Analysis of the Bone Stresses for a Posterior Stabilized Knee Prosthesis with an Endomedullary Stem

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Abstract

Being not only the largest and the most complex joint of the human body, the knee is also the most vulnerable, therefore, particular unfavourable conditions, may lead to its structural damage, resulting in the limitation, or even loss of its function. In these situations, replacing the damaged joint by an artificial one, through the Total Knee Arthroplasty (TKA), enables the restablishment of its mechanical functionalities and stability, while relieving patient's pain, and giving them back their independency in mobility, and their life quality. However, postoperative complications can occur, requiring then a revision TKA, which consists of replacing the damaged components, and adding metallic stems in the bones to enhance the stability and fixation of prosthetic components, and consequently guarantee the stability of the knee joint as well, thereby leading to better outcomes and improved implant durability. Particularly for this study, the Fixed Bearing Posterior Stabilized (PS) Condylar Knee Prosthesis and the Fluted Stem model, were the chosen implants to create the Finite Element (FE) models of the knee after a TKA. For that, SolidWorks®, ABAQUS® and MATLAB®, were the main software used. Thus, different configurations of the joint were then numerically simulated under distinct mechanical conditions, and a comparison of the stress values along the knee bones was made between the four FE models under study. By doing this, it was possible to verify that the results obtained point to the fact that inserting a stem in the endomedullary canal of a bone, has a real influence on the stress values, not only of the bone where the stem was implanted, but also of the complementary one.

Keywords: Total Knee Arthroplasty, revision surgery, knee joint, stem, Von Mises Stress

1. Introduction

According to Jules P'eau (1830-1898), arthroplasty may be defined as "the creation of an artificial joint for the purpose of restoring motion" [1], and in fact, since the very early stages of the 19th century that several approaches have been developed and investigated to restore joint movements, while maintaining its stability and relieving pain. Furthermore, with the evolution of time and technology, together with the increased knowledge regarding knee biomechanics and kinematics, new approaches for knee replacement have been showing progressively better postoperative results [1]. Therefore, and despite being already a long history of repetitive cycles of new discoveries and failures, it was only during the 70s that important developments were made for the Total Knee Arthroplasty (TKA), having the Total Condylar Replacement demonstrated the most successful results. As a consequence, initial approaches such as hinged, unicondylar, and duocondylar prostheses ended up being abandoned [2].

Besides that, over the last decades arthritis has become one of the most common chronic diseases worldwide, affecting the patient's knee joint through joint stiffness, pain and decreased motion range. Therefore, when these symptoms become so severe that they become disabling, and consequently diminish patient's life quality, TKA is currently the standard treatment. Particularly in the United States (US), more than 600000 of TKA are performed annually, with a proven effectiveness and success in 80% of the cases [3]. Regarding implants life length, it has also been showing some positive outcomes, reaching 10 to 15 years [4]. However, TKA failure rates, and hence, the indication rates for revision TKA surgeries, remain greater than the desired. Due to the occurrence of complications deriving from the primary TKA (such as decreased bone quality or even severe bone loss), to increase the quality and consistency of the revision outcomes, the surgeons end up replacing the knee prosthesis that was previously implanted, and adding metallic stems in the medullary canal of the knee bones. Accordingly, the stems are often used in revision TKA surgeries to provide additional fixation to the implant, while ensuring a better alignment between the femoral and tibial components. Furthermore, by implanting these stems, the loads previously supported by the bone and the femoral and tibial

components decrease, which in a short-term analysis can lead to an increased implant longevity [5]. However, a long-term analysis, specially concerning the bone remodelling processes, would still be required to confirm this statement.

Since the quality of the revision postoperative results, as well as the durability of the implants, are currently a major concern in the world of the TKA surgeries, the main goal of this study is to understand whether the use of a femoral or a tibial stem, or even both at the same time, has a real impact and influence on the stresses of the knee bones. Precisely, a comparison between these three joint configurations after a revision TKA, and the one resulting from the primary procedure (without stems) will be done, in order to understand whether the fixation of the knee prosthesis with an endomedullary stem, has an influence on the stress values of the complementary bone. Additionally, it is intended to investigate which configuration may be a better option for the revision procedure, or at least, the less harmfull, to the joint and implant, in a short-term analysis.

2. Anatomy of the Knee Joint

The knee joint is the largest and the most complex joint in the human body that enables normal and daily activities such as standing, sitting and walking. Being described as a hinge joint, it is responsible for the flexion/extension movements, and for limiting lateral/medial rotations and gliding, while being articulated and stabilized by several muscles, ligaments and tendons. Nevertheless, it is still one of the most vulnerable joints of the skeleton, therefore, unfavourable conditions as aging, overuse and traumatic injuries, may lead to its structural damage, resulting in the limitation, or even, in the loss of its function.

Particularly, it consists of two distinct joints: the tibiofemoral (between the femur and tibia) and the patellofemoral (between the femur and patella). Additionally, the proximal ends of the tibia and fibula form the ligamentous tibiofibular joint as well.

