NUMERICAL ANALYSIS OF REMODELLING ON A PROSTHETIZED FEMUR

Andrea Corvi, Bernardo Innocenti

Dipartimento di Meccanica e Tecnologie Industriali, Università degli Studi di Firenze, Firenze, Italy

bernardo.innocenti@unifi.it

Abstract: The aim of the present work is to develop a procedure that allows to estimate the effects of the bone remodelling on a prosthetized femur. This procedure, that has been implemented in a FEM system, is based on the application of a phenomenological bone remodelling model that associates to a strain variation, from the physiological situation to the prosthetized one, a variation of the bone geometry. The procedure, inside of which the remodelling function is placed, executes in automatic manner a succession of steps which, from the physiological to the pathological model FEM solution, iteratively applies the remodelling function producing every time a new geometry and therefore a new strain state until the process reaches the numerical convergence. The proposed model has allowed to estimate the entity of the induced bone remodelling from a prosthetic stem. The results have been compared both with the results present in the literature and with data obtained using radiographs of prosthetized patients showing a good performance.

Introduction

Bone is an intelligent material which, on the contrary of the most common building materials, has the ability to adapt itself to the load condition to which it is subjected. For example, it has been demonstrated that an insufficient use of the limbs or o long exposure to a weak gravitational fields involve a decrement of the bony mass. Inversely, if a skeletal zone is subjected to an excessive mechanical stimulation, we have or a bone deposition or a transformation of the bony tissue in fibrous tissue. This phenomena is called bone remodelling, and induces bone geometric changes which cannot be considered like a normal growing and development of the bony tissue, but which are finalized to maintain the stresses into specific ranges [1-3].

As already asserted by Wolff in 1892 [4], bony morphology is influenced from the applied load and to every variation of the functional requirement corresponds a variation of the bony tissue architecture.

From these affirmation it can be deduced that whatever factor which modifies the stress distribution inside the bone can induce a new bony configuration. For example, with the presence of a prosthetic implant, we have the substitution of bone in the proximal femur whit a structure with different mechanical characteristics, so the transmission of the loads on the residual bone is in a different manner respect to the physiological one.

Such non physiological distribution of the loads around the prosthesis is one of the main cause of failure of the prosthetic implants because it can induce the aseptic mobilization of the system, that is the mismatching of the prosthesis inside the femoral channel characterized by resorption and apposition phenomena inside the bone.

In literature there are many works which try to define a model of the physiological process of the bone remodelling assuming connections between stress or strain and geometric variation or inducing a variation of the mechanical properties (density or modulus of elasticity) in relation to the loads acting on the bone. However the biological processes that involved in the remodelling are not still completely characterized and various theories have been developed [5-13].

The aim of the present work is to elaborate a procedure in order to evaluate the effect of bone remodelling on a prosthetic femur. The procedure, which can be implemented in a FEM system, is based on the application of a bone remodelling phenomenological model which associates to a strain variation, from the physiological situation to the prosthetic one, a bone geometry variation.

Materials and Methods

The present work has been divided into the following three step:

1. development of the geometric model of the femur, of the prosthesis stem and their connection;
2. implementation of the physiological model and of the prosthetic model in an analysis FEM code and their solution;
3. implementation of a remodelling procedure to the aim to evaluate the value induced by the stem insertion.

<u>*Step 1 Realization of the geometric model of the femur, of the prosthesis stem and their connection*</u>

The developed procedure is based on the use of a femur model and of a prosthesis model realized with Solid Works 2003 software. The femur model is composed by a cortical part and a cancellous part to the aim to determine the differences between the two typologies of tissues. The used prosthesis has been a cementless ABG (Striker). Connecting the cortical part model with the cancellous one we obtain the physiological model, on which it's possible to realize the cut of the head, following the technique that the surgeon executes before the insertion of the prosthetic stem. The resection plane, according to the commen surgical procedure, realizes an angle of approximately 55° respect the stem axis and begins from a distance of about 10 mm to the lesser trochanter. Because in a cementless implant the prosthesis size corresponds substantially with that of the femoral rasp used to realize the housing, in the developed model we consider the prosthesis external shape coincident with the external profile of the rasp. It's specified that the interest zone has not been the integral femur but only the proximal diaphyseal zone because it's in this zone that the greater variation of stresses induced by the prosthetic implant and the most significative remodelling values are localized.

Step 2:implementation of the physiological model and of the prosthetic model in an analysis FEM code and their solution

The solid model has been used only as a start geometry reference: in fact only the initial coordinate of the section of interest are localized on it. I.e. such model has not been directly imported on a finite element code because the geometry of the model demands to be upgraded in relation with bone remodelling so it's necessary to have a reconstruction of the femur remodelled geometry by means of the new acquired data.

