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Study and improvement of a trapezium–metacarpal prosthesis in relationship with the bone remodeling phenomena

Daniel Lamblin, Guy Guerlement, Laurent Van Parys,
Sélim Datoussaïd
Department of Civil Engineering and Structural Mechanics, Polytechnic Faculty of Mons
rue du Joncquois 53 , B-7000 Mons, Belgium

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Simulation of the long term behavior of a metacarpal bone with a prosthesis is presented with the help of an evolutive 3D finite element model taking into account "the stress shielding" phenomena. The same model allows to improve the shape of the prosthesis.

Keywords: prosthesis, bone remodeling, optimisation, iterative method

1. Introduction

Total arthroplasty of the trapezium–metacarpal articulation. This expression is used by the orthopaedic surgeons in order to qualify the surgical intervention that leads to implant a prosthesis instead of the articulation located, at the basis of the thumb, between the metacarpal bone and the trapezium bone (Fig. 1). This operation is quite usual and remedies to traumatisms or sharp and painful arthritic situations that forbid any movement of the articulation. Different types of prosthesis exist and are used with satisfaction. However, acquired experience shows the appearance of some new progressive pathologies as the consequence of the intervention. The damages may lead, in a few years of time, to a total and irreversible loss in the use of the articulation.

2. Dr Ledoux's prosthesis

The Dr Ledoux's prosthesis is shown in Fig. 2. Its particularly anatomical design allows to fill in exactly the previously prepared central channel in the metacarpal bone and provides a maximal area for the contact between implant and bone. It is composed of three different parts:

- The non axisymmetrical stem implanted (uncemented anchor) in the metacarpal bone and made of embossed titanium in order to improve the settlement by the bone.
- The axisymmetrical socket (uncemented anchor) made of titanium too and set up in the trapezium bone.
- The joint located between the two previous parts. It is a ball fixed on the stem and moving inside a polyethylene winged housing fitted into the socket.