Concerning the muscles, there are two main groups of it, the extensors and flexors of the leg. Then, regarding the ligaments, two main sets reinforce the joint structure: the lateral and medial collateral ligaments (LCL and MCL), and the anterior and posterior cruciate ligaments (ACL and PCL), respectively responsible for stabilizing the extended and the flexed joint. Particularly, the collateral ligaments work as support providers and as lateral/medial movements preventers, while the cruciate ligaments constitute the central pivot of the joint, connecting both femur and tibia within the joint capsule, and helping to maintain the proper alignment of the knee [6].

3. Fixed Bearing Posterior Stabilized Condylar Knee Prosthesis

For an improved design, longevity, stability and range of motion, the Hospital for Special Surgery (HSS) has developed the Posterior Stabilized (PS) Condylar Knee Prosthesis, which required the PCL substitution, and enabled the femur rollback during flexion, as well as the change in the femoral condyles center of curvature [7].

Precisely, for the primary TKA, the components of the PS prosthesis (Figure 1 (a)) consist of three main parts: the femoral component, the tibial tray, and the tibial insert (Figure 1 (b)). Particularly for this study, the P.F.C.® SIGMA® PS Femoral Component was made of cobalt-chromium-molybdenum (Co-Cr-Mo) alloy, and asymmetrically designed in left and right configurations. Concerning the tibial tray, it was of the fixed bearing type and made of highly polished Co-Cr, making the smooth surface more polyethylene friendly, while minimizing wear. The P.F.C.® SIGMA® PS Tibial Insert, was made of ultra high molecular weight polyethylene (UHMW) with improved wear properties, and enhanced resistance to the multidirectional stresses [8][9]. Finally, both tibial tray and femoral component can be fixed to the bone using cement made of polymethylmethacrylate (PMMA) [7].

(a) (b)

Figure 1: (a) SIGMA® PS Fixed Bearing knee prosthesis and (b) its components (Femoral component, Tibial Insert and Tray from the left to the right) [10]

Then, to further enhance joint stability and prosthesis fixation, additional stems that constrain even more the flexion and extension motions were developed for the revision TKA [7]. Moreover, as it is available in a variety of lengths and diameters, the P.F.C.® SIGMA® Modular System of stem components is then adaptable to each patient by choosing the modular component size that best fit the joint. Particularly for this study, the P.F.C.® SIGMA® Fluted Femoral and Tibial Stems used were, respectively, 175mm and 75mm long, and both with a diameter of 14mm, and made of titanium Ti-6Al-4V alloy F-136 [8].

4. Methods

New implant design methods, based on the use of medical images of patients' hard and soft tissues, were used in this study. For that, segmentation, computer assisted design (CAD) and Finite Element (FE) soft-

(a) (b) (c) (d) (e)

Figure 2: 3D models of the (a) Femoral Component, (b) Tibial Insert, (c) Tibial Tray, (d) Femoral and (e) Tibial Fluted Stems of the Fixed Bearing PS Knee Prosthesis.

ware, were used to build the three-dimensional (3D) bones and prosthesis models. Following this, numerical simulations were ran, enabling the study of the knee joint biomechanical behaviour.

4.1. Bone Image Acquisition, Segmentation, Smoothing and Modelling

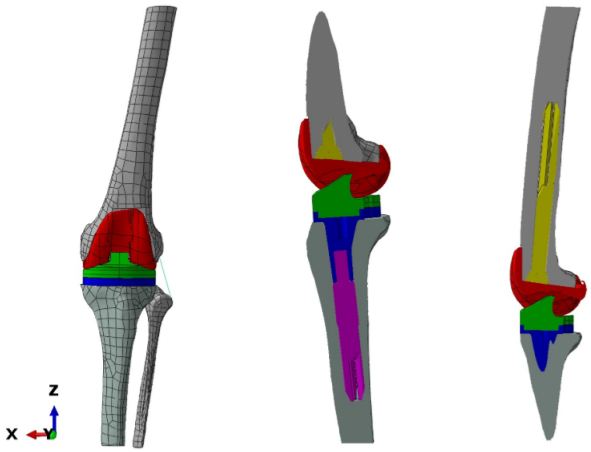
Primarily, 750 CT images from a female's left knee, provided by [11], were used for bone segmentation on ITK-SNAP. Additionally, MeshLab 2016.12 was then used to correct or at least attenuate the surface defects introduced by the 3D surface mesh generated by the segmentation. Then, the resulting bone models were imported to SolidWorks®, and its ScanTo3D® Toolbox was used to create the intended solid bone models. Finally, ABAQUS®, was used to generate their volume mesh.

