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Computational Study on Bone Remodeling and Osseointegration for a Hip Replacement using a Conservative Femoral Stem

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In this work a computational model to simulate the osseointegration fixation of cementless femoral stems is described. This biological fixation and the bone remodeling are not independent. Thus this model combines the osseointegration analysis with a bone remodeling model. The osseointegration process is modeled based on the relative displacement between bone and stem as well as on the interface stress level, i.e., the osseointegration depends on the mechanical stability of the stem. The law of bone remodeling is derived from a material optimization problem, via the minimization of a function that takes into account structural stiffness and the metabolic cost related with bone mass maintenance. The problem is solved considering contact conditions on the interface between bone and implant.

The model is used to analyze a conservative stem (Mayo, Zimmer Inc.) and to compare its performance with a tapered one (Trilock, Depuy Orthopaedics, Johnson&Johnson). These conservative stems require minimal bone removal, and are suitable to apply minimally invasive surgery techniques. However, the fixation and the stability of the stem is one aspect of major concern. The model developed in this paper allows us to investigate the performance of such stems with respect to stability and compare them with conventional stems.

Keywords: *Biomechanics, Hip, Prosthesis, Stability, Osseointegration, Bone remodeling*

1. INTRODUCTION

Nowadays, Total Hip Arthroplasty (THA) is one of the most important and common orthopedic procedures. This fact is related with population ageing and also with the development achieved in the medical technology.

The main reason for a THA is hip joint diseases such as primary osteoarthritis, but it is also a successful treatment in several other situations such as fracture repair [1]. The THA consists on the replacement of the natural hip joint by an artificial one. This artificial hip joint has two components, the acetabular cup and the femoral component. In a total hip arthroplasty, femoral head is removed and the stem component is inserted into the femoral canal removing a considerable amount of host bone (figure 1).

Recently the interest on conservative stem design for minimally invasive hip replacement surgery has been increased. These conservative stems are smaller than conventional ones, and require less bone removal thus leaving intact many elements of fixation, that would otherwise be lost in a traditional primary arthroplasty. However, one aspect of major concern with these stems is its fixation and stability.

Stem stability plays a decisive role in the success of a Total Hip Arthroplasty and thus in order to assure the long term stability, the cementless stems should be designed to promote biologic fixation, *i.e.*, the bone attachment into the stem surface. This osseointegration (or bone ingrowth) is achieved coating the surface with hydroxyapatite (HA) or with a porous coating layer. However, even with such a special coated surfaces, several factors can inhibit or destroy the biologic fixation. In fact clinical studies show that retrieved stems do not present uniform osseointegration all over the porous coated surfaces and some stems are surrounded by a layer of soft tissue [2]. Among the factors that may inhibit or destroy the osseointegration are the mechanical ones, such as large displacements and high stresses in the bone/implant interface. Thus, for a long term stability it is required a satisfactory initial stability, that is, in order to have a stable and well osseointegrated stem it is necessary that the interface displacements and stresses are within admissible biological values. An adverse interface condition can be induced by a severe or inconvenient load, but the stem properties (shape, size, material and coating) also play a role in the process. The biologic limits of relative displacement and stresses to have osseointegration are not definitely known. For instance, Viceconti *et al.* [3] reports that the limit value for the tangential relative displacement, in order to have bone ingrowth, is a value between 30 and 150 mm, and that displacements between 150 and 220 mm lead to the formation of a soft tissue fibrous layer, circumventing a complete fixation. Concerning the decision of employing cement versus cementless stems, there are several factors of influence but among them is the patient age the [1]. In fact cementless stems have some advantages, mainly in the event of revision surgeries and this is especially significant for younger patients.

Although interface conditions and the osseointegration process have been studied in several research works (see e.g. [4, 5]), the only method that integrates osseointegration analysis and bone remodeling is presented in Fernandes *et*

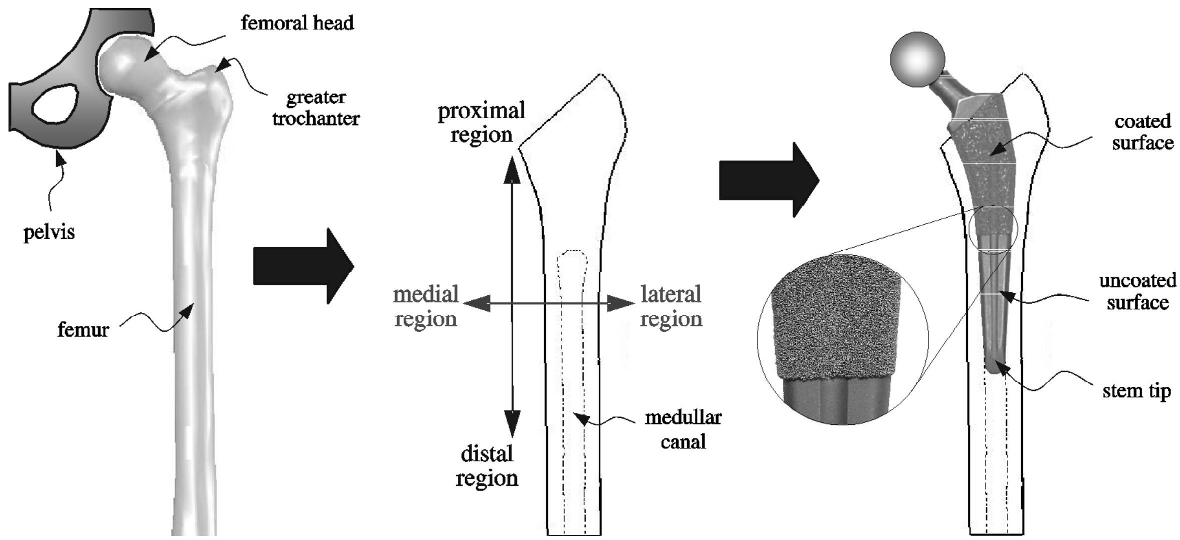

Figure 1: Cementless Total Hip Arthroplasty.

al. [6]. The work reported herein extends this work improving the model by incorporating the interface stress influence.

