Journal of Orthopaedic Science 25 (2020) 132–138

Contents lists available at ScienceDirect

Journal of Orthopaedic Science

journal homepage: http://www.elsevier.com/locate/jos

Original Article

Should early weightbearing be allowed after intramedullary fixation of trochanteric femur fractures? A finite element study

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ARTICLE INFO

Article history:
Received 10 June 2018
Received in revised form
14 November 2018
Accepted 5 February 2019
Available online 7 March 2019

ABSTRACT

Background: This study aims to investigate the effects of early weightbearing after intramedullary fixation of trochanteric fractures.

Methods: Femurs with different types of trochanteric fractures were modeled according to AO/OTA classification. Fractures were ideally reduced with one mm gap between fragments and fixed with intramedullary nails. Forces were applied simulating single- (Body weight: 60 kg, joint reaction force: 1999.2 N, abductor muscle force:1558.8 N) and double-leg standing positions (Joint reaction force: 196 N). In another model, a 500 Nm rotational force was applied as a simulation of a fall.

Results: A higher level of stress was determined at the calcar femorale, the fracture site, the holes for the lag screws, and the hole for the proximal locking screw on the nail, the threadless parts of the lag screws, and the mid-portion of the nail. During the single-leg stance, up to 3 mm displacement was observed with the reverse oblique type of fractures. In the simulation of the fall, 1.5 mm displacement occurred at the fracture site. No displacement was measured at stabile and type 31A2 fracture models. In addition, higher levels of stress were measured at the body of the nail (up to 133 MPa), proximal screws (up to 133 MPa) and at the bone distal to the nail (up to 84.3 MPa), but all values were under the limit of the yield stress of the bone and the titanium.

Conclusion: Full weightbearing after intramedullary fixation of trochanteric femur fractures may be allowed except in obese patients and patients with 31A3 type fractures according to the AO/OTA classification. The use of support is recommended in order to prevent complications. Implant removal can be discussed with patients after fracture union in order to prevent possible periprosthetic fractures.

1. Introduction

Trochanteric fractures are mostly seen in elderly patients and are related to increased mortality and morbidity. The aim of the treatment is to achieve stable fixation, which allows early mobilization and healing [1]. Various implants can be used in fracture fixation, but the proximal femoral nail (PFN) is one of the most popular. Despite the continuous debate, especially in unstable fractures (AO/OTA classification 31A2 and A3 type), intramedullary nailing is accepted as the leading technique by some authors due to its biomechanical advantages [2–5].

The finite element method (FEM) is widely used in mechanical engineering and became more popular in biomechanics [6]. It is a numerical method for solving problems, and divides the whole large structure into small parts. The results of those small parts are combined in order to make an approximation for the whole structure. Several types of software were reproduced to use this method in biomechanical experiments. Situations that could not be studied in clinical investigations can be generated, and various types of possibilities can be taken into account by using computer models. FEM has been frequently used in orthopedic studies over the past two decades [6–8].

There are numerous studies that have investigated the biomechanical properties of PFN; however, most of them studied just single or limited types of fractures [7,8]. This study aims to evaluate the biomechanical analysis of a simple intramedullary nail used for the fixation of subtypes of pertrochanteric fractures by FEM. The

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https://doi.org/10.1016/j.jos.2019.02.011
0949-2658/

effects of a fall and early weightbearing in osteoporotic and non-osteoporotic bones were also investigated.

2. Materials and methods

2.1. Finite element models

All finite element models were constructed using CATIA V5R18 and MSC PATRAN 2008 software. All analyses were performed using MSC MARC 2008 software packages in the Istanbul Technical University Biomechanics Laboratory.