4.2. Prosthesis Computational Modelling

For the creation of the 3D prosthesis model, SolidWorks® was used once again. Particularly, instead of modelling the entire knee prosthesis only based on the physical models, both femoral component and tibial insert were previously 3D scanned due to their complex geometries, thereby establishing a starting point for the following modelling process in CAD. By doing this, more accurate and better fitting models were then created. Having said that, the final 3D models are shown in Figures 2 (a)-(c). Finally, the 3D model of the stems (with the dimensions referred in section 3) were also created on SolidWorks® (Figures 2 (d) and (e)). Remarkably, small changes had to be done on the femoral stem design and size, so that it could fit the previously modelled femur.

4.3. Assembly Process

Following the bone scaling and the alignment correction between the bone models, the tibial resection cut was performed, perpendicularly to its mechanical axis, and 10mm distal from its proximal end. Then, the assembly process was initiated, with the insertion of both tibial tray and polyethylene component on the tibia, following the femoral component insertion on the femur, and the femoral resection cuts. Lastly, after being imported to ABAQUS®, the fibula and ligaments were finally assembled to the model, as well as the fluted stems, resulting in the complete 3D anatomical model of the knee joint presented in Figure 3.



(a) (b) (c)

Figure 3: (a) Anterior, and (b) Lateral and (c) Medial cut views of the final 3D knee joint model with insertion of the prosthesis. Bones in grey, femoral component in red, tibial insert in green, tibial tray and the ligaments, respectively, in dark and light blue, fluted tibial and femoral stems in purple and yellow.

4.4. Numerical Model

As a starting point for the numerical analysis through the Finite Elements Method (FEM), the creation of the numerical models of the bones and implant was primarily done on ABAQUS®. Thus, it started with the mesh generation on both original and final 3D bone models (without and with implant insertion, respectively), precisely, with linear tetrahedral elements. Then, the ABAQUS® Bonemapy plug-in was used to extract the apparent density values of each mesh node of the original bones (μ_i), followed by their convertion into values of the final FE model (μ). For that, previous operations were performed on MATLAB®, including the mesh mapping between these two FE models, and the following weighted average density calculation of all original nodes contained in a small sphere with a radius of

2mm (r_{min}) (and at a distance $dist_i$ from the final node), according to the equations 1 and 2 presented below:

$$\mu = \frac{\sum_{i=1}^{n} w_i * \mu_i}{\sum_{i=1}^{n} w_i} \tag{1}$$

$$w_i = r_{min} - dist_i \tag{2}$$

The obtained μ values were then attributed to the femur and tibia. Remarkably, for their surface nodes, the maximum density value of 1.8g/cm^3 was attributed to, to guarantee the surface cortical bone. Beyond these, FE models of the ligaments were also created (with a wire shape and a truss section), and assembled to the final FE model. The geometric mesh characterization of the bones and ligaments is included in Table 1.

Having the meshes, the material parameters were then attributed to each component of the FE model. Therefore, the Young's Modulus (E) of the bones was primarily calculated, according to the equation 3 presented below [5], and the remaining values were given according to the literature (Table 2):

$$E = 3790 * \mu^3 \tag{3}$$

Particularly for the fibula, an elastic, isotropic and homogeneous bone material was considered, with the same parameters as the ones indicated in Table 2.

Besides these, interactions and constraints were defined between the bones and the prosthetic components. Precisely, for the interfaces femur/femoral component, tibial insert/tibial tray, and tibia/tibial tray, a tie constraint was applied, while for the interactions femur/femoral stem, femoral component/tibial insert, tibia/tibial keel, and tibia/tibial stem, a surface-to-surface contact was defined. Additionally, for the surface contacts, a value of 0.30 was considered for the friction coefficient, excepting for the interface femoral component/tibial tray, to which a value of 0.04 was attributed to [15]. Furthermore, for the ligaments insertion points, coupling constraints between their ends and the bone surfaces were also defined, as well as their influence radius, as indicated in Table 3. Moreover, to further constraint the models, two pinned boundary conditions were applied on the distal ends of the tibia and fibula, and five springs were introduced in the center of the joint to avoid anterior/posterior and medial/lateral displacements (both with stiffness $k = 10^{10}$), and excessive bending, lateral flexion and torsion in the corresponding directions (being $k_x = 10^6$, $k_y = 10^8$ and $k_z = 10^{10}$).

Finally, to simulate the normal walking situation, three load conditions were included: the ligaments were subjected to a predefined field of 1MPa; a concentrated force of 400N was applied on the femur proximal end, with the same direction as its anatomical axis; and a moment of 10kNm was also applied

Table 1: Characterization of the FE mesh generated on the numerical models of the knee bones and ligaments.

	Number of Nodes / Number of Elements			
	Femur	Tibia	Fibula	Ligaments
Without Stems	16664 / 85585	21362 / 111367	5473 / 24453	2 / 1
Only Tibial Stem	28096 / 148228	21945 / 110935	5485 / 24504	
Only Femoral Stem	34068 / 176993	21487 / 111904	5485 / 24504	
With Both Stems	20247 / 102142	21808 / 110204	5485 / 24504	

Table 2: Material type and the corresponding properties and parameters, Young's Modulus and Poisson's Ratio attributed to the final FE model of the knee joint, prosthetic components and stems.