In the present model, the osseointegration process is modeled based on the relative displacement between bone and stem as well as on interface stress level. The biological fixation is not dissociated with the surrounded bone remodeling. Thus the model combines osseointegration with bone remodeling. The law of bone remodeling is derived from a material optimization criterion, based on the minimization of a function that takes into account structural stiffness and bone mass maintenance metabolic cost, and where bone is modeled as a porous material with variable relative density. The problem considers contact conditions on the interface between bone and implant. During the remodeling process, the mechanical interface conditions are updated according with the osseointegration algorithm: if the displacement and stress conditions required for bone attachment are satisfied, then a connection between bone and implant is established. Consequently, the bone behavior is fully simulated from the immediate post operative condition to a long term condition. The osseointegration process emphasizes the behavior of the bone/stem interface, addressing the problem of prosthesis stability.

The model developed is used to analyze a conservative stem (*Mayo*, Zimmer Inc.) comparing its performance with a conventional one (*Trilock*, Depuy Orthopaedics, Johnson&Johnson). Results allow the computational model validation and appraise the performance of these two distinct stems.

2 METHODS

An iterative procedure is developed to simulate the bone behavior from the immediate post operative condition until a long term condition. This iterative procedure includes two biomechanical models, the osseointegration model and the bone remodeling model. The osseointegration model is presented in first place followed by the remodeling model and by the concurrent computational model.

2.1 Osseointegration Model

A characteristic of a porous coated cementless stems is the biological fixation by bone and metal interaction [2]. After insertion, the bone starts to attach to the stem surface stabilizing the prosthesis (figure 2).

This process is enhanced by the high coefficient of friction of a coated surface [7]. However and despite the coating, high relative displacements can occur in certain regions resulting in inhibition of osseointegration [4]. Furthermore, this early bony attachment can be destroyed [8] and among possible reasons one can postulate that the stress level is the most relevant. In fact, in the osseointegration model proposed in Viceconti *et al.* [5] the interface stress level is taken into account. However, clinical experience states that stem failure is observed to result from failure of initial ingrowth attachment rather than deterioration of osseointegration [6]. Notwithstanding one can consider that a large and well established ossointegrated zone can hardly be disrupted, in the early stage the existing spot weld sites of bony attachment can be broken, and that can affect the overall pattern of osseointegration. Therefore, in this work the osseointegration process is modeled based on the relative displacement between bone and stem as well as on the interface stress level. The model proposes an evolutional or iterative procedure to determine where osseointegration occurs.

The implanted femur is considered under the action of several load cases, simulating the patient activity. In the immediate post-operative situation no osseointegration is considered. In fact, after the insertion of the stem into the bone one should considerer, for the bone/stem interface

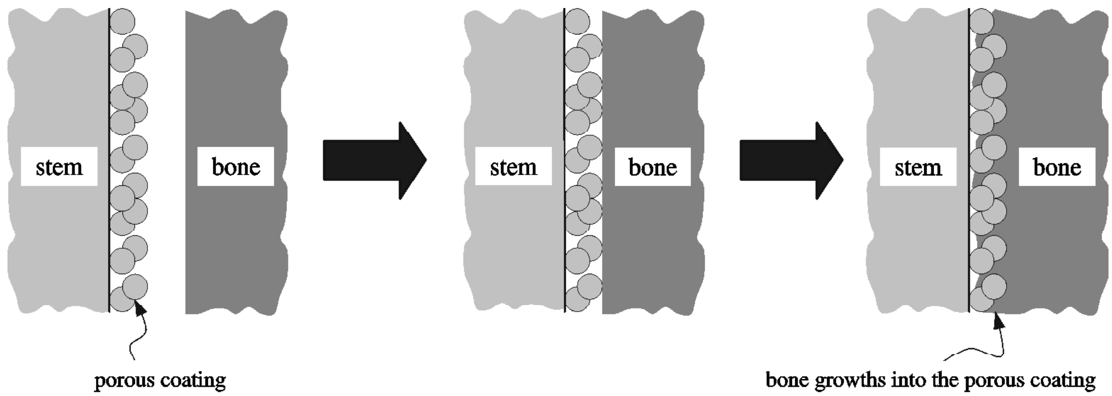

Figure 2: Osseointegration

condition, the contact between bone and stem. Thus, the initial interface conditions are contact with friction in the coated surface and contact without friction in the smooth uncoated surface. In each iteration (or time step) of the osseointegration algorithm, the interface relative displacement and interface stresses are computed. If, at a certain point in the contact interface, contact actually happens and the bone/stem relative displacement is less than a threshold value, than a connection between bone and stem is established. Osseointegration is assumed and the connection is set to completely bounded. If, at a certain point where osseointegration was already achieved, the interface stress level is too severe, than the connection between bone and stem is removed. In the next iteration that point will belong to the bone/stem contact interface. The osseointegration algorithm is illustrated in figure 3.

