The Effect of Changing the Shape of Tibial Component on the Performance of Total Knee Replacement

دراسة تأثير التغير فى شكل مكون الساق على أداء مفصل الركبة الصناعى

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الملخص

خلل توزيع الاجهادات Stress Shielding احد اهم اسباب فشل جراحة تغير مفصل الركبة نتيجة الاختلاف الكبير بين مادة المفصل الصناعي و مادة العظام الطبيعية مما يؤدي الى حدوث هشاشه في العظام والتى تسبب الالم للمريض وفقدان الجزء الصناعى لوظيفته. الهدف الرئيسى للبحث هو دراسة التصميم الأمثل الذى يعمل على اعادة توزيع الاجهادات بالشكل الطبيعي للمفصل وذلك من خلال تغيير أبعاد المفصل الصناعي. ويسعى البحث المقترح إلى زيادة عمر المفصل الصناعى وتخفيف الالم عن المريض وتقليل فرص الاحتياج لعملية مراجعه جراحية للركبه.

Abstract

Stress shielding of the tibial component has remained one major cause of failure after total knee replacement surgery (TKR), which a relatively high stiffness of the prosthesis can cause most of the load to be transferred through the prosthesis rather than the bone. It leads to the resorption of the bone and a decrease in the bone strength and stiffness. The geometry of the tibial component has a strong effect on the bone compared to its material, so the present study investigates the optimum design of uncemented tibia tray to relieve the stress shielding by changing the dimensions of the metal tibial tray (shape optimization). The results of the optimization process for uncemented tibial component of TKR indicate a trend toward using a shorter cylindrical stem with a smaller diameter and longer metal tibial tray height to house the polymer insert compared with the initial design. This optimal shape model increases the maximum von-mises stress value five times the initial model on medial cancellous bone region. The maximum von-mises stress value of the optimal shape model is very close with that obtained using the natural bone with about 6.5% reduction in the maximum von-mises stress value on medial cancellous region. Also the maximum von-mises stress value of the optimal shape model is increased by 3% compared to the initial model on lateral cancellous bone region. Stress shielding is reduced related to increase stresses on medial and lateral regions. In addition, the maximum interface shear stress value on lateral region for the optimal shape model is decreased by 4% compared to the initial model. Aseptic loosening is decreased related to this reduction in shear stress on lateral side. This leads to reduce patient's pain and increases the implant life and stability.

Keywords

Total knee replacement TKR, shape optimization, von-mises stress, interface shear stress, stress shielding, stress concentration.

1. Introduction

The main key to improve the knee replacement is dealing with prosthesis shape design (peg or stem), material selection (cobalt chrome alloy, titanium alloy, etc), and fixation technique (press fit or cement). However, many reasons of the implant failure after TKR surgery occurs such as stress shielding, aseptic loosening, tibial tip stress concentration, tibial femoral instability, patellar instability, wear of the ultra-high molecular weight polyethylene UHMWPE, etc. Through stress shielding of the tibial component a relatively high stiffness of the prosthesis can cause most of the load to be transferred

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through the prosthesis rather than the bone. It leads to resorption of the bone and a reduction in the bone strength and stiffness. As well as the high stress concentrations within the distal region of the metallic stem cause patient's pain [1, 2].

Peek et al. illustrated that, over-sizing the tibial component may lead to painful feeling; however under-sizing leaves the tibial tray resting on cancellous bone which leads subsidence of the tibial tray [4]. Completo et al. showed that, short stems produce a minor effect in bone relatively too long stem cause stress shielding and stress concentration at tip region [5]. N. Fouda carried out a shape optimization of a two-dimensional cemented tibia stem to improve the stress shielding at the proximal tibia and the stress concentration at the stem tip. The result showed that, a short stem with small diameter at the distal stem tip is the optimal shape. Von-mises stress at the lateral and medial cancellous bone increases by 130% and 140%, respectively compared to initial stem shape. The stress concentration at stem tip is reduced by 25% compared to initial shape. However, the optimal stem shape is reduced the stresses in cement at stem tip by 28% at lateral side and 17% at medial side compared to the initial stem shape [3].

