SCIENTIFIC ARTICLE

Femoral Stress and Strain Changes post-Hip, -Knee and -Ipsilateral Hip/Knee Arthroplasties: a Finite Element Analysis

Zhen-hui Sun, MD¹, Yue-ju Liu, MD^{2,3}, Heng Li, MD^{2,3}

¹Center for Joint Diseases, Tianjin People's Hospital, Tianjin, ²Department of Orthopaedics, Third Hospital of Hebei Medical University, and ³Biomechanical Key Laboratory of Hebei Province, Shijiazhuang, China

Objective: To identify the optimal ratio of free femur for minimizing the risks of periprosthetic fracture.

Methods: Three dimensional models of the femur with hip and knee stem elongation were constructed. With the distal femoral condylar surface fixed in a three dimensional model, the femoral head loading was performed according to the methods described by Huiskes and van Rietbergen in the models of hip replacement, knee replacement with or without hip stem or knee stem elongation. The maximum principal stress (MPS) and maximum principal elastic strain (MPES) of the femur were recorded and their relationships to the free femur ratio were analyzed using Pearson's correlation analysis.

Results: There were no obvious changes in MPS and MPES with hip stem elongation from 100 to 180 mm. In ipsilateral hip and knee replacement, the MPS and MPES had a tendency to decrease with knee and hip stem elongation. The MPS and MPES were mainly located in the anterior medial side of the middle to distal femur post-hip replacement and distalized with stem elongation. When the knee stem had been elongated more than 120 mm, the stress and strain concentrated strongly in the middle of the femoral shaft. There was a positive correlation between MPS and MPES to the free femur ratio ($P < 0.01$); however, no optimal ratio of free femur that would minimize the risks of periprosthetic fracture was identified.

Conclusion: Positive correlations were found between implant free femur and stress and strain changes in total knee arthroplasty, total hip arthroplasty and ipsilateral hip and knee replacement.

Key words: Arthroplasty; Finite-element analysis; Periprosthetic fracture

Introduction

Total hip arthroplasty (THA) and total knee arthroplasty (TKA) are well established standard methods for treating end stage of hip and knee diseases that can relieve pain and restore joint function. With prosthesis modification and improvement in surgical techniques, satisfactory clinical results and long term implant survival have been extensively reported¹⁻⁴. However, as with all surgical procedures, THA and

TKA may be accompanied by complications, including infection, instability, aseptic loosening and periprosthetic fracture. Among these complications, periprosthetic fracture is relatively common and challenging⁵.

After isolated THA, femoral fractures occur in approximately 0.1% of cases⁶; whereas after isolated TKA, there is reportedly a 1% to 2% incidence of supracondylar fractures of the femur⁷. Higher periprosthetic fracture ratios have also been

Address for correspondence Heng Li, MD, Department of Trauma Center, Third Hospital of Hebei Medical University, 139, Zi-qiang Road, Shijiazhuang, Hebei, China 050051 Tel: 0086-311-88603682; Fax: 0086-311-87023626; Email: liuyueju1983@gmail.com, Lihengdoctor@163.com
Yue-ju Liu and Zhen-hui Sun contributed equally to this paper.

Disclosure: The authors confirm that there are no known conflicts of interest associated with this publication and there has been no significant financial support for this work that could have influenced its outcome.

Received 28 October 2013; accepted 19 February 2014

Fig. 1 Interprosthetic femoral fracture. Lateral femoral radiography indicating the distal femoral fracture post ipsilateral hip and knee replacement.

reported: Berry *et al.* reported an overall incidence after THA of 0.1% to 6%⁸ and Rorabeck *et al.* reported fracture ratios from 0.3% to 5.5% after TKA⁹. Because of the escalating number of joint replacements and increased activity and survivorship of older subjects, the ratio of periprosthetic fractures is expected to increase^{10,11}. Among the various periprosthetic fractures, one particular one, namely interprosthetic fracture (IPF), has recently attracted the attention of surgeons (Fig. 1). Although IPFs is not a commonly complication in clinical practice, Kenney *et al.* followed a series of 320 limbs with ipsilateral hip and knee arthroplasties, and 4 cases suffered IPFs, making the incidence 1.25%⁷.