	Model Part	Material Type	Material Property	Young's Modulus (MPa)	Poisson's Ratio	References
Bones	Femur Tibia	Bone	Elastic, Isotropic and Non Homogeneous	0.00379 - 22103.28	0.30	
Ligaments	LCL	Ligament	Elastic and Isotropic	183.5	0.30	[12]
	MCL			332.2		[13]
	PTL			259		[14]
Prosthesis	Femoral Component	Co-Cr-Mo Alloy	Elastic and Isotropic	200	0.30	[15]
	Tibial Insert (Polyethylene)	UHMW Polyethylene		780	0.40	[16]
	Tibial Tray	Co-Cr-Mo Alloy		200	0.30	[15]
	Femoral Stem Tibial Stem	Titanium Ti-6Al-4V Alloy		115000	0.33	[17]

Table 3: Description of the FE models of the ligaments of the knee joint, including their cross-section area, the interaction of their insertion points with the corresponding bones, and their influence radius.

Ligament	Bone	Ligament Cross-section Area (mm^2)	Influence Radius of the Insertion Point (mm)	References
LCL	Femur	11.9	7	[18]
	Fibula		6.5	[12]
MCL	Femur	25	9	[19]
	Tibia		15	
PTL	Tibia	7	6.4	[14]
	Fibula		9.7	

Table 4: Characterization of the four FE models of the knee joint analyzed and compared in this study.

Knee Joint Model	Simplified Designation	Stem Dimensions	Number of Elements	Number of Nodes
Femoral Component	Reference		47844	10433
Tibial Insert			24733	7508
Tibial Tray			15758	3984
Femoral Component	Tibial Stem		60502	13168
Tibial Insert			7051	10659
Tibial Tray			96146	20860
Tibial Stem		Ø14mm x 75mm	13486	3193
Femoral Component	Both Stems		60502	13168
Tibial Insert			7051	10659
Tibial Tray			16673	4241
Tibial Stem		Ø14mm x 75mm	12238	2985
Femoral Stem		Ø14mm x 175mm	20413	5570
Femoral Component	Femoral Stem		60502	13168
Tibial Insert			7051	10659
Tibial Tray			17455	4441
Femoral Stem		Ø14mm x 175mm	22151	6184

on the same location, but on the x direction.

5. Results and Discussion

In the following sections, the effects of inserting a stem in the intramedullary canal of a bone, and its further influence on the stresses of the complementary bone, will be analyzed using FEM and ABAQUS®. Therefore, different knee joint configurations will be numerically simulated and compared.

5.1. FE Knee Joint Models

Having in mind the study done by Completo [5], and the fact that the need for a tibial stem insertion is the most frequent indication leading to a revision TKA, the knee joint configuration with insertion of the femoral stem only, was not considered at the initial stages of this study. Therefore, the knee joint models that were primarily analyzed and biomechanically compared were: the one resulting from the primary TKA, without stems (Reference model); and two possible configurations resulting from the revision TKA, particularly, with insertion of the tibial stem only (Tibial Stem model), and with the insertion of both femoral and tibial stem (Both Stems model). Notably, the considered complementary bone will be the femur, except for the subsection 5.1.4. Finally, the characterization of these FE models is presented in Table 4.

5.1.1. Analysis of the Von Mises Stress Distribution

Being the load and stress distribution an important concept to investigate the performance and effects of implants on the human body structures, the Von Mises stress distribution along the femur was analyzed. Remarkably, the same numerical conditions were applied to every model (section 4.4), so that a consistent and coherent biomechanical comparison of the results could be made.

Thus, recalling the equation 3 and the Hooke's Law (equation 4), it is clear that the higher the bone density, the higher the Young's Modulus (stiffness), meaning higher stress (σ) levels for the same deformation (ε):

$$\sigma = E * \varepsilon \tag{4}$$

Accordingly, it was then expected to observe higher Von Mises stress values in the regions with greater bone density and E [20]. In fact, the higher stresses observed in the cortical bone regions (shown in Figure 4), and the consequent lower stresses in the trabecular bone were in agreement with the expectations.

Following the same principles, when inserting a stem, it is expected that the stress previously absorbed by the bone, is then transferred to the stem, as it has greater stiffness. In fact, by particularly observing the Both Stems model case, it is clear that the expected results were obtained, having the stem per-