Jazrawi et al. illustrated that, there is a reduction in motion of the tibial tray with increasing press fit stem length and stem diameter. The short cemented stems produce tibial tray stability similar to long press fit stems. There is a trend for increasing the proximal tibial stress shielding with the use of cement and longer, wider stems. Press fit stems achieve tibial tray stability similar to a smaller cemented stem [6]. Chong et al. introduced that, longer stems reduced bone and bone/cement interfacial stresses thus minimizing the loosening, but with increasing bone resorption [7]. Saravanos et al. illustrated that, shape optimization shows a trend toward wider and tapered posts to reduce the maximum stress in the bone near the implant and produce more uniform stress distributions [8].

2. Material and method

2.1 Model geometry

a. Natural bone

This study is based on using solid model software which is a three-dimensional (3D) mathematical representation of natural tibia bone anatomy without any ligaments and muscles. This anatomy is taken from GRABCAD website as a SolidWork 3D model (Solid modeling CAD software produced by Dassault Systèmes SOLIDWORKS Corp.). The design is based on laser 3D scans (Device which analyses a real object to collect data on its shape and medical imaging data) for a real left human tibia, then it was constructed digitally by Mimics Innovation Suite (Software for medical image segmentation and 3D model creation) [9]. After that, this model was imported into a CATIA software program to divide the natural tibia bone into its two main bone parts; the cortical and cancellous bone sections. The tibia is simplified into four sections: cortical diaphyseal, cortical metaphyseal, cancellous epiphyseal, and cancellous diaphyseal bones [7, 10]. The positions of these sections have been considered theoretically, as shown in Fig. (1).

Linear elastic behavior, isotropic, and heterogeneous bone properties are assumed for the cortical and cancellous bone. Mechanical properties of the natural tibia bone materials are illustrated in Table (1) [10, 11]. The applied load of this model corresponds to a three times body weight of 70 kg, where a 2000 N vertical force is applied to the femoral component. In which the joint reaction force is distributed to 40% on the lateral condyle (800 N) and 60% on the medial condyle (1200 N). This load distribution is for a stance phase during normal level walking, as shown in Fig. (2). The distal end of the tibia is fixed in all directions [12, 13].

b. Initial artificial tibial model

In this analysis a 3D model is used to represent a section of the coronal plane of uncemented artificial tibial model, as shown in Fig. (3). The tibial model consisted of the natural tibial bone and the tibial component. The bone is divided into four sections: cortical diaphyseal, cortical metaphyseal, cancellous epiphyseal, and cancellous diaphyseal bone. The positions of these sections have been considered theoretically. The upper cortical diaphyseal part is removed, and a hole is produced in the cancellous diaphyseal part to insert the tibial component implant. The tibial component consisted of the titanium alloy tibial tray (stem) and the ultra-high molecular weight polyethylene UHMWPE insert [7, 10].

The tibia component represents tibia with a medial–lateral width of 74 mm, the UHMWPE insert height equals 8mm. The initial design of the metal tibial tray (H) is 4mm stem height with 2mm inner groove to house the UHMWPE insert and the stem length (L) is 40mm with diameter (D) equals 12mm, as shown in Fig. (4) [7, 10].

Loads of 2000N corresponds on the plastic insert, as shown in Fig. (5). The distal end of the tibia is fixed in all directions [12, 13]. These loads are the same loads that applied on the natural tibia bone as illustrated in Fig. (2).

2.2 Finite element model

The artificial tibia prosthesis was created in CATIA V5R18, and the assembly was set with input parameters in SolidWorks 2012. Then it was exported to ANSYS 14.5 (static structural application) for the finite element analysis. The 3D model was meshed with 2mm element size and patch-conforming tetrahedral elements in order to create multiple triangular surface mesh then volume mesh. The 3D model was consisted of approximately 1400962 elements and 2138971 nodes for all model components.

2.3 Optimization Technique

a. Objective functions

The aim of this analysis is to optimize the shape of the metallic tibia stem in order to solve the problems appeared after knee replacement operation. The first problem is stress shielding which appeared as a result of transferring the load from the natural bone to the metallic tibial stem. This will cause stress shielding at the proximal part of the cancellous bone. The second problem is increased stress distribution along the interface between the model and the surrounding cancellous bone. This second problem will cause aseptic loosening in the artificial knee. In order to solve these two conflicting problems, the optimization procedure was carried out using ANSYS 14.5 program. The objective of the optimization analysis is to maximize the von-mises stress under the metal tibial tray on cancellous bone.

b. Design Variables

1. Change the stem diameter D within values obtained from literature
   $12mm \le D \le 16mm$
2. Change the length L of the stem within values obtained from literature
   $40mm \le L \le 120mm$
3. Change the metal tibial tray height H within values
   $3mm \le H \le 6mm$

c. Constraints

1. The maximum von-mises stress σ_o in the stem tip equals or less than the maximum von-mises stress with the initial shape of titanium tibia σ_i. However, its minimum value equals or greater than the value of von-mises stress at this part of natural tibia bone σ_n.
   $\sigma_n \le \sigma_o \le \sigma_i$

2. The maximum interface shear stress τ_o in the medial and lateral cancellous diaphyseal bone equals or less than the maximum interface shear stress using the initial shape of titanium tibia τ_i. However, its minimum value equals or greater than the value of shear stress at this part of natural tibia bone τ_n.