The gradually increasing ratio of revision arthroplasties of hip and knee combined with additional long hip stem or stemmed knee components to maintain prosthesis stability^{12,13} will likely lead to a higher incidence of IPFs with proximal and distal stem fixation. Soenen *et al.* proposed adding a category D to the Vancouver classification, corresponding to IPFs after TKAs with diaphyseal extension stems¹⁴. With the availability of intramedullary (IM) metal stem elongates, the stress and strain on the “implant free femur” will theoretically increase, making patients more prone to fractures. Lehmann *et al.* reported four cases of IPFs that had occurred after minimal trauma in 6 years and proved that in two cases IM implants had reduced the strength significantly¹⁵. IPFs are more difficult to treat than other periprosthetic fractures due because of the proximal and distal metal implants around the fracture line.

Because it is well known that the “implant-free femur” plays an important role in the occurrence of periprosthetic fractures, when selecting implants to treat fractures, surgeons should consider the likely prognosis. Some surgeons advocate short hip femoral stem prostheses to preserve host bone and avoid “stress riser” on the femur, anticipating that using these will decrease the periprosthetic fracture ratio¹⁶. However, the optimal length for “implant free bone” has not been established. The aims of this study were to evaluate changes in the distribution of stress and strain in the femur post-TKA, -THA and -ipsilateral hip/knee replacements with or without stem elongation. In this study, we subjected hip and knee prostheses to finite-element analysis (FEA) under simulated daily physical loading in an attempt to identify the optimal ratio of free femur to minimize the risks of periprosthetic fractures.

Materials and Methods

Finite Element Models

This study was approved by the Ethics Committee of our hospital. Using 3D model reconstruction software (MIMICS 10.01, Materialise, Leuven, Belgium), a digital right femur was reconstructed using four node hexahedron isometric elements from two dimensional computed tomography (CT) images. The total femoral length was 377 mm measured from the femoral head center to the distal femoral intercondylar line. The CT image dataset was obtained by scanning the lower limb of a single 41-year-old female subject, with a weight of 60 kg and a height of 160 cm (Fig. 2). 3D models of femoral stems and knee prostheses were scanned by 3D laser scanner (ATOS Std 600, Atos, München, Germany). The prostheses selected for this study were: i) hip femoral stem and femoral head, Ribbed Anatomic with proximal anatomic contours and ribs (Waldemar Link GmbH, Hamburg, Germany); and ii) knee prosthesis and knee femoral stems, Sigma RP-F Knee System with closed intercondylar box (Depuy, Warsaw, IN, USA). Because the elements and nodes of the finite element models would change depending on bone resection and implantation of prostheses with different lengths of stem or rod, we constructed only some representative models for testing, including primary free femur, knee prosthesis, knee prosthesis with 16 mm rod elongation, hip prosthesis with 10 mm stem and hip prosthesis with 16 mm stem (Table 1).

Next, the finite element models of the femur and prosthesis were transported and polished in Geomagic Studio software 10.0, after which the STP format files were saved and transported to Unigraphics NX 6.0 (Siemens, Erlangen, Germany). Osteotomies and joint replacements were performed as real arthroplasty surgery in the workbench of Unigraphics, after which the paraSolid format files were saved and prepared for FEA.

Finite Elementary Analysis

The models were subsequently imported into ANSYS Workbench v13.0 (ANSYS, Canonsburg, PA, USA). The properties of femoral bone; hip/knee prostheses and stems were assumed

Fig. 2 Three dimensional model of the femur and axial CT images of proximal, mid shaft and distal femur.

to be isotropic and linearly elastic. To simulate the mechanical behavior of the femur and prosthesis used as a benchmark, the following material properties were assumed (Poisson's ratio $\nu = 0.3$ for titanium alloy, cortical and spongy bone)¹⁷. The elastic modulus of the components for cortical bone was $E = 14,200$ MPa; for spongy bone $E = 69$ MPa; and for titanium alloy $E = 107,000$ MPa¹⁸. A fixed bond between the bone and the implant along the interface was presumed.

The femoral head loading was performed according to the methods described by Huiskes and van Rietbergen¹⁹. The following three versions of daily loading cycles were assessed: (i) hip 35° flexion, representing the loading in of heel strike of stance²⁰, $N = 2132$, coronal angle $\alpha = 23.4^\circ$ and sagittal angle $\beta = 5.7^\circ$; (ii) hip 0° flexion (45% of walking cycle), $N = 1586$, $\alpha = 21.9^\circ$, $\beta = -4.6^\circ$ ²⁰; and (iii) hip 70° flexion (ascending stairs), $N = 1690$, $\alpha = 25.0^\circ$, $\beta = 15.0^\circ$ ²¹. The loading time was 1 second for all models.