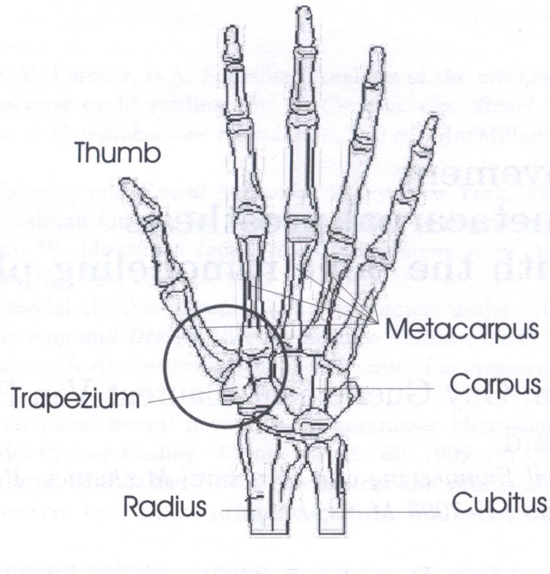

Fig. 1. Localization of the surgical intervention

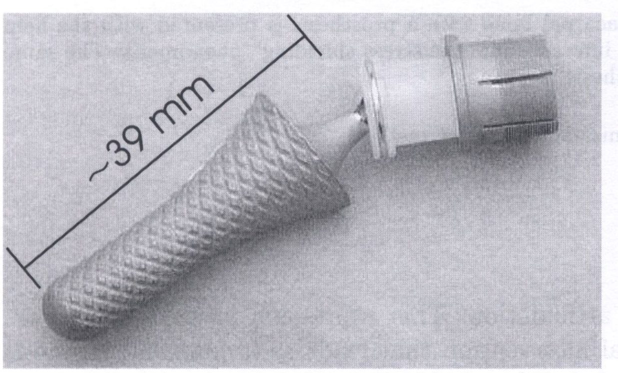

Fig. 2. Dr Ledoux's prosthesis

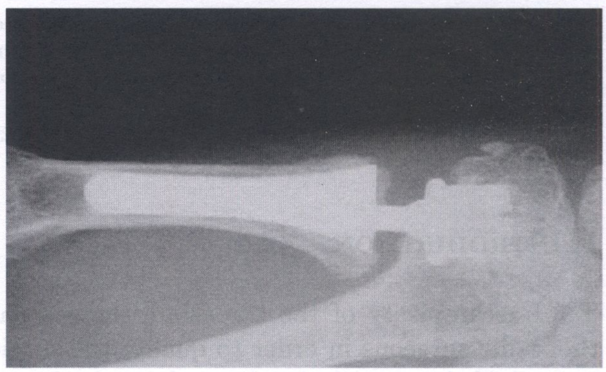

Fig. 3. Dr Ledoux's prosthesis after implantation

The first part of this paper describes the theoretical studies contributing to a better understanding of the damaging mechanisms affecting the bone welcoming the prosthesis. The second part presents some ways of improvement leading to interesting proposals, uniting structural mechanics with science of new materials.

Figure 3 shows the prosthesis after implantation.

3. Remodeling phenomena

3.1. Pathology

The most important pathology found with the considered prosthesis is not related with strength but with the developing by stages evolution of the welcoming metacarpal bone due to "stress shielding" [7]. When the prosthesis is missing, a given stress distribution we call "natural configuration", exists inside the metacarpal bone under any loading. Implantation of the prosthesis as a foreign body of imposed geometry and quite different stiffness creates, for any same loading, a new different stress distribution we call "configuration with prosthesis". The bone, being a living material, is sensitive to these modifications of the stress states. In parts where stresses are increased, bone specific mass and stiffness are going up; in parts where stresses are reduced, bone specific mass and stiffness are going down.

We call respectively "bone neoformation" and "bone lysis" these adapting behaviors of bone to the new stress states. Bone neoformation can give birth to unwanted and uncontrolled displacements of the prosthesis . Ultimate state of the lysis is the total vanishing of some parts of the bone inducing a complete and disastrous immobility of the articulation for the patient and requiring the removing of the implant.

3.2. Model and adaptive mechanism of bone

The adaptive mechanism of bone has been studied since several years and different models have been proposed for which bibliography and a good state of the art review can be found in [4, 9–11]. Present authors are thinking the choice of a given model is always a compromise from the point of view of geometry, sophistication of behavior, experimental possibilities to determine involved parameters, convenient use for numerical calculations and accuracy of predicted results. Taking into account all the preceding considerations, the authors decided:

- to model the bone with and without implant with a 3D finite element model based on actual geometry and realistic boundary conditions. Plane stress bone models were excluded due to the non axisymmetrical character of the stem and the welcoming channel in the bone.

- to accept a perfect settlement of the implant by the bone without any gap and any relative displacement at the interface. This hypothesis was judged as a reasonable one due to the embossed work of the adequately coated contact area between stem and bone.

- to use, under loading, isotropic elastic laws to model the behavior of the bone in which the inhomogeneity in initial and subsequent (after remodeling) states is only represented by an evolutionary distribution in space and time of elastic modulus and a constant and remaining always unchanged Poisson ratio.

- to adopt the concept of the mechanical stimulus [2, 3, 11] including an equilibrium zone defined by threshold levels to govern the remodeling process. Following this concept, a link is accepted between the stimulus defined as $S - S_{\text{ref}}$ and the derivative of the specific mass of the bone ρ versus time t. S is identified as the Mises effective stress per unit of bone mass eventually averaged for different load cases in the "configuration with prosthesis" and S_{ref} is the same quantity in the "natural configuration" (without prosthesis). Figure 4 shows the linear law used in this investigation and characterized by s and A for calculating the evolution of the local specific mass ρ versus time. Additionally to the concept of the mechanical stimulus, the remodeling process needs definition of the local stiffness versus local specific mass. When the bone is seen as a linear and isotropic material, with constant Poisson ratio, the local stiffness is only dependent on the