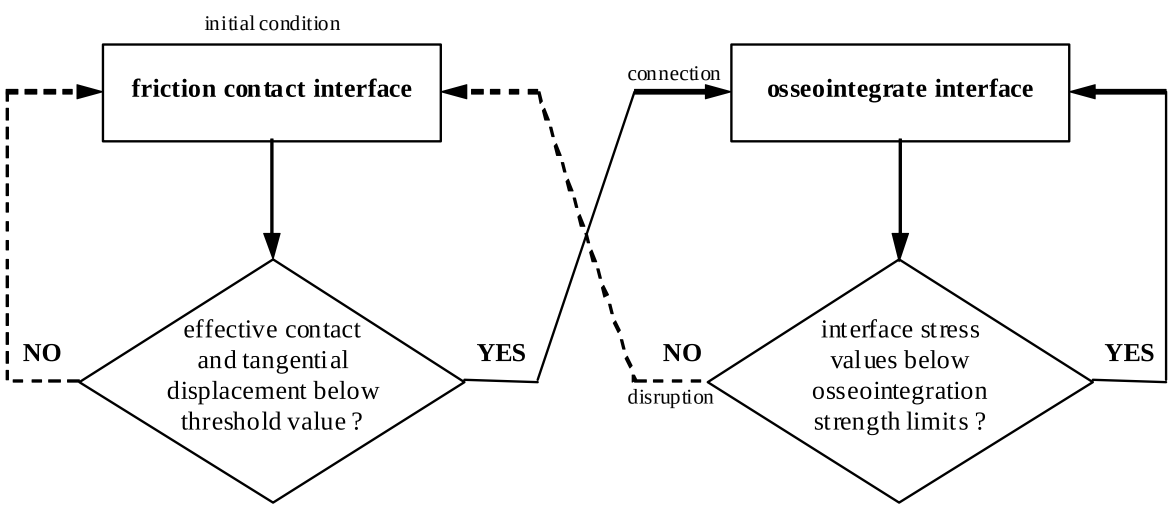

Figure 3: Osseointegration model

Note that in order to achieve or to maintain the osseointegration, the above conditions must be verified for every load case.

A consequence of the model is that, on the coated surface, we simultaneously have regions where contact with friction occurs and bonded regions, as determined by the relative displacement and interface stress level at each location. Furthermore, these conditions can change at each iteration (or time step) depending upon the instantaneous relative displacement and interface stresses.

Another issue is the choice of the threshold displacement value and the strength limit of the osseointegration. An experimental study with dogs relates occurrence of osseointegration for displacement values of 0 and 20 μm, but for 40 and 150 μm bone ingrowth does not emerge or it is not totally defined [8]. For humans, some authors reference values between 50 and 150 μm for the threshold displacement value [9, 10]. In the examples presented in this work it is used a threshold value of 50 μm. In relation to the strength limit, it depends on the type of the coated surface as well as the time that have passed after osseointegration was initiated. In Svehla *et al.* [11] five different types of coat surfaces for a titanium stem are tested. One of the coat surface that is tested is *Porocoat* (Depuy Orthopaedics, Johnson&Johnson), the porous coat that is used in the tapered stem. Shear stress limits are presented for 4, 8 and 12 weeks after the insertion and for *Porocoat* shear stress limits are 18 ± 10 MPa after 4 weeks, 33 ± 5 MPa after 8 weeks and

35 ± 5 MPa after 12 weeks. Concerning the tensile limit value, Viceconti *et al.* [5] reference a value between 0.7 MPa and 0.9 MPa. In order to perform a long term analysis, it was select the osseointegration stress limits corresponding to a longer period of time after the insertion. Thus, in the examples presented in this work it is used a shear stress limit of 35 MPa and a tensile stress limit of 0.8 MPa.

When one compares the extended model derived in this work and the original model presented in Fernandes *et al.* [6], the difference is the disruption condition. The present model considers not only the displacement but also the interface stress level. In fact, the original model [6] is based just in the displacement and once a point is set to bounded it will remain bounded until the end of process – no disruption condition. This "small" change may have large influence in the overall pattern of osseointegration.

2.2 Bone Remodeling Model

To obtain the law of bone remodeling, a material model for trabecular bone is introduced, with a variable density (the opposite of porosity) from point to point. This porous material is obtained by the locally periodic repetition of a unitary micro cell, with a parallelepiped hole of dimensions $\boldsymbol{a} = \{a_1, a_2, a_3\}$, and where density can be obtained by $\mu = 1 - a_1 . a_2 . a_3$ (figure 4).

The selection of this material model leads to an orthotropic porous material. Thus, at each point bone is characterized by the microstructure parameters a_1, a_2 and a_3, which define the local relative density. The apparent material properties are calculated through an asymptotic homogenization method [12].

The bone remodeling model consists on the computation of relative bone density, at each point, by the solution of an optimization problem formulated in the continuum mechanics context and assuming contact conditions for the bone/stem interface. Assuming bone adapts to the mechanical environment in order to obtain the stiffest structure for the applied loads, the optimization problem consists of minimizing a linear combination of structural compliance and the metabolic cost to the organism of maintaining bone tissue. The design variables are the hole dimensions of the microstructure defined above. These variables have values in the interval $a_i \in [0,1]$, i=1,2,3, where the extreme values, $\boldsymbol{a} = \mathbf{0}$ and $\boldsymbol{a} = \mathbf{1}$, correspond to compact bone and void respectively, while intermediate values correspond to trabecular bone with a given apparent density. One assumes trabecular bone tissue (cell walls) has the mechanical properties of compact bone.

