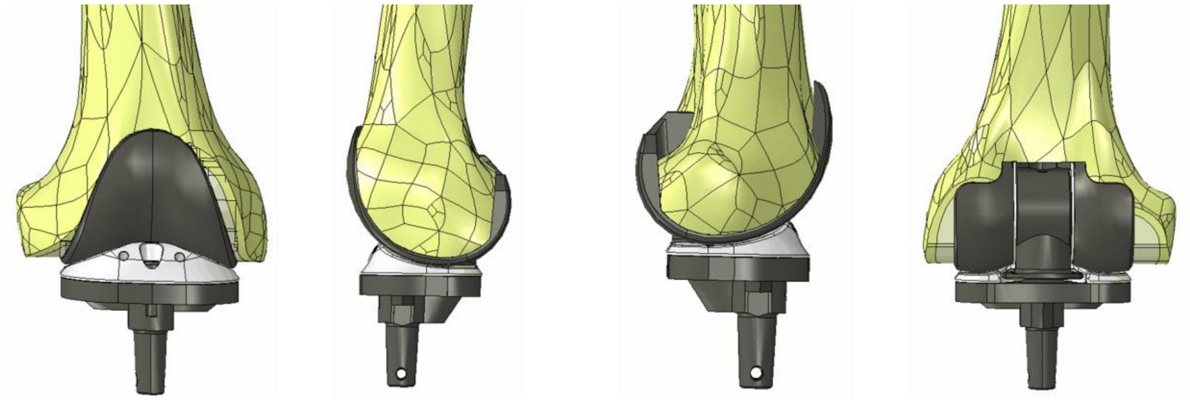
Fig.39 Medial view of the prosthesis with uncemented stems with a length of 50mm (left) and 120mm (right)

Finally, to complete the models, the femur was added and assembled to the prosthesis. The bone geometries were obtained by 3D reconstruction from CT images.
Once the femur has been correctly positioned (**Fig.40**), what happens during the surgery was simulated, i.e. the bone was cut to fit the femoral component (**Fig.41**).



Frontal View Lateral View Medial View Back View

Fig.40 Femur positioned



Frontal View Lateral View Medial View Back View

Fig.41 Femur cut

Ten configurations were made and for each of them a static simulation was made:

- Two for models without femoral stem insertion: at 0° and 90° of flexion;
- Two for models with uncemented femoral stem insertion with a length of 50mm: at 0° and 90° of flexion;
- Two for models with uncemented femoral stem insertion with a length of 120mm: at 0° and 90° of flexion;
- Two for models with cemented femoral stem insertion with a length of 50mm: at 0° and 90° of flexion;
- Two for models with cemented femoral stem insertion with a length of 120mm: at 0° and 90° of flexion.

2.3.2 Loads

0° Model ***90° Model***

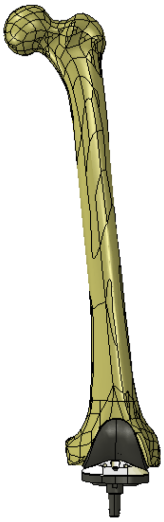


Fig.42 Configuration at 0° of flexion

Fig.43 Configuration at 90° of flexion

In the configuration at 0° of flexion (**Fig.42**) an axial force of 700N was applied to the lower surface of the tibial component in the caudal-cranial direction.
In the configuration at 90° of flexion (**Fig.43**) an axial force of 2144.12N was applied to the lower surface of the tibial component in the caudal-cranial direction.
These values were obtained from an experimental and numerical study [16].

2.3.3 Constraints

In all the models the proximal part of the femur was completely constrained as is shown below:

Fig.44 Constraint of the Femur

2.3.4 Mesh

The size of the prosthesis mesh does not present excessive variations compared to the dynamic model, but, having been added the femur, it had to be meshed (**Fig.45**).
A virtual topology was done to improve some geometries of the femur, then a mesh size that was a compromise between being small (more precise results, but slower simulation) and big (less precise results, but faster simulation) was chosen.

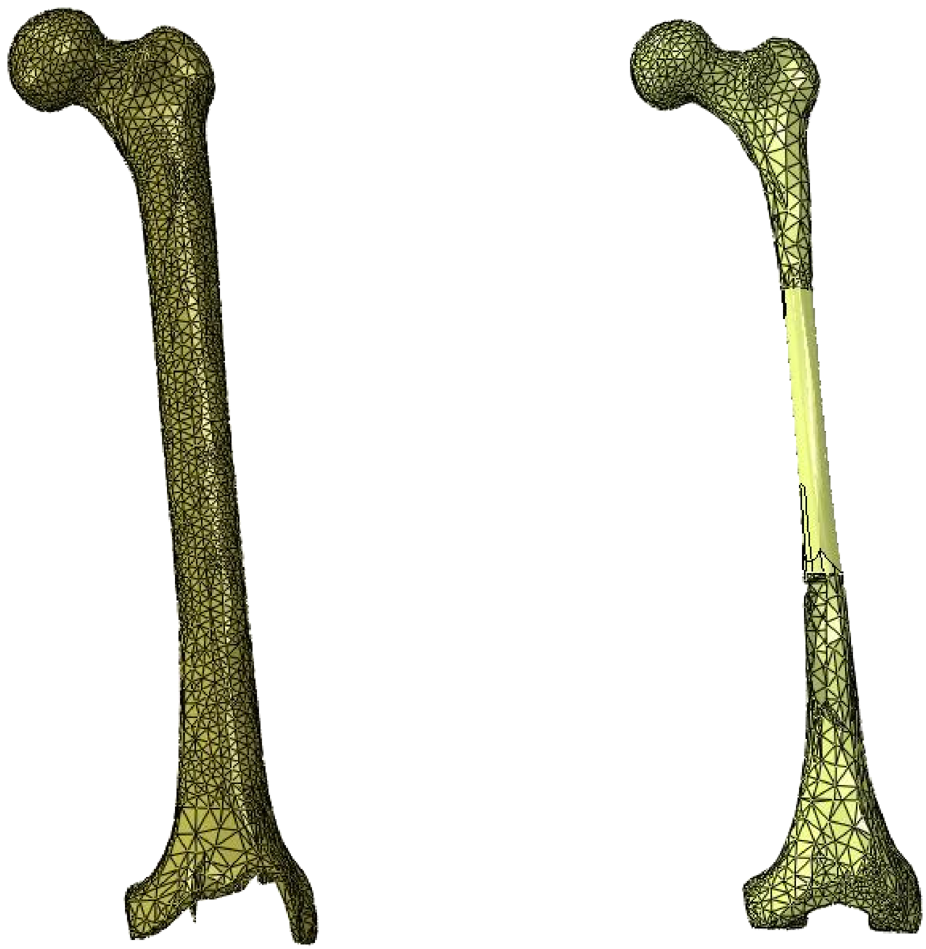


Fig.45 Femur meshed: Cortical Bone (left) and Cancellous Bone (right)

3 Results

3.1 Results part I

An explicit dynamic simulation was made with a duration of 10s and a time increment of 10^{-4} (according to the mesh size).
In the simulation results, the following parameters have been represented:

- Total area in contact;
- Center of the total force due to contact pressure;
- Total force due to contact pressure.

