Finite element analysis of cemented prostheses for hip replacement in elderly patients with comminuted intertrochanteric fracture

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BACKGROUND: *To determine postoperative stress distribution after cemented arthroplasty in elderly patients with comminuted intertrochanteric fracture, and assist in determining a rational prosthetic stem length.*
METHODS: *A three-dimensional (3D) model of intertrochanteric fracture was established using the Mimics and Unigraphics modeling software, which included the 3D model of comminuted intertrochanteric fracture, two long-stem(#4,#5) and one short stem(#3) prostheses, and the mantle layer of cement. The bone defect of the calcar femorale was replaced with a 5-mm thick cement. Then, the 3D finite element model of those materials was established, boundary conditions of force were imposed, and material parameters were set. Accordingly, a finite element analysis was performed to this model in stress.*
RESULTS: *(1) The stress of the femur in the three-stem replacement prosthesis models increased from proximal end to distal end in the same pattern, while a stress concentration region was found at 5 mm from within the distal tip of the short-stem prosthesis (#3), which had a peak value of 67.85 MPa. However, no stress concentration was found on the long-stem prosthesis model. (2) For the short-stem prosthesis, the stress distribution of the cement-prosthesis interface was significantly concentrated in the distal region around the prosthesis end, in which the peak value of the lateral interface exceeded the fatigue strength of the bone cement. However, the biomechanics for the long prosthesis was better appreciated.*
CONCLUSION: *Long prosthesis stems may theoretically be a better option for comminuted intertrochanteric fractures in elderly patients. However, the application of exceedingly long prosthesis stems would not be a better option.*

KEY WORDS: Comminuted intertrochanteric fracture, Elderly patients Finite element analysis, Prosthetic replacement

Introduction

Intertrochanteric fracture in elderly patients has become more and more common with the arrival of the aging society, and the main reason resulting in hip fracture in elderly patient is osteoporosis. Therefore the unstable intertrochanteric hip fractures which are comminuted because of poor bone quality account for approximately one quarter of all hip fractures in the elderly, and are increasing in frequency [1]. Present available options for treating intertrochanteric fractures include conservative traction, open reduction and internal fixation (ORIF), and prosthetic replacement [2]. Although the ORIF with good reduction and fixation quality could have satisfactory curative effects and better levels of activity as prosthetic replacement in elderly patients with unstable intertrochanteric fractures [3-5]. However more and more failure cases such as cases with fracture nonunion and screw looseness have been reported, the majority of

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patients with severe osteoporosis, inflammatory arthritis, or unstable fracture patterns with poor bone quality may benefit much more from prosthetic replacement [6,7]. The definite advantages of arthroplasty include shorter bed stay, earlier recovery, lower failure rate and less postoperative complications [8]. At present most of orthopedic surgeons opt to take cemented prosthesis because of osteoporosis with or without calcar replacemen [7].

However, many factors influence the result of this approach such as length of prosthetic stem, the contour of the proximal stem, the treatment of calcar femorale and bone quality. Present studies on the biomechanical features of prosthetic replacement in elderly patients with comminuted intertroehanteric fractures have been rarely reported. It remains to be determined whether prosthesis stability could be ensured after a cemented arthroplasty, whether longer stem prosthesis was superior to shorter stem prosthesis, and whether a extra long stem would make it more stabilized.

As a numerical method for solving partial differential equations, finite element method has the advantages of generality, practicality and easy application. Since Courant [1] put forward the basic idea of finite element method in 1943, the finite element method has developed rapidly in its theoretical and applied research. Since the traditional experimental methods (mechanical method, electrical measurement method and optical measurement method) cannot fully and accurately reflect the stress characteristics of femur [2], the finite element numerical simulation of the biomechanical behavior of femur has become an effective means to deepen the understanding of the human body. In 1972, Rybick [3] first applied the finite element method to the biomechanics of femur, laying a foundation for the finite element research of femur. Finite element modeling is an important part of finite element analysis. With finite element analysis, we attempted to explore the biomechanics of cemented prosthesis replacement in elderly patients with comminuted intertroehanteric fractures.

Materials and Methods

This study was conducted in accordance with the Declaration of Helsinki. This study was conducted with approval from the Ethics Committee of the Second Hospital of Shandong University. Written informed consent was obtained from the participants.

Equipment and software: Siemens Somatom Balance Spiral computed tomography(CT); Image processing software Mimics10.0; Integrated modeling software for computer aided design(CAD) system: Unigraphics (UG) and finite element analysis software ABAQUS6.5.