The solution of this problem yields the law of bone remodeling,

$$\sum_{P=1}^{NC}\left[\int_{\Omega}\frac{\partial E_{ijkl}}{\partial \boldsymbol{a}} e_{kl}\left(\boldsymbol{u}^P\right) e_{ij}\left(\boldsymbol{v}^P\right) d\Omega\right] + \kappa \int_{\Omega} \frac{\partial \mu}{\partial \boldsymbol{a}} d\Omega = 0 \qquad (1)$$

in the sense that when this equation holds the remodeling equilibrium is achieved and at this point the bone correspond to the stiffest structure with total mass regulated by the parameter $\boldsymbol{k}$ that quantifies biological factors [13]. Thus, this law reflects both mechanical advantage and metabolic cost. In equation (1) NC is the number of load cases, E_{ijkl} is the material properties tensor (homogenized properties for trabecular bone), e_{ij} is the strain field, and $\boldsymbol{u}^P$ and $\boldsymbol{v}^P$ are the set of state and adjoint virtual displacements, respectively. A detailed description of the derivation of optimality conditions is presented in Fernandes *et al.* [6].

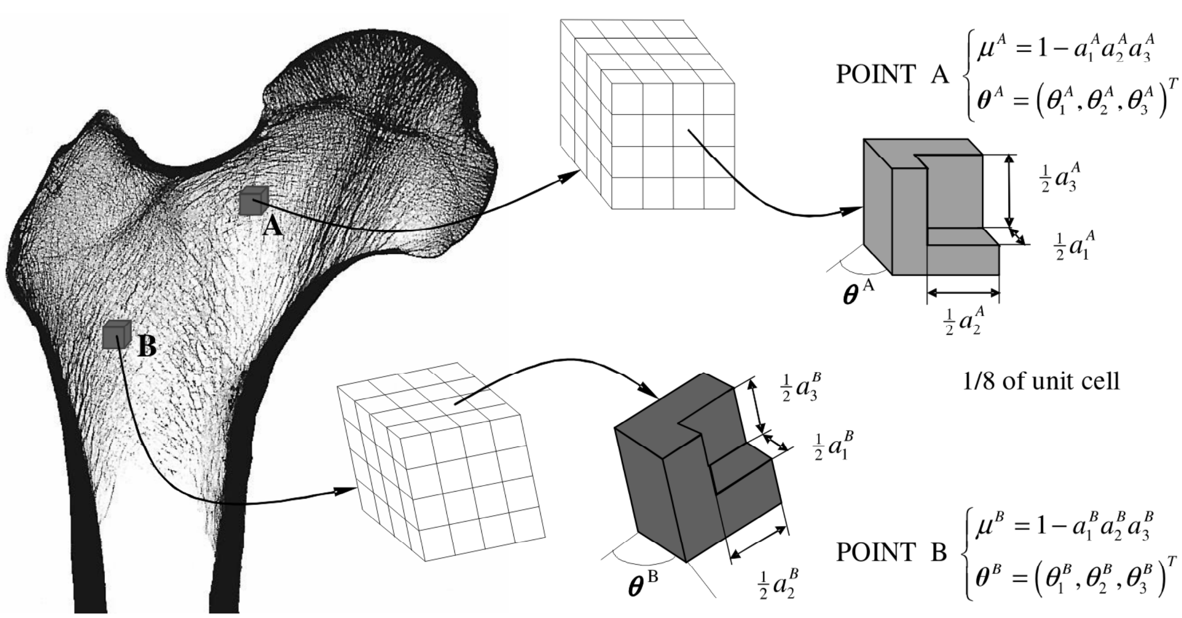

Figure 4: Bone material model.

2.3 Computation Model

The computational model contains the osseointegration model and the bone remodeling model. Both models are solved in iterative procedures and, in this work, iterations in the two biomechanical models are perform simultaneously. The computational procedure is based in a finite element model of an implanted femur. Succinctly the computational procedure is as follows: First the bone homogenized elastic properties are computed for an initial solution. Next, one computes the set of displacement fields $\boldsymbol{u}^P$ and the set of adjoint displacements $\boldsymbol{v}^P$ using the finite element method. The convergence conditions for osseointegration and remodeling are checked and if they are not satisfied, improved values of densities (cell parameters) are computed, the interface conditions are updated and the process restarts (figure 5).

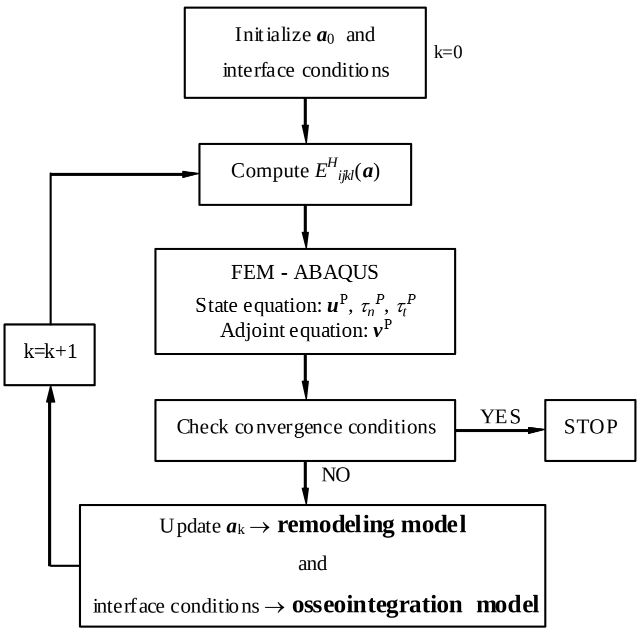

Figure 5: Computational model.