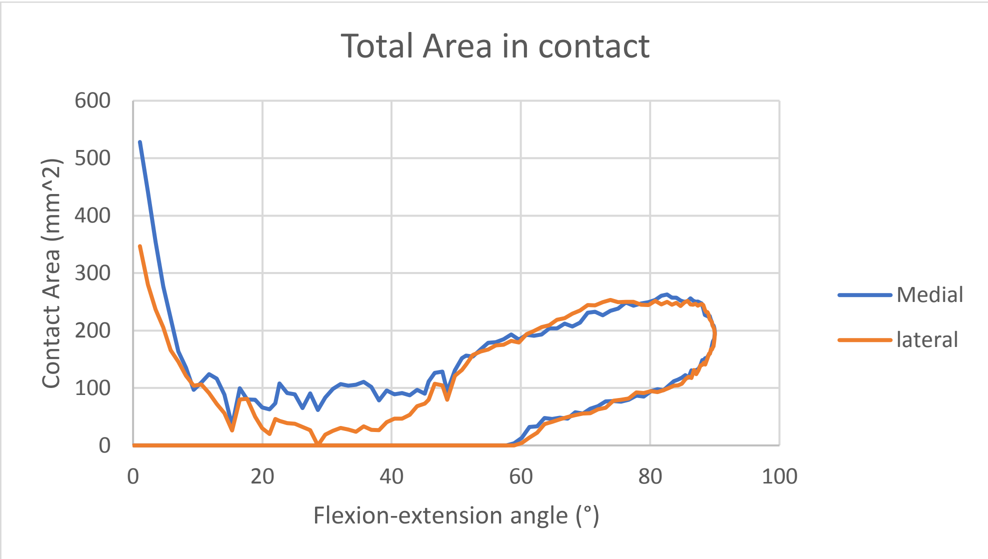


Fig.46 Trend of the Total Area in contact over the Flexion Angle

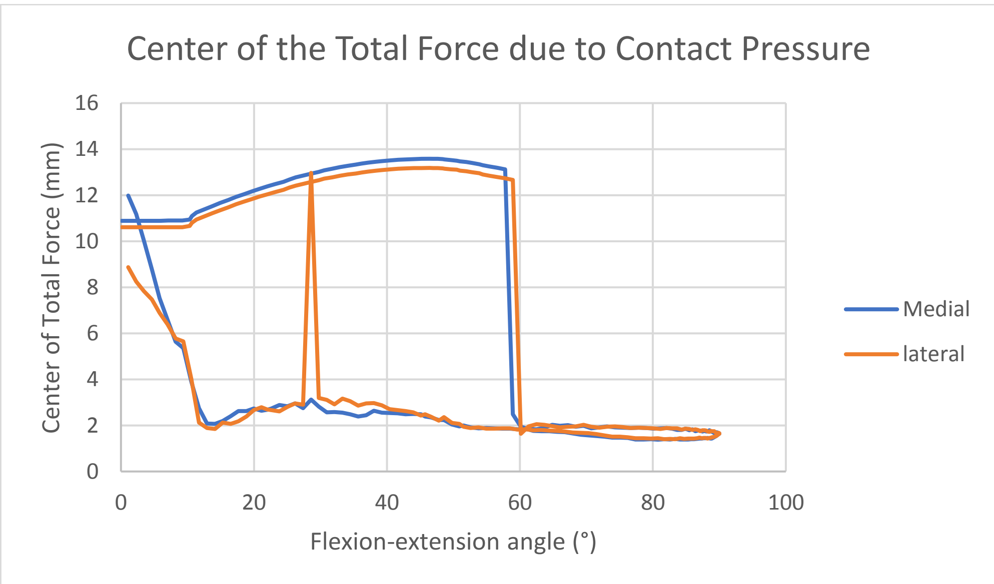


Fig.47 Center of the Total Force due to Contact Pressure over the Flexion Angle

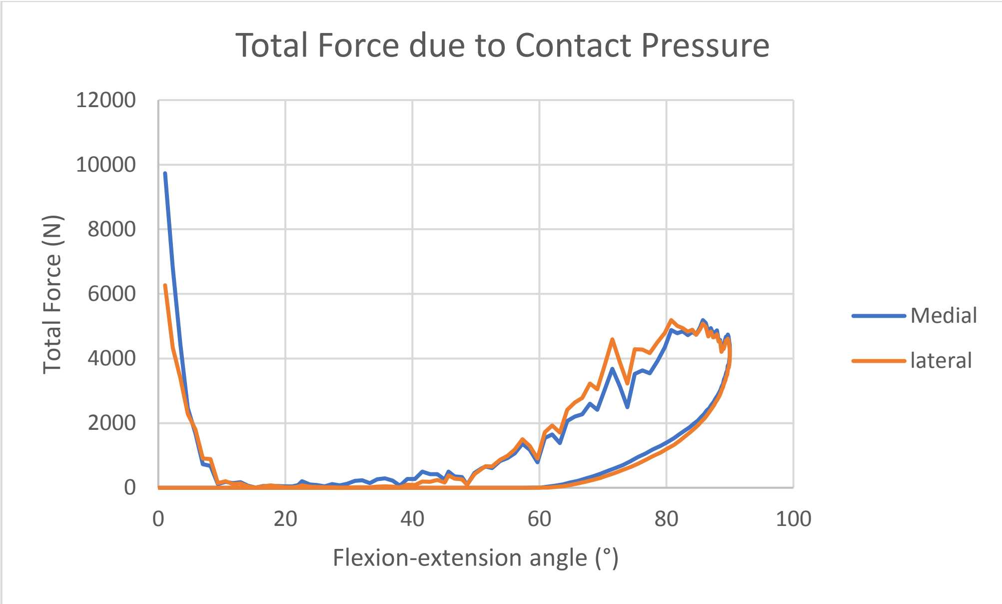


Fig.48 Total Force due to Contact Pressure over the Flexion Angle

In the first graph (**Fig.46**) the trend of the contact area of the medial and lateral part of the prosthesis is represented as a function of the flexion-extension angle.
There is a first section in which the area (both medial and lateral) is almost zero, after which 60° of flexion begins to increase up to 90° of flexion. In this last stretch the two areas have almost the same values.
When the extension movement begins, the lateral area has values that are initially not very different from the medial area, but then tend to be less than medial area values, with a trend that grows to more than $300mm^2$ for the lateral area and more than $500mm^2$ for the medial area.

In the second graph (**Fig.47**) the trend of the center of total force is a function of the flexion-extension angle. The center is initially positioned in the central area of the plateau, after which it advances more and more towards the front of the plateau, until it is moved to the lateral area, which means that a small extra-rotation occurred.

In the third graph (**Fig.48**) the trend of the total force is a function of the flexion-extension angle and is like the trend of the total area. The lateral total force ends with a value greater than 6000N, instead the medial total force ends with a value less than 10000N.

To better understand the graphs, sequential images that represent the contact pressure on the plateau are shown below:

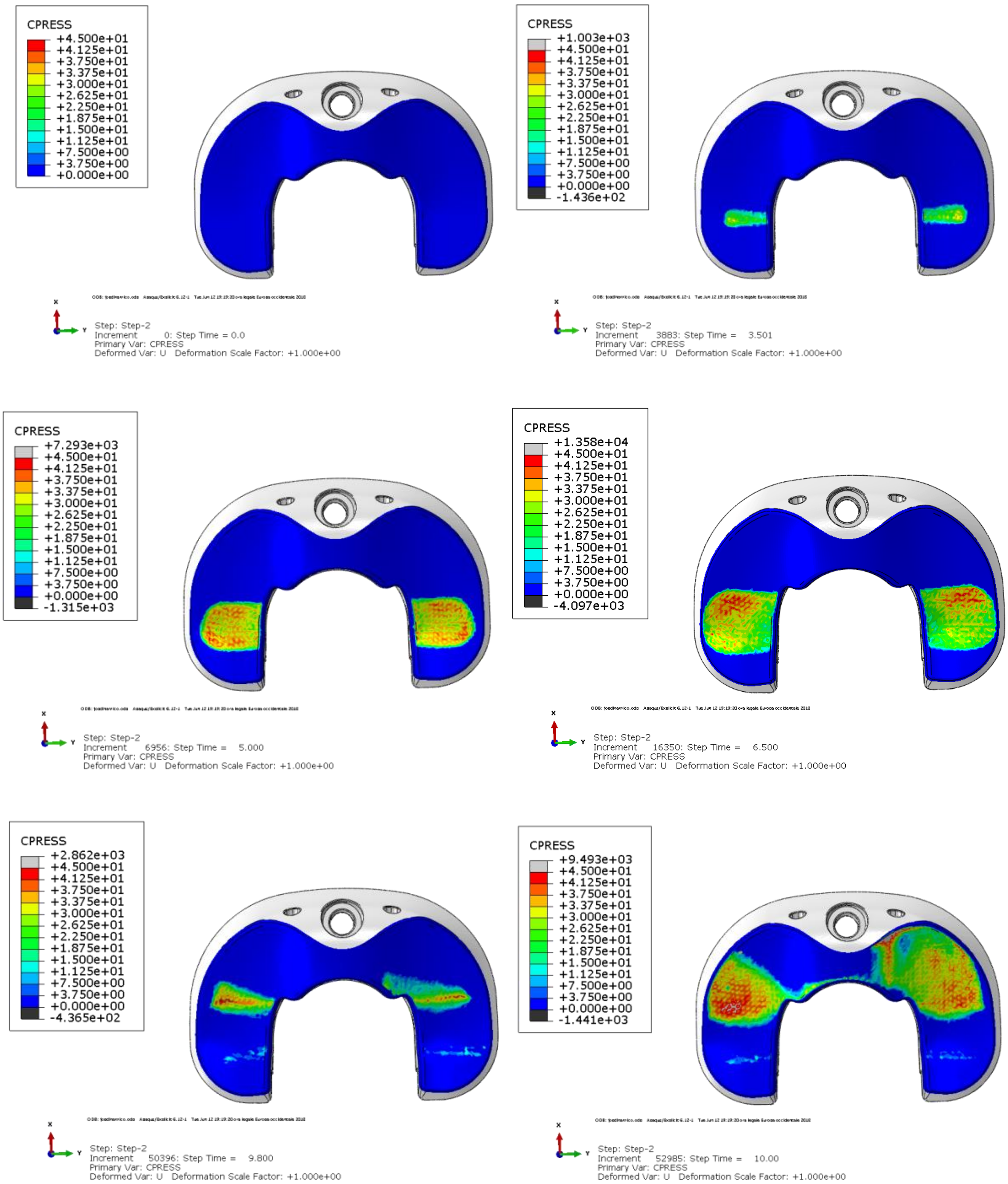


Fig.49 Qualitative images of the distribution and the amplitude of the Contact Pressure on the Plateau Component

3.2 Results part II

In the static simulation results, the average and maximum Von Mises and compressive stresses have been represented as a function of the partition planes made on the femur, for each of the five configurations, respectively in the knee extension and flexion configuration.
In the representative graphs of the compressive stresses as a function of the partition planes, only the negative values have been represented since they are indicative of the compression and therefore opposite to the direction in which the axial force acts on the model.
In the extension configuration the Von Mises and compressive stresses tend to decrease gradually towards the proximal part of the femur (**Fig.50, Fig.52, Fig.54, Fig.56**), as opposed to the flexion configuration, in which they tend to increase (**Fig.51, Fig.53, Fig.55, Fig.57**). Furthermore, the maximum is always towards the partition that is located above the area where the stem tip is located (partition plane at 12cm from the distal partition of the femur).

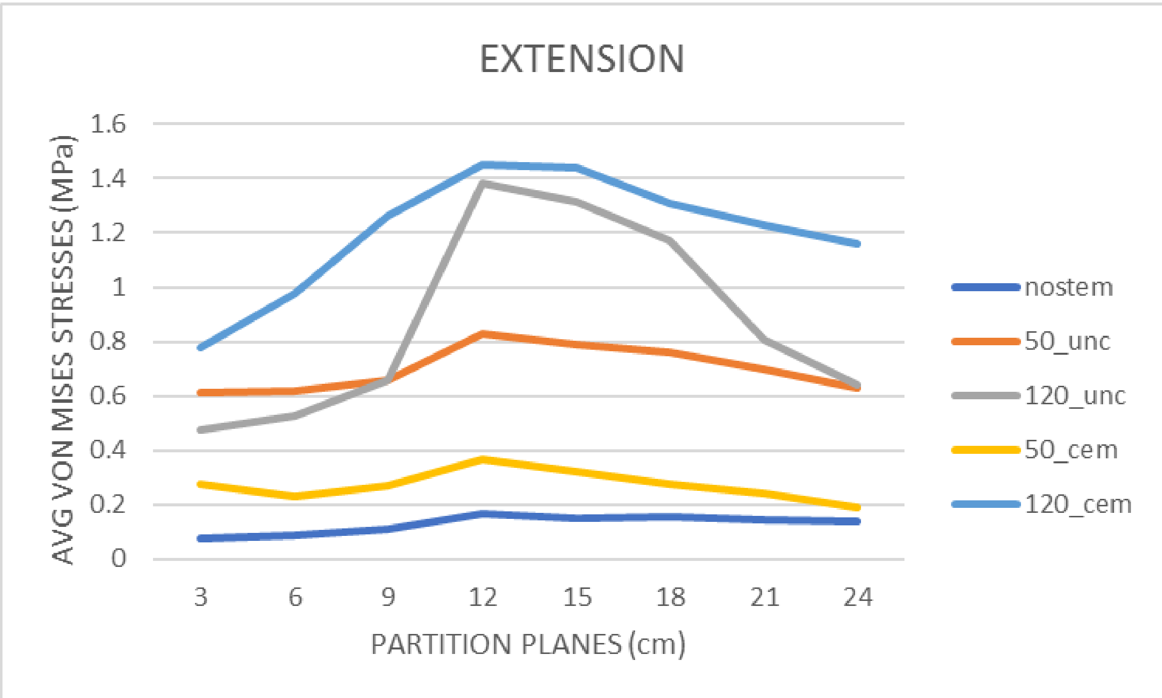


Fig.50 Average Von Mises Stresses for the five configurations as a function of the Partition Planes of the Femur in the extension configuration