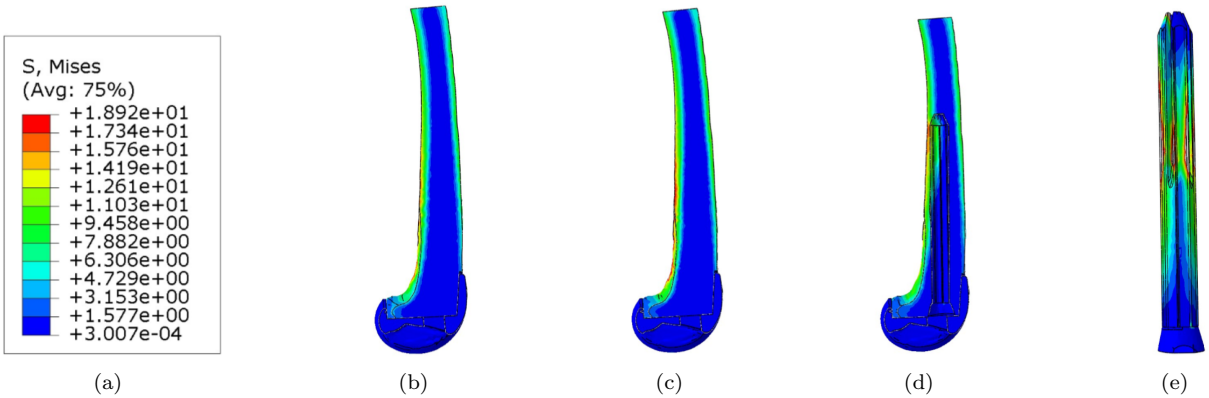


Figure 4: Von Mises stress distribution along the femur, femoral component and stem, for the three first models referred in Table 4. (a) In [MPa]. Sagittal cut views of the (b) Reference, (c) Tibial Stem (d) Both Stems (without the femoral stem displayed), and (e) Femoral Stem used in the Both Stems Model.

formed its role absorbing the major stress, thereby relieving the trabecular bone. Additionally, both stress shielding and stem tip stress phenomena were observed as well, having a decrease in stress around the majority of the stem length, and an increase around its ends. To conclude, the results obtained clearly point to the fact that inserting a stem actually influences the stress distribution, not only on the bone where it was implanted, but also on the complementary bone.

5.1.2. Stem Influence on the Bone Stress Values

To further investigate the stem influence on the nodal stress values of the complementary bone, a comparison of the Tibial Stem and Both Stems model, with the Reference one was done. For that, the absolute and relative deviations of the stress nodal values (σ_f) of each final FE model (Tibial Stem and Both Stems), in relation to the ones of the Reference model (σ_i), were analyzed and calculated according to the equations 5 and 6:

$$AbsoluteDeviation = \sigma_f - \sigma_i \tag{5}$$

$$RelativeDeviation = \frac{(\sigma_f - \sigma_i) * 100}{\sigma_i} \tag{6}$$

Absolute Deviation of the Von Mises Stress Looking first to the Figures 5 (b) and (c), it is clear that, the insertion of the tibial stem caused an increase in the Von Mises stress values of the large majority of the diaphysis nodes, as well as around the femoral component. Additionally, the decrease in stress has mainly occured on the posterior femoral cut regions and on both lateral and medial sides of the femur. Finally, looking to the Figure 5 (d), it is evident that, in the proximal half of the femur, the stress values have mainly decreased in the medullary canal region, and increased in the cortical bone regions.

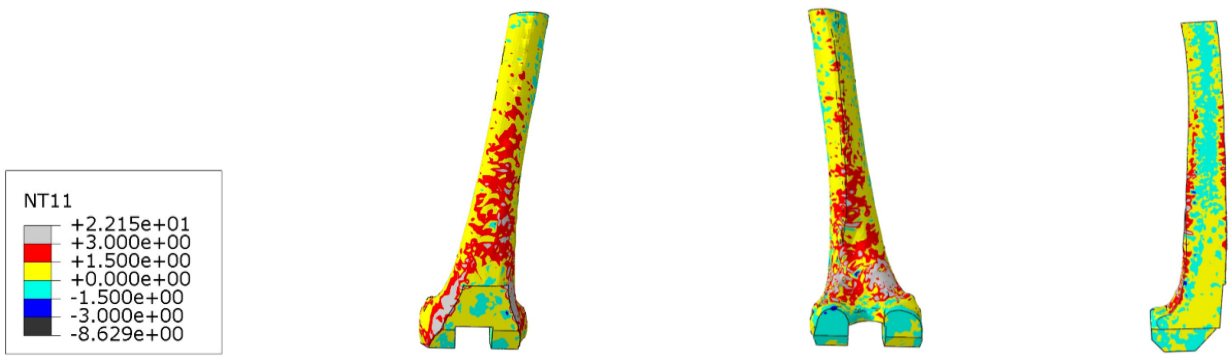
Turning to Figures 6 (b) and (c), it is visible that the insertion of both tibial and femoral stems, has mainly caused a decrease in the stress values of the surface nodes. Particularly on the posterior surface, the stress decrease is greater close to the lateral side. Additionally, the expected stress shielding and stem tip stress phenomena can also be verified in Figure 6 (d).

To conclude, from the analysis done it seems clear that inserting one (tibial) or both stems, actually causes several changes not only on the general stress distribution, as seen in the subsection 5.1.1, but also on its nodal values.

