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FINITE ELEMENT MODELING AND MECHANICAL EVALUATION OF ANTEVERSION IN FEMUR AND FEMORAL PROSTHESIS

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ABSTRACT

We studied the mechanical effects of anteversion of hip prosthesis, using a geometrical model of the femoral bone derived from computed tomography sections.

Linear elastic mechanical behavior was assumed for both cortical and trabecular bone. The results of finite element modeling were studied qualitatively. Stress distribution was evaluated during one-legged stance loading of models with various degrees of anteversion.

The undeniable impact of anteversion raises questions about the interpretation of bone remodeling. Anteversion modifies mechanical stresses, most notably those through the

femoral shaft; they are rotated around the vertical axis in the same direction as the anteversion.

Changes in Von Mises stresses were larger than changes in longitudinal stresses, suggesting that anteversion may have a more significant impact on shear stresses, which may cause failure at the prosthesis-cement interface, particularly toward the middle of the femoral shaft.

INTRODUCTION

Anteversion of the femoral bone or of a femoral prosthesis is defined as the angle made by the femoral or prosthetic neck with the long axis of femur, in the horizontal plane [1]. This

angle, which is open medially and anteriorly, ranges from 0° to 30° in normal individuals. In our study, we selected 15° as the standard angle of anteversion and studied anteversion angles of up to 40°.

The recent literature contains studies of proximal femur stresses and deformations under various loading conditions [2-4]. These studies explored particular loading conditions and did not focus on torsion, especially due to femoral neck anteversion.

Anteversion of the neck of the femur influences rotation of the femoral head within the acetabulum. However, the shift in the points of application of articular forces generated by these specific positions also affects the analysis of stresses [5, 6].

We used finite element modeling to investigate femoral stresses and deformations. Taking physiology, anatomy, and the mechanical behavior of bone components into account, we constructed models of the femur before and after femoral prosthesis implantation [7].

We investigated the distribution of femoral stresses according to the degree of anteversion, to the maximum and minimum values of longitudinal stresses, and to the maximal Von Mises stresses. Our goal was to determine how increasing anteversion affected longitudinal and shear stresses, two factors likely to influence bone remodeling and to cause alterations at the cement-prosthesis interface [8].

MATERIAL AND METHODS

1. MODEL OF THE FEMORAL BONE

Femoral geometry was modeled based on volume generation logic used in various finite element analysis computer programs (Abaqus®, Nastran®, and Algor®).

The gross structure of the femur is composed of two main components : cortical bone and trabecular bone. Cadaver femurs were modeled by delineating the contours of each component.

We used a digitizer (Wacom®) to analyze the magnified images of computer tomography cross sections of femur, for the intervals of 5 mm, to define inner and outer contours of each femoral component.

Using these contours, we carried out the mesh generation of femur, between neighboring sections. By combining the meshes obtained at each level, we constructed a geometric model of the femurs (Fig. 1). The model consisted of 1056 brick elements for the cortical bone and 532 for the trabecular bone.

Deformation was analyzed under standard one-legged stance conditions [9]. The **Z** axis was the long (longitudinal) axis of the femur, which was directed toward the proximal femur. The **X** axis was perpendicular to **Z** in the frontal plane and was directed toward the opposite leg. The **Y** axis was perpendicular to the surfaces of the **Z** and **X** axes ($Y = Z \times X$).

The cortical bone and trabecular bone were assumed as linear elastic materials, with modulus of elasticity of 16 Gpa and 1 Gpa, consecutively. This simplification was used to conduct a strictly qualitative analysis of deformation.