A healthy 75-year-old Asian male volunteer was introduced into the modeling process, with the height of 175 cm and weight of 70 kg.

Modeling Process

(1) *Three-dimensional (3D) CAD model of the femur shaft*
The whole length of the femoral shaft, with a slice width of 1.0 mm, a layer space of 0.8 mm, a window width of 1,000 and a window level 400 was scanned by CT. The Dicom data acquired was submitted to Mimics10.0 for primary processing, and was converted to the initial graphics exchange specification (IGES) format. Finally, these IGES files were imported to the UG modeling software to establish the computer simulated model.

(2) *Simulation of intertrochanteric fracture prosthesis with long or short stems, and bone cement*
Comminuted intertrochanteric fracture was simulated with the osteotomy plane from the greater trochanter's tip to the lesser trochanter's bottom, including the partial calcar femorale region (Fig. 1). Both long- and short-stem prostheses were simulated based on the parameters provided by Beijing Aikang products (Short-stem prosthesis #3: stem length, 120 mm; Long-stem prosthesis #4: stem length, 170 mm; Long-stem prosthesis #5: stem–length, 210 mm. All three types of prostheses have flat cone-shaped stems with a thickness of 13 mm and a neck-shaft angle of 135°). By Boolean operation, the prosthesis was modulated into the femoral medullary canal with an anteversion angle of 15°, and the center of the prosthetic neck top was kept parallel with the greater trochanter tip. The thickness of the intramedullary bone cement surrounding the prosthesis stem was simulated using a 3-mm cylinder [9], and the extramedullary cement (calcar femorale) was set at 5 mm. The upper boundary of the bone cement was set in accor-

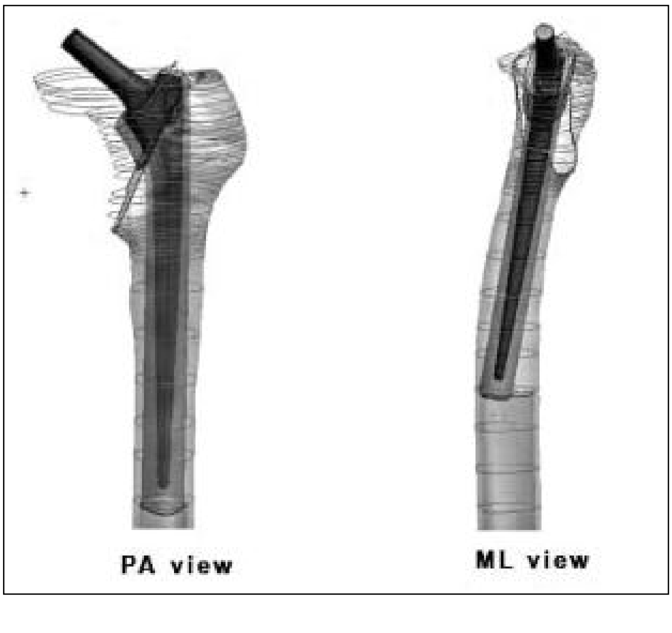

Fig. 1: PA view and Lateral view of three-dimensional Establishment (Red: prosthetic stem, Green: cement mantle, White: cortical bone)

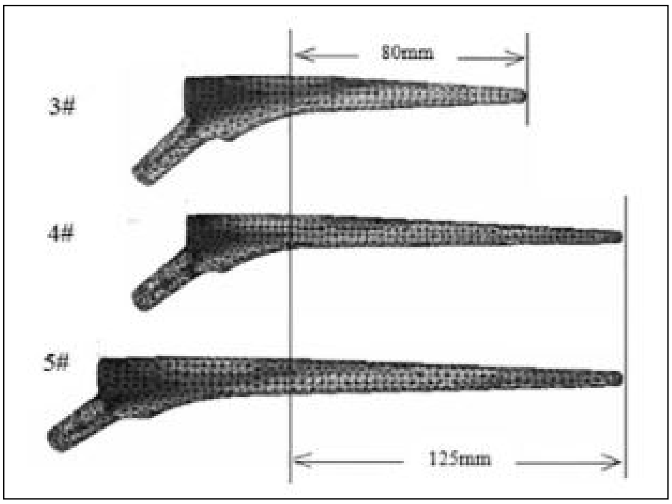

Fig. 2: The research scope of three different length prosthetic stem, matching with the femur and bone cement (The stress increases gradually from blue to red, the same below)

dance with the osteotomy plane in the regular total hip arthroplasty, and the lower boundary was 10 mm from the prosthesis end tip.