Relative Deviation of the Von Mises Stress Comparing the analysis done for the absolute deviation values with the Figures 7 and 8, it is possible to conclude that, for both knee joint configurations, the regions presenting greater variations on the Von Mises stress values, in module, are also the ones presenting more significant changes, when comparing to the initial stress values (on the Reference model). Therefore, the positive and negative relative deviation regions are in accordance with the positive and negative absolute deviation regions. Thus, the final conclusion stated before, based on the absolute deviation, can now be verified once again, by the relative deviation results.

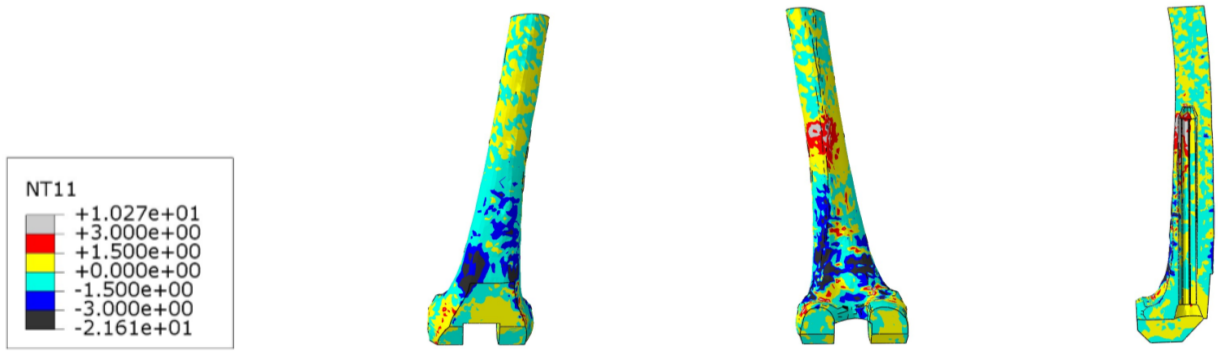
Nodal Analysis To verify the apparent conclusions stated above, the same deviation analyses were performed once again, but now only on six individual nodes of the femur surface. Notably, the chosen nodes had approximatelly the same location on every FE models, so that a coherent comparison between their stress values could be made.

By simply looking to the nodal Von Mises stress values, it was clear that they have changed after the insertion of one or two stems, in comparison to the Reference model. Then, by the nodal absolute de-



(a) (b) (c) (d)

Figure 5: Absolute Deviation of the Von Mises stress values between the Tibial Stem and the Reference models. (a) In [MPa]. (b) Anterior, (c) Posterior and (d) Sagittal views of the femur.



(a) (b) (c) (d)

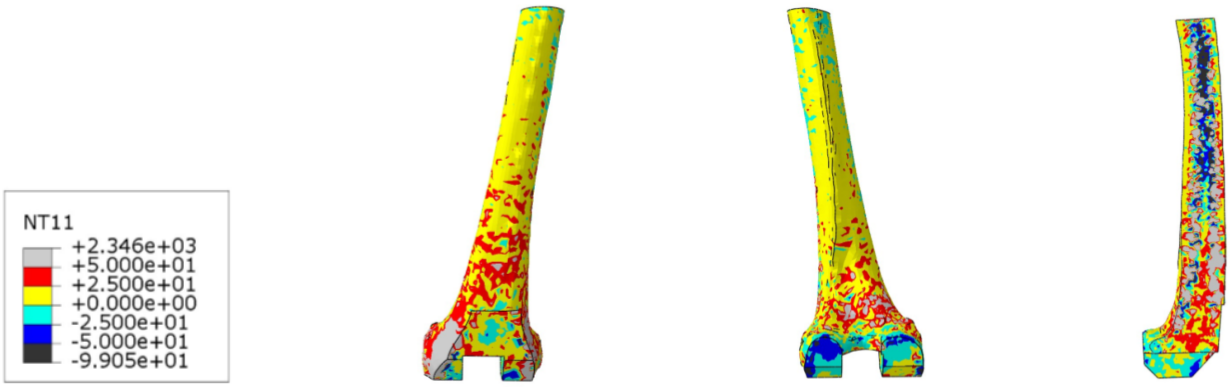
Figure 6: Absolute Deviation of the Von Mises stress values between the Both Stems and the Reference models. (a) In [MPa]. (b) Anterior, (c) Posterior and (d) Sagittal views of the femur.

viation values, it was also clear that these changes in stress were not constant between the cases under study (Tibial Stem and Both Stems), meaning that, different deviation values were obtained depending not only on the node location, but also on the FE model. Therefore, inserting only one or two stems in the bone medullary canal seems to actually produce distinct effects on the complementary bone. Precisely, when analyzing the relative deviation values, it was evident that the impact of these changes were also distinct between them, presenting significant variations in their orders of magnitude. Still, in most cases a significant influence on the bone stress was observed.