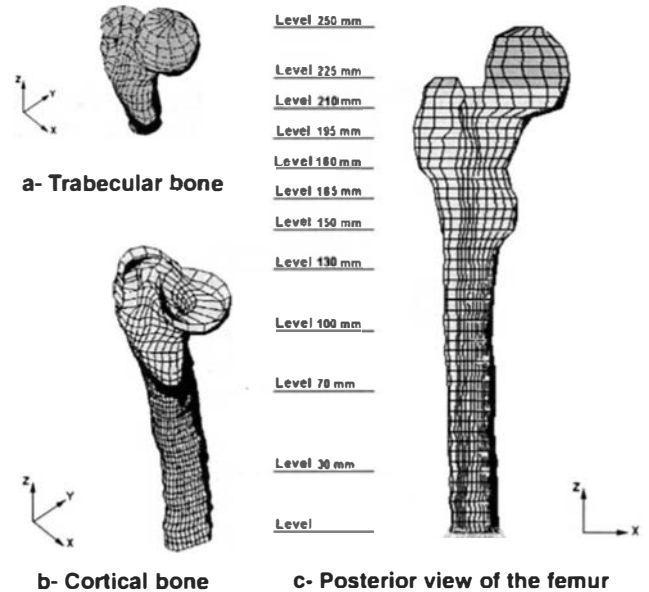


Fig. 1- Geometrical modelling of a left femur.

2. MODEL OF THE IMPLANTED FEMUR

We simulated implantation of a titanium femoral prosthesis (ALIZE®-F. H.) into the femoral model as described above.

Computed tomography images obtained after implantation were unsatisfactory as a result of X-ray refraction and diffraction by the metal implants. Medical imaging techniques capable of providing satisfactory images were not available.

We used the industrial plans of the femoral prosthesis to select the best prosthesis position and size. Implant selection relied on the principles listed below:

1. Implantation of the prosthesis should cause the least possible change in femoral structure.
2. The geometry of the proximal part of prosthesis should be adapted to the anatomy of the femur, most notably at the femoral neck, to avoid medialization or lateralization.
3. The position of the center of the prosthetic head should be similar to that of femoral head.
4. The size of the implant was selected as to provide the best fit with the smallest volume of cement (the thickness of the cement layer varies between different levels).

After defining the optimal position of the prosthesis, we delineated the outer contours of the prosthesis. These contours were drawn within each previously modeled cross section of

femoral bone. The space between the prosthesis and the femoral bone was considered to be optimally filled by surgical cement.

Using the same method as for the femoral bone model, we constructed a model composed of 941 brick elements for cortical bone, 437 for trabecular bone, 760 for the surgical cement, and 756 for the femoral prosthesis (Fig. 2). The mechanical behavior of the cement and prosthesis was also assumed to be linear elastic, with modulus of elasticity of 2.8 Gpa and 200 Gpa, consecutively.

Strains on the implanted femur were evaluated using loads identical to those applied to the femoral bone.

We neglected the micromotion in our model.

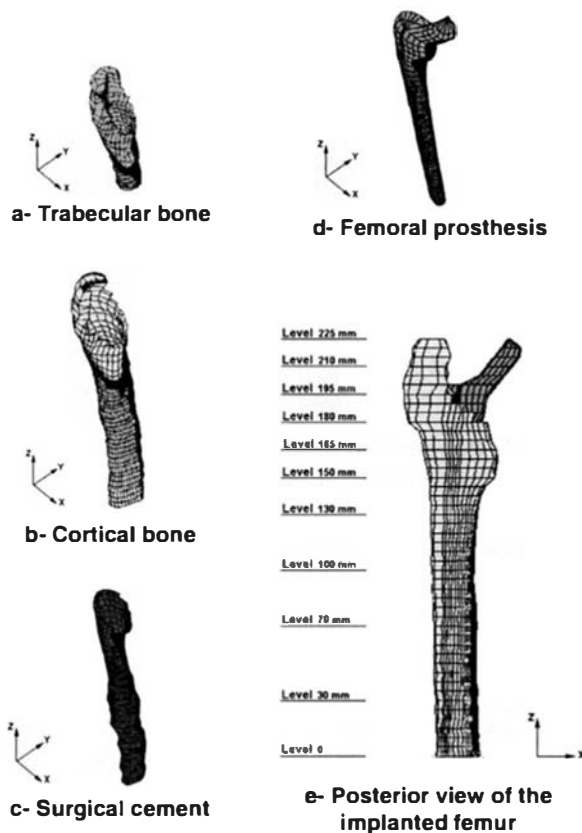


Fig. 2- Geometric modeling of the femur after implantation of a femoral prosthesis.

RESULTS

1. FEMORAL BONE

Analysis of the posterior view of the proximal femur with 40° of anteversion demonstrated a reduction in the compression zone, as compared to standard anteversion (compare Fig. 3a-1 and 3a-2). The conversion of compressive stresses to traction forces at the femoral head was significant. Also, greater anteversion was associated with a decrease in the maximum stresses of femoral neck at the intertrochanteric line.