The PFN was modelled on the Trigen Trochanteric Nail (Smith&Nephew, Fort Worth, TX, USA). The nail was 150 mm long and 11.5 mm in distal diameters. The lag screws were 6.4 mm in diameter and 95 mm (distal) and 90 mm (proximal) long. The locking screws were 5.0 mm in diameter and 40 mm long. The fourth generation standardized femoral model was used in the analyses. This was modeled by Rizzoli Orthopedic Institute, which is freely available over the internet for academics [9]. The material properties of porotic and non-porotic bones, cortical and spongious bones, and titanium were determined according to the literature [10,11] (Table 1). In order to make the analyses, 31A1.1, 31A2.1, 31A3.1, 31A3.2, 31A3.3 fracture types were modelled according to the AO/OTA classification system [12] (Fig. 1). Because of the difficulties in analyses, comminuted fracture types (31A2.2 and 31A2.3) were not modelled. Two non-porotic models without fracture, (one with a nail and one without) and an osteoporotic 31A2 type fracture model were also used in the analyses. It was accepted that fractures were ideally reduced with 1 mm space between the fragments. The nail filled the whole medullary canal distally. Two lag screws and two locking screws were used for fixation.

2.2. Loading conditions

Loads that occurred due to the weight of a 60 kg man were applied to all models in the single- and two-leg stance positions according to the biomechanical principles [10]. During loading, femur models were fixed distally from the medial and lateral condyles to mimic standing positions. The application point of joint reaction force (JRF) was accepted as the superomedial portion of the femoral head. The abductor muscle force was applied to the greater trochanter at the point of the gluteus medius and minimus muscles' attachment (Fig. 2). In the double-leg stance position, the half weight of the body (W)—other than the two lower extremities—was applied as JRF (~2/6 W = 196 N). In the single-leg stance position, JRF (~3,4 W = 1999,2 N) and abductor muscle force (~2.6 W = 1558.8 N) were applied. The JRF was applied on the femoral head at an angle of 15.6° with the frontal plane, and a 6° angle with the sagittal plane. The abductor muscle force was applied at a 24° angle with the frontal plane, and a 15° angle with the sagittal plane [8,10].

As a simulation of fall, a 500 N force was applied to the greater trochanter in order to evaluate the effects of rotational moments. The experiment was performed on a 31A2.1-type unstable fracture model and a model without fracture but with a nail (mimicking a healed fracture). The femur was fixed distally and force was applied at the level of the greater trochanter. The other forces were applied as in the single-leg stance position. The results of all experiments were evaluated according to the yield strength values of bone (128 MPa) and titanium (830 MPa) [10,11,13,14]. Since this was a computer model study and the applied forces and application points were fixed, the results would be the same for repetitive experiments. Therefore, only one experiment was performed for each model as in the literature [6–8].

The study had local ethical committee approval.

Table 1
The elasticity moduli of the materials. The Poisson ratio was assumed to be constant (n = 0.3) for all materials.

Material	Elasticity modulus (E) GPa
Titanium	110
Healthy cortical bone	14.2
Healthy spongious bone	1
Osteoporotic cortical bone	11
Osteoporotic spongious bone	0.1

3. Results

Following the application of forces, stress accumulation was determined at different points but especially in the calcar femorale in all models. However, the magnitude of stress values in the femoral head were relatively low. In the fracture models, stress accumulation especially occurred around the fracture sites. Other sites where higher amounts of stress had been measured were the holes for the lag screws, the hole for a proximal locking screw on the nail, the threadless parts of the lag screws, and the upper portion of the nail (Fig. 3). In unstable fractures (31A2 and 31A3 types), the amount of stress around the fracture site was higher compared to the stable fracture (31A1 type). In addition, due to the forces acting on the models, compressive stress accumulation on the medial femoral cortex and tensile stress accumulation on the lateral femoral cortex were determined. In the single-leg stance position, both the JRF and the abductor muscle forces were acting, and stress accumulation was higher than in the two-leg stance position in which only the joint reaction force acts. In all 31A3 type fracture models, displacement was observed in the fracture site in the single-leg stance position. The stress values can be found in Table 2.

3.1. The models without fracture

Two healthy femur models were studied: (i) a model without a fracture; and (ii) a model without a fracture but with a nail. The second model was assumed to represent a completely healed fracture. In both models, the stress accumulation was higher close to the calcar femorale and the diaphyseal area. In the single-leg stance position the maximum value for stress accumulation was determined in the distal lateral cortex in both models as 84.3 MPa. In the double-leg stance position it was 2.18 MPa. These values were quite a bit lower than the yield stress value of the bone. When we checked the model with the nail, the highest stress value on the nail was observed on the proximal part that intersected with the locking screws, especially the proximal screw (51.7 MPa).