5.1.3. Influence of changes in the Numerical FE Model on the Bone Stresses

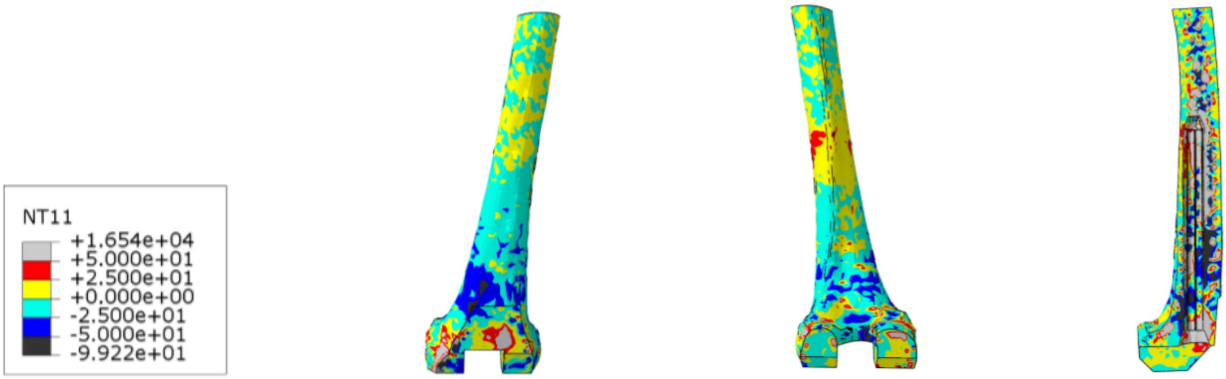
Being aware that the FE models that have been created to perform the previous analyses, were very simple and minimalist versions of the real knee joint, the influence of the assumed simplifications as well as the impact of the variables applied to circumvent them, were then investigated. Therefore, some changes were performed in the constraints and load conditions previously applied, and the exact same nodal analyses described above were then repeated and compared to the original ones. As the most evident simplification was the fact that not all ligaments, neither muscles, have been included in the FE models, the spring constraints, as well as the concentrated force and the moment applied, were the ones that were subjected to the numerical changes. Thus, the torsion springs stiffness was primarily increased, ended up with $k_{x,y,z} = 10^{13}$. Precisely, for the Tibial Stem model, no significant changes were observed on both absolute and relative deviation results. Still, a very small reduction in the stress increments previously observed was verified, which may be explained by the actual increase in the springs stiffness. Concerning then the Both Stems model, no remarkable changes were observed on both deviation results.

Following this, a second change was done in the concentrated force magnitude, having increased from 400N to 1000N. Remarkably, the springs stiffness, as well as the moment magnitude, were all kept the same as the ones applied on the original knee joint models. Actually, and according to what was expected, the load magnitude increment caused a general increase in the nodal Von Mises stress values. Consequently,



(a) (b) (c) (d)

Figure 7: Relative Deviation of the Von Mises stress values between the Tibial Stem and the Reference models. (a) In [%]. (b) Anterior, (c) Posterior and (d) Sagittal views of the femur.



(a) (b) (c) (d)

Figure 8: Relative Deviation of the Von Mises stress values between the Both Stems and the Reference models. (a) In [%]. (b) Anterior, (c) Posterior and (d) Sagittal views of the femur.

increments in the absolute and relative deviation values, in module, were also verified. Finally, and comparing the Tibial Stem and Both Stems models results, it was also evident by the nodal analysis that, for the majority of the nodes under study, the load increment resulted in greater absolute and relative deviations values (in module) for the Tibial Stem, than for Both Stems.

Finally, the third change was to investigate the moment influence, and a change on its magnitude value was not performed this time, but two additional ones were introduced in the model, on the y and z directions, being both equal to 0.250Nm. Additionally, and contrarily to what was previously done, the springs stiffness had to be slightly changed this time, for the simulation to converge. Hence, it was possible to conclude that the additional lateral flexion and torsion moments did not cause significant changes on the stress values of the femur. Particularly, the values were actually kept approximately the same or in some cases they have only presented a slight decrease. Furthermore, it was also clear by the nodal analysis that the changes between the new relative deviation values and the original ones were much higher, in module, for the Tibial Stem than for the Both Stems, meaning that, as previously observed for the load change analysis, the additional moments had greater impact on the Tibial Stem model.

5.1.4. Final Remarks considering also the Femoral Stem Model

After analyzing the Tibial Stem and Both Stems models, and as an attempt to verify the results presented by [5], the Femoral Stem model (referring to the femoral stem insertion only, as indicated in Table 4) was then analyzed as well. Therefore, the influence of the femoral stem insertion was studied on the complementary bone, which now refers to the tibia instead of the femur. Thus, to perform the same nodal analyses described above but now for this new case, other six nodes were chosen on the tibia surface, and the absolute and relative deviation of the Von Mises stress values were obtained (comparing again with the Reference model). Hence, the deviation results on the tibia pointed to the fact that the femoral stem has also caused several significant changes on

its nodal stress values.