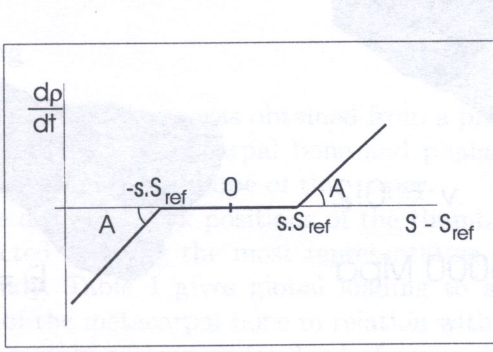

Fig. 4. Linear law for bone remodeling

elastic modulus. Huiskes [8] proposed a relation between the specific mass ρ and the modulus E under the form

$$E = C\rho^3, \tag{1}$$

C being a constant equal to 3790 MPa cm^3/gr (international unit system).

3.3. Aim of this research

In relation with the described pathology and the accepted model for remodeling of bone, our research aims to simulate with finite elements models the behaviour of the bone with the prosthesis versus time and to develop, if possible, an improved geometry for the prosthesis diminishing the "stress shielding".

4. Simulation with a finite element model

4.1. Geometry of the models

From the geometry of cadaveric samples, a first three dimensional finite elements model (with more or less 12000 nodal points) of the metacarpal bone in the "natural configuration" has been prepared. This model has to respect some strong geometrical constraints. It has:

- to model with sufficient accuracy the initial inhomogeneity of the metacarpal bone made of an external homogeneous stiff cortical part surrounding an inhomogeneous internal trabecular part for which local specific mass can be approached from X ray data.
- to permit simulation of the surgical intervention during which a part of the metacarpal bone will be eliminated.
- to include the geometry and the exact volume of a part of the stem of the future prosthetic implant and a defined shape for the interface between implant and bone.
- to allow the loading in a realistic manner.