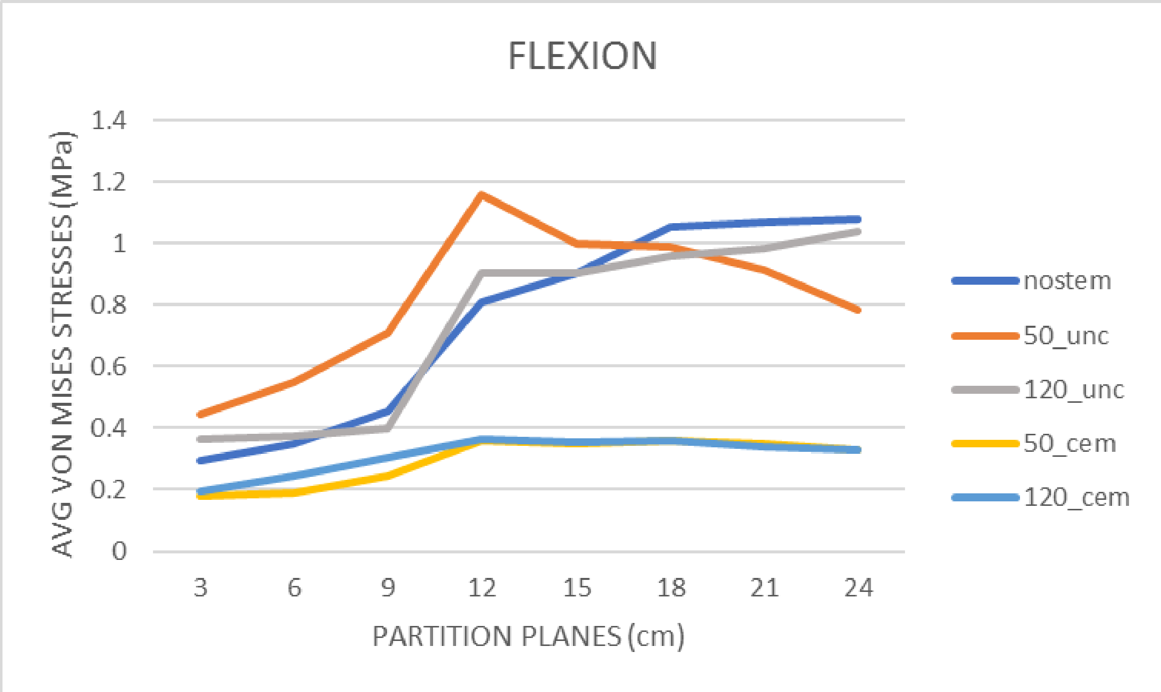


Fig.51 Average Von Mises Stresses for the five configurations as a function of the Partition Planes of the Femur in the flexion configuration

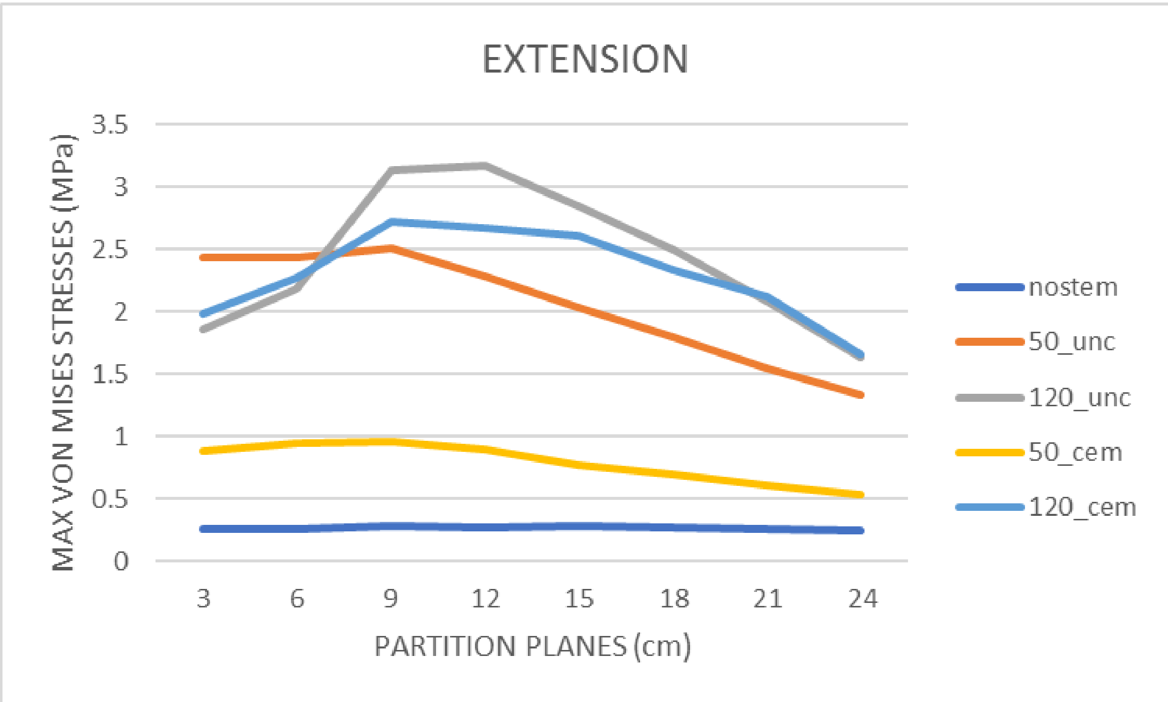


Fig.52 Maximum Von Mises Stresses for the five configurations as a function of the Partition Planes of the Femur in the extension configuration

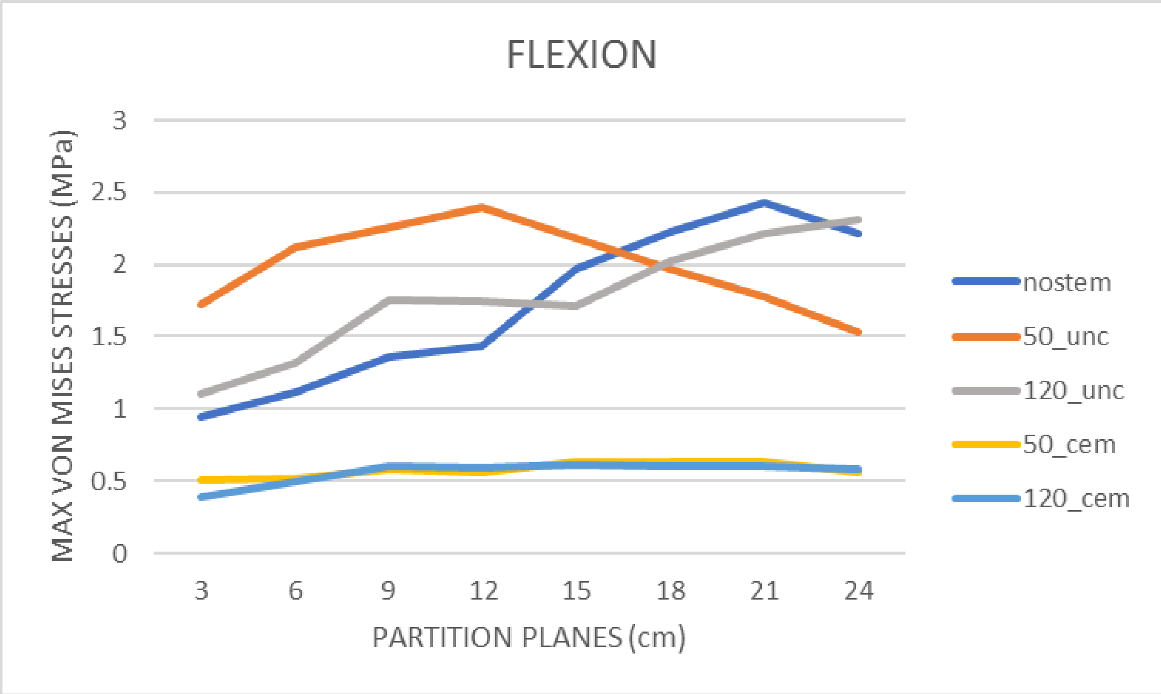


Fig.53 Maximum Von Mises Stresses for the five configurations as a function of the Partition Planes of the Femur in the flexion configuration