(3) *Establishment of the three-dimensional finite element model*
Using ABAQUS6.5, the modified quadratic subdivision of tetrahedron unit C3D10M was adopted for meshing. Then, the sharp edges were polished to improve the accuracy of analysis in some critical sections such as the interface between the prosthesis and bone cement, and the interface between the bone cement and femur.

Definitions of Material Properties

The material properties of the anatomical properties were determined based on previous reports [10,11] on finite element analysis, which is shown in Tables I, II.

Table I - *Material property parameter*

Material	Elastic modulus (MPa)	Poisson ratio
Vitallium	220000	0.3
Bone cement (PMMA)	2620	0.35
Cancellous bone	2130	0.3

Table II - *Cortical bone property parameter*

Material	Young's modulus of elasticity (MPa)	Shear elastic modulus (MPa)	Poisson ratio
Cocortical bone	17000	3300	0.31

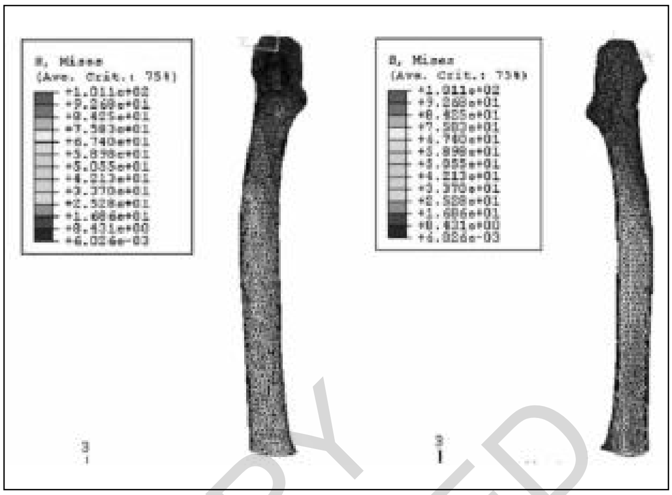

Fig. 3: The stress nephograms of femur model(medial and lateral)

It was assumed that the prosthesis was firmly fixed into the femoral canal by bone cement. Boundary conditions were applied on the distal plane of the femur, and the degrees of freedom were completely restrained. Loading conditions were applied to simulate the instantaneous phase of standing on one leg with normal gait (body weight: 70 kg). When a healthy person walks, the load on the femoral head is about 3.2 times that of a human body weight (about 70 kg). With the muscular loading neglected, stress distribution among the prosthesis, bone cement and femur was analyzed. According to Bergmann's femoral loading criteria, the values of the loads applied instantaneously on the femoral head on the X-axis, Y-axis, and Z-axis are 728.12 N, 196.46 N, and 2072.02 N, respectively (positive values indicate that the force component is in the same direction as the coordinate axis), loaded to the center of the femoral head.
Stress distributions on the interface of the femur-bone cement and bone cement-prosthetic stem were analyzed using the ABAQUS6.5 software. This study set Von mises stress as the main indicator for measuring stress level, which is a kind of integrated stress. The area of stress alterative concentration was chosen as study object (Fig. 2), the study region was defined as 80 mm from #3 tip to the proximal end, and 125 mm from #4 and #5 tip to the proximal end. We mainly determined the values by the equivalent stress distributions curve of the study object, the X direction was the variable Arc length, and the Y direction was the variable Von mises equivalent stress, and mean inter-group(#3 *vs.*#4) and intra-group(#4 *vs.*#5) stress values were respectively compared. Furthermore, the stress nephogram and stress distributions curves were drawn. The peak value and regional mean value were used to depict stress alteration, and the latter was calculated using a mean stress value of four corresponding points 5 mm from within the peak value area.