With the distal femoral condylar surface fixed in a three dimensional model, the FEA tests were performed in the following four steps: (i) the loading on the free femur model; (ii) the loading on the femur with hip prosthesis (hip stem elongated from 100 to 180 mm); (iii) loading on femur with knee prosthesis and stem elongation from 100 to 160 mm; and (iv) loading on the femur with both hip and knee stem elongation

simultaneously from 100 mm to 160 mm. The maximum principal stress (MPS) distributions and the maximum principal elastic strain (MPES) were recorded and analyzed.

Statistical Analysis

The relationships between free femur ratio and MPS and MPES were analyzed using Pearson's correlation analysis. Statistical analyses were performed using SPSS version 17.0 (SPSS, Chicago, IL, USA), and P values < 0.01 (two-tailed) were considered significant.

Results

Values of and Distribution Changes in MPS

The changes in MPS under different loading are shown in Figure 3. In the free femur, the MPS loading under 1586 N, 1690 N and 2132 N were 98.54, 147.36 and 143.51 MPa, respectively. After hip replacement with a 100 mm stem, the MPS were 94.20, 145.49 and 137.29 MPa, respectively, which do not differ significantly from those of the free femur. With hip stem elongation from 100 mm to 180 mm, the MPS did not change significantly. After knee replacement, MPS loading with 1586 N and 213 N increased to 98.63 and 143.61 MPa, respectively; whereas under 1690 N loading it decreased to 142.67 MPa. When the knee stem was elongated, MPS loading with 1690 N decreased gradually. However, with stems of 110 and 120 mm, MPS loading with 1586 N and 2132 N had a platform similar to the MPS in the knee model without a stem. Thereafter, the MPS decreased gradually with stem elongation. With 160 mm stems, the MPS under loading with 1586 N, 1690 N and 2132 N decreased to 84.74, 121.13 and 126.16 MPa, respectively.

With varying loading on the femur with knee and hip stem elongation, the MPS had a tendency to decrease. For femurs with 100 mm knee and 100 mm hip stems, the MPS were 93.80, 128.70 and 137.71 MPa, which are similar to those for TKA without stems. However, in femurs with 160 mm knee and 160 mm hip stems, the MPS were 85.40, 110.21, and 114.80 MPa, which are lower than those found for any type of femoral model. As shown in Figure 3, the MPS under loading with 1586 N was lower than that under loading with 1690 and 2132 N.

TABLE 1 Number of nodes and elements in the FE models

Items	Nodes	Elements
Free femur	1,020,477	692,536
Hip prosthesis with 10 mm stem	66,101	39,840
Hip prosthesis with 16 mm stem elongation	88,834	53,406
Knee prosthesis	150,427	96,554
Knee prosthesis with 16 mm rod elongation	237,280	143,510

Fig. 3 MPS loading with 1586 N, 1690 N and 2132 N and corresponding changes in MPS (MPa; Y axis) for various femoral models (X axis). FF, free femur; H, hip stem; K, knee prosthesis with stem; K0, knee prosthesis without stem; numbers indicate the intramedullary stem lengths in cm.

The locations of changes in MPS were similar under different loading. Figure 4 shows a representative example: under loading with 1690 N the areas of MPS concentration clearly differ between knee and hip arthroplasties. Post-hip replacement, the MPS mainly distributed in the anterior medial side of the middle to distal femur and distalized gradually with stem elongation. After knee replacement and stem elongation from 100 mm to 120 mm, the MPS location gradually moved proximally. However, when the stem was elongated more than 120 mm, the stress concentrated strongly in the middle femoral shaft. When the hip and knee stem were elongated simultaneously, the changes in MPS location were similar to those in knee replacement with stem elongation,

indicating that the MPS concentrates strongly in the implant free femur zone after implantation of a 120 mm knee stem.