The implementation of the model (both the physiological and the prosthetic) has been obtained using the coordinates of the point which define the geometry as parameter. Assuming that in all the points, described in a cylindrical coordinate system with Z axis coincided with the femoral axis, the remodel varies only along in radial direction, it's possible to express the coordinates of a generic point P by means of a variable (the radius) and two constants (the angle and the height assumed by the point in the chosen coordinate system). This choice results to be a problem simplification but it has been done both for making more simply the calculation and also because it's in agree with works found in literature [14], moreover it's permit to make a parametric model exclusively in the single radial coordinate of the points.

Once all the coordinates of the points are defined, for all of the three considered geometries (cortical bone, cancellous bone and prosthesis stem) the model links each other the points with a cubical spline, subsequently it reconstructs the areas and the volumes (fig. 1).

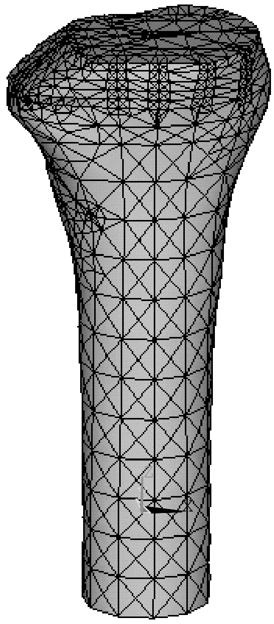

Figure 1: Finite element model

Once the volumes are implemented in FEM calculus environment, in order to perform an analysis of the behaviour both the physiological femur and prosthetic femur, the mechanical properties of the three components of the model are chosen. In both cases a elastic isotropic material has been chosen. The used values of the mechanical properties, according to literature works, are shown in Table 1[15-22].

Table 1: Mechanical properties of the materials used in the models

Material	E (MPa)	Poisson Ratio
Cortical bone	12000	0.3
Cancellous bone	100	0.3
Prosthesis Titanium alloy Ti6Al4V	110000	0.33

A set of forces have been later applied on the model in order to reproduce the phisiological stress in wich the femur are subject, both in modulus, and in direction. Generally, forces on the femur are variably in time and depends from several factors, for example the analized subject and the walking typology. To conduce a finite element analysis we chose a static analysis with reference to Viceconti [18]. The used values, which are referred to the joint reaction, to the adductor, gluteus and vastus muscle, are shown in Table 2.

Such loads have been applied in the muscle considered insertion points and the joint reaction has been applied, in the physiological case, on the femoral head, while in the prosthetic femur case it has been applied in the prosthetic head.

The model has been finally constrained fixing the distal section.

Table 2: applied load values. α angle is the inclination angle of the force relative to the vertical direction and β the risultant inclination angle relative to the X axis in XY plane. R represents the modulus of the complissive resultant subdivided in the three components. XZ plane is referred to the anterior plane, Z axis is the femur axis.

Applied loads	R [N]	R_X [N]	R_Y [N]	R_Z [N]	α [°]	β [°]
Joint reaction	3750	1588	-425	-3370	26	15
Longus adductor muscle	160	-64	-23	145	25	20
Magnus adductor Muscle	160	-101	18	123	40	10
Gluteus maximus muscle	950	-364	-170	861	25	25
Gluteus medium muscle	500	-271	-227	354	45	40
Gluteus minimum muscle	350	-235	-164	201	55	35
Vastus intermedius muscle	320	-3	-5	-320	1	63
Vastus lateralis muscle	320	-3	-5	-320	1	54
Vastus medialis muscle	264	1	-4	-264	1	77

Step 3 Implementation of a remodelling procedure to the aim to evaluate the value induced by the stem insertion

In this work we think that it is necessary to use two different remodelling functions to study separately the bone evolution in the zones subjected at compression stress and in the zones subjected at tensile stress. Frost [2] in fact deduces that the answer of the bone is different according to mechanical stress to which it is subjected, going beyond the pure proportionality described and proposed from other authors. In general terms, it's defined a variable range ($S_1 - S_2$), different from compressive and tensile, called dead zone (several authors called it inertial zone) in which even with the presence of a strain variation the remodelling is not present. Externally to this range it has been assumed a remodelling rate proportional to the strain variation along the femoral axis between the physiological condition and the prosthetic condition.