3.2. Stable fracture: 31A1.1 type

The stress values in the fracture site were between 2.7 and 29.9 MPa for the single-leg stance position and between 0.27 and 2.72 MPa for the double-leg stance position. In the single-leg stance position, the highest amount of stress accumulation was measured on the lateral cortex distal to the nail as 84.3 MPa. The stress accumulation on the nail was especially observed on the threadless portions of the lag screws (with a maximum value of 130 MPa). No displacement was observed at the fracture site.

3.3. Unstable fracture: 31A2.1 type

The stress values at the fracture site were between 2.7 and 32.6 MPa for the single-leg stance position and between 0.27 and

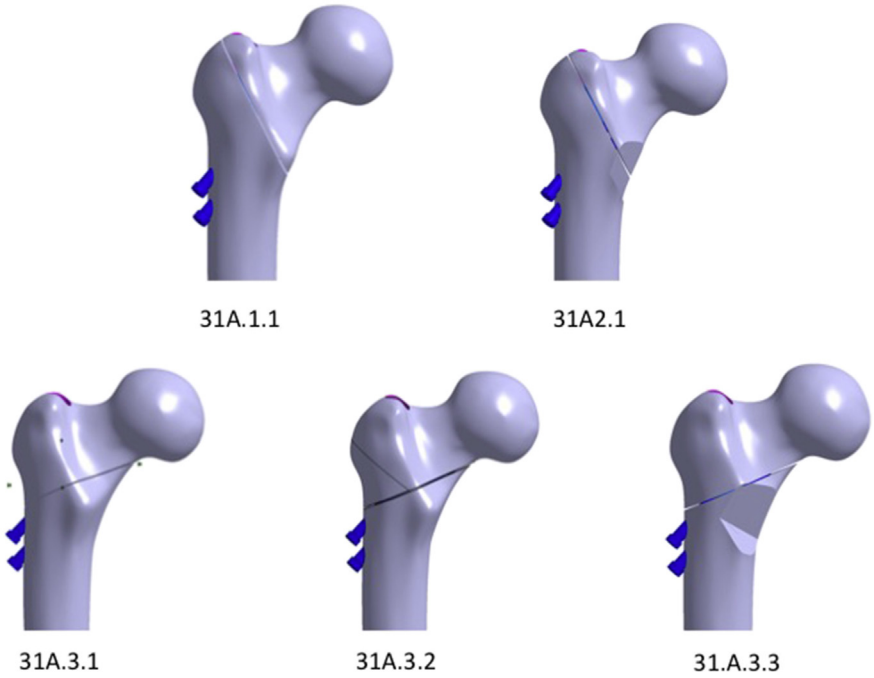

Fig. 1. Fracture models according to AO/OTA classification system.

3.81 MPa for the double-leg stance position. In the single-leg stance position, the highest stress values were measured as 84.3 MPa on the lateral femoral cortex distal to the nail, and 130 MPa on the proximal screws. No displacement was observed at the fracture site.

3.4. Unstable fractures: 31A3.1, 31A3.2, and 31A3.3 type

These types of fractures are called “reverse oblique fractures” and are accepted as highly unstable [12]. The localization of stress accumulation was similar in these three types of fractures, but the stress values increased from subtypes 1 to type 3. The maximum amount of stress measured in fracture sites for subtypes 1, 2, and 3 were 4.6, 4.9, and 5.1 MPa for the double-leg stance position, and 35.4, 89.8, and 106 MPa for the single-leg stance position, respectively. Similar to the other models, the threadless parts of the lag