Values and Changes in Distribution of MPES

The changes in MPES loading with 1586 N, 1690 N and 2132 N are shown in Figure 5. In the free femur, the MPES under loading with 1586 N, 1690 N and 2132 N were 6.877×10^{-3} , 10.074×10^{-3} and 10.066×10^{-3} mm/mm, respectively. Overall, the MPES changes in different models were similar to those in MPS. Post-hip replacement and hip stem elongation, the MPES did not change significantly and was similar to the strain of the free femur. After knee replacement and knee stem elongation to 100, 110 and 120 mm, strain loading with

Fig. 4 Distribution of changes in MPS loading with 1690 N. (a) Free femur; (b) knee prosthesis without stem; (c) 100 mm hip stem; (d) 160 mm knee stem; (e) 160 mm hip stem; (f) 160 mm hip stem and 160 mm knee stem. Red indicates maximum stress and blue minimum stress.

Fig. 5 Changes in MPES under loading with 1586 N, 1690 N, 2132 N and corresponding changes in MPES (MPa; Y axis) for various femoral models (X axis). FF, free femur; H, hip stem; K, knee prosthesis with stem; K0, knee prosthesis without stem; numbers indicate the intramedullary stem lengths in cm.

1586 N and 2132 N was greater than occurred with hip replacement. However, the strain under loading with 1690 N had a tendency to gradually decline. In ipsilateral hip and knee replacements, the model with 120 mm hip stem and knee prosthesis without stem had a similar strain value as knee replacement alone. With elongation of the hip and knee stem, the the maximum principal elastic strain declined gradually. With 160 mm stems, the MPES under loading with 1586 N, 1690 N and 2132 N were 5.295×10^{-3} , 7.781×10^{-3} and 8.109×10^{-3} mm/mm, respectively.

The locations of changes in MPES loading with 1586 N, 1690 N and 2132 N were similar to the changes in MPS; an illustrative example of location changes with 1690 N loading is shown in Figure 6). As can be seen in Figure 5, the locations of

the MPES changed significantly between knee and hip arthroplasties. The center area of the MPES was located in the anterior medial side of the middle to distal femur. Post-hip replacement and stem elongation, the focus had a tendency to migrate distally. After knee replacement and stem elongation from 100 mm to 120 mm, the MPES location gradually moved proximally. However, when the stem was elongated more than 120 mm, the strain concentrated strongly in the middle of the femoral shaft (Fig. 7). When the hip and knee stem were elongated simultaneously, the changes in MPS locations were similar to those after knee replacement with stem elongation, indicating the MPES concentrates to the implant free femur zone strongly after increasing knee stem elongation (to more than 120 mm).

Fig. 6 Distribution of changes in MPES under loading with 1690 N. (a) Free femur; (b) knee prosthesis without stem; (c) 100 mm hip stem; (d) 160 mm knee stem; (e) 160 mm hip stem; (f) 160 mm hip stem and 160 mm knee stem. Red indicates maximum stress and blue minimum stress.

Fig. 7 Distribution of changes in MPES in knee models under loading with 1690 N. (a) Femur with knee prosthesis; (b) 100 mm knee stem; (c) 120 mm knee stem; (d) 130 mm knee stem; (e) 140 mm knee stem; (f) 160 mm knee stem. Red indicates maximum elastic strain and blue minimum strain.

Correlations between MPS and MPES and Free Femur Ratio

As shown in Table 2, the MPS and MPES under loading 1586 N, 1690 N and 2132 N were all positively correlated with the free femur ratio ($P < 0.01$). As the free femur ratio declined, the MPS and MPES both decreased. In the model with 160 mm knee and hip stems (implant free femur ratio: 0.07), the MPS and MPES under different loading were significantly lower than with the free femur. The scatter distribution of the MPES under loading 2132 N is shown in Figure 8: there is a linear correlation between the strain and free femoral ratio ($R = 0.760$, $P = 0.000$).

Discussion

The most important finding of our study is that knee stems significantly influence stress and strain changes in the femur under daily loading. In addition, short knee stems can even increase the femoral principal stress and strain in some types of loading; whereas the length of hip stems has no obvious effect on stress and strain in the femur.

When we analyzed MPS and MPES changes in hip replacement models under different daily loading, we found that stress and strain did not change significantly with hip stem elongation from 100 mm to 180 mm. However, the maximum stress and strain distribution gradually and minimally distalized. Therefore, provided it is not anticipated that a TKA will be required in the future, in primary or revision THA the

choice of femoral stem can be relatively free because of the primary stability of the different types of stem and fixation method. In this FEA study of hip stems with fixed bonds to the femur bone, we found no advantages for short or long hip stems in regard to decreasing stress and strain mechanism.