Analytically such conditions, shown in fig.2, can be expressed from the following relations:

$$\begin{cases} \dfrac{\partial x}{\partial t} = \alpha\,\varepsilon & \Delta\varepsilon < S_1^T \text{ or } \Delta\varepsilon > S_2^T \\ \dfrac{\partial x}{\partial t} = 0 & S_1^T < \Delta\varepsilon < S_2^T \end{cases} \tag{1}$$

$$\begin{cases} \dfrac{\partial x}{\partial t} = \alpha_1\,\varepsilon & \Delta\varepsilon < S_1^C \\ \dfrac{\partial x}{\partial t} = 0 & S_1^C < \Delta\varepsilon < S_2^C \\ \dfrac{\partial x}{\partial t} = \alpha_2\,\varepsilon & \Delta\varepsilon > S_2^C \end{cases} \tag{2}$$

where in each formula:

- $\partial x / \partial t$ it's the radial increase during time, with X radial coordinate of the point;
- S_1 e S_2 represent the threshold values of the dead zone and they have been localized in relation to the average value of the physilogical strain respectively in the compressive zones and in the tensile zones;
- α, α_1 and α_2, are proportionality coefficients which substantially depend by the speed of bone remodelling and which have been calibrated in relation to the model convergence.

It may be observe how, in present work, we assumed the same value of the proportional coefficient for the behaviour of the tensile zones (formula 1) while it is different in the compression zones (formula 2). The values chosen to define the dead zone result to be different for each zones.

In the present work the values chosen for S_1, S_2, α_1 and α_2, are shown in Table 3.

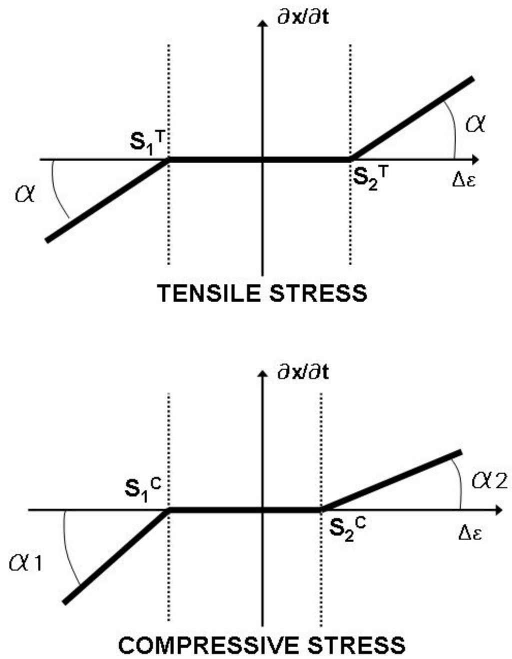

Figure 2: remodelling model behaviour for tensile

stress zones e for compression stress zones

The procedure is then based on the calculation of the difference between the local strain estimated in the physiological femur and the strain in the same points in the prosthetic femur. In relation to this difference ($\Delta\varepsilon$) it is also possible to evaluate the possible radial variation (Δx) and so it's finally possible to reconstruct the modified geometry according to the remodelling model and the successive strains calculations on new bone geometries. Accordingly to the new possible strain variation from this case and the physiological case, it is possible to apply again the remodelling procedure until the convergence, which is obtained when all the points (or a predetermined number) are inside the dead zone and so they don't induce bone remodelling.

Table 3: Constant values used in the remodelling function

Parameter	Tensile stress	Compression stress
S_1	-197 μm	0 μm
S_2	197 μm	197 μm
α_1	0.1E-4	0.1E-4
α_2	0.1E-4	0.5E-3

Results

The procedure, once the model reaches the convergence, permits to visualize, and therefore to compare, the initial physiologically geometry with the prosthetic remodelled geometry (fig. 3)

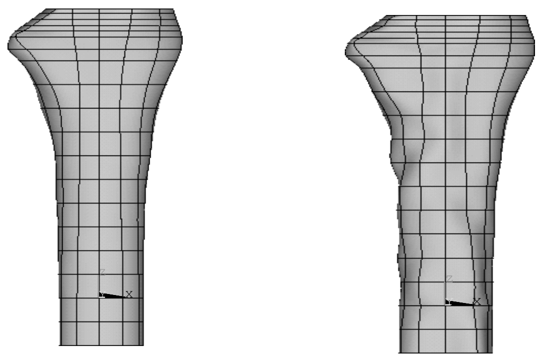

Figure 3: FEM mode:, physiological model on the left and remodelled model on the right

Data obtained from the procedure permits moreover to estimate the radial percentile variation along the femoral axis (fig. 4).

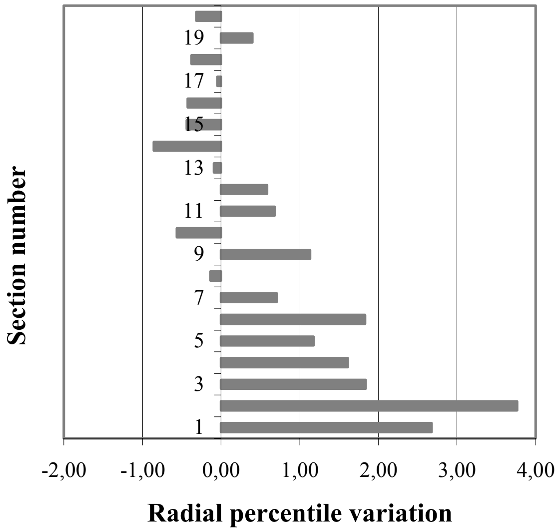

Figure 4: Radial percentile variation obtained along the section of the model

It is moreover possible to estimate and visualize the trend of the physiological and prosthetic solution for each section (fig. 5).