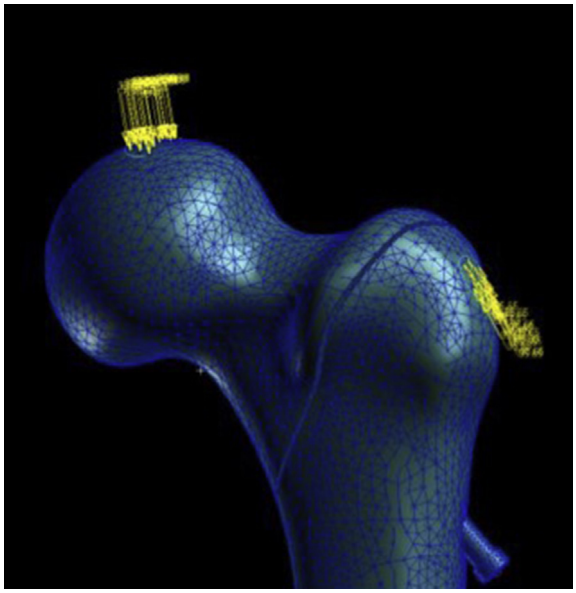

Fig. 2. Joint reaction force was applied superomedial portion of femoral head. The abductor muscle force was applied to the greater trochanter at the point of gluteus medius and minimus muscles' attachment.

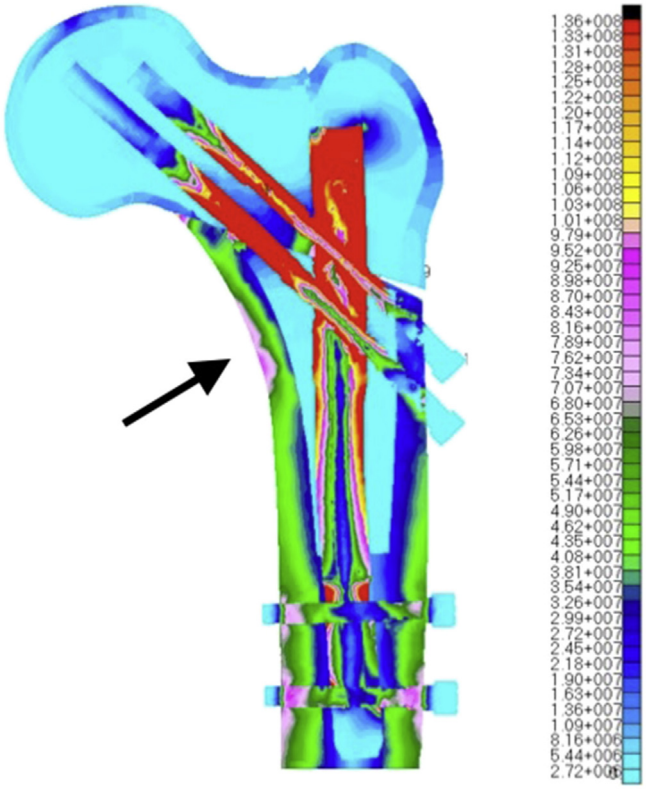

Fig. 3. Higher stress values were measured at calcar femoral (black arrow), threadless parts of the lag screws, proximal portion of the nail and contact points of the distal screws and nail.

Table 2
Maximum stress values measured on the specific regions of the bones and nails.

Model	Standing position	Proximal femur (MPa)	Fracture site (MPa)	Subtrochanteric region (MPa)	Distal femur (MPa)	Body of the nail (MPa)	Proximal screws (MPa)	Distal locking screws (MPa)
Intact femur	Single leg	13.6	–	32.6	84.3	–	–	–
	Double leg	2.18	–	2.72	2.18	–	–	–
Healed femur	Single leg	5.44	–	24.5	84.3	51.7	16.3	38.1
	Double leg	1.9	–	2.18	2.18	4.62	2.99	4.08
31A-1.1	Single leg	5.44	29.9	29.9	84.3	128	130	54
	Double leg	1.9	2.72	1.9	2.18	40.8	73.4	5.4
31A-2.1	Single leg	5.44	32.6	32.6	84.3	128	130	59.8
	Double leg	1.9	3.81	3.26	2.18	43.5	103	5.4
31A-3.1	Single leg	5.44	35.4	32.6	84.3	133	131	62.6
	Double leg	1.9	4.6	3.54	2.18	49	32.6	5.4
31A3.2	Single leg	13.6	89.8	78.9	84.3	131	131	87
	Double leg	2.18	4.9	4.35	2.18	54.4	54.4	5.44
31A-3.3	Single leg	16.3	106	106	84.3	133	133	92.5
	Double leg	2.45	5.1	4.62	2.18	57	54.4	10.0
Osteoporotic fracture	Single leg	5.44	35.4	38	84.3	128	133	62.6
Simulation of fall	Fracture	8.16	38.1	29.9	112	130	136	59.8
	Intact femur	5.44	–	19	112	49	24.5	32.6

screws and the hole for the proximal locking screw were the areas where more stress accumulated on the nail. In addition, the proximal segment of the nail was exposed to a high level of stress in these three models. We observed 3 mm displacement in the fracture site during the single-leg stance (Fig. 4).