$\tau_n \leq \tau_o \leq \tau_i$

3. Results and discussion

3.1 Natural tibia bone and initial model

High stiffness of the prosthesis can cause most of the load to be transferred through the prosthesis rather than the bone. It leads to the resorption of the bone and a decrease in the bone strength and stiffness. That's what happens after using TKR surgery. Stress shielding appears at initial model compared to the natural bone on medial cancellous bone region, as shown in Fig. (6), and on lateral cancellous bone region, as shown in Fig. (7). A comparison between natural bone and initial model illustrated that, the maximum von-mises stress value is decreased from 1.5MPa for the natural bone to 0.28MPa for the initial model on medial cancellous bone region, as shown in Fig. (6). Also the maximum von-mises stress value is decreased from 0.2MPa for the natural bone to 0.12MPa for the initial model on lateral cancellous bone region, as shown in Fig. (7).

A comparison between natural bone and initial model showed that, the maximum von-mises stress value is increased on the diaphyseal bone for the stem tip region from 0.1MPa for the natural bone to 0.5MPa for the initial model, as shown in Fig. (8). This increase in stress concentration at the stem tip region causes a lot of patient's pain.

The maximum interface shear stress value is increased around the stem at the bone/stem interface on cancellous bone, as shown in Fig. (9) on medial region and Fig. (10) on lateral region. This will cause aseptic loosening which leads to micro-movement of the prosthesis. The comparison between natural bone and initial models showed that, the maximum interface shear stress value is increased on the diaphyseal bone from 0.07MPa for the natural bone to 0.3MPa for the initial model on medial region, as shown in Fig. (9). Also the maximum interface shear stress value is increased from 0.05MPa for the natural bone to 0.24MPa for the initial model on lateral region.

3.2 The effect of shape optimization on solving stress distribution problem

The optimization runs started using 20 trials (design points) as Design of Experiments DOE, it's found that 6mm metal tibial tray height, 40mm stem length and 12mm stem diameter is the optimal Shape. The comparison between the stresses obtained using the initial and optimal shapes are illustrated through Figs. (11 to 15).

A comparison between the maximum von-mises stress values for initial and optimal shape models on medial cancellous bone region is illustrated in Fig. (11). The optimal shape model increases the maximum von-mises stress value five times the initial model on medial cancellous bone region, this value is very close with that obtained using the natural bone with about 6.5% reduction in the maximum von-mises stress value on medial cancellous bone region. As well as the maximum von-mises stress value of the optimal shape model is increased by 3% compared to the initial model on lateral cancellous bone region, as shown in Fig. (12). These results reduce the stress shielding of the optimal shape model compared to the initial model for the proximal cancellous bone.

A comparison between maximum von-mises stress value at the stem tip region of cancellous diaphyseal bone for initial and optimal shape is illustrated in Fig. (13). The maximum von-mises stress value at the stem tip region for the optimal shape model is decreased by 1% compared to initial model. This result relieves the patient's pain.

Another comparison between shear stress around the stem at the interface of the stem/diaphyseal cancellous bone for both medial and lateral regions for initial and optimal shape models is illustrated in Figs. (14, 15). There are no noticeable differences in maximum interface shear stress values for medial cancellous region for both initial and optimal shape models, as shown in Fig. (14). The maximum interface shear stress value on lateral region for the optimal shape model is decreased by 4% compared to the initial model, as shown in Fig. (15).