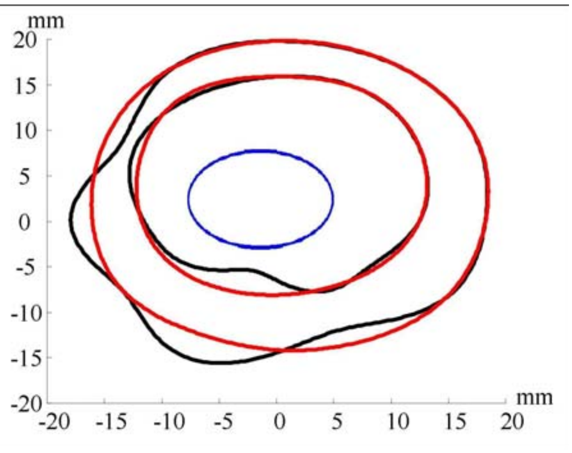

Figure 5: Representation of a section of the model, the start model is in red, the remodelled model is in black, the prosthesis is in blue

Discussion

If we observe the bone remodelling in a regular surgery course (fig. 6) it is possible to note that the bone apposition increase passing between the proximal zone to the diaphyseal zone coming down along the femoral axis. Such behaviour is easily justifiable because the prosthetic stem lean itself, during its use, in proximity of the lesser trochanter, make lever in that point and begin to load the outside diaphyseal zone. This new load distribution induce therefore a resorption which brings to a reduction of the bony mass in the greater trochanter zone, which unloaded itself, and an increase of the diaphyseal zone. Such phenomena has been even found even in the numeric model, as shown in fig. 4, in fact we notice an increment of the radial variation in the diaphyseal zone and a radial reduction in the epiphyseal zone.

By the observation of the experimentally evaluated bone remodelling it's possible moreover to note how it is not homogeneous along the section. In fact in the fig. 6 on the right we look an increase of the bony section only toward the exterior and not toward the interior. Even such result has been found and it can be visualized in fig. 5 where we can see how the bone remodelling doesn't distribute along all the section but only in proximity of some zones.

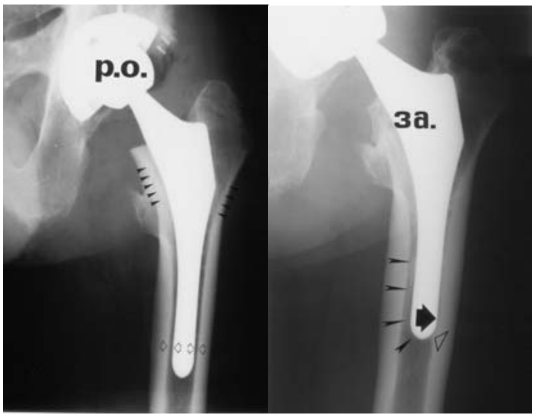

Figure 6: Xray of a prosthetic femur, post operative on the left and 3 years follow up on the right. It's possible to observe a remodelled area on the lower zone on the right of the femur in the second radiogram.

Moreover the obtained results agree with the literature [9,11,23,24].

Conclusions

In this works a procedure which permits to estimate the effect of the bone remodelling in a prosthetized femur has been presented. The procedure, which can be implemented in a FEM system is based on the application of a phenomenological bone remodelling model which permit to associate a variation of the geometry of the bone to a strain variation, from the physiological case to the prosthetized case.

This developed remodelling function allows, differently from others present in literature, a differentiation from the remodelling induced by a strain variation in the tensile zone and that induced by a strain variation in the compressive zone, considering moreover an inertial range in which there isn't any remodelling. The amplitude of such interval has been considered dependent by the average physiological strain.

The choice to have differential rates of variation for the strain level is justified because it's prevalently an increment in the compression (and not of tensile) which induce the apposition of new bony material. The model is based on the hypothesis to assume the rate of increase, or decrease of the bone tissue, proportional to the local strain on the prosthetized bone.

The procedure, inside which is collocated the remodelling function, executes in automatic manner a succession of steps which, from the FEM solution of the physiological model and the prosthetized model, iteratively applies the remodelling function producing every time a new geometry and therefore a new strains state until the convergence.

The proposed model has permitted to estimate the entity of the bone remodelling induced by a prosthetic stem. The results have been compared both with the results present in literature and with data obtained by prosthetized patients radiographs, showing in each cases a good comparison.

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