After knee replacement and stem elongation, the changes in MPS and MPES were different for walk cycle loading (1586 N, 2132 N) than for stair ascent loading (1690 N). With no stem and stems of 110 and 120 mm, the MPS and MPES in walk cycle loading had a platform similar to that of the free femur; with further stem elongation both decreased gradually. However, the MPS and MPES under loading with 1690 N post-TKA decreased similar to those for the free femur and also decreased smoothly with knee stem elongation. This means that knee stems longer than 120 mm can decrease the MPS and MPES loaded on the femur. However, taking the mechanism as reflected by the site of concentration of MPS and MPES into account, the strain concentrated strongly in the middle of the femoral shaft when the stem was longer than 120 mm, which would markedly increase the likelihood of periprosthetic fractures occurring. Therefore, considering stress and strain factors comprehensively, we recommend that knee stems no longer than 120 mm be used to optimize knee prosthesis stability after primary or revision TKA.

When ipsilateral hip and knee stems were elongated simultaneously, the changes in MPS, MPES and their location

TABLE 2 Pearson correlation between MPS, MPES and ratio of implant free femur

Statistical value	Maximum principal stress			Maximum principal elastic strain		
	1586 N	1690 N	2132 N	1586 N	1690 N	2132 N
R value	0.751	0.608	0.757	0.749	0.591	0.760
P value	0.000	0.002	0.000	0.000	0.003	0.000

Fig. 8 Changes in maximum principal elastic strain according to free femur ratio. A significant linear correlation was found.

were similar to the changes after knee replacement with stem elongation, which means that with 120 mm knee stem elongation, the MPS and MPES concentrated strongly in the implant free femur whereas the absolute stress and strain declined gradually. The MPS changes were contrary to what we commonly believe; that is, the peak MPS in the implant free femur increased as the size of the gap between the two ipsilateral femoral stems declined. Even with 160 mm knee and hip stems (implant free femur ratio: 0.07), the MPS values under different loadings were significantly lower than those of the free femur and we detected no stress riser mechanism. These findings are similar to those of Iesaka *et al.*, who established a FEA model with a 300 mm long hollow cylinder femur with two solid 14 mm cylinders metal stems fixed intramedullarily and analyzed it by FEA²². They found that the size of the gap between two well-fixed stem tips did not affect the maximum tensile stress experienced in the femur. However, these researchers did not assess strain changes between the two stems. In our study, we found a positive correlation between MPS, MPES and free femur ratio, which indicates that shorter implant free femurs are under less stress and strain. In addition, the distribution is not concentrated. However, we were unable to identify an optimal free femur ratio to prevent the stress and strain concentration leading to IPF, meaning that we did not confirm our primary hypothesis.

Therefore, even if future epidemiological data indicates a higher rate of femoral periprosthetic fracture than for the free femur under daily physical loading, the mechanism for this could not be explained simply by post-arthroplasty stress and strain changes. The only finding of our study that can explain the propensity to periprosthetic fractures is the concentration of the MPS and MPES, which is mainly influenced by knee stems longer than 120 mm. However, because periprosthetic fractures can occur in many different position^{23,24}, such as Van-

couver B and C, their propensity to occur after arthroplasties cannot be attributed to the MPS and MPES distribution concentrating phenomenon we identified in the present study.

Many other factors related to the periprosthetic fractures have been reported. They are commonly divided into two groups: design of the prosthesis and individual host variables²⁵. Hu *et al.* reported that hip stems with ribs with sharp corners are associated with significant stress increases that can lead to intraoperative periprosthetic fractures²⁶. Iesaka *et al.* reported that tips of loose stems can act as stress risers particularly with thinner cortices (3 mm thick)²². Kenny *et al.* reported that three of their four case of IPF had rheumatoid arthritis and had received prolonged courses of steroid therapy⁷. Pre-injury radiographs in these three patients showed osteolysis and evidence of stress risers in the area between the prostheses. Gough *et al.* also showed that generalized osteoporosis makes patients more susceptible to periprosthetic fractures²⁷. Femoral cortical defects created by reaming or broaching of the femoral canal in cemented stem or long hip stem implantation has been reported as another important factor that can induce periprosthetic fractures^{28–32}. Recently, Wilson *et al.* showed that bypass of cortical windows with a revision femoral component may not reduce the risk of periprosthetic fracture³¹. In addition to these bone and prosthesis factors, the mechanism of trauma and internal bone stress at the moment of fracture should be considered when explaining periprosthetic fractures.