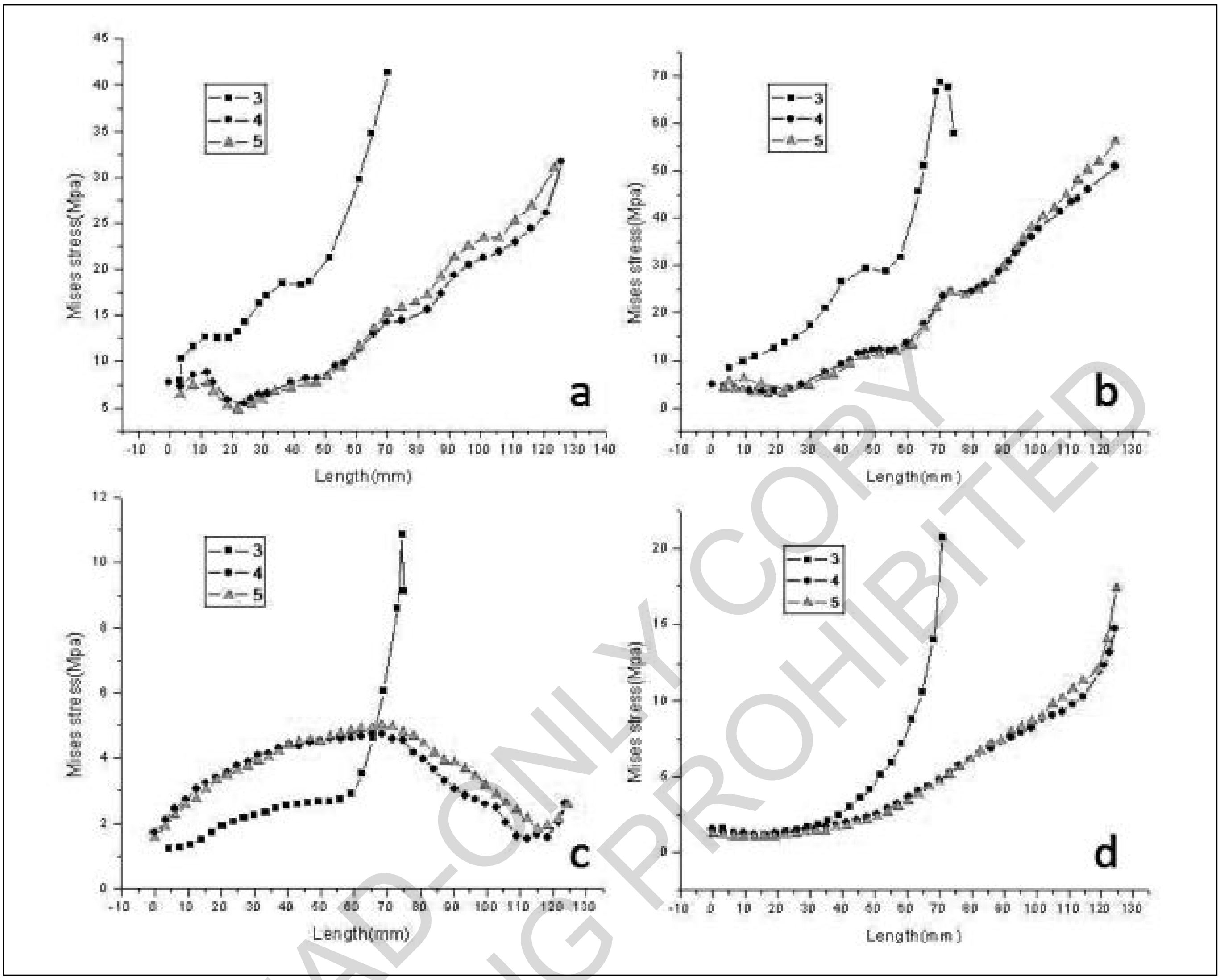

Fig. 4: The stress distribution curves of different object of study. A) The stress distribution curves of interface of femur-bone cement corresponding to the prosthesis taper (medial); B) The stress distribution curves of the interface of femur-bone cement corresponding to the prosthesis taper (lateral); C) The stress distribution curves of the interface of bone cement-prosthetic stem corresponding to the prosthesis taper (medial); D) The stress distribution curves of the interface of bone cement-prosthetic stem corresponding to the prosthesis taper (lateral).

Results

Stress Distribution of the Femur

As shown in the stress nephograms of the femur model (Fig. 3), the stress distribution of the femur gradually increased from the proximal end to the distal end (the color changed gradually from blue to red). Stress concentrated most significantly at the lower 1/3 part of the femur shaft, and decreased gradually towards the distal end. The red region of the lateral stress nephogram was significantly larger than that of the medial.