3.5. Simulation of fall

A 500 N force was applied perpendicular to a femur model without fracture and to a model with a 31A2.1 type fracture. In the fracture model, we measured stress accumulation on the medial (up to 95.2 MPa) and the lateral (up to 112 MPa) surfaces of the femur distal to the nail (Fig. 5). These values were identical in the non-fractured model. In the nail, the highest stress values were measured on the threadless parts of the proximal lag screws as 136 MPa in the fracture model. This value was measured as 24.5 MPa in the non-fractured model. We observed 1.5 mm displacement at the fracture site (Fig. 6).

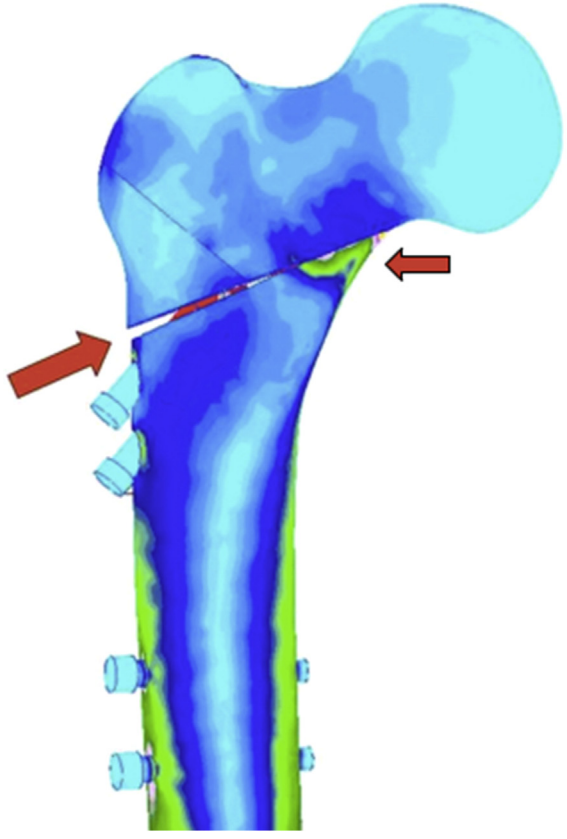

Fig. 4. Single leg standing position caused displacement at the fracture site.

3.6. Osteoporotic fracture

In order to investigate the effect of osteoporosis, we modeled an osteoporotic femur with a type 31A2 fracture. The elasticity modulus of the cortical and spongious bones were decreased to mimic osteoporosis. We applied forces as in the single-leg stance position and compared the results with the same type of non-osteoporotic fracture model. In the stress analysis, we observed similar results in the femur model; however, in the osteoporotic model, a higher level of stress was measured on the proximal screws (133 MPa vs. 130 MPa).

4. Discussion

The FEM is widely used in biomechanical investigations. The simulation of conditions with various factors and repeatability are the advantages of this method [6–8,15]. Performing such an experiment with real patients, and investigating the effects of weight bearing on femoral fractures in the early postoperative period would have ethical problems. Therefore, we think that FEM is the ideal way to research this issue. In this study, we generated different types of intertrochanteric femur fractures and fixed them with a simple intramedullary nail. During analysis we investigated the effects of body weight and abductor muscles while standing on a single leg and on double legs. There are several studies related to the finite element analysis of intertrochanteric fractures; in these studies, the authors generally studied single types of fractures and simplified models [7,8,16–18]. To our knowledge, there have been no prior studies that investigated and compared the subtypes of trochanteric fractures in terms of AO classification together.