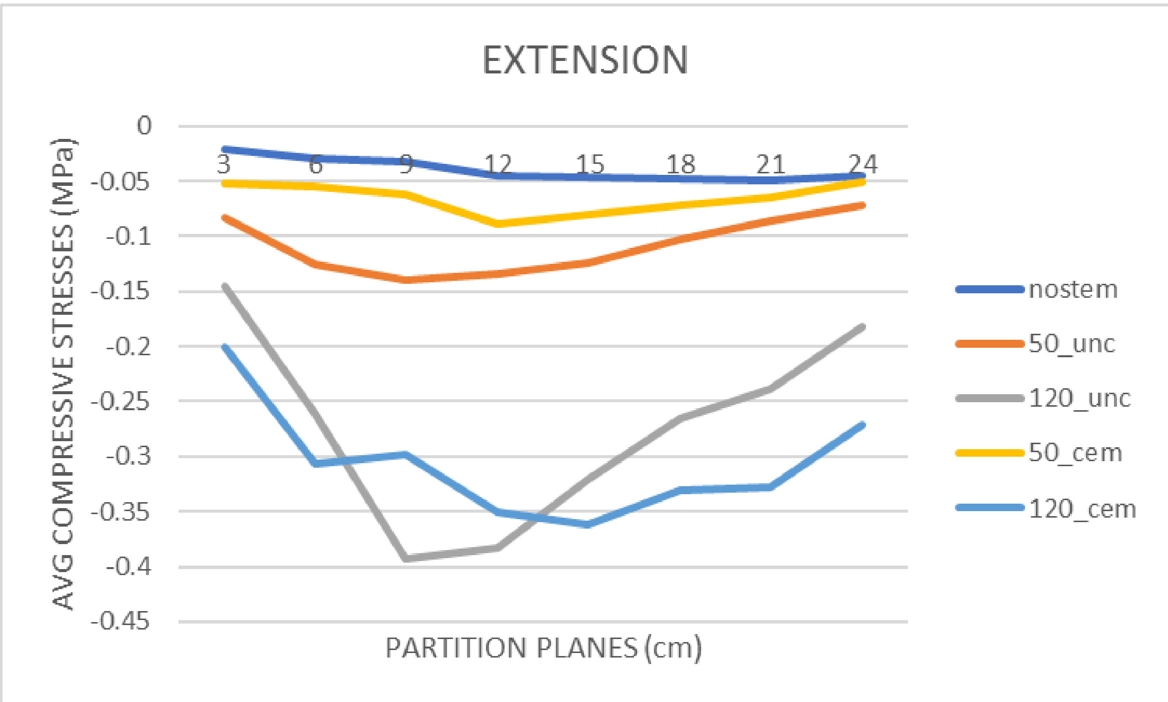


Fig.54 Average Compressive Stresses for the five configurations as a function of the Partition Planes of the Femur in the extension configuration

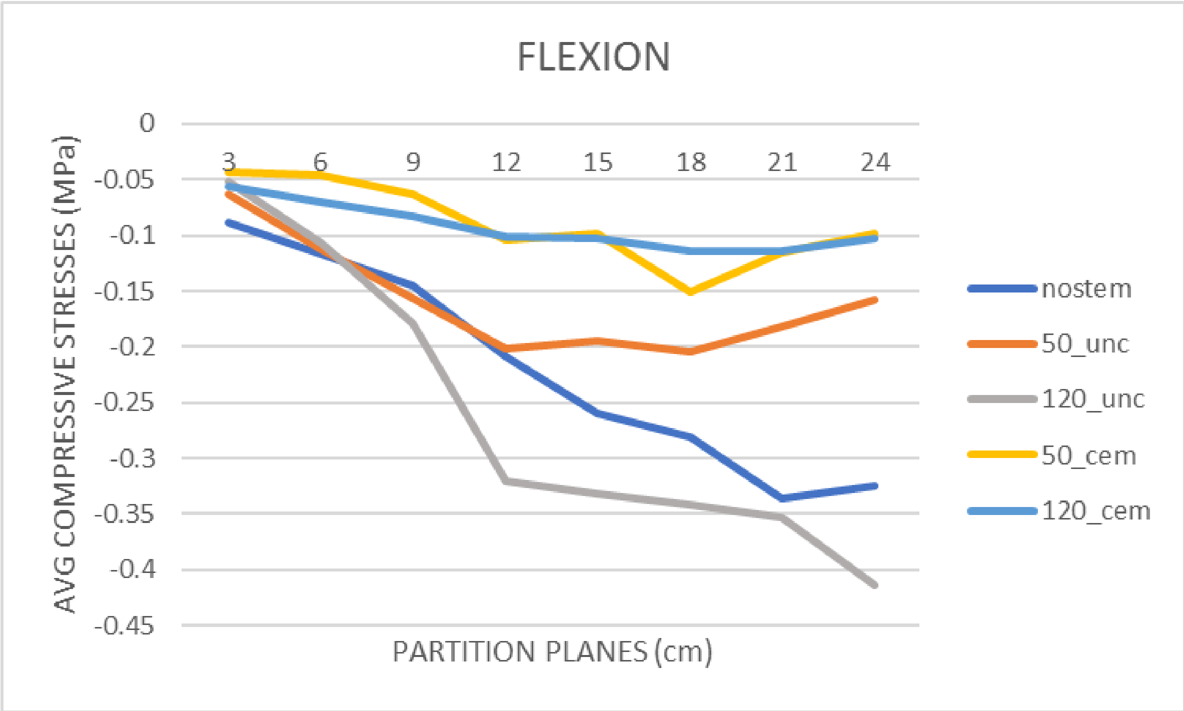


Fig.55 Average Compressive Stresses for the five configurations as a function of the Partition Planes of the Femur in the flexion configuration

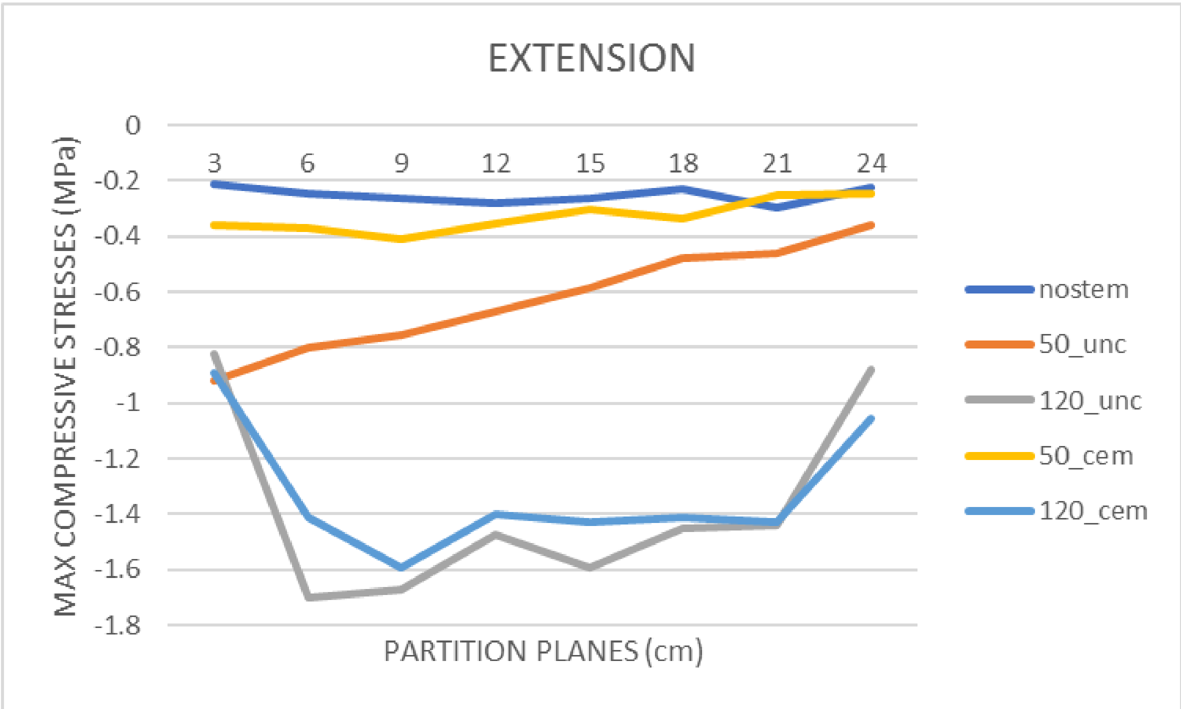


Fig.56 Maximum Compressive Stresses for the five configurations as a function of the Partition Planes of the Femur in the extension configuration

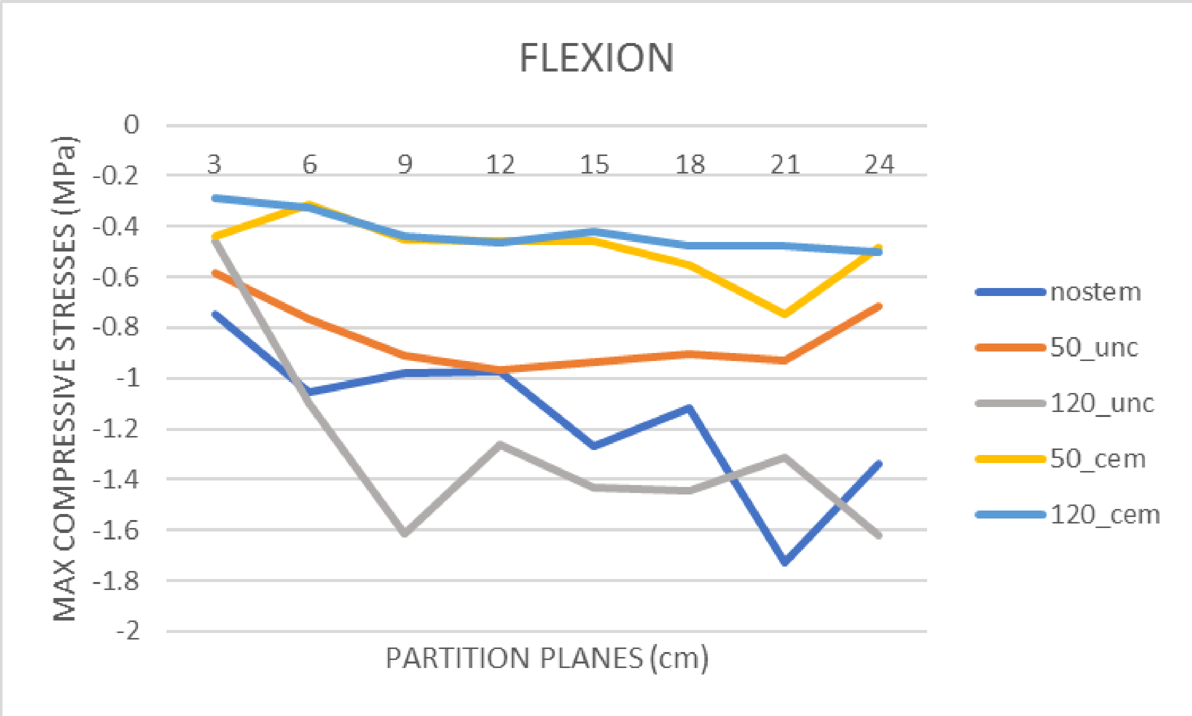


Fig.57 Maximum Compressive Stresses for the five configurations as a function of the Partition Planes of the Femur in the flexion configuration

Below are shown the representative histograms of the average Von Mises stresses (**Fig.58, Fig.62, Fig.66, Fig.70, Fig.74**), maximum Von Mises stresses (**Fig.59, Fig.63, Fig.67, Fig.71, Fig.75**), average compressive stresses (**Fig.60, Fig.64, Fig.68, Fig.72, Fig.76**), maximum compressive stresses (**Fig.61, Fig.65, Fig.69, Fig.73, Fig.77**) as a function of the partition planes of the femur, for each of the five configurations. In each histogram the values present in the extension and flexion configuration have been compared.