The neutral surface shifted medially when anteversion was increased. This shift is due to rotation around the vertical axis (Fig. 4 and 5).

To clarify stress distribution, we evaluated the results of the finite element analysis on the femoral cross sections. The stress distribution curves were represented by the polar angle defined on the center of gravity of each section (Fig. 4). This method allows to evaluate the stress changes spatially.

The posterior and anterior views of the femoral shaft (Fig. 3-b and 4) confirmed that the position of the neutral surface changed as a result of rotation around the vertical axis. Also, maximum compression and traction stresses were diminished.

Maximum traction and compression stresses of femur varied with the degree of anteversion. These variations were more marked at the femoral shaft than at the proximal femur (Fig. 4).

To gain insight into the effects of anteversion on applied stresses, we conducted a section-by-section comparison of stress distributions for a femur with standard versus 40° of anteversion (Fig. 5). Results were as follows :

Greater trochanter (Fig. 5a)

Maximum traction stresses showed little change. Conversely, there was a significant increase in maximum compression stresses. The shift in the position of the neutral surface was small.

Lesser trochanter (Fig. 5b)

None of the changes were significant. Changes were small for both stress magnitudes and stress distribution. However, the neutral surface showed a marked and visible medial shift.

Femoral shaft (Fig. 5c)

Maximum traction and compression stresses were reduced. The neutral surface rotated around the vertical axis, coming closer to the medial region.

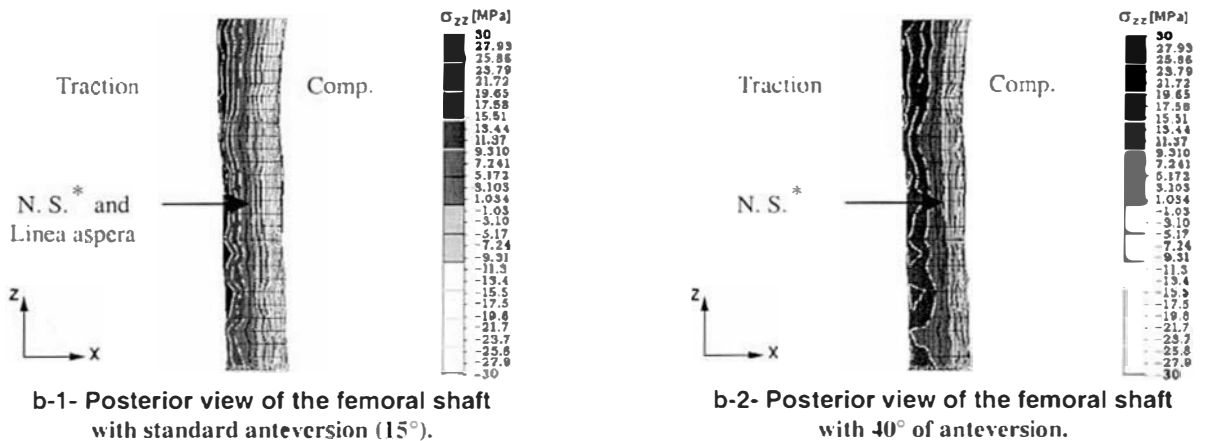
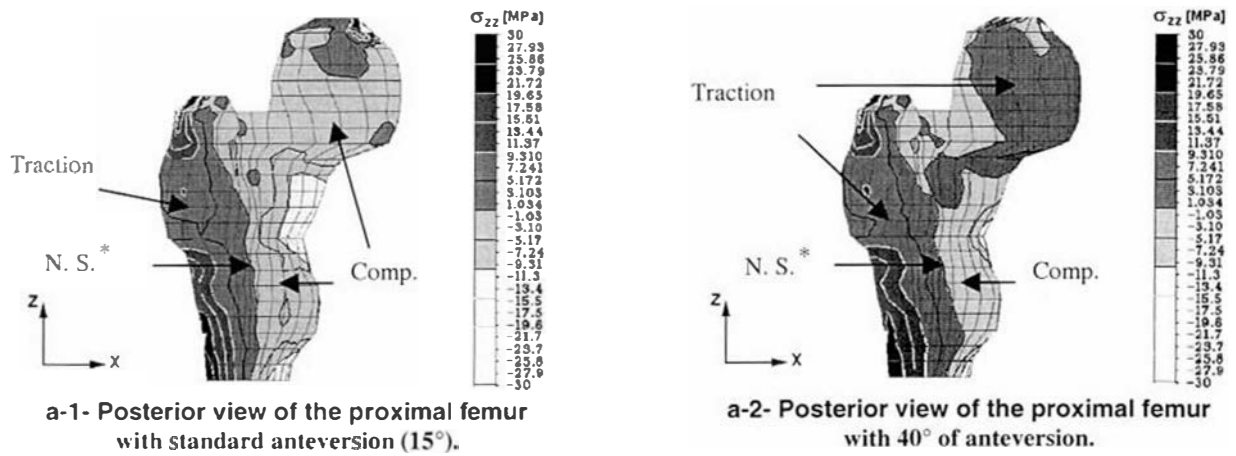