The stress distribution curves of the medial and lateral of the femur corresponding to the prosthesis taper are illustrated in Figs. 4a, b. The X direction of stress distribution curves is the variable Arc length, and the Y direction is the variable Von mises equivalent stress. The medial stress of the femur in the three-stem replacement prosthesis models increased from the proximal end to distal end in the same way, without an obvious stress concentration region. Similarly, the lateral stress of the femur also increased from the proximal end to distal end, while a stress concentration region was found at 5 mm from within the distal tip of short-stem prosthesis #3, with the peak value of 67.85 MPa. However, no stress concentration was found on the long-stem prosthesis model.

The medial and lateral peak values of the femur after the replacement of short-stem prosthesis #3 were 83.35 MPa and 96.86 MPa, respectively, and the corresponding peak values of the femur after the replacement of long-stem prosthesis #4 were 82.11 MPa and 95.88

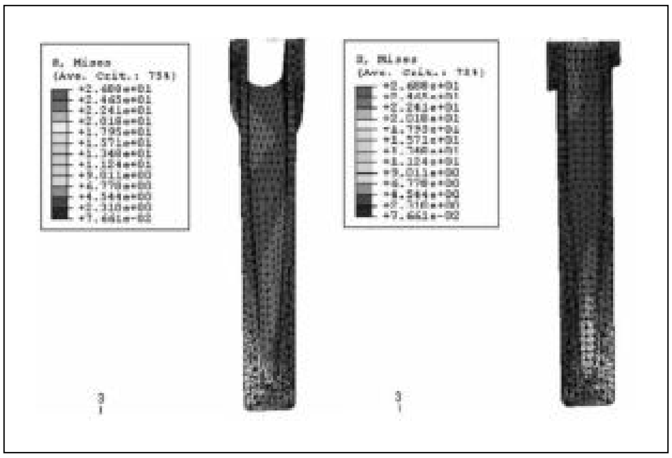

Fig. 5: The stress nephograms of the interface of bone cement model(medial and lateral)

MPa, respectively. Furthermore, the medial and lateral regional mean values after the replacement of short-stem prosthesis #3 were 78.25±4.93 MPa and 89.54±6.69 MPa, respectively, and the corresponding mean values after the replacement of long-stem prosthesis #4 were 77.12±4.10 MPa and 86.75±7.18 MPa, respectively. *T*-test revealed that there was no significant difference in regional mean stress of the medial and lateral femur between the replacement of short-stem #3 and long-stem #4 prosthesis (medial t=0.395, P>0.05; lateral t=0.636, P>0.05).

The medial and lateral peak values of the femur after long-stem #5 prosthesis replacement were 82.11 MPa and 95.90 MPa, respectively, and the corresponding mean values were77.06±3.98 MPa and 87.85±6.88 MPa, respectively. There was no significant difference between long-stem #4 and long-stem #5.