4. Conclusion

The optimal shape for uncemented tibial component of TKR is to be a shorter cylindrical stem equals 40mm with a smaller diameter equals 12mm at the distal stem tip and a longer metal tibial tray height equals 6mm to house the polymer insert. The optimal shape model increases the maximum von-mises stress value five times the initial model on medial cancellous bone region. This value is very close with that obtained using the natural bone with about 6.5% reduction in the maximum von-mises stress value on medial cancellous bone region. Also the maximum von-mises stress value of the optimal shape is increased by 3% compared to the initial model on lateral cancellous bone region. These reductions mean less stress shielding. As well as, the maximum interface shear stress value on lateral diaphyseal bone region for the optimal shape model is decreased by 4% compared to the initial model. This reduces the aseptic loosening. These results trend to reduce revision surgery needs as prevent fast implant failure. These reduce the patient's pain and bone loss. Also, these stable the implant and increase the artificial prosthesis life.

5. Future work

The future work is recommended to be on studying the effect of changing tibial tray material on solving stress distribution problem. The future study is classified to three trials:

1. Changing the full material of the tibial stem using three different homogenous materials.
2. Designing the tibial stem from binary materials, metallic tibial tray and polymer material (plastic tip).
3. Using a metallic tibial component with hydroxyapatite coating material.

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Table (1) The mechanical properties of the prosthesis components and the natural bone

Material	Elastic modulus (MPa)	Poisson's ratio
Diaphyseal cortical bone	17000	0.3
Metaphyseal cortical bone	5000	0.3
Cancellous eiphyseal bone	400	0.3
Cancellous diaphyseal bone	100	0.3
UHMWPE insert	1000	0.3
Titanium alloy tibial tray	110000	0.33

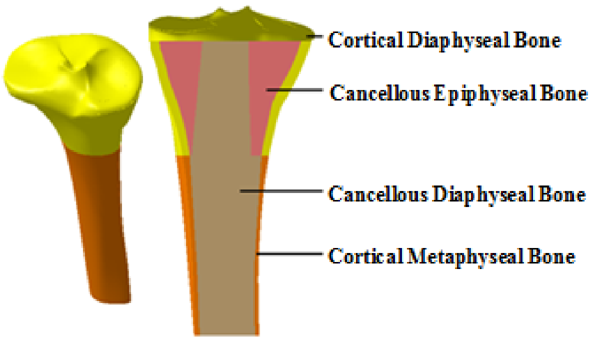

Fig. (1) A three-dimensional model for a natural tibia bone

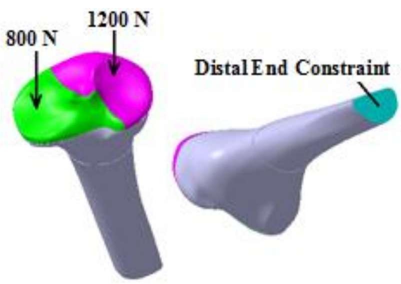

Fig. (2) The loading conditions applied at the natural tibia model

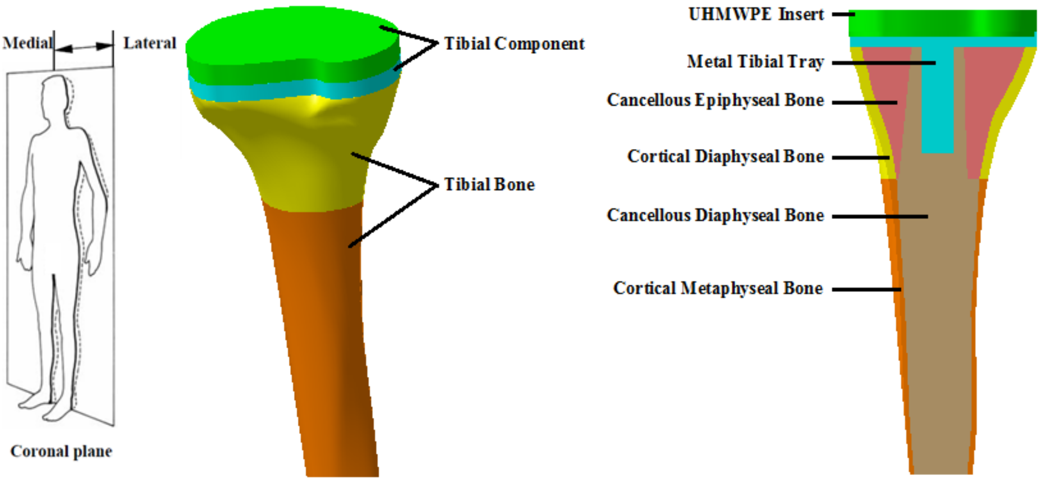

Fig. (3) A three-dimensional initial tibial prosthesis model & a coronal plane section of the artificial tibia model