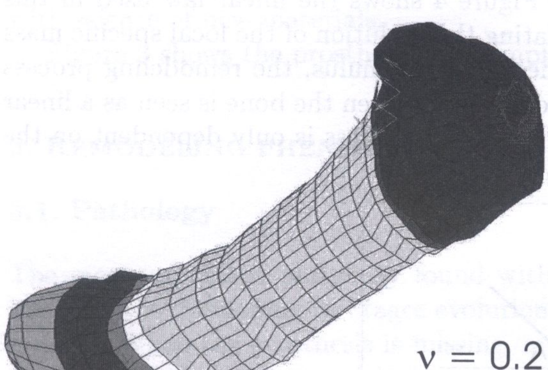

Fig. 5. Cortical part of the metacarpal bone

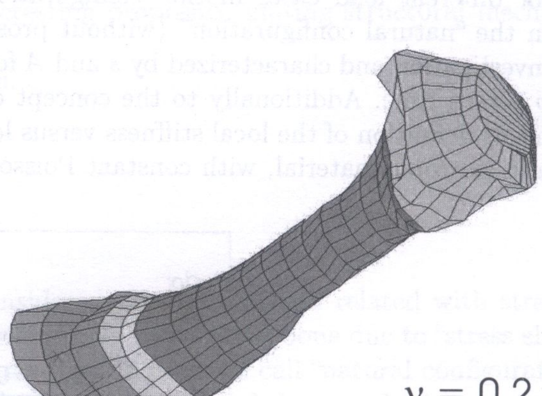

Fig. 6. Trabecular part of the metacarpal bone

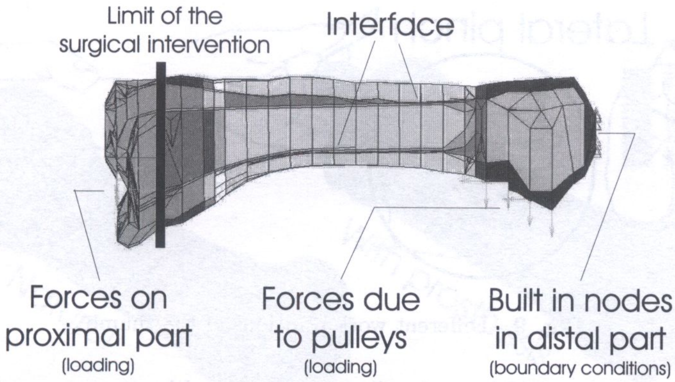

Fig. 7. Sagital view of the metacarpal bone

Such constraints allow to build from the first model a second one for the "configuration with prosthesis" with a "nodal point to nodal point" correspondence in the common parts of the two models, necessary to evaluate the difference between S and S_{ref} inside the same given element. Figures 5 and 6 show cortical (external) and trabecular (internal) parts of the metacarpal bone in natural configuration with initial or bounds for initial properties. Figure 7 is a particular cross section (sagital view) helping to precise roughly the limits of the surgical intervention, the volume of the implant, the loading on the proximal part of the bone, the loading due to the pulleys (see later) and the boundary conditions (built in nodes on distal part of the bone).

Figure 8 shows the implant itself with ball, neck, complete non axisymmetrical geometry and properties. In the finite element model with implant, loading in the proximal part of the bone is applied to the ball.

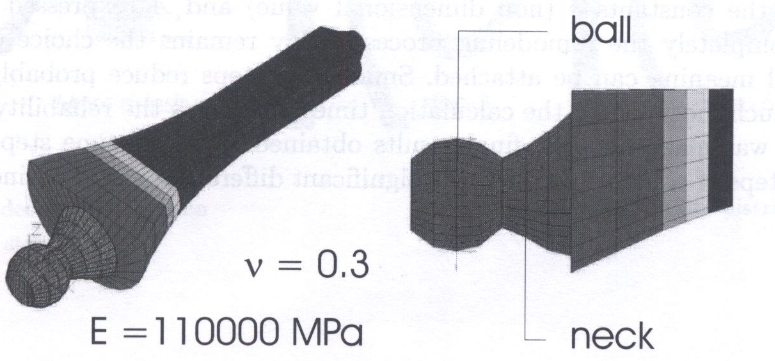

Fig. 8. Details of Dr Ledoux's implant

4.2. Intensity of the loading

The loading to apply to the metacarpal bone was obtained from a previous complete finite element model of the thumb including not only metacarpal bone and phalanx but muscles, tendons and pulleys too [12]. Such a model is outside the scope of this paper.

Considering (Fig. 9) all the different work positions of the thumb, the lateral pinch developing a load equal to 50 N was selected as being the most representative. Only this work position was taken into account in this study. Table 1 gives global loading to apply on proximal and distal parts (loading due to pulleys) of the metacarpal bone in relation with the considered load case. Let us remark that the loading is mainly a compression load of course inducing some bending in the metacarpal bone.

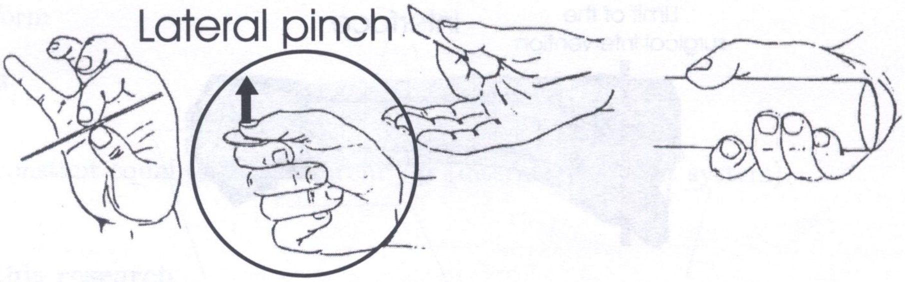

Fig. 9. Different work positions of the thumb

Table 1. Loading on metacarpal bone

Lateral pinch (50 N)	Value [N]
Normal force in metacarpal bone	250
Shear force in metacarpal bone	10
Normal load in pulleys	20