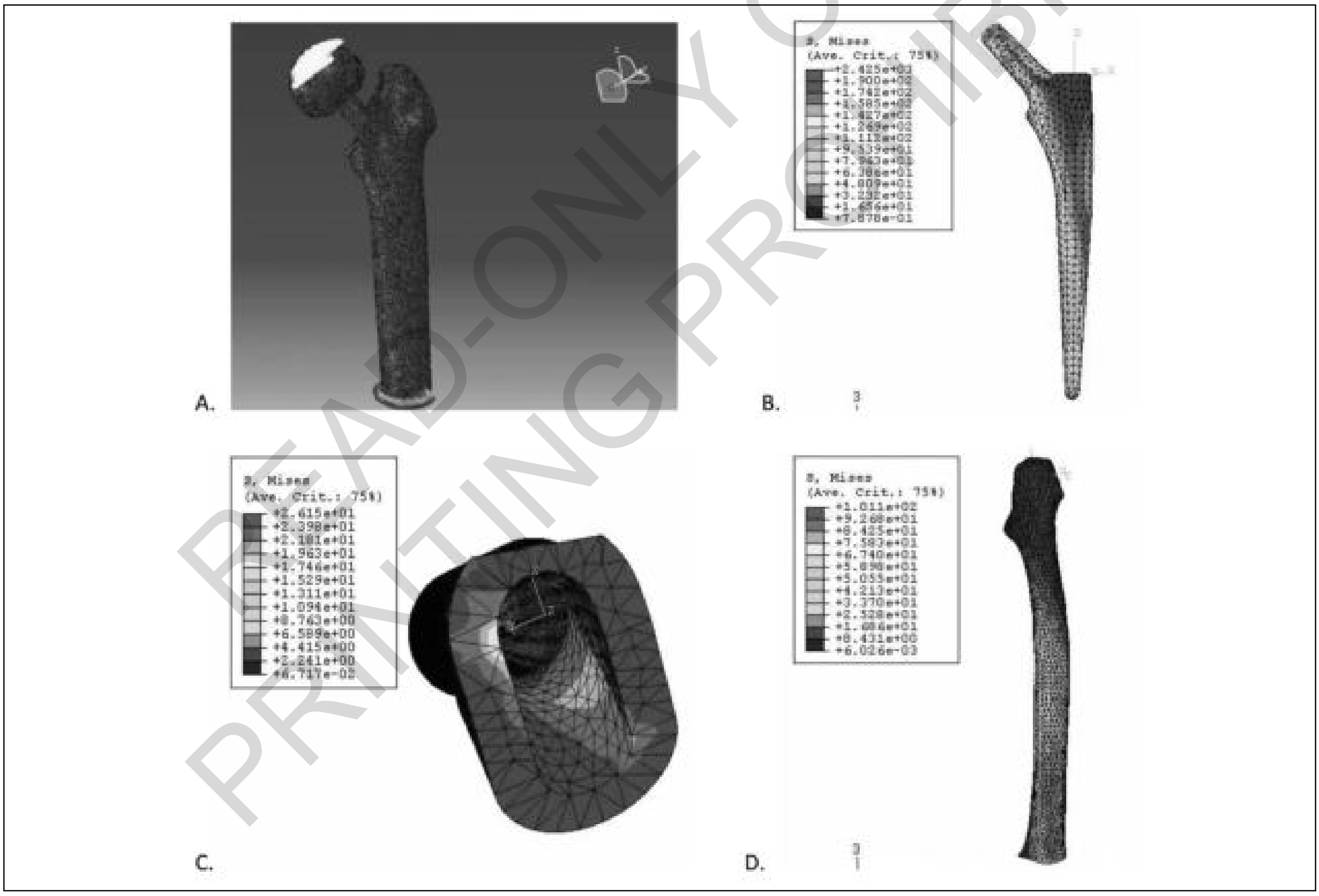

Fig. 6: The schematic diagram about the force of 3-axis. A) the simulation diagram of triaxial loading; B) the stress cloud diagram of the prosthesis; C) the stress cloud diagram of the bone cement jacket (top view); D) the femoral stress cloud diagram.

The calculation formula:

$$\bar{\sigma} = \sqrt{\frac{1}{2}[(\sigma_x^2-\sigma_y^2)^2+(\sigma_y^2-\sigma_z^2)^2+(\sigma_z^2-\sigma_x^2)^2+6(\tau_{xy}^2+\tau_{yz}^2+\tau_{zx}^2)]}$$

($\bar{\sigma}$: equivalent stress, σ_{ij} :Stress component)

Stress Distribution of the Interface Between the Bone Cement and Prosthetic Stem

The stress distribution curves of the cement-prosthesis interface and stress nephograms are presented in Figs. 4c, 4d and 5, 6.
For the replacement of short-stem prosthesis #3, the stress of the medial interface slowly increased from the lower part of the calcar region to the distal end. Then, this rapidly increased to a peak value of 10.9 MPa at 5 mm from the distal end of the prosthetic stem, and decreased. The mean value of the stress concentration region was 6.75±3.03 MPa. Similarly, the stress of the lateral interface gradually increased from the proximal end to distal end, and reached a peak value of 21.3 MPa in the region of distal end of the prosthetic stem. Nevertheless, this value exceeded the fatigue strength limit of bone cement, which was 17 MPa, as reported by literatures[12,13]. The mean value of this region was 12.56±5.12 MPa.
For the replacement of long-stem prosthesis #4, the stress of the medial interface slowly increased from the lower part of the calcar femorale region, and reached its first peak value of 4.78 MPa at the middle part of the prosthesis; and the regional mean value was 4.59±0.15 MPa. Then, the stress decreased towards the distal end, but increased again at the distal end and reached the second peak value of 2.48 MPa; and the regional mean value was1.87±0.38 MPa. Compared with the regional mean value of the short stem prosthesis at the corresponding section, a statistical significance was revealed (t=3.56, *P*<0.05). Similarly, the stress of the lateral interface gradually increased from the proximal end to distal end, and reached a peak value of 14.83 MPa; and the regional mean value was 12.49±1.57 MPa at the distal end of prosthetic stem. Both the peak value and regional mean value were under the fatigue strength limit of the bone cement. In comparing the regional mean value with short-stem prosthesis #3 at the corresponding section, no statistical significance was detected (t=0.029, *P*>0.05).
Both long-stem prosthesws #4 and #5 had the same rule of stress distribution, the regional mean values of medial stress concentration for these two were 4.81±0.12 MPa and 1.92±0.34 MPa, the peak value of lateral stress was 17.25 MPa, and the mean value was 12.85±1.88 MPa. In comparing the regional mean values of long-stem prosthesis #4 with long-stem prosthesis #5, no statistical significance was detected. Nevertheless, the lateral stress peak value of long-stem prosthesis #5 exceeded the fatigue strength limit.