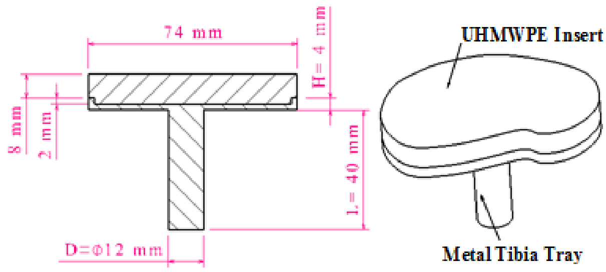

Fig. (4) The dimensions of the initial tibial components

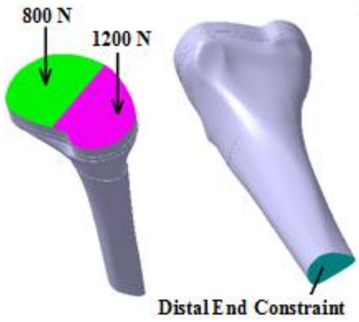

Fig. (5) The loading conditions applied at the initial tibia prosthesis

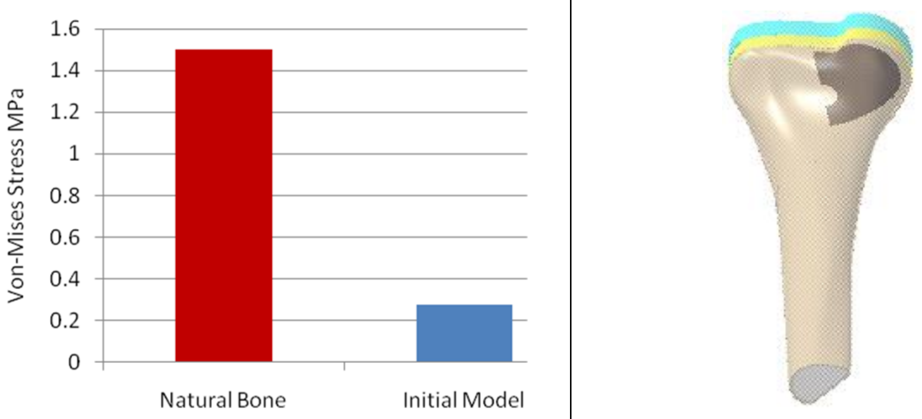

Fig. (6) A comparison between von-mises stresses for natural bone and initial model for cancellous medial side

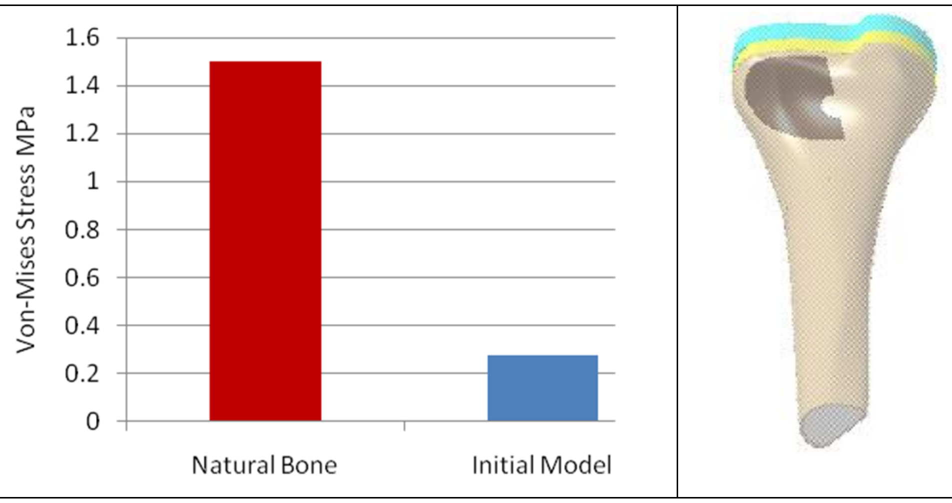

Fig. (7) A comparison between von-mises stresses for natural bone and initial model for cancellous lateral side