To minimize the computational cost, a mesh of homogenized coefficients is previously computed using PREMAT [12]. Then the homogenized properties are calculated by interpolation of the previously computed values.

The equilibrium and the adjoint problem are solved by the finite element method, making use of the commercial code ABAQUS [14]. The contact problem is solved using standard parameters of ABAQUS with an infinitesimal-sliding formulation and Lagrange multipliers to compute the tangential force.

The density (*i.e.*, the sizes of the micro cell holes) is interpolated in a constant mode in each finite element, which let us to write the optimal condition independently for each finite element. The solution for the cell parameters, a_i, is obtained by an iterative process based on a first order Lagrange method. The design parameters for the element e in the iteration k are obtained from the solution of the law of remodeling equation (1).

Concerning the interface conditions, as was previously mention, initially the bone/stem interface conditions are set to contact with friction on the coated area and contact without friction on the non-coated zone. On the coated area, the contact surface in ABAQUS is modeled as nodes against to surface and the analysis describe in the osseointegration model is performed in the slave nodes of the bone/stem interface (see [14] for details). After the initial contact analysis, the interface conditions are updated based on the absolute value of the relative displacement and interface stresses. For each contact node in the friction contact interface, the relative displacement is computed and if the value verifies the condition for osseointegration, bone and stem are bounded in that point. Otherwise that node remains in the friction contact interface. For each “contact” node osseointegrated the interface stresses are computed and if the stresses are below the strength limits, bone and stem remains bounded in that point. Otherwise, the connection between bone and stem is removed and the node that was in the osseointegrate interface goes to the friction contact interface.

Thus, for every iterations of the computational procedure, the interface condition must be updated for each node of the coated interface, as well as the densities for each element of bone.

3. GEOMETRIC AND FINITE ELEMENT MODEL

The computational model was applied to a three-dimensional model of an implanted left femur. The finite element mesh was created using the bone geometry of the “Standardized Femur” [15] and two different stem geometries were considered. One is based on a conservative stem (*Mayo*, Zimmer Inc.) and the other, a conventional tapered one, based on the tapered stem (*Trilock*, Depuy Orthopaedics, Johnson&Johnson). Figure 6 shows the computational geometric model for the implanted femur with the conservative stem.

In the proximal part (see figure 1), before the stem reaches the medullar cavity, it is assumed a perfect adjust between bone and stems. Both stems are proximal half coated stems, *i.e.*, just a portion on the upper part of the stem surface is porous coated (figure 7). So, in the computer model, above a certain point all the stem surface is modeled as coated. This mimics conveniently the tapered stem but it is not the case for the *Mayo* stem, where uncoated stripes coexist with the coated surface. Thus, in the computer model the conservative stem has, in a certain sense, a larger coated surface than the one existing in the *Mayo* stem.

The femur is fixed on the lower extremity, the articulation load is applied on the top of the stem (F_h) and a global muscles force is applied in the greater trochanter zone (F_a). Three load cases (table 1) are taken into account to mimic two situations of walking (load case 1 and 2) and a stair climbing situation (Kuiper [16]).

Figure 6: Computational geometric model for the implanted femur with the conservative stem.

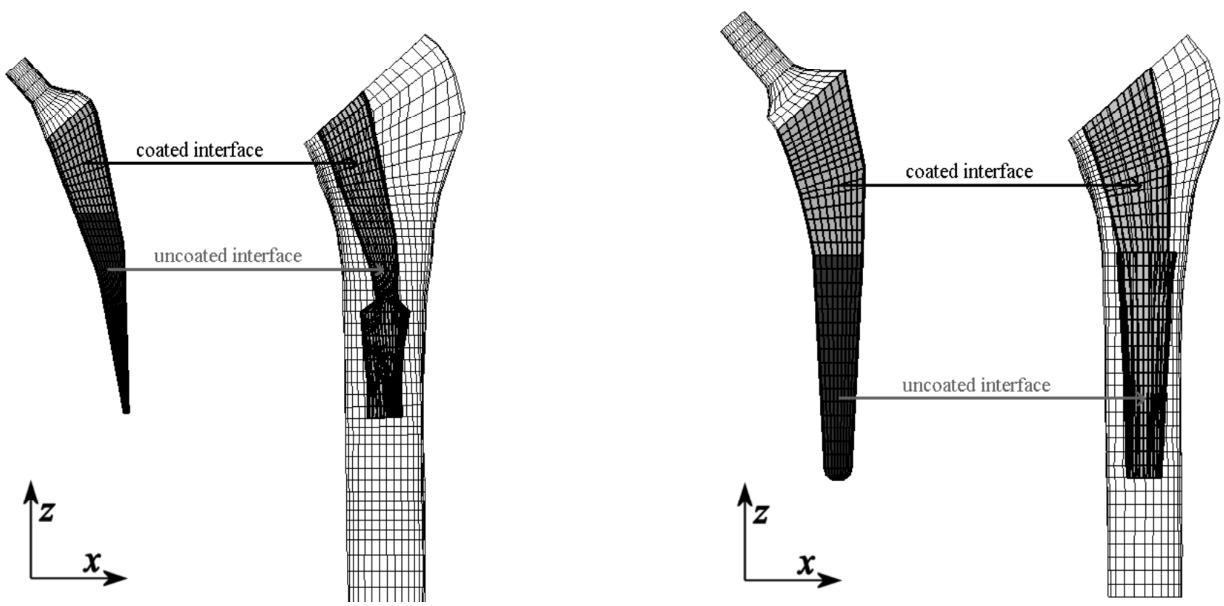

Figure 7: Finite element model for the implanted femur with the conservative stem (left) and with the tapered stem (rignt).