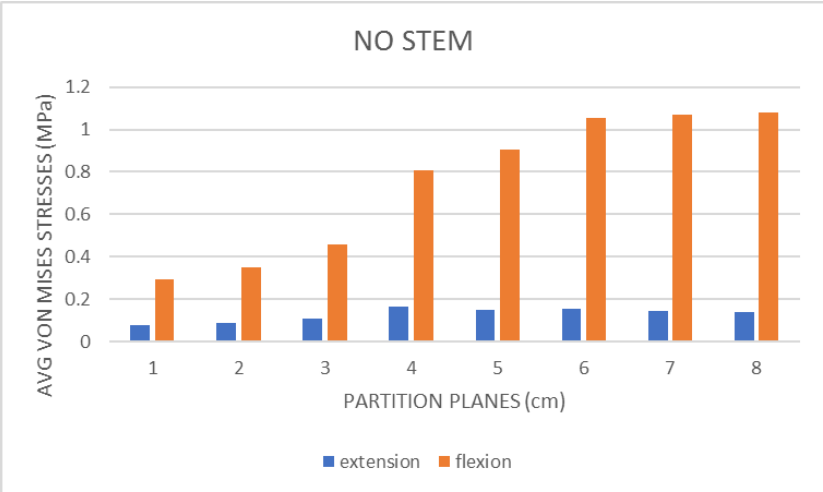


Fig.58 Average Von Mises Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration without the Femoral Stem

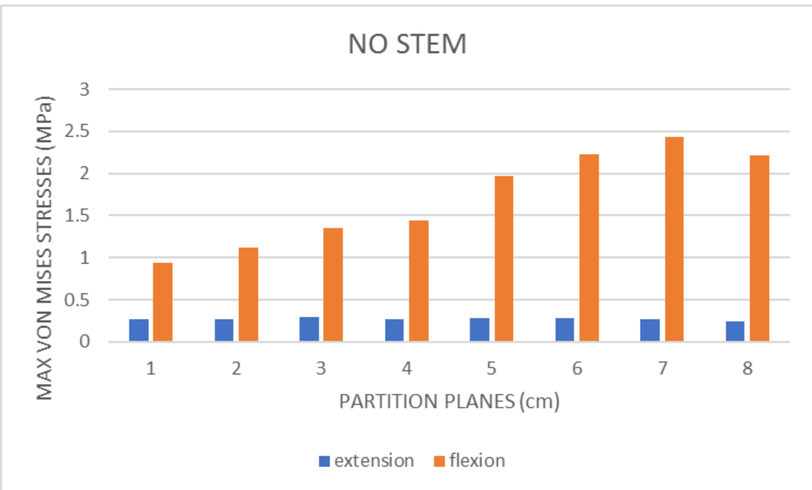


Fig.59 Maximum Von Mises Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration without the Femoral Stem

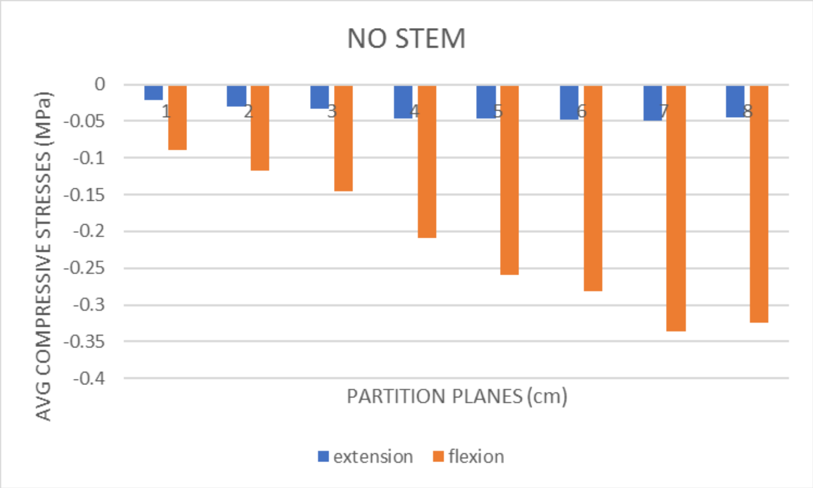


Fig.60 Average Compressive Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration without the Femoral Stem

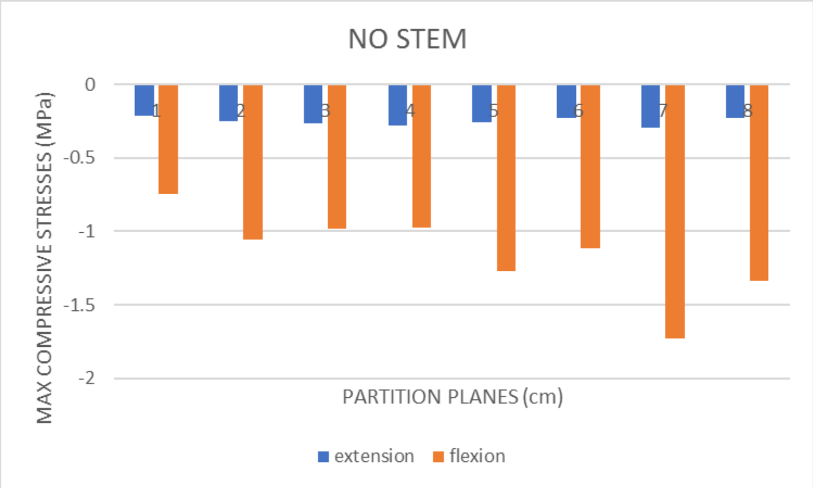


Fig.61 Maximum Compressive Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration without the Femoral Stem

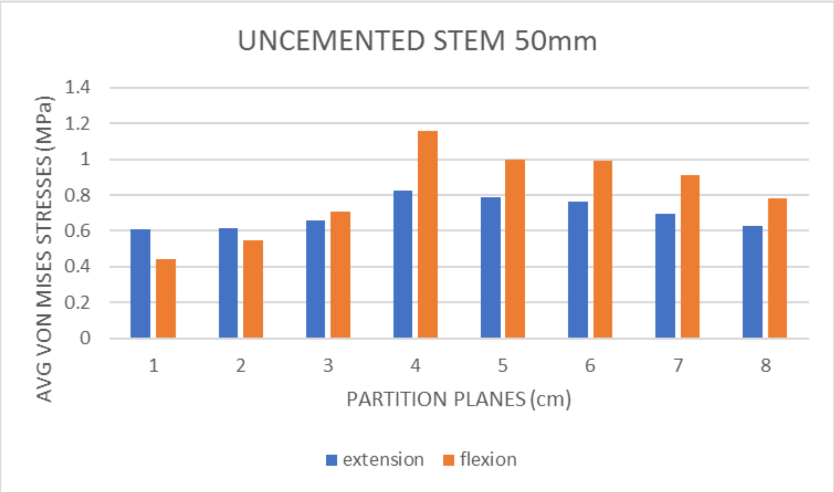


Fig.62 Average Von Mises Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration with the Uncemented Femoral Stem of 50mm

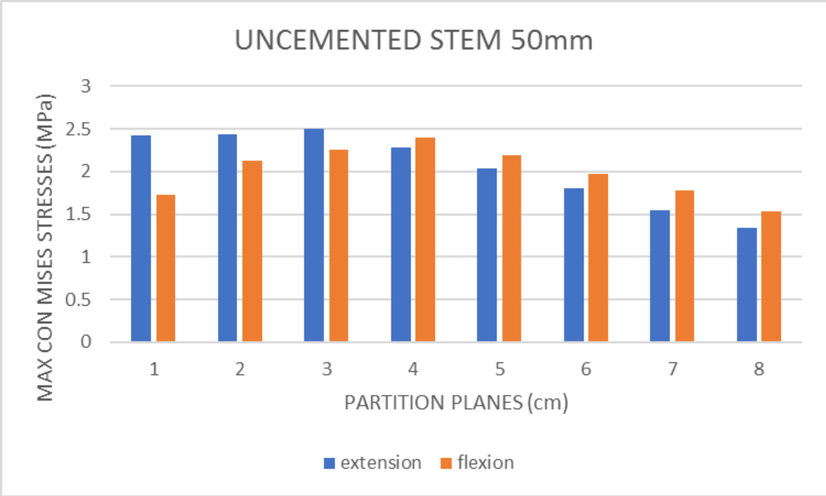


Fig.63 Maximum Von Mises Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration with the Uncemented Femoral Stem 50mm

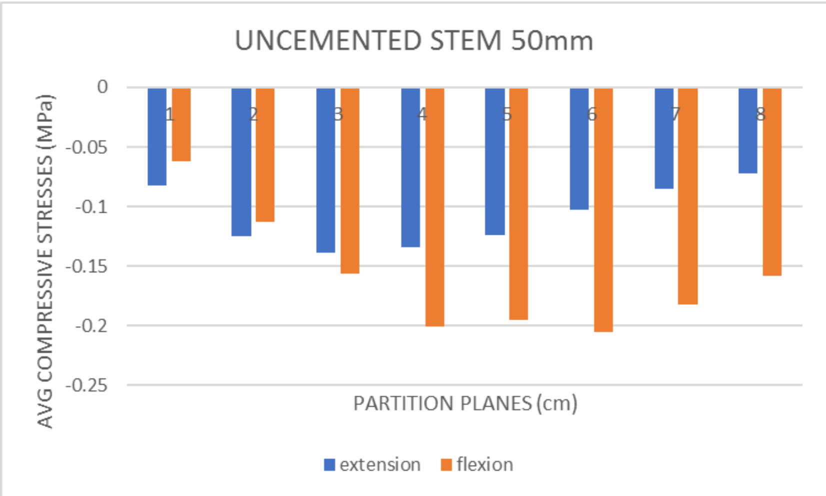


Fig.64 Average Compressive Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration with the Uncemented Femoral Stem 50mm