Fig. 3- Distribution of longitudinal stresses on the left femur with 15° and with 40° of anteversion. (* Neutral Surface)

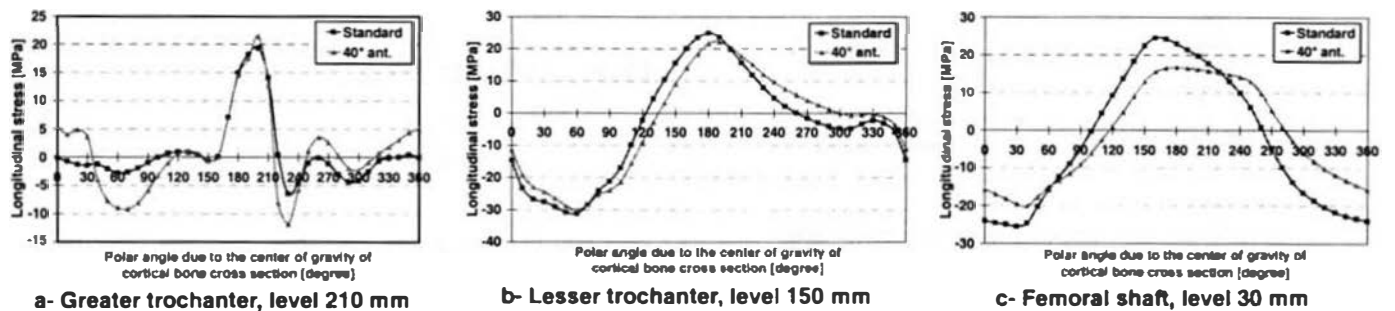
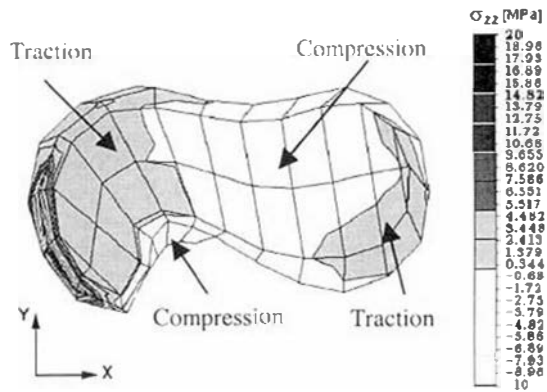
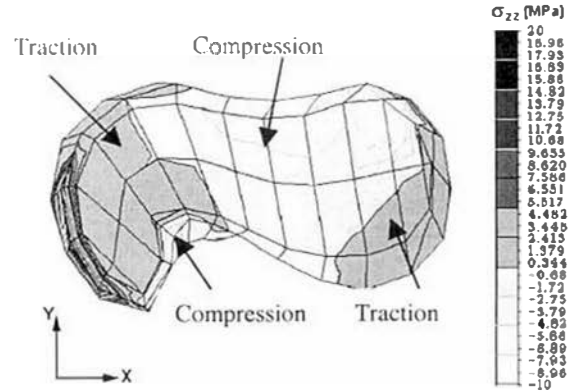