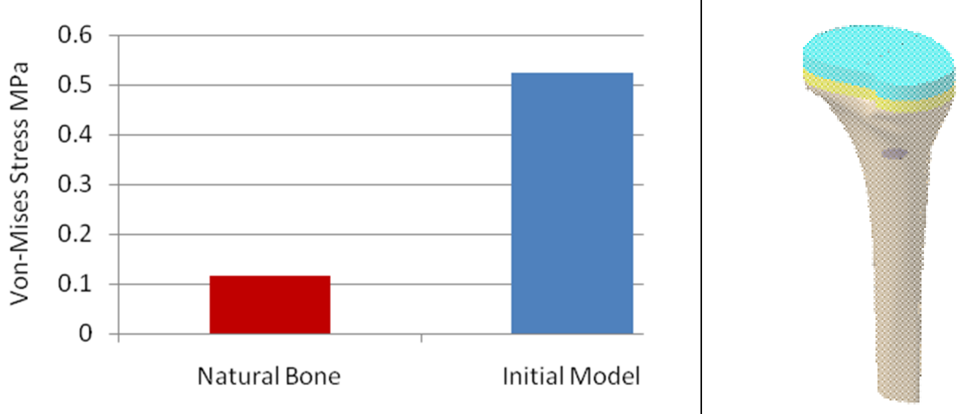

Fig. (8) A comparison between von-mises stresses on the stem tip region for natural bone and initial model

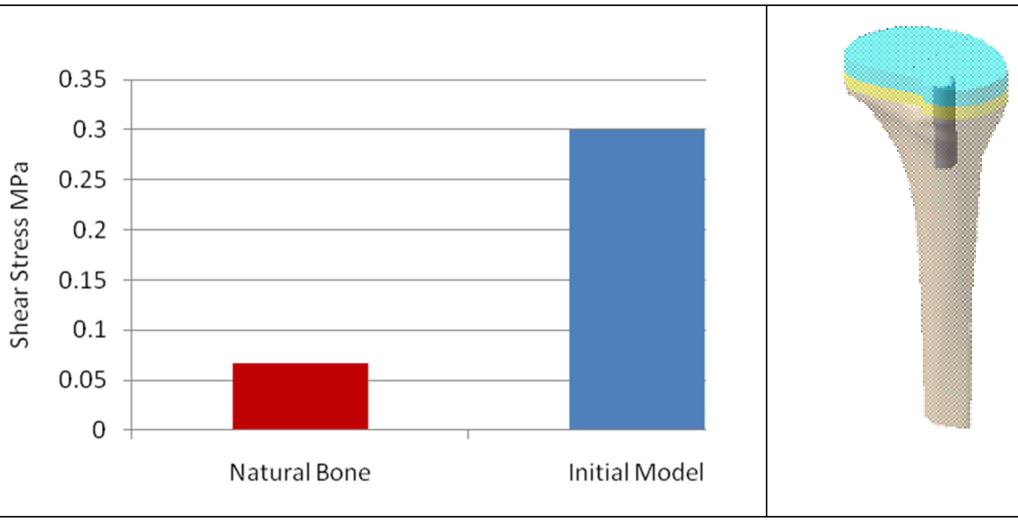

Fig. (9) A comparison between interface shear stresses for natural bone and initial model for cancellous medial side

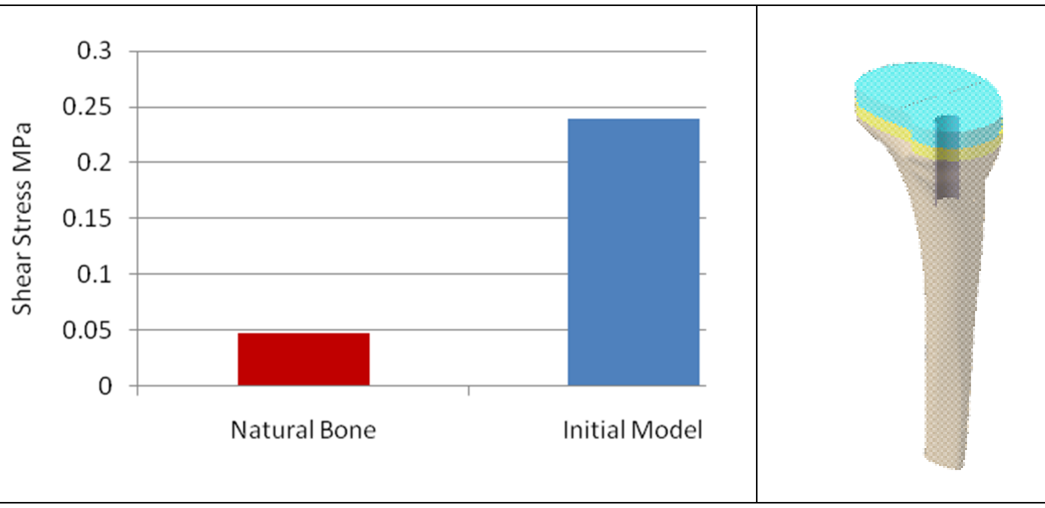

Fig. (10) A comparison between interface shear stresses for natural bone and initial model for cancellous lateral side