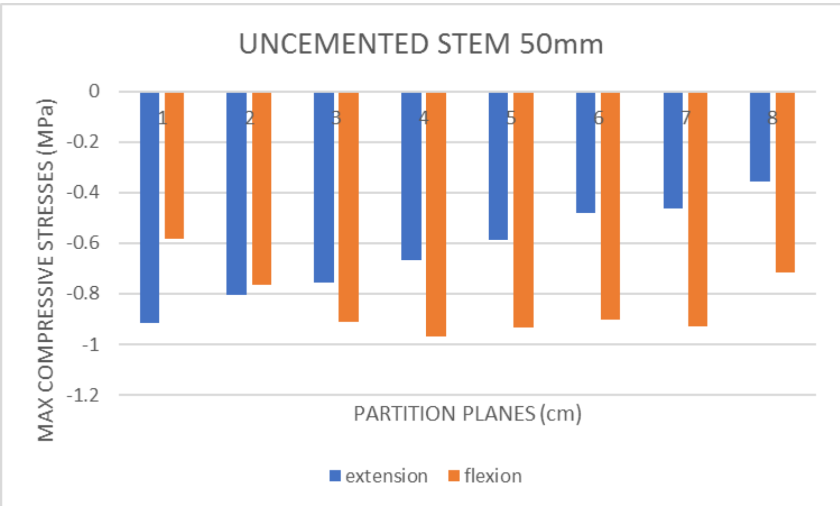


Fig.65 Maximum Compressive Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration with the Uncemented Femoral Stem of 50mm

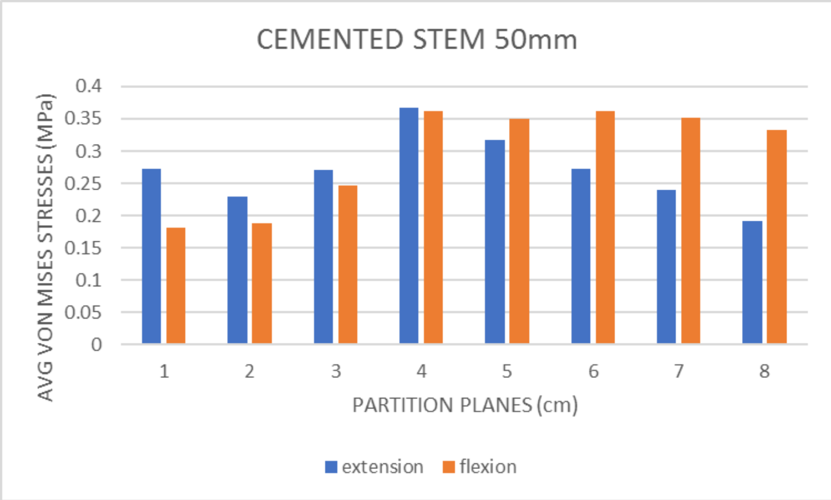


Fig.66 Average Von Mises Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration with the Cemented Femoral Stem of 50mm

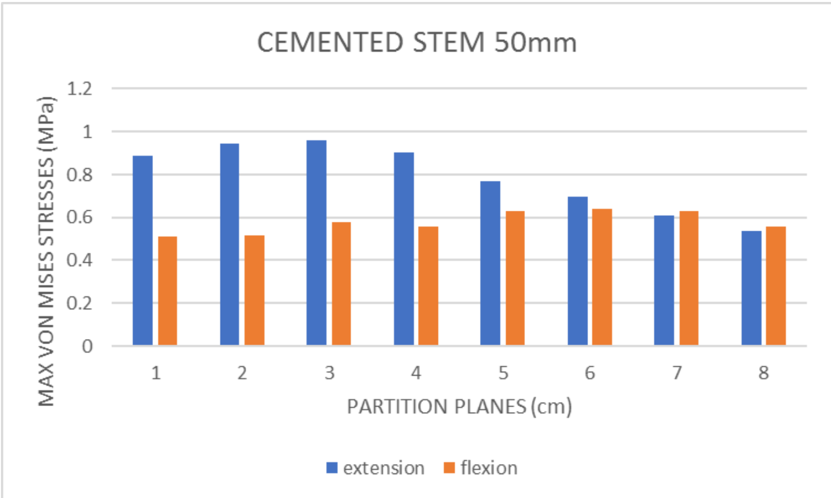


Fig.67 Maximum Von Mises Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration with the Cemented Femoral Stem of 50mm

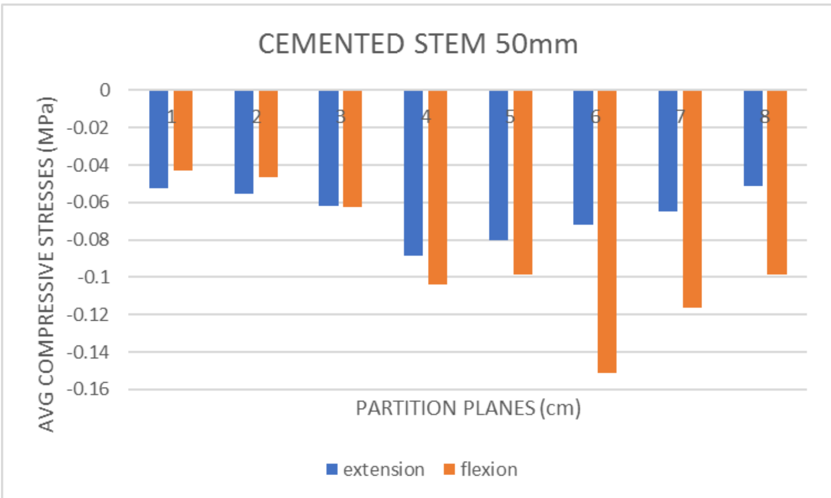


Fig.68 Average Compressive Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration with the Cemented Femoral Stem of 50mm

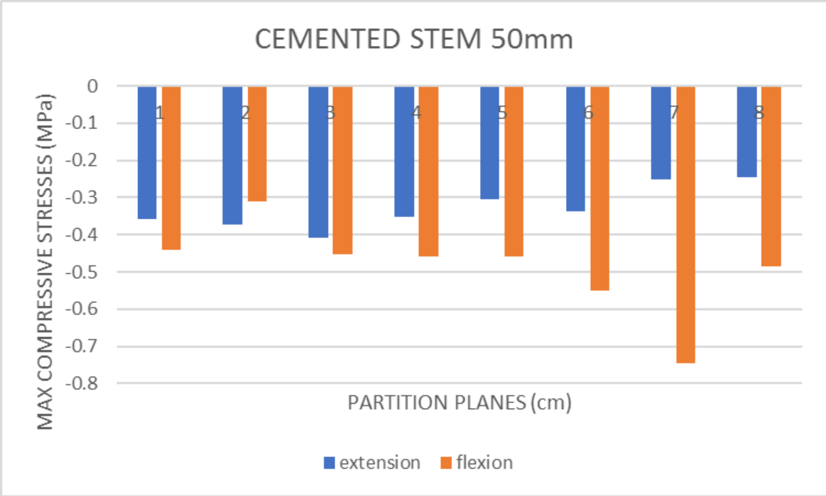


Fig.69 Maximum Compressive Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration with the Cemented Femoral Stem of 50mm

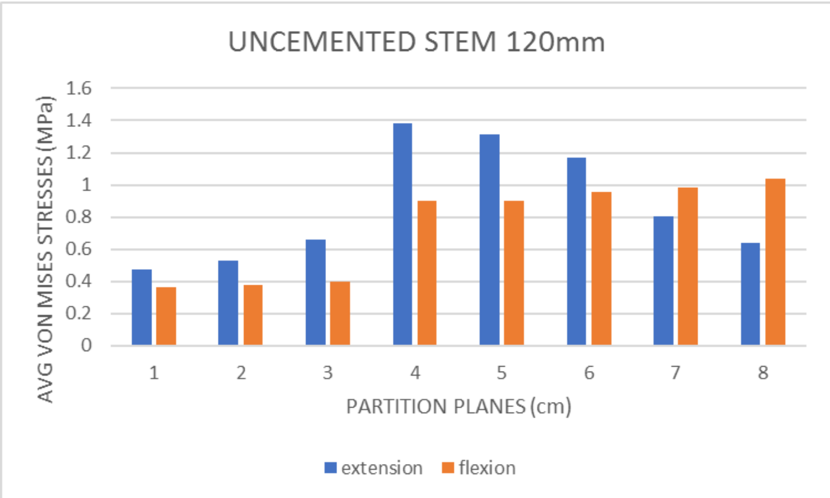


Fig.70 Average Von Mises Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration with the Uncemented Femoral Stem of 120mm

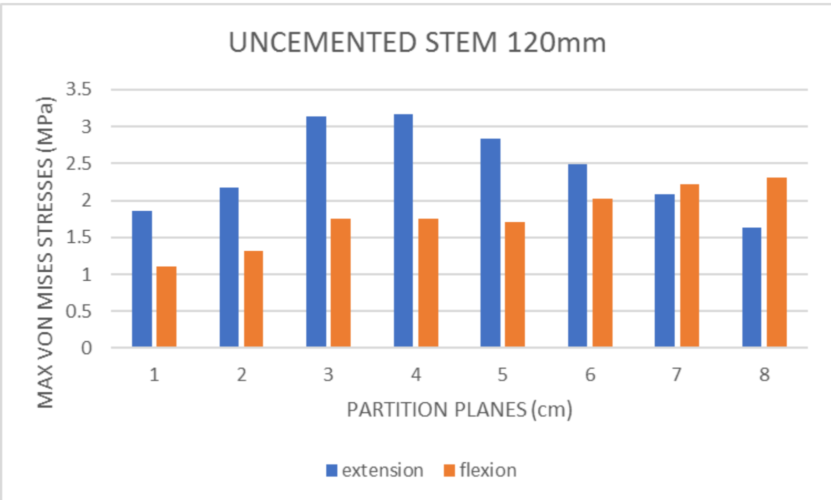


Fig.71 Maximum Von Mises Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration with the Uncemented Femoral Stem of 120mm