The present study has some limitations. First, our findings are limited to a finite element model of the femur in which the knee side is fixed. Our model may not replicate the behavior of bone and muscle *in vivo*. Second, to facilitate the FEA analysis, we assumed that the contact between bone and prosthesis was a full-range fixed bond, which is not true of real arthroplasties *in vivo*, where micro motion occurs. Third, bone quality, cortical bone thickness, type of hip and knee prosthesis and characteristics of the metal alloy are all factors that can influence the occurrence of IPF; we did not assess these factors and presumed all the metal in our study was equivalent to titanium alloy. Fourth, we did not examine the time-dependent effects of loading and relaxation. The increased stress on the femoral bone would be expected to become less significant with the progressive relaxation of stress³³. Further study is needed to evaluate the effect of different loading regimens and the specific effects of muscle forces and to compare data from testing models with clinical data.

In conclusion, we found positive correlations between implant free femur and changes in stress and strain after TKA, THA and ipsilateral hip and knee replacement. We did not identify any stress riser mechanism because of the gap in decline between proximal and distal stem tips found in this study. Because the length of the hip stem had no significant effect on stress and strain in the femur; longer stems can be selected to increase prosthesis stability provided there is no distal knee stem. If implanting of a knee prosthesis with stem elongation with or without proximal hip prosthesis is planned, we recommend the knee stem be no longer than 120 mm to avoid concentration of strain and stress.