Taking now into account all four knee joint configurations, a more general analysis of the stem influence on the complementary bone was done. To primarily compare the Tibial and Femoral stem models, both femur and tibia had to be respectively observed, resulting on the verification that the impact of the stress changes caused by the stems was much less significant for the Tibial Stem model (being the relative deviation mainly between 0 and 25%) than for the Femoral Stem (which was mainly of the order of 50% or higher). Therefore, these may lead to the apparent conclusion that in fact, the Femoral Stem configuration may not be a good choice for the revision TKA, as stated in the Completo's study [5], since it causes much more significant changes on the stress values of the complementary bone.

Additionally, when taking into account the Both Stems model as well, and when comparing it with the Tibial Stem and Femoral Stem models, it was possible to verify that, for the majority of the chosen nodes, the Both Stems configuration have shown what in a short-term analysis can be considered as less harmful variations to the tibia, meaning that, the increments in stress are not so significant, or not significant at all. In addition to this, it was also visible that the decrease in stress was also more significant for the Both Stems model, which may be considered as a better result, again in a short-term point of view. Therefore, the Completo's statement that, whenever a femoral stem is required, it is better to proceed to the insertion of both femoral and tibial stems than to the insertion of the femoral stem only, regardless of the tibia condition [5], seemed to be once again verified, but now concerning the Von Mises Stresses. However, this may be stated only for a short-term analysis, thereby requiring a long-term analysis to enable the confirmation of this statement with certainty, namely, a bone remodelling analysis.

6. Main Conclusions

Following the analyses described in the subsections 5.1.1 and 5.1.2, it was possible to verify that the results obtained point to the fact that inserting a stem in the medullary canal of a bone influences the stresses, not only on the bone where the stem was inserted, but also on the complementary one. Particularly, for the case of the Both Stems model, the impact of the observed stress changes was stronger than for the Tibial Stem, which in turn may be explained by the fact that, in that case, the femur has actually a stem inserted as well, while for the Tibial Stem model, the bone is kept intact. Despite concluding that these changes are in fact quite significant, it seems that these may not be totally harmful to the femur, at least in a short-term analysis, since they evolve towards what is bearable by the bone, given its constitution and the relations suggested by the equations 3 and 4, meaning that the major stress increments occur in the cortical bone regions, thereby releaving the trabecular regions that are more susceptible to deformation and damage. However, a long-term analysis would still be required to further confirm this statement, particularly, an analysis concerning the changes produced on the bone remodelling processes.

Beyond these, and regarding the attempt to verify whether the simplifications assumed in the FE model could badly influence the numerical results, it was possible to conclude that only the increase in the load magnitude was able to produce significant variations on the Von Mises stress values. Furthermore, when comparing the same effects on both Tibial Stem and Both Stems models, it was clear that whenever some variations were observed, these were actually more significant on the configuration with only the tibial stem inserted, which may be once again explained by the femoral stem included on the Both Stems models.

Turning to the last analysis described in subsection 5.1.4, it was clear that, similarly to what was previously observed, implanting only a femoral stem produces also some variations in the Von Mises stress values of the complementary bone, being actually quite significant in some regions. Furthermore, the obtained results have also pointed to the fact that the Both Stems configuration may be a better option when comparing with the Femoral Stem model, as stated by the Completo's work [5]. Additionally, when comparing the four knee configurations under study, and their influence on both femur and tibia, the Both Stems models seemed to be once again the best choice for the revision TKA (at least in a short-term point of view), since it was the only one mainly causing a decrease in stress on the bone surfaces, while simultaneously causing the lower stress increments. Still, a long-term analysis would be once again required to confirm this statement with certainty.

7. Further Improvements and Future Work

As a suggestion for further improvements on the FE models, working on the parameters related to the simplifications assumed on their creation process would be a priority. Therefore, the geometric modelling of the anatomical elements that naturally constraint the knee joint should be the first step, in order to obain more accurate numerical results. Furthermore, the prosthesis modelling should also be improved, particularly, the scaning process, so that no post-processing of its geometry, in CAD, would be required. As a result, the contact surfaces and their associated constraints and interactions defined on the FE models, would also be improved, becoming the numerical results more accurate and reliable, once again. Finally, the insertion of the stems could also be ameliorated, thereby improving their final position in the medullary canal of the bones.

Thinking now on possible future analyses, they would be more in line with the evaluation of the

stress-strain states of the bones and the deformations suffered by them. Moreover, and as long-term analysis, the changes caused in the bone remodelling processes should also be investigated, since the stem insertion may badly influence these processes, thereby influencing the implant performance and longevity. Accordingly, the conclusions stated in this study may actually change as well, with these new long-term results. Finally, it would be extremelly relevant to try to bring together all the analyses done for each configuration, and compare as much parameters as possible, such as bone stresses, deformation, displacements, stress-strain states, and bone remodelling processes, in order to find out which model brings more advantages, or at least which one is, in a long-term analysis, less harmfull to the patient and to the implant itself.

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