We studied the effects of early weightbearing after intramedullary fixation of trochanteric fractures. There is no consensus about postoperative rehabilitation protocol after these procedures. Most surgeons prefer early weightbearing as tolerated by the

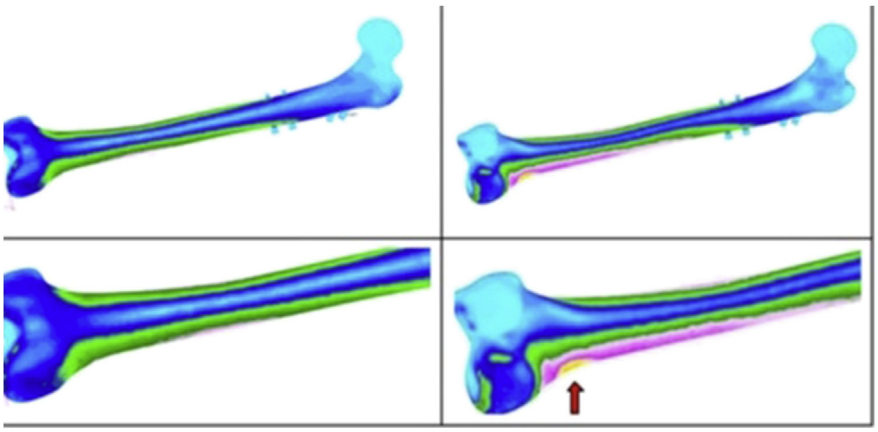

Fig. 5. Rotational forces increased stress values at the femur distal to the nail.

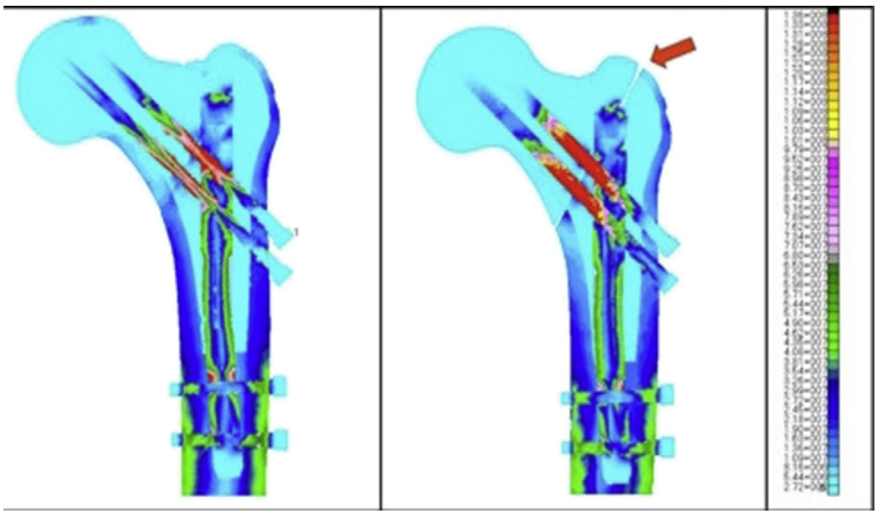

Fig. 6. Displacement was observed at the fracture site at simulation model of the fall.

patient, but some prefer to wait for radiologic proof of fracture union [19,20]. In our daily practice, we prefer to allow early weightbearing with crutches as tolerated by the patient. In this study, we investigated the effect of this protocol. Our results support this, except in obese patients and those with reverse oblique fractures. Makki et al. [21] reported a failure rate of up to 22% with proximal femoral nailing in patients with AO 31A3 type fractures. The authors claimed that the failures were not related to fracture malreduction or screw positions, but to the severity of the fractures. Our experiment is consistent with this argument.

In all models, the amount of stress on the bone was lower than on the intramedullary nail. This was obvious, especially in intact bones and bones with stable fractures. When the forces were applied to the intact femur, the level of stress measured on the calcar femorale was higher than on other parts of the femur. This condition was more obvious in the fracture models and supports the importance of this area in terms of load carrying function (Fig. 3). Regardless of the position, whether standing on one or two legs, the magnitude of stress measured on neither bone nor nail was greater than the fracture limits.