Table 1
Applied Load Cases

Load		F_x *(N)*	F_y *(N)*	F_z *(N)*
1	F_a	-768	-726	+1210
	F_h	+224	+972	-2246
2	F_a	-166	-382	+ 957
	F_h	-136	+630	-1692
3	F_a	-383	-669	+ 547
	F_h	-457	+796	-1707

Stem material is titanium with a Young modulus of 115 GPa and Poisson coefficient of 0.3. In each finite element of bone, the mechanical properties depend on the density value (cell parameters). In this work the initial distribution density is homogenous with a value of 0.7. For compact bone a Young modulus of 20 GPa and Poisson coefficient of 0.3 is considered. It was assumed that bone tissue, which forms wall cells of trabecular bone, has the mechanical properties of compact bone.

The finite element mesh uses 8 nodes hexahedron elements and the maximum number of iterations is set to 100.

With respect to the "biological parameter" presented in the remodeling model, it was used a value $\kappa = 0.1\times10^5$. Concerning the coating, it was assumed that both stems have the same coating mechanical properties. This simplification is a convenient approach to compare the different shape and size of the stems. For the friction coefficient it was used a value of 0.6, corresponding to a stem coated by micro spheres [17]. In relation with the osseointegration strength limits, for the shear stress limit was used a value of 35 MPa. This value is the average value presented for a titanium coated stem with *Porocoat* (Depuy Orthopaedics, Johnson & Johnson), 12 weeks after the stem insertion [11]. For the tensile stress limit it was used the value of 0.8 MPa [5]. Finally, for the displacement threshold value required in the osseointegration model, it was used a value of 50 mm [9, 10].

4. RESULTS

4.1 Osseointegration Results

In order to analyze the influence of the interface stress condition in the osseointegration, we compare results based only in the interface displacement condition and results that include the interface displacement condition and the interface stress condition. The model derived in Fernandes *et al.* [6] does not considerer a disruption condition, it is based only in the interface displacement between bone and stem. The extended model derived in the present work includes the interface displacement condition as well as the interface stress condition, and consequently it has a disruption condition. In figures 8 to 11 are presented osseointegration results in the end of the iterative process.

Figures 8 and 9 shows the osseointegration patterns for the conservative stem and figures 10 and 11 shows the osseointegration patterns for the tapered stem. The results presented in figures 8 and 10 just take into account the displacement condition–model without a disruption condition. The results presented in figures 9 and 11 take into account the interface displacement condition and the interface stress condition–model with a disruption condition. The uncoated surface is presented in white, coated surface in light gray and the osseointegrate surface presented in dark gray. In table 2 it is presented the percentages of the osseointegrate surface, when compares with the all coated surface. Percentages are computed in terms of the number of nodes. The data is presented for the overall coated surface, as well as for each partial region of the stem (rotating anticlockwise; medial, anterior, lateral and posterior region). For each partial region it is indicated in parentheses the contribution for the overall surface.

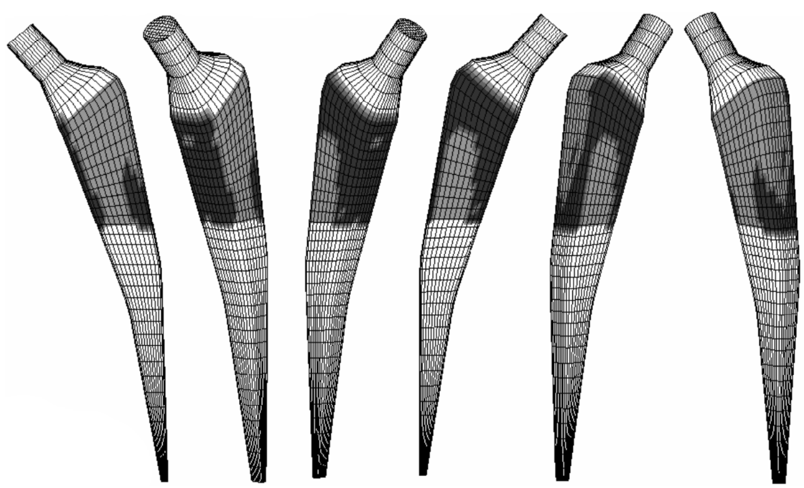

Figure 8: Osseointegration pattern for the conservative stem – model without **a disruption condition.**