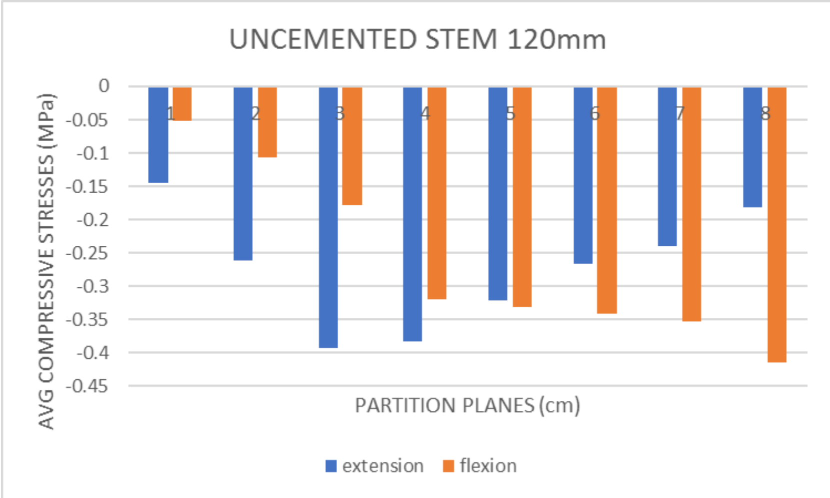


Fig.72 Average Compressive Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration with the Uncemented Femoral Stem of 120mm

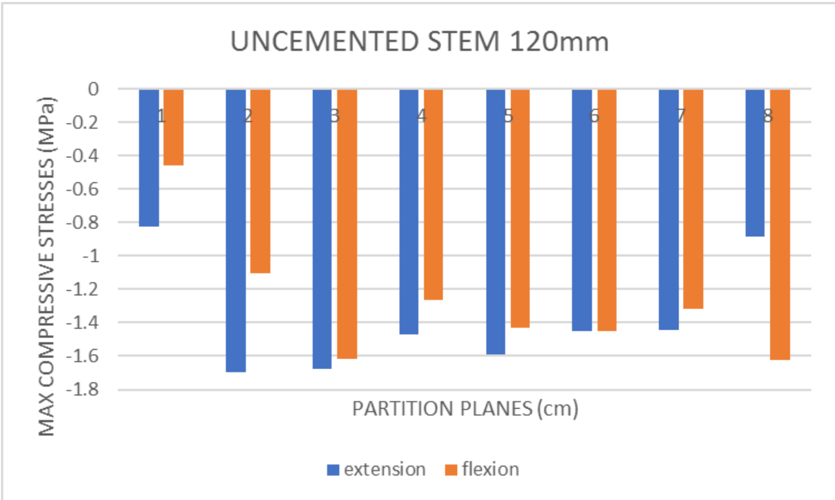


Fig.73 Maximum Compressive Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration with the Uncemented Femoral Stem of 120mm

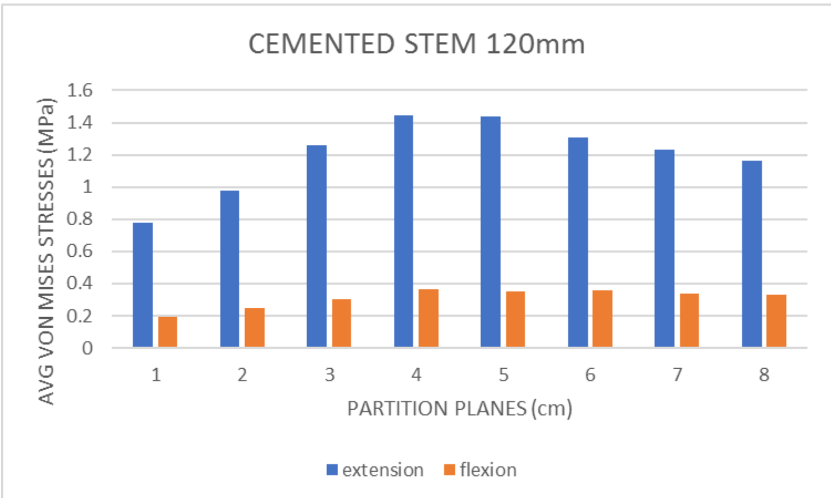


Fig.74 Average Von Mises Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration with the Cemented Femoral Stem of 120mm

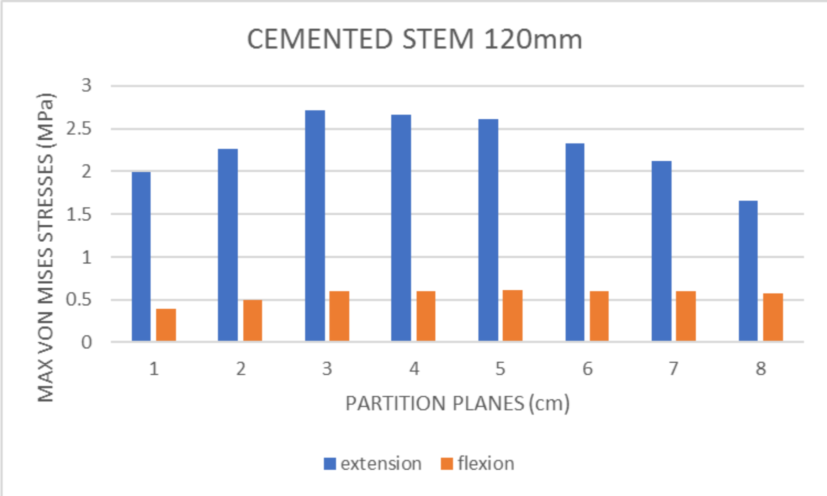


Fig.75 Maximum Von Mises Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration with the Cemented Femoral Stem of 120mm

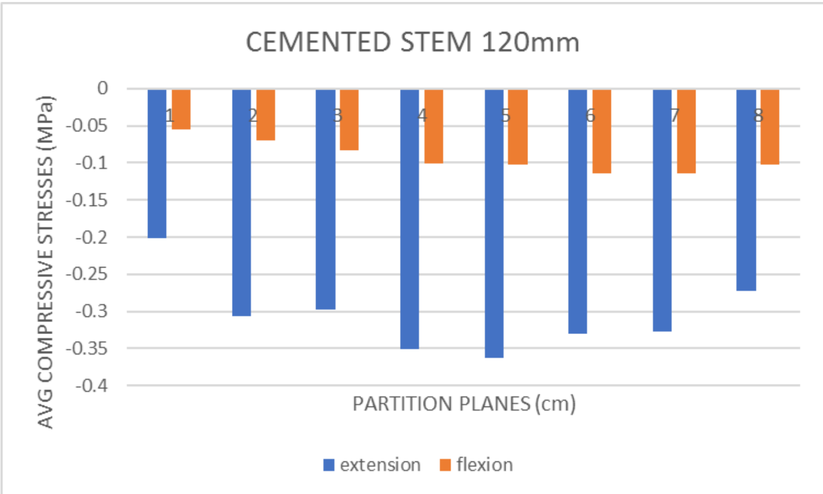


Fig.76 Average Compressive Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration with the Cemented Femoral Stem of 120mm

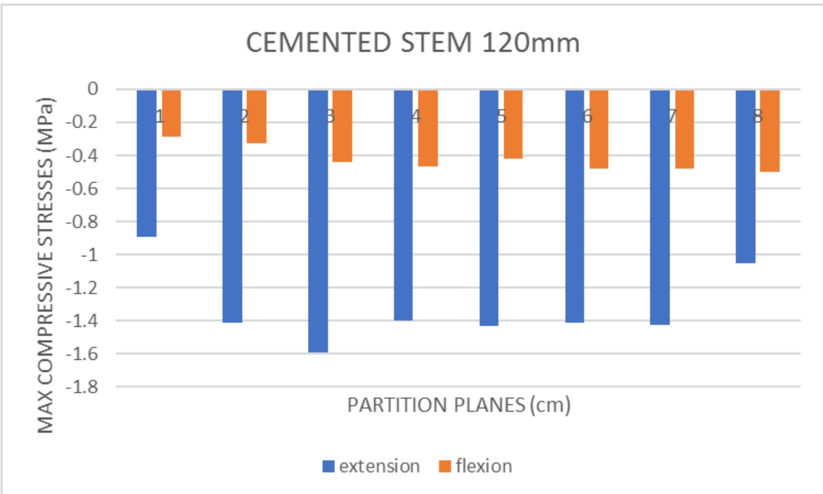


Fig.77 Maximum Compressive Stresses for the extension and the flexion configuration as a function of the Partition Planes of the Femur in the configuration with the Cemented Femoral Stem of 120mm

To better understand the quantitative results, the qualitative results have been shown below. They represent Von Mises stresses (**Fig.78, Fig.79, Fig.80, Fig.81**) on the femur and on a section of the femur, for each of the five configurations, respectively in the knee extension and flexion configuration.