The stress distribution was not equal in all parts of the nail. The proximal part of the nail, the threadless portions of the lag screws, and the contact points of the distal screws and screw holes sustained more stress than other parts. Seral et al. [8] found similar results with different types of nails and reported high levels of stress accumulation on the proximal lag screws. The level of stress on the threadless part of the lag screws measured up to 136 MPa depending on the fracture type in our study. The yield strength of titanium is much higher than those values. These findings prompted us to consider that weightbearing for a short period does not cause implant failure, but we did not study a continuous load application; therefore, we could not comment on the fatigue of the materials. Strengthening the parts of the nail that sustain higher stress levels may allow early weightbearing.

In stable and 31A2 type unstable fractures, we did not detect a loss of reduction both in single- and double-leg stance positions. In 31A3 type fractures, distraction was observed on the lateral cortices during the single-leg stance. This causes varus angulation of the proximal femur and malunion during follow-up. Therefore, early full weightbearing is not suitable for these fracture types.

In order to simulate a fall, we applied a 500 N force perpendicular to the axis of a femur model with a 31A2 type fracture and a femur without a fracture. The model without a fracture was designed with an intramedullary nail to mimic a healed fracture. We detected high stress values on the medial and lateral femoral cortices distal to the nail. These values were less in the model

without a fracture. The stress values on implants were less than the yield strength of the titanium. According to our results, a single fall may not cause the screws to break. Although measured stress values were less than the yield strength of the bone, in patients with higher body weights, rotational movements may cause periprosthetic fractures. Therefore, patients with a risk of falling should be advised to use support. We measured less stress values in a non-fractured femur, but this would also increase the fracture risk in overweight patients. These results made us consider implant removal in selected patients but patients should be warned and precautions should be taken for possible fractures through the remaining screw holes.

We compared osteoporotic bone with non-osteoporotic bone. The models had the same type of fractures. The elasticity modulus was changed during the modelling of osteoporosis. Some other studies also used this technique in order to generate osteoporosis [22]. In the osteoporotic fracture model, the amount of stress was higher than in the non-osteoporotic femur model with the same type of fracture. This may increase the risk of failure in patients with low bone quality. Weightbearing should be allowed with support in order to prevent complications.

All the fractures were modeled as ideally reduced, and the implants were positioned perfectly. This concept is compatible with the literature [7,8,23]. The effects of non-anatomic reduction and inadequate implant positioning may change the results. We also investigated the early post-operative period and did not model callus tissues. The addition of callus tissue may prevent displacement and change the stress distribution, but this needs to be investigated in a different study.

In FEM models, analysis of the comminuted fractures is difficult. Unfixed fragments of the model would cause confusion. Therefore we did not model fracture types of 31A2.2 and 31A2.3. However when we check the figures, it can be seen that, except the reverse oblique fractures, weightbearing causes stress accumulation on the calcar femorale and does not cause too much displacement. So we can assume that in case of ideal reduction weightbearing may not cause problems but if anatomic reduction can not be achieved this would cause excessive stress on nail and failure. We can not make exact comments about these fracture types. Studies can be performed with advanced softwares.

The main limitation of this study was the simplification of the models. Some soft tissue components, other than abductor muscles, did not contribute to the models. However, the addition of all other muscles and ligaments to the model would make the analysis process inextricable for the computer. Our aim was to investigate the effects of standing positions on the model. First, we applied forces according to the biomechanical principles and took only bone, abductor muscles, and the effect of body weight into account. Second, we did not have any data to comment on fatigue. Third, we did not apply cyclic forces simulating daily life. The addition of such a complex analysis would give more reliable results. Because of these limitations, our model cannot be accepted as a match of real-world situations, and an in vivo investigation of such a study would not be ethical; therefore, FEM models are good options for real-world simulations. In addition, the AO/OTA classification of proximal femur fractures had been revised in 2018 [24]. Since we designed this study and modeled the fracture types before revision of the classification, this study consists only fractures in the former classification.

5. Conclusion

Full weightbearing after intramedullary fixation of trochanteric fractures can be allowed except in obese patients and patients with AO/OTA 31A3 type fractures. The use of support is recommended in order to prevent complications. Implant removal can be discussed with patients after the fracture union in order to prevent possible periprosthetic fractures.

Conflict of interest

All authors in this study confirm that there is no conflict of interest.

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