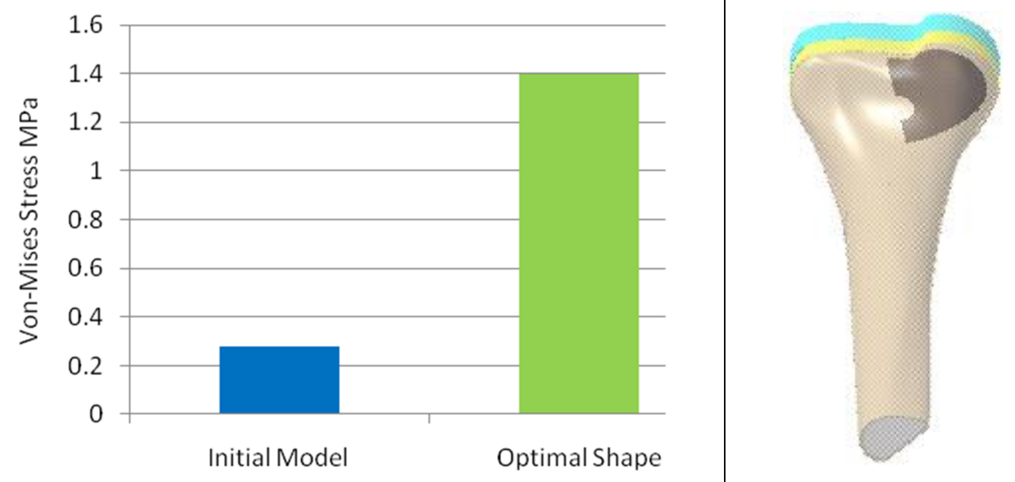

Fig. (11) A comparison between von-mises stresses for initial and optimal shape models for cancellous medial side

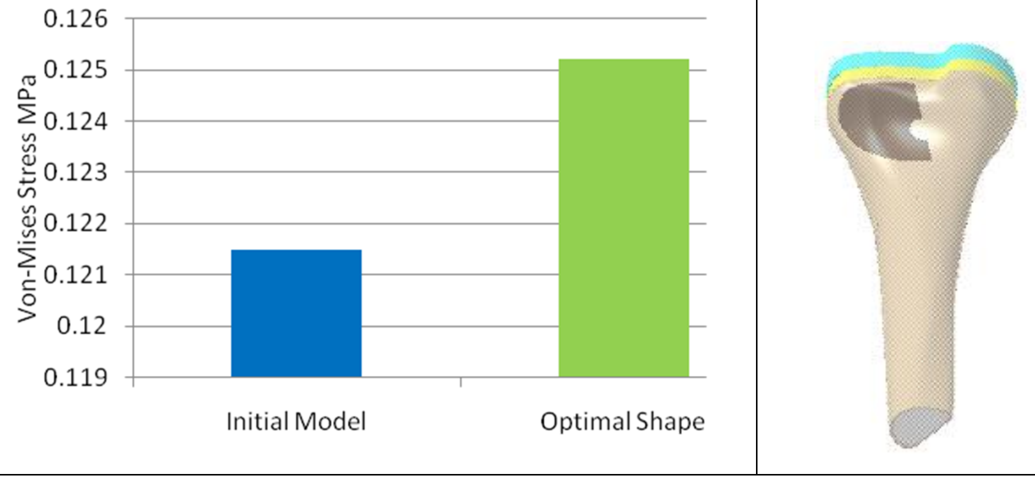

Fig. (12) A comparison between von-mises stresses for initial and optimal shape models for cancellous lateral side