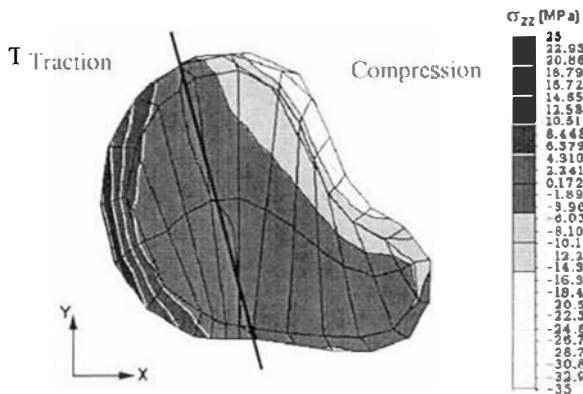
Fig. 4. Comparison of longitudinal stresses on the outer perimeter of the femoral cortical bone with standard (15°) and with 40° anteversion. (The levels are those shown in Figure 1c; ant. = anteversion)



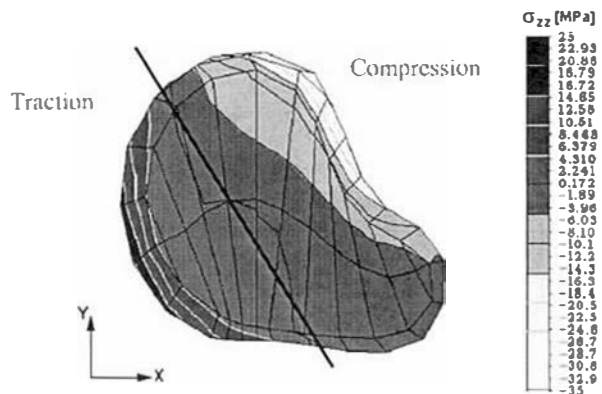
a-1- Greater trochanter (level 210 mm)
standard anteversion (15°)



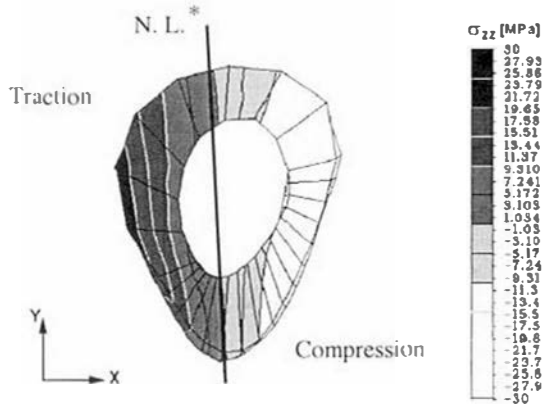
a-2- Greater trochanter (level 210 mm)
40° of anteversion



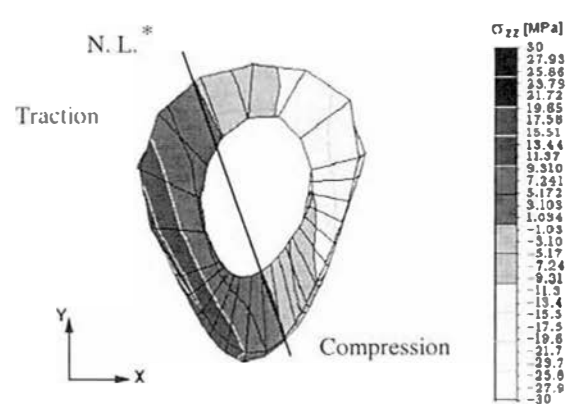
b-1- Lesser trochanter (level 150 mm)
standard anteversion (15°)



b-2- Lesser trochanter (level 150 mm)
40° of anteversion



c-1- Femoral shaft (level 30 mm)
standard anteversion (15°)



c-2- Femoral shaft (level 70 mm)
40° of anteversion

Fig. 5 - Longitudinal stress distribution on femoral cross sections with 15° and with 40° of anteversion. (The levels are those shown in Figure 1c; * Neutral Line)

2. IMPLANTED FEMUR

Evaluation of longitudinal mechanical stresses on the implanted femur showed a distribution similar to that demonstrated for the femoral bone prior to implantation. The posteromedial region remained subjected to compression and the posterolateral region to traction. The neutral surface shifted across the anterior region of the femur to a location near the linea aspera.