4.3. Algorithm

First, a study of the metacarpal bone without prosthesis is prepared to define S_{ref} in each element of the model. Second, a simulation of long term behavior of bone with prosthesis is made step by step. At step i, and with associated properties, the stimulus S_i is calculated in each element. The difference $S_i - S_{\text{ref}}$ and a given step of time allow to determine a new distribution of specific mass and stiffness giving a new model from which S_{i+1} is calculated. The iterative process stops when a critical bone lysis is obtained.

In this study, the constants s (non dimensional value) and A (expressed in consistent units system) define completely the remodeling process. Only remains the choice of the time step to which no physical meaning can be attached. Small time steps reduce probably the discretization error but make much more longer the calculation times. To assess the reliability of the chosen time step, comparison was made between final results obtained after five time steps of equal value Δt and fifteen time steps of equal value $\Delta t/3$. No significant differences were obtained between the final results.

4.4. Results

Figure 10 shows the differences in Mises stress between the two finite elements models for the same loading immediately after the surgical intervention and only due to the implantation of the prosthesis. It summarizes, by itself, all the problematic of the design of a prosthesis.

To go further in presentation of results, we have to compare the spatial distribution of bone mass given by Fig. 11 immediately after the intervention ($t = 0$) to similar results obtained after an increasing number of time steps (Fig. 12). We can conclude to:

- a rather fast global destruction of bone significantly visible from step 3 and going from the external area to the inside of the bone.
- a more important development of the lysis on one side.
- a quasi-symmetric development of the lysis in the proximal cross-section. This axial symmetry was unexpected because axis of bone and prosthesis are not coincident and loading and boundary conditions have no symmetry.