Discussion

Accuracy and Rationality of the Established Model

By Dicom 3.0 network transmission, the digital information collected by CT was imported to the image processing software to help in computer modeling, which is currently the standard approach for finite element analysis [14,15] The accuracy of the model was correlated with meshing, and the size, shape and amount of units. More units and a denser mesh lead to better model quality. In our procedures, the total units of the short-stem prosthesis model was 60,548, while the total units of the long-stem prosthesis model was 65,686, which was almost 50% more than those of the models from other peer studies [16,17]. Moreover, we performed the convergence test using different mesh densities to test the accuracy and rationality of the established model, the result proved to be reasonable.
The bone density significantly affects the bone-cement strength, especially intertrochanteric fracture of elderly patients often had osteoporosis. We reconstructed bone model based on a healthy subject who had no serious medical problems such as heart disease and could be free to move. But the bone density T value of this subject was -2.0, his bone mass was reduced, although it did not meet the diagnostic criteria for osteoporosis, but basically represented the average bone condition of this age group.
Intertrochanteric fracture of elderly patients used to be complicated by osteoporosis, and the comminuted calcar also brings challenges and controversy to arthroplasty management. In view of this, we reconstructed the calcar with thickened bone cement. Then, in establishing the finite element model, we chose the osteotomy line from the greater trochanter's tip to the lesser trochanter's bottom (the iliopsoas adhesion remained intact) to simulate a comminuted intertrochanteric fracture with the calcar involved. The thickness of the bone cement forreconstructing calcar was set at 5 mm, and the thickness of the intramedullary prosthesis cement mantle was 3 mm[9].

Stress Distribution of the Femur After Prosthesis Replacement

The stress of the hip joint was transferred to the femur through the prosthesis and cement, which is different from the pattern under a normal physiological condition. Previous researches have demonstrated that the stress of a normal human femur concentrated on the lower 1/3 part of the femur shaft and calcar, and the stress value of the shaft was higher than that of the calcar.
The present study indicated that after prosthesis replacement of the comminuted intertrochanteric fracture, the stress concentration region of the femur shaft was consistent with the normal human femur. This may be because the prosthesis replacement did not change the anatomical shape and stress transfer pattern of the femur. However, the stress distribution curves of the femur corresponding to the prosthesis taper revealed different results between the long- and short-stem prostheses. In

addition, since there was no apparent stress concentration region surrounding the prosthesis, the long stem was a better fit for the biomechanical requirements of the human femur.

Stress Distribution of the Interface Between the Bone Cement and Prosthesis

A series of researches [18-20] have indicated that the stress of the bone-cement interface was less than that of the cement-prosthesis interface. Furthermore, the osteolysis caused by the debris from the interface between the bone cement and prosthesis is mainly responsible for the loosening of the prosthesis. Therefore, it is very important to analyze the stress distribution of the interface between the bone cement and prosthesis.

Our analysis on the stress distributions of the interfaces between the bone cement and prosthesis were apparently different between long and short stem types. The peak value of the stress of the lateral side of the distal end of the interface between the bone cement and short stem prosthesis reached 21.3 MPa, and the range of the fatigue strength limit of the bone cement was thought to be 8-17 MPa in literatures [12,13]. Updated techniques such as the vacuum mixture and the pre-heating of the stem could reinforce cement fatigue resistance [21,22]. Furthermore, compared with long-stem prosthesis, the probability of osteolysis and loosening was higher in the short stem type. In addition, we found that two different lengths prosthesis (170 mm and 210 mm) had a similar rule of stress distribution after comparison. However, if the length of the prosthesis is increased too much, the stress on the bone cement would be too high; and this would cause the stress to be very near the cement fatigue strength (limit of stress required for bone fracture) or led to cement fragmentation. It was also considered that it might be correlated to the block of the anterior arch of the femur.