The magnitude of stresses applied to the proximal femur showed no significant changes. In contrast, stresses on the distal part of the femoral shaft were increased. Maximum compression and traction stresses were shifted distally.

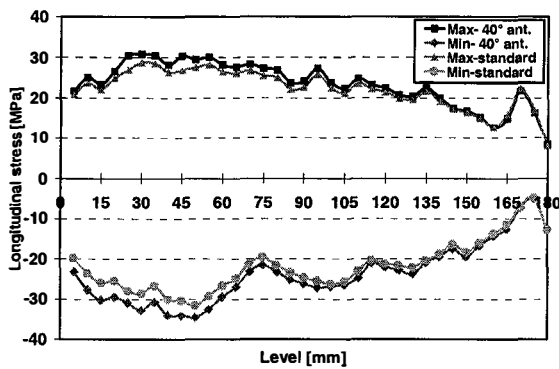
Evaluation of the distributions of stresses on the femoral implant with 40° of anteversion showed results similar to those obtained prior to implantation.

The maximum and minimum longitudinal stresses on the cortical bone of the implanted femur and on the femoral implant were compared across various degrees of anteversion (Fig. 6).

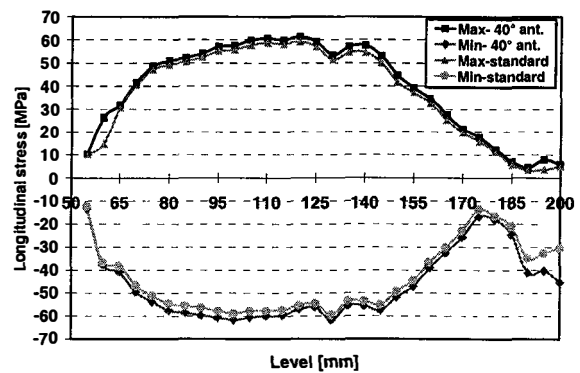
Greater anteversion was associated with an increase in the stresses on the distal part of the cortical bone. Changes in stresses on the implant were consistent from one site to the other. However, changes in longitudinal stresses on the cortical bone and femoral implant were small in absolute value.

We evaluated maximum Von Mises stresses for various degrees of anteversion (Fig. 7). The changes were larger, particularly at the distal part of the cortical bone, at the middle of the implant, and at the neck of the implant.

The fact that Von Mises stresses increased more than longitudinal stresses demonstrates the substantial impact of shear and torsion associated with exaggerated anteversion of the implant inserted in the femur.

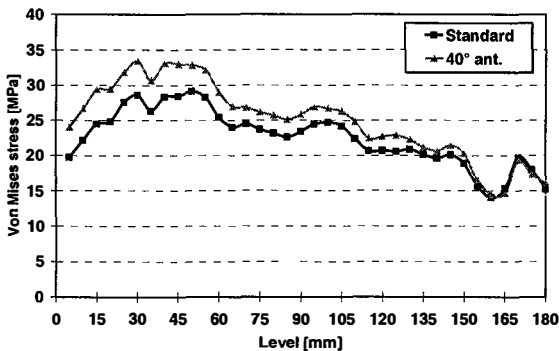


a- Cortical bone

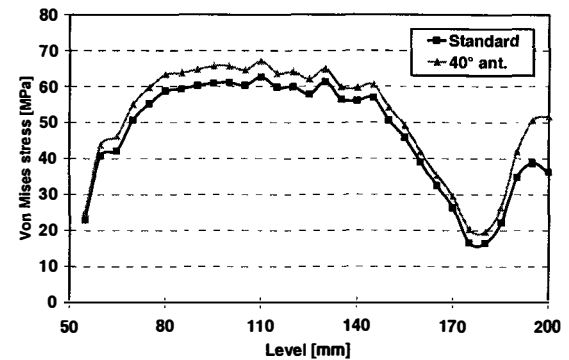


b- Femoral prosthesis

Fig. 6- Comparison of maximum and minimum longitudinal stresses based on all cross sections of the implanted femur, with 15° and with 40° of anteversion. (The levels are those shown in Figure 2e; ant. = anteversion)



a- Cortical bone



b- Femoral prosthesis

Fig. 7- Comparison of maximum Von Mises stresses based on all cross sections of the implanted femur, with 15° and with 40° of anteversion. (The levels are those shown in Figure 2e; ant. = anteversion)

DISCUSSION

The impact of femoral implant anteversion has not yet been the focus of extensive modeling studies. Soyer et al. [10] and Garcia-Cimbrelo et al. [11] have drawn attention to the deleterious effects of exaggerated anteversion.