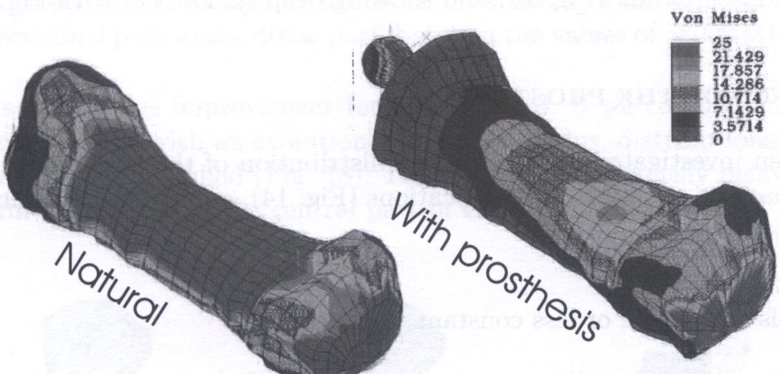

Fig. 10. Mises stresses from finite elements models

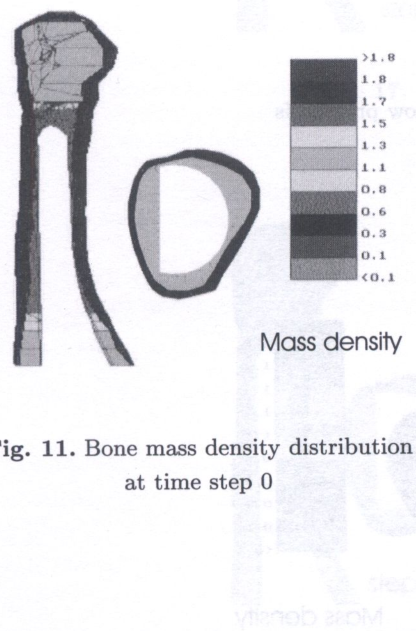

Fig. 11. Bone mass density distribution at time step 0

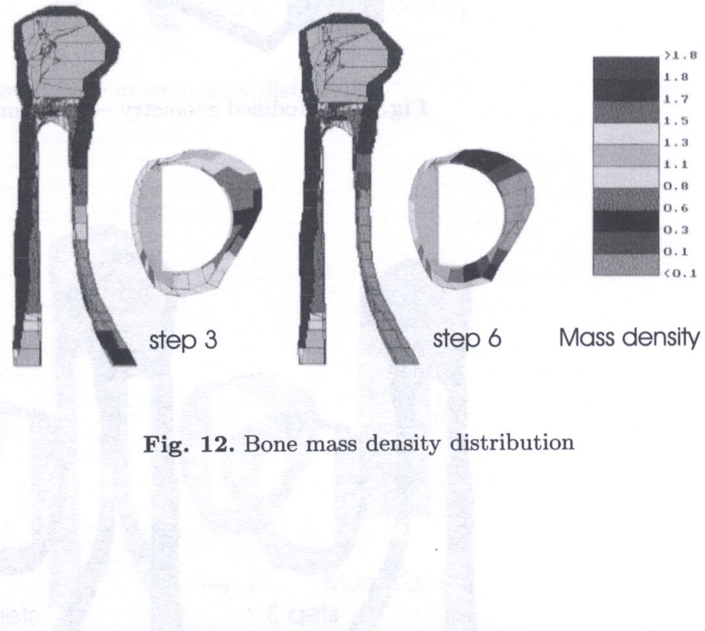

Fig. 12. Bone mass density distribution

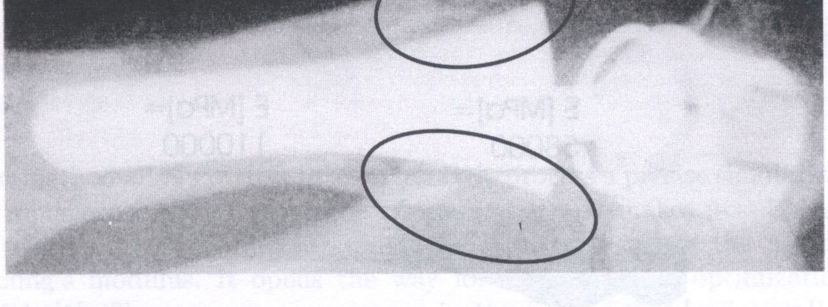

Fig. 13. Bone lysis shown on X ray photographies

These results are totally confirmed by some X-ray photographies taken from some patients, ten years after surgical intervention (Fig. 13) and showing deep similarities with Fig. 12 at time step 3.

5. Improvements of the prosthesis

Two ways have been investigated to improve the distribution of the stiffness inside the prosthesis. First one is only based on geometrical modifications (Fig. 14), applied to a titanium prosthesis, like:

- modified neck,
- hollow prosthesis with more or less constant thickness,
- development of an internal cavity near the neck to spread the stresses in the cortical bone.

Figure 15 shows the distribution of the mass density at step 3 and 6. An obvious improvement can be observed.