The reason why we focus our research on the stress values of different subjects is because the existence of stress concentration will determine the stability of the prosthesis and even affect the service life. If the stress exceeds the range of the fatigue strength limit of this material, damage will soon occur. If the interface stress concentration is obvious, it means that the probability of loosening of the prosthesis is greatly increased, and this situation should be avoided. In the treatment of intertrochanteric fractures, the length of the prosthesis will affect the stress changes at each interface under other conditions. Our study found that stress concentrations at the ends of common short-item prostheses meant increased risk of failure after replacement, and the optimal prosthesis length was recommended to be 170 mm.

The treatment for comminuted intertrochanteric fracture included conservative treatment and surgical treatment. In surgical treatment, the common treatment has side nail board, artificial prosthesis replacement and Intramedullary nail which has become the preferred fixation method for comminuted unstable fractures. Stoffel K et al.[23] had reported that from a biomechanical point of view, the femoral neck system was a valid alternative to treat unstable femoral neck fractures, representing the advantages of a minimally invasive implant with comparable stability to the 2 DHS systems and superior to cannulated screws. Saini P et al. [24] reported that biological fixation of comminuted subtrochanteric fractures with PF-LCP provided stable fixation with high union rate and fewer complications. Chinzei N et al. [25] compared the ability of the different types of HFDs to prevent femoralhead rotation and found that the ability to stabilize femoral head appeared to be greater with blade-type materials than with screw-type materials.

Limitations

There are many limitations in this study. For example, The sample size was limited, the choice of subjects should be multiple rather than one. Patients with osteoporosis should be selected, which may be more representative. We have attempted to establish an intertrochanteric fracture model, but there may be some deficiencies in this model, such as thickened bone cement to reconstruct the calcar femorale, although this is sometimes the case in clinical work. The calculation formula and schematic diagram about the force of 3-axis were lacking in recent study. Besides, another important factor on affecting the successful rate of surgery was the primary stability after implantation and need further research. These limitations will continue to be improved in the next study.

Conclusions

In conclusion, by finite element analysis, we found that the stress distribution of the femur did not significantly change after prosthesis replacement, and the peak value of the stress of the lateral side of the bone-cement interface for short-stem prosthesis significantly exceeded the fatigue strength limit of the bone cement. Hence, the probability of loosening would be theoretically higher after short-stem prosthesis replacement. Therefore, we come into a conclusion that long-stem prosthesis may be a better fit for the treatment of comminuted introehanteric fracture in elderly osteoporotic patients. However, rational stem length for the prosthesis should be chosen, and not the longest one.

Riassunto

Per determinare la distribuzione dello stress postoperatorio dopo artroplastica cementata in pazienti anziani con

frattura intertrocanterica comminuta e cercare di determinare una lunghezza razionale dello stelo protesico, è stato adottato un modello tridimensionale (3D) di frattura intertrocanterica utilizzando il software Mimics e Unigraphics, comprendente il modello 3D della frattura intertrocanterica comminuta, due protesi a gambo lungo (n. 4, n. 5) e una a gambo corto (n. 3) e lo strato di cemento del mantello.
Il difetto osseo del calco del femore è stato sostituito con un cemento spesso 5 mm. Quindi, è stato definito il modello 3D degli elementi finali di quei materiali, sono state impostate le condizioni delle forze di confine e sono stati inseriti i parametri del materiale. Di conseguenza è stata fatta un'analisi degli elementi finiti su questo modello in condizioni di stress con i seguenti risultati.
(1) Lo stress del femore nei modelli di protesi sostitutiva a tre steli è aumentato dall'estremità prossimale all'estremità distale nello stesso schema, mentre una regione di concentrazione dello stress è stata trovata a 5 mm dall'interno della punta distale dello stelo corto protesi (n. 3), che aveva un valore di picco di 67,85 MPa. Tuttavia, nessuna concentrazione di stress è stata trovata sul modello di protesi a stelo lungo.
(2) Per la protesi a stelo corto, la distribuzione delle sollecitazioni dell'interfaccia cemento-protesi era significativamente concentrata nella regione distale attorno all'estremità della protesi, in cui il valore di picco dell'interfaccia laterale superava la resistenza a fatica del cemento osseo.
Tuttavia, la biomeccanica per la protesi lunga è stata considerata migliore.
In conclusione: gli steli protesici lunghi possono teoricamente essere un'opzione migliore per le fratture intertrocanteriche comminuta ei pazienti anziani. Tuttavia, l'applicazione di steli protesici estremamente lunghi non rappresenterebbe l'opzione migliore.

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