Configuration at 0° of flexion

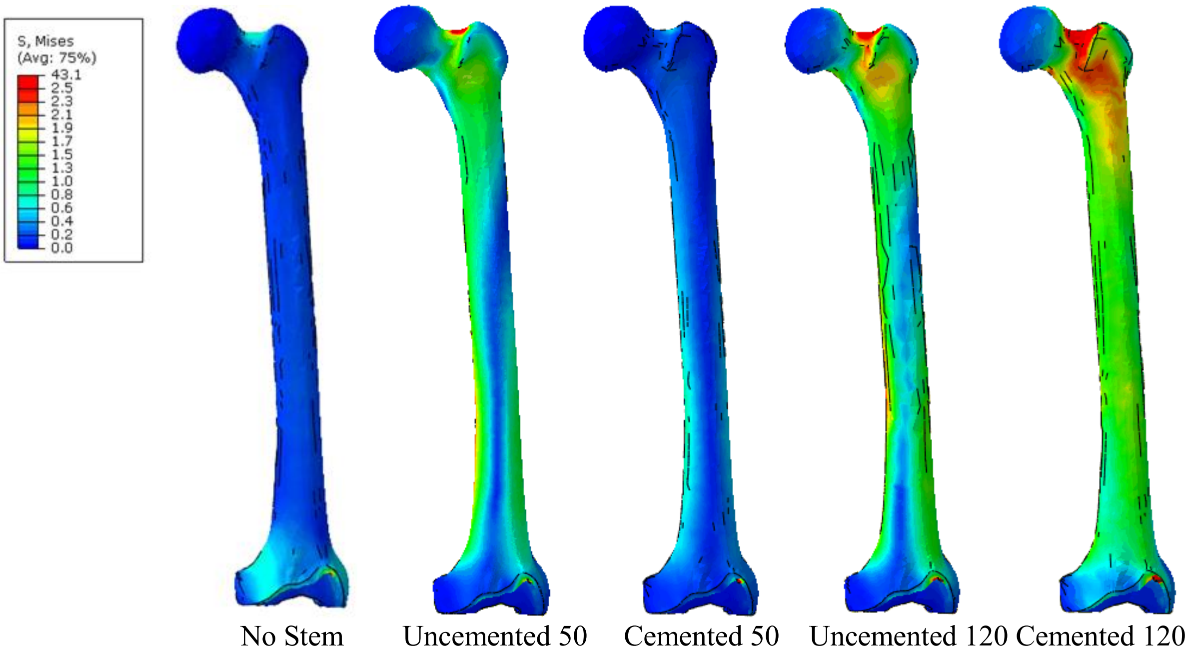


Fig.78 Qualitative Results: Von Mises Stresses on the Femur in the extension configuration

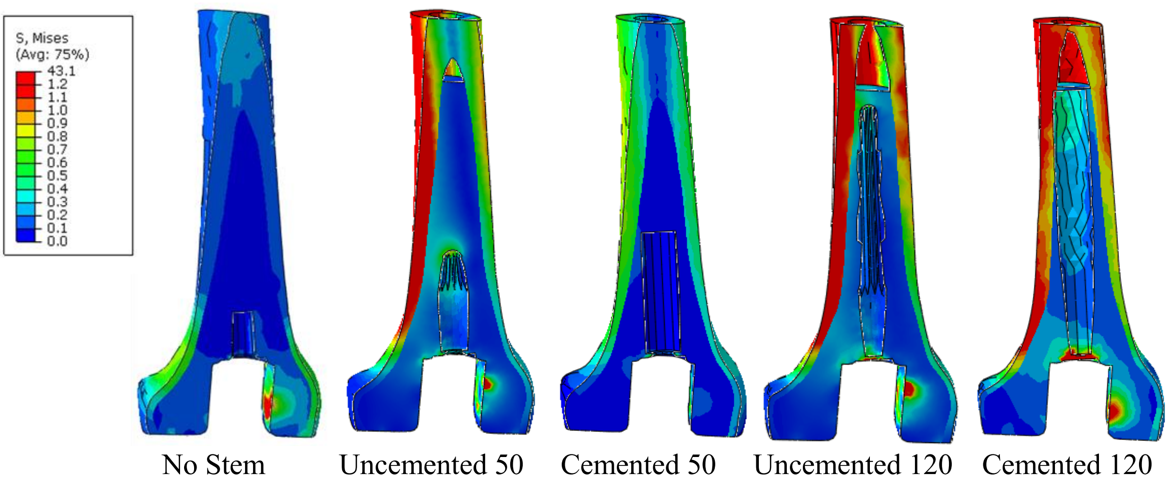


Fig.79 Qualitative Results: Von Mises Stresses on a section of the Femur in the extension configuration

Configuration at 90° of flexion

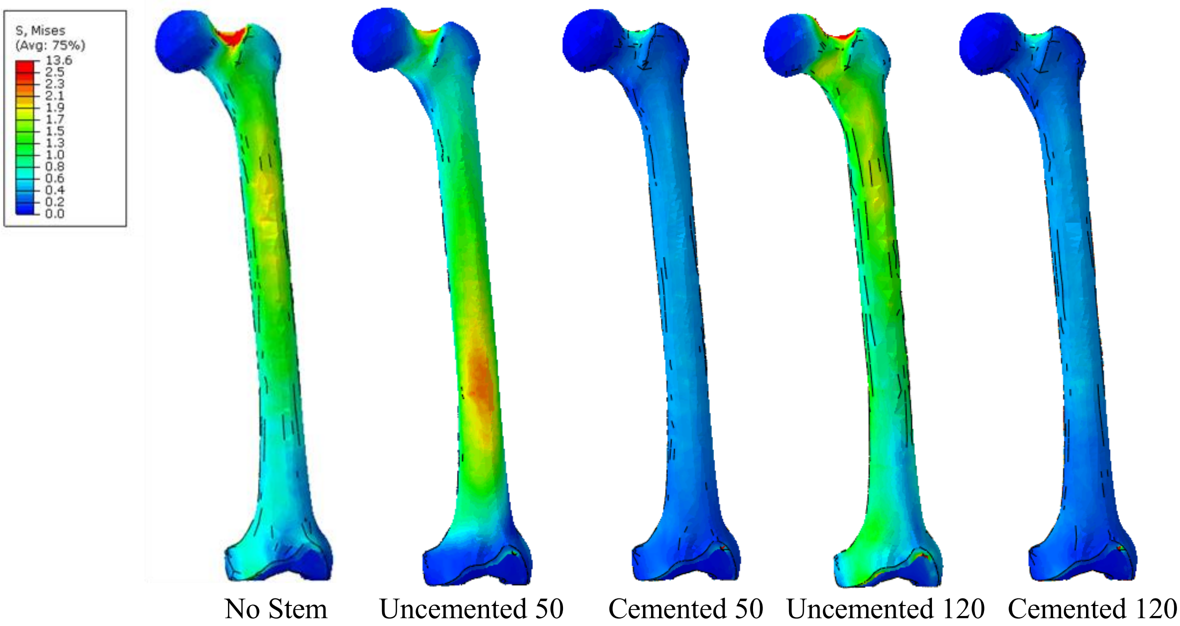


Fig.80 Qualitative Results: Von Mises Stresses on the Femur in the flexion configuration

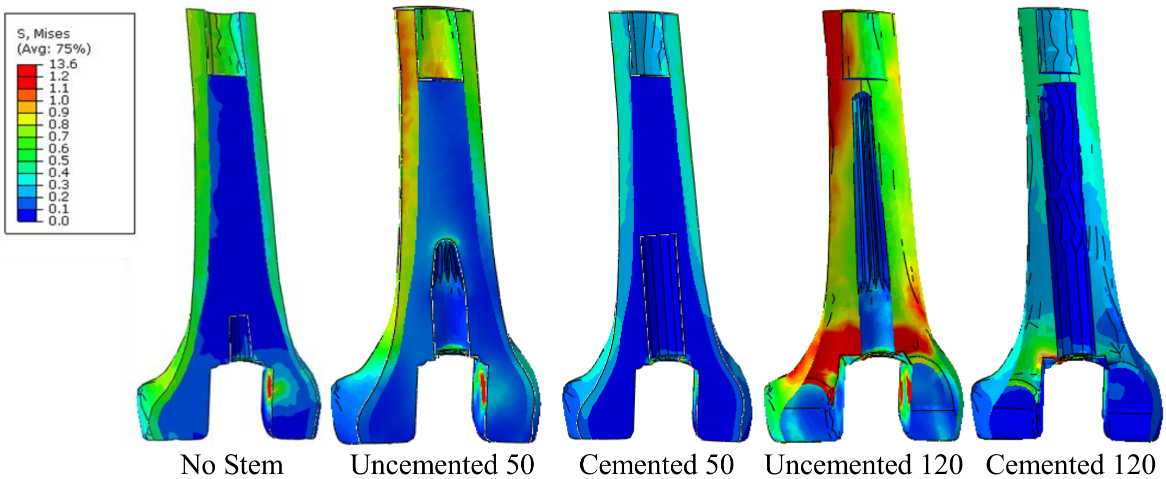


Fig.81 Qualitative Results: Von Mises Stresses on a section of the Femur in the flexion configuration

4 Conclusions and Future Developments

In the results, even if both stems with a length of 50mm and 120mm offer acceptable stress levels in the femur, the uncemented configurations, compared to the cemented ones, show higher Von Mises and compressive stress levels in the femur. In particular, the presence of the cement affects more the configuration with the stem with a length of 50mm, since the area of interaction between the femur and the stem is smaller than the configuration with the stem with a length of 120mm.

In the extension configuration the lowest Von Mises and compressive stresses appear in the configuration without the femoral stem, while in the flexion configuration the lowest Von Mises and compressive stresses appear in the configuration with the cemented stem with a length of 50mm.

Having said that, in the extension configuration the applied axial load is not such as to require the insertion of the femoral stem; while in the flexion configuration the lowest stresses appear when a short-cemented stem is used (in this study a stem with a length of 50mm).

Since today does not exist a guideline that shows the appropriate length stem and whether to use cement in a rTKR, these data will support the surgeon's decision making for the choice of stem length and fixation technique.

This, because the evaluation of stresses in different regions of the femur after revision is difficult in vivo as well as cadaveric models.

Consequently, these results seem to be best determined by Finite Element Analysis (FEA) with standardized sawbones having defined material properties for cortical and cancellous bone.

Possible developments could consider:

- How the Hinged Knee Prosthesis behaves in other movements other than the squat;
- Micromovements between the uncemented stem and the femur and between the cemented stem and the cement (more realistic study);
- Other measures of stem lengths (for example 90mm and 160mm), to get more results and a more complete view of the study;
- A pathological bone, affected by osteoporosis;
- Patellar ligament and patella in the model.

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Federica Armaroli

Lunedì 10 dicembre 2018

5 Figures

6 Tables

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