References

1. Diduch DR, Insall JN, Scott WN, Scuderi GR, Font-Rodriguez D. Total knee replacement in young, active patients. Long-term follow-up and functional outcome. *J Bone Joint Surg Am*, 1997, 79: 575–582.
2. Hopley CD, Crossett LS, Chen AF. Long-term clinical outcomes and survivorship after total knee arthroplasty using a rotating platform knee prosthesis: a meta-analysis. *J Arthroplasty*, 2013, 28: 68–77.
3. Synder M, Drobniewski M, Sibinski M. Long-term results of cementless hip arthroplasty with ceramic-on-ceramic articulation. *Int Orthop*, 2012, 36: 2225–2229.
4. Tai CC, Cross MJ. Five- to 12-year follow-up of a hydroxyapatite-coated, cementless total knee replacement in young, active patients. *J Bone Joint Surg Br*, 2006, 88: 1158–1163.
5. Parvizi J, Vegari DN. Periprosthetic proximal femur fractures: current concepts. *J Orthop Trauma*, 2011, 25 (Suppl. 2): S77–S81.
6. Sah AP, Marshall A, Virkus WV, Estok DM 2nd, Della Valle CJ. Interprosthetic fractures of the femur: treatment with a single-locked plate. *J Arthroplasty*, 2010, 25: 280–286.
7. Kenny P, Rice J, Quinlan W. Interprosthetic fracture of the femoral shaft. *J Arthroplasty*, 1998, 13: 361–364.
8. Della Rocca GJ, Leung KS, Pape HC. Periprosthetic fractures: epidemiology and future projections. *J Orthop Trauma*, 2011, 25 (Suppl. 2): S66–S70.
9. Rorabeck CH, Taylor JW. Periprosthetic fractures of the femur complicating total knee arthroplasty. *Orthop Clin North Am*, 1999, 30: 265–277.
10. Kim KI, Egol KA, Hozack WJ, et al. Periprosthetic fractures after total knee arthroplasties. *Clin Orthop Relat Res*, 2006, 446: 167–175.
11. Su ET, DeWal H, Di Cesare PE. Periprosthetic femoral fractures above total knee replacements. *J Am Acad Orthop Surg*, 2004, 12: 12–20.
12. Turner RH, Mattingly DA, Scheller A. Femoral revision total hip arthroplasty using a long-stem femoral component. Clinical and radiographic analysis. *J Arthroplasty*, 1987, 2: 247–258.
13. Completo A, Simoes JA, Fonseca F. Revision total knee arthroplasty: the influence of femoral stems in load sharing and stability. *Knee*, 2009, 16: 275–279.
14. Soenen M, Migaud H, Bonnomet F, Girard J, Mathevon H, Ehlinger M. Interprosthetic femoral fractures: analysis of 14 cases. Proposal for an additional grade in the Vancouver and SoFCOT classifications. *Orthop Traumatol Surg Res*, 2011, 97: 693–698.
15. Lehmann W, Rupprecht M, Hellmers N, et al. Biomechanical evaluation of peri- and interprosthetic fractures of the femur. *J Trauma*, 2010, 68: 1459–1463.
16. Lombardi AV Jr, Berend KR, Ng VY. Stubby stems: good things come in small packages. *Orthopedics*, 2011, 34: e464–e466.
17. Sakai R, Itoman M, Mabuchi K. Assessments of different kinds of stems by experiments and FEM analysis: appropriate stress distribution on a hip prosthesis. *Clin Biomech (Bristol, Avon)*, 2006, 21: 826–833.
18. Viceconti M, Muccini R, Bernakiewicz M, Baleani M, Cristofolini L. Large-sliding contact elements accurately predict levels of bone-implant micromotion relevant to osseointegration. *J Biomech*, 2000, 33: 1611–1618.
19. Huiskes R, van Rietbergen B. Preclinical testing of total hip stems. The effects of coating placement. *Clin Orthop Relat Res*, 1995, 319: 64–76.
20. Graichen F, Lendrzynski H. The influence of floor materials and shoes on the hip joint loading. The Netherlands, Proceedings 7th meeting European Society of Biomechanics, Aarhus: B40, 1990.
21. Kotzar GM, Davy DT, Goldberg VM, et al. Telemeterized *in vivo* hip joint force data: a report on two patients after total hip surgery. *J Orthop Res*, 1991, 9: 621–633.
22. Iesaka K, Kummer FJ, Di Cesare PE. Stress risers between two ipsilateral intramedullary stems: a finite-element and biomechanical analysis. *J Arthroplasty*, 2005, 20: 386–391.
23. Friessecke C, Plutat J, Block A. Revision arthroplasty with use of a total femur prosthesis. *J Bone Joint Surg Am*, 2005, 87: 2693–2701.
24. Hou Z, Moore B, Bowen TR, et al. Treatment of interprosthetic fractures of the femur. *J Trauma*, 2011, 71: 1715–1719.
25. Jakubowitz E, Seeger JB, Lee C, Heisel C, Kretzer JP, Thomsen MN. Do short-stemmed-protheses induce periprosthetic fractures earlier than standard hip stems? A biomechanical *ex-vivo* study of two different stem designs. *Arch Orthop Trauma Surg*, 2009, 129: 849–855.
26. Hu KZ, Zhang XL, Wang CT, Ji WT. The contour of cementless femoral stem has minor effect on initial periprosthetic von Mises stress distribution. A 3-dimensional finite element analysis. *Saudi Med J*, 2009, 30: 947–951.
27. Gough AK, Lilley J, Eyre S, Holder RL, Emery P. Generalised bone loss in patients with early rheumatoid arthritis. *Lancet*, 1994, 344: 23–27.
28. Edgerton BC, An KN, Morrey BF. Torsional strength reduction due to cortical defects in bone. *J Orthop Res*, 1990, 8: 851–855.
29. Hipp JA, Edgerton BC, An KN, Hayes WC. Structural consequences of transcortical holes in long bones loaded in torsion. *J Biomech*, 1990, 23: 1261–1268.
30. Missakian ML, Rand JA. Fractures of the femoral shaft adjacent to long stem femoral components of total hip arthroplasty: report of seven cases. *Orthopedics*, 1993, 16: 149–152.
31. Wilson LJ, Richards CJ, Irvine D, Tillie A, Crawford RW. Risk of periprosthetic femur fracture after anterior cortical bone windowing: a mechanical analysis of short versus long cemented stems in pigs. *Acta Orthop*, 2011, 82: 674–678.
32. Beals RK, Tower SS. Periprosthetic fractures of the femur. An analysis of 93 fractures. *Clin Orthop Relat Res*, 1996, 327: 238–246.
33. Shultz TR, Blaha JD, Gruen TA, Norman TL. Cortical bone viscoelasticity and fixation strength of press-fit femoral stems: finite element model. *J Biomech Eng*, 2006, 128: 7–12.