Fig. 14. Modified geometry — titanium hollow prosthesis

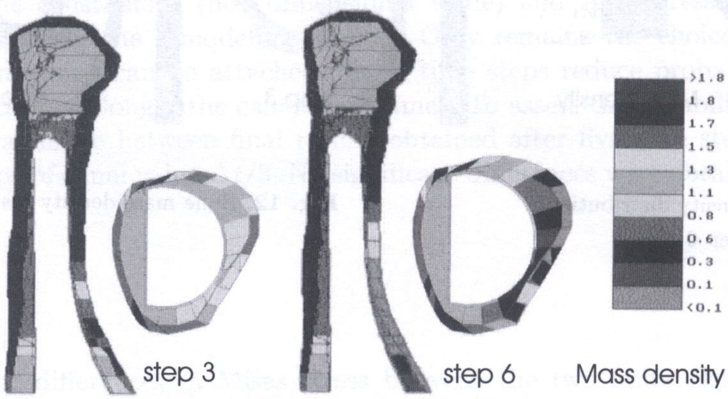

Fig. 15. Improved bone mass density distribution

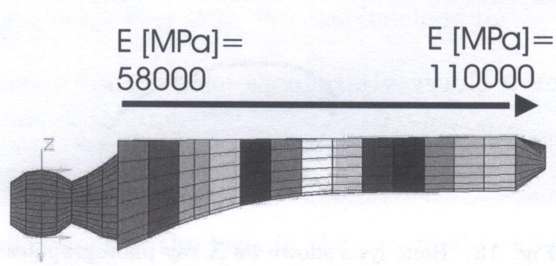

Fig. 16. Initial geometry and hypothetic material

A second way of improvement we examined (Fig. 16) consists in realization of the prosthesis with the initial geometry and in an hypothetic material allowing a discrete variation of the elastic modulus from the proximal part to the distal part, between the values of 58000 MPa and 110000 MPa (titanium).

Figure 17 shows an obvious improvement for step 3 and 4. If we combine in a same model all the geometrical modifications with an evolution of young modulus, distributions of mass density at step 3 and 6 are very deeply modified (Fig. 18). The bone lysis is seriously delayed (from step 3 to step 6) in the proximal part and in the central part of the metacarpal bone.

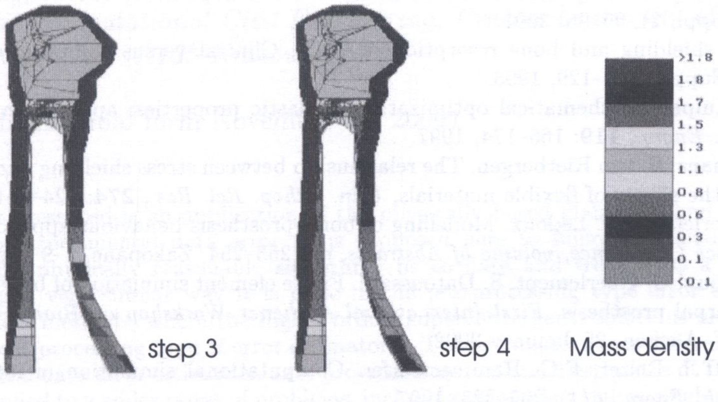

Fig. 17. Improved bone mass density distribution

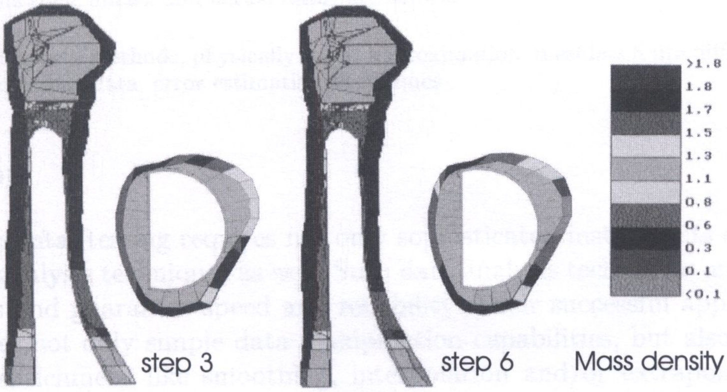

Fig. 18. Improved bone mass density distribution

6. Conclusions

The finite element method allows with help of an iterative process a precise simulation for remodeling phenomena of a metacarpal bone in presence of a prosthesis. It makes possible a fast theoretical investigation in relationship with a new geometry of the prosthesis or the use of new materials with variable Young's modulus. It opens the way for application of optimization methods and optimized materials [1]. The important progresses in the field of powders' metallurgy give to our project a promising future. Some researches in this direction were initiated in relationship with the Department of Metallurgy of the Polytechnic Faculty of Mons.

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