Studies reported to date investigated the magnitude of anteversion-related shear and torsion in a qualitative and segmental manner [12]. Shear and torsion are believed to adversely affect cortical bone remodeling and to cause the development of failure at the cement-prosthesis interface.

One of the main limitations of available studies is inherent in the mesh generation technique, which is widely used. Our experience with 3-D mesh generation (Nastran®, and Algor®) suggests that this method fails to accurately replicate the geometry of a bone as complex as the femur. The meshes are sometimes found to show substantial differences as compared to bony anatomy, particularly when the reconstructions are analyzed in the horizontal plane.

Reported methods for mesh generation remain unsatisfactory either because they require a very large number of elements [13, 14] or because they are not sufficiently flexible for use in individual patients [15, 16].

The complexity of the geometry of the proximal femur, particularly in the area between the greater and lesser trochanter, is the main stumbling block for mesh generation techniques. Between the two trochanters, the femur is tilted along two different directions; in addition, the femoral metaphysis is rotated from the lesser trochanter to the greater trochanter. These characteristics are major sources of distortion [17] that limit contour segmentation for automatic mesh generation.

Nevertheless, finite element modeling remains an effective and appealing method for evaluating and designing femoral implants, as well as for performing preoperative evaluations [18, 19].

A linear elastic mechanical behavior is frequently assumed for the areas supposed to represent the cortical bone and trabecular bone. This simplification is valid only for qualitative deformation analysis. If more realistic results are desired, preference should be given to nonlinear models [20-22] or to use of several modulus of elasticity for the bone, most notably its trabecular component [21, 23].

Another limitation of modeling studies is inherent in the representation of the space between the implant and the femoral bone. Our model assumed that this space was optimally filled by cement, i.e., disregarded any imperfections in filling [20, 24].

We found that increasing the degree of anteversion caused a larger change in Von Mises stresses than in longitudinal stresses. This finding demonstrates that anteversion affects shear stresses, which may adversely affect cortical bone remodeling and cause the development of cracks at the cement-implant interface.

We believe that this phenomenon, which is frequently disregarded, should be systematically investigated. At present,

anteversion is evaluated only as a factor affecting prosthesis stability [25] and dependent on the orientation of the acetabulum [26] and the design of the femoral implant. We suggest that it should be studied specifically because of its major impact on the implanted femur.

CONCLUSION

We constructed geometric femoral models before and after implantation of a cemented femoral prosthesis, based on computed tomography cross sections of femoral bone. A linear elastic mechanical behavior was assumed for both cortical and trabecular bone. Mesh generation was performed semiautomatically.

Using finite element modeling of the femur before and after prosthesis implantation, we performed a qualitative analysis of stress distribution. By comparing the effects of one-legged stance loading across various degrees of implant anteversion, we determined the impact of anteversion on mechanical stresses.

We found that implant anteversion affected mechanical stresses, suggesting that it may also affect femoral bone remodeling in the medium and long term. When anteversion was increased, the neutral surface rotated around the vertical axis in the same direction as the anteversion. In our model, changes in mechanical stresses were greater at the femoral shaft than at the proximal femur.

Our comparison of stress distribution curves for cortical bone obtained at various levels, defined relative to the center of gravity of each cross section, confirmed that greater anteversion resulted in rotation; the pattern of stress distribution suggested that segmental femoral remodeling may occur subsequently.

With our model and with the implant used in our study, stress distribution for the implanted femur was similar to that for the bony femur, most notably as regards the position of the neutral surface, which was very close to the *linea aspera*.

Increasing the degree of anteversion of the prosthesis increased the mechanical stresses on the distal part of the cortical bone and affected the distribution of stresses on the implant according to the same law.

Comparison of the changes in the maximum Von Mises stresses to the longitudinal stresses showed that anteversion affected shear more than longitudinal stresses. The effect of anteversion was most marked in the middle and distal parts of the femur, resulting in adaptation of the implanted bone.

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