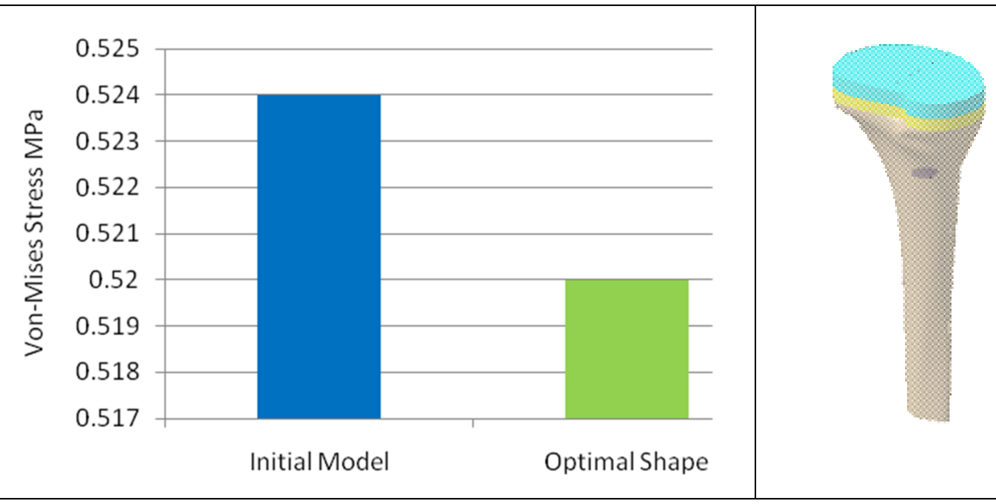

Fig. (13) A comparison between von-mises stresses on the stem tip region for initial and optimal shape models

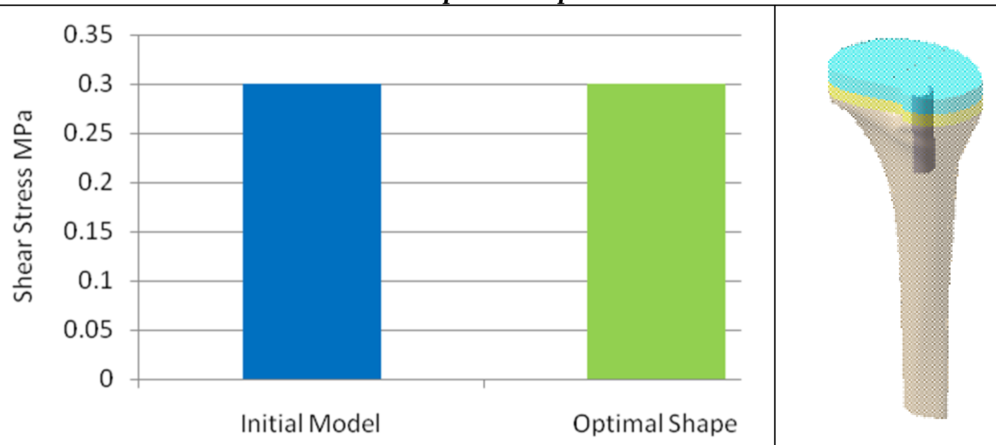

Fig. (14) A comparison between interface shear stresses for initial and optimal shape models for cancellous medial side

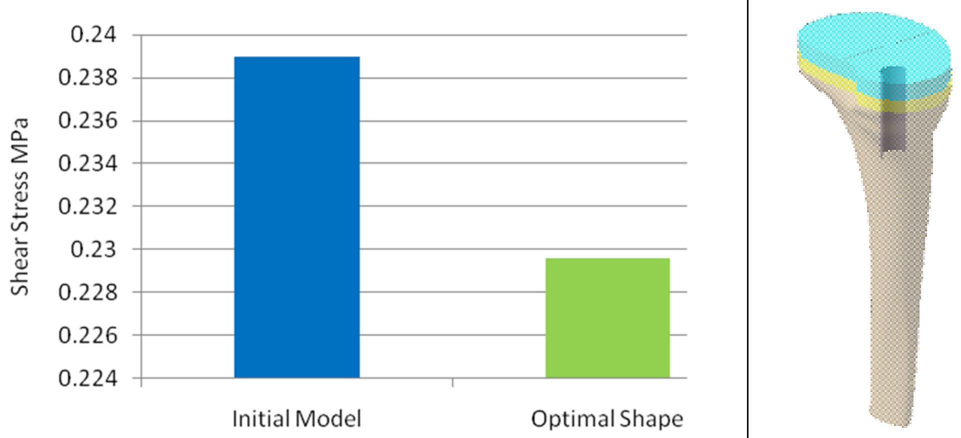

Fig. (15) A comparison between interface shear stresses for initial and optimal shape models for cancellous lateral side