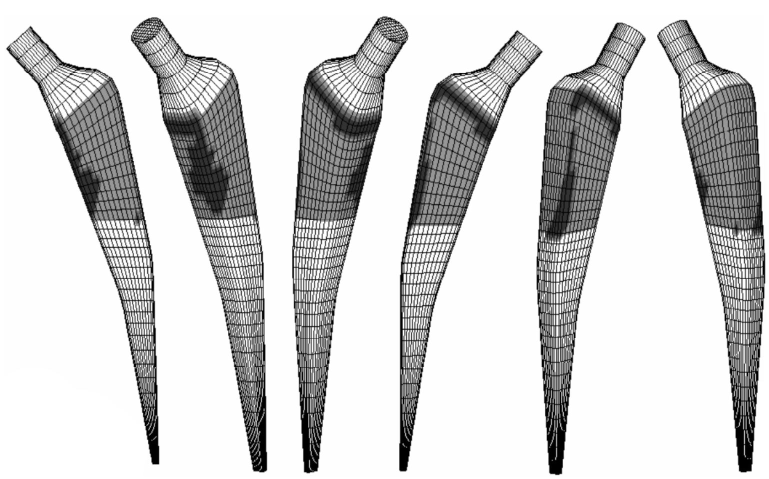

Figure 9: Osseointegration pattern for the conservative stem–model with **a disruption condition.**

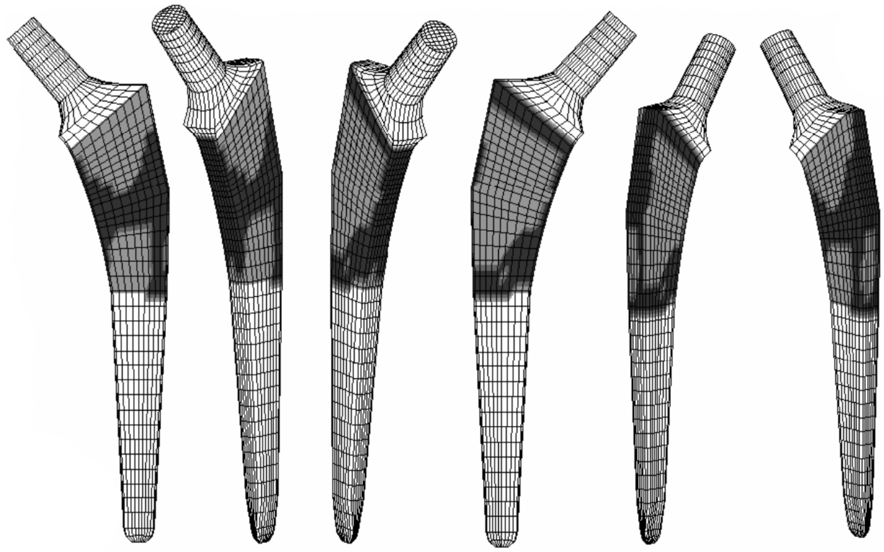

Figure 10: Osseointegration pattern for the tapered stem – model without **a disruption condition.**

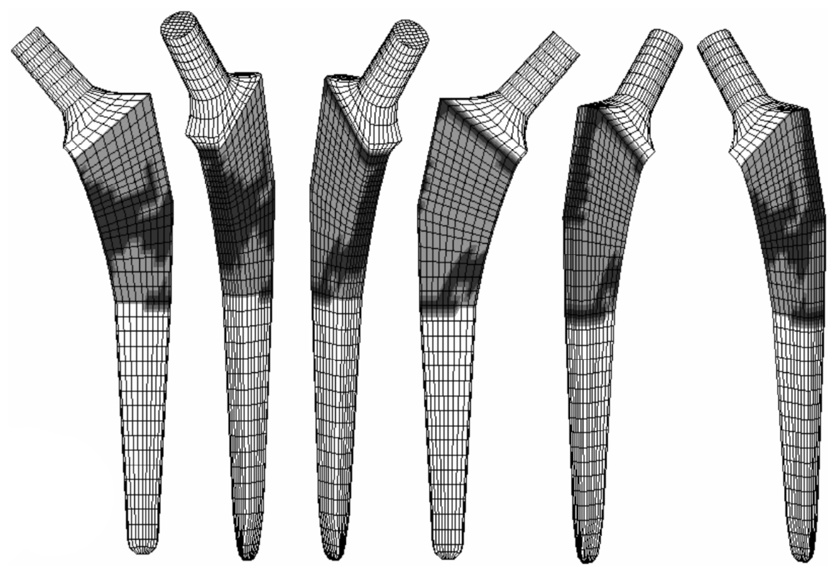

Figure 11: Osseointegration pattern for the tapered stem–model with **a disruption condition.**

Table 2
Percentage of Nodes Osseointegrated

stem	*model **without** a disruption condition*					*model **with** a disruption condition*				
	medial	*anterior*	*lateral*	*posterior*	*overall*	*medial*	*anterior*	*lateral*	*posterior*	*overall*
conservative	95.4% (21.5%)	25.0% (6.8%)	41.7% (9.4%)	61.4% (16.9%)	54.6%	50.0% (11.3%)	15.9% (4.4%)	13.0% (2.9%)	22.0% (6.0%)	24.6%
tapered	93.0% (17.4%)	47.9% (15.0%)	78.9% (14.8%)	33.2% (10.4%)	57.6%	81.6% (15.3%)	40.5% (12.6%)	54.4% (10.2%)	20.0% (6.3%)	44.4%

Results show the influence of the disruption condition; computational osseointegration patterns are less extent for the model that takes into account a disruption condition. This influence is more pronounced for the conservative stem and fairly noticed for the tapered stem. Nevertheless, and independently of the model that is used, osseointegration is attained for both stems, even if for the conservative stem with the model with a disruption condition the extent of the osseointegration is minimal. Furthermore, in this case, the most part of osseointegration is achieved in places where *Mayo* stem has uncoated stripes. Thus, there is some concern with this result for the conservative stem and it requires further investigation.

Comparing the osseointegration attained in each region of the two stems there are similarities but also some differences. For both stems one major part of the osseointegration occurs in the medial zone of the stem (see figure 1). The tapered stem obtain more osseointegration on the anterior part of the stem than on the posterior part, in agreement with clinical results. The conservative stem obtain more osseointegration on the posterior part than on the anterior, at least for the model without a disruption condition.

4.2 Bone Remodeling Results

Figure 12 shows the remodeling results, for both stems, obtained in the end of the iteration process. Results are presented in a gray scale where black represents compact bone, white represents void zones and gray represents the trabecular bone with intermediate density. For each stem are show a anterior and a posterior cross cut.

For both stems, compact bone is obtained in the distal region while in the proximal region one can see the formation of trabecular bone enclosed by a cortical bone shell, reproducing the morphology of the femur.

One can verify that the femur implanted with the conservative stem has less absorption than the one implanted with the tapered stem (the remodeling solution for conservative stem presents more bone mass than the solution for the tapered stem). This can be justified by the fact that the conservative stem has a minor length and a minor size of the coated surface, when compare with the tapered stem. Consequently, the tapered induces a more pronounced bone atrophy than the conservative stem.

Furthermore, for the tapered stem, one can see the densification in the proximal anterior part of the femur in the adjacent zone of the osseointegrate interface (compare figure 11 and figure 12). This result is an evidence of the interconnection between osseointegration and bone remodeling phenomenons.

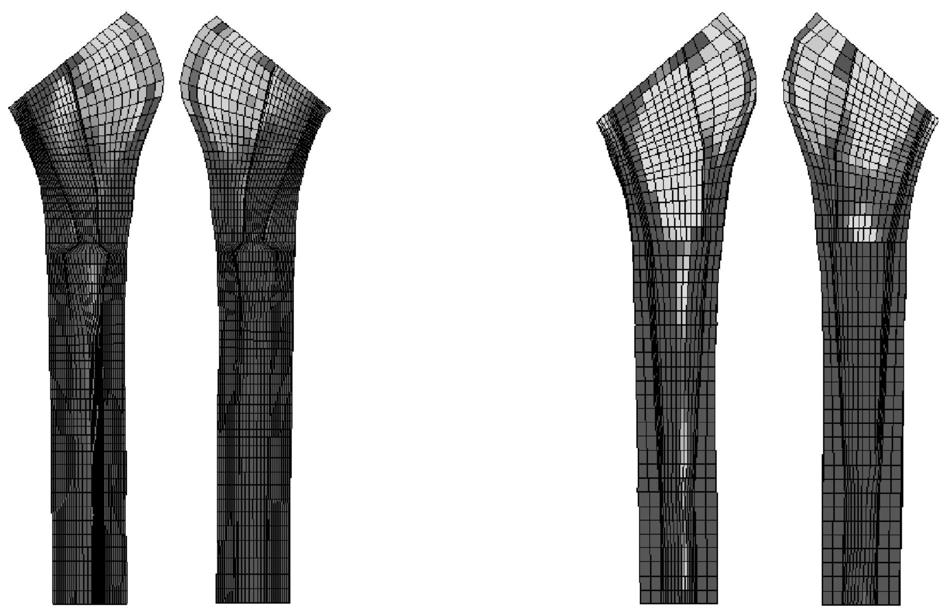

Figure 12: Remodeling results for conservative stem (left) and tapered stem (right).

5. CONCLUSION

In this work a computational model to simulate the osseointegration and the bone remodeling processes in a cementless femoral stem was developed. The osseointegration process is modeled based on the relative displacement between bone and stem as well as on interface stress level. The model combines the osseointegration analysis with a bone remodeling model where the law of bone remodeling is derived from a material optimization problem (Fernandes *et al.* [6]). The bone behavior is fully simulated from the immediate post operative condition until a long term condition, where the osseointegration process emphasizes the behavior of the bone/stem interface, addressing the problem of stability of the prosthesis. The model was applied to analyze a conservative stem (*Mayo*, Zimmer Inc.) comparing its performance with a conventional one (*Trilock*, Depuy Orthopaedics, Johnson&Johnson). To test the influence of the disruption condition included in the derived osseointegration model, results are compared with a model without a disruption condition.

Obtained results show that osseointegration was attained for both stems. Results also show the influence of considering the interface stress level in the osseointegration process. Actually, the disruption condition based on the interface stress reduces the amount of the osseointegrated region. This reduction is more moderated for the tapered stem than for the conservative stem. Further investigation should be performed to confirm these results.

In the model, the disruption condition depends on the strength of the osseointegration. This strength is represented by one value for the normal stress and other for the shear stress. However, *in vivo* the osseointegration strength depends on the elapsed time after the beginning of the osseointegration process [11]. In order to perform a long term analysis, it was selected the osseointegration stress limits corresponding to the longest period of time after the insertion. On the other hand, most of the disruption happens immediately after the bone attachment, that is, in the iterations immediately after the connection is established. So, an improvement can be thought for the computational osseointegration model: the value considered for the osseointegration strength can increase with time. Furthermore, results are sensitive to geometric modeling of the implanted femur. Thus, an adequate control of the modeling, including a geometry and initial density distribution obtained from medical images can concur for the results reliability.

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