Biomechanical analysis of an FNS® fixation construct for femur neck fractures and clinical implications: a finite element method

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Abstract

Background

Despite widely use of femoral neck system system (FNS®), there is little evidence for mechanical property according to type of femoral neck fracture. This study is to assess the structural/mechanical stability of fixation constructs with a femur neck system by using the finite-element (FE) analysis after simulating the femur neck fractures and to introduce the clinical implications.

Methods

We simulated the fracture models of subcapital, transcervical, basicervical, and vertical types by using the right femur model (SAWBONES®) and imported the implant model of FNS® to ANSYS® to place the implant in the optimal position. The distal end of the femur model was completely fixed and was abducted 7°. The force vector was set laterally at an angle of 3° and posteriorly 15° in the vertical ground. We did the analysis using Ansys® software with the von Mises stress (VMS) in megapascal (MPa).

Results

The max VMS of the fracture site was 67.01 MPa for a subcapital fracture, 68.56 MPa for a transcervical fracture, 344.54 MPa for a basicervical fracture, and 130.59 MPa for a vertical fracture. The max VMS of FNS® was 840.34 MPa for a subcapital fracture, 637.37 MPa for a transcervical fracture, 464.07 MPa for a basicervical fracture, and 421.01 MPa for a vertical fracture. The max VMS of the implant corresponded to the value of the entire fixation construct and thus, FNS® mainly functions as a load-bearing implant. When we compared the basicervical and vertical fractures, the stress distribution between the implant and the fracture sites differed significantly, and the basicervical fracture had higher VMS in the bone, implant, and fracture sites.

Conclusion

Considering the stress distribution of the assembly model, FNS® fixation should consider the osseous anchorage between the proximal bolt and cancellous bone of femoral head and might be appropriate for vertical fractures. Regarding the VMS of fracture site, FNS® might be applied cautiously just in the basicervical fracture of anatomical reduction without gap and comminution.

Background

Considering that fracture site and orientation affects management modality and fixation construct in the treatment of young femoral neck fractures, trauma surgeons need convenient and reproducible standards
to help them choose the best surgical implant and predict fracture-related complications [1, 2]. Of femur neck fractures in young adults, vertically oriented femur neck fracture (Pauwels type III) should be distinguished, because the high shearing force could explain the relatively high rate of nonunion and fixation failure [3–7]. Anatomic reduction and choice of optimal implant are crucial for minimizing complications of femur neck fracture in young adults. Although the reduction adequacy was dependent on the surgeon’s experience and tactics, the implant choice is based on preoperative planning with an accurate assessment of fracture morphology, especially in high-energy injuries.

Although various treatment options exist for femoral neck fractures, traditionally, these fractures are often stabilized with multiple cannulated screws (MCS), dynamic hip screws (DHS) with or without an antirotation screw. Recently, the new minimally invasive implant femoral neck system (FNS®; Synthes GmbH, Oberdorf, Switzerland) developed for dynamic fixation of femoral neck fractures. Owing to the advantages of angular stability with a minimally invasive surgical technique [8], the indications for FNS® have been significantly broadened and have led to an increase in the use for various femur neck fractures, although there is little evidence for clinical outcomes.

One of the computational methods that has received wide acceptance in orthopedic research is the Finite Element (FE) Analysis [9–11], which is the preferred method for numerical problems occurring in fracture fixation by dealing with the stress and strain analysis of bone and load-bearing implants as the structure-mechanical aspects. Hence, we did this study to find out the structural-mechanical stability of FNS® by using the FE method after simulating various femur neck fractures and to introduce the clinical implications.

**Methods**

**Development of the FE model**

This study did not need approval by the Institutional Review Board, because its three-dimensional (3D) computer-aided design (CAD) model was from the commercially available high-resolution file of a right femur model: the standard fourth-generation composite bones (SAWBONES®, Vashon, WA, USA). Given the commercially available FNS®, we modeled the 3D implant at actual size by using the 3D CAD software of SolidWorks 2019® (Dassault Systems SolidWorks Co, MA, USA). Both the 3D femur and FNS® were imported to SolidWorks® for further polishing and were meshed using 1.0-mm tetrahedral mesh (Table 1).
Table 1

<table>
<thead>
<tr>
<th>materials</th>
<th>Density (g/cm³)</th>
<th>Elastic modulus (MPa)</th>
<th>Poisson’s ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cortical bone</td>
<td>1.5</td>
<td>7200</td>
<td>0.35</td>
</tr>
<tr>
<td>Cancellous bone</td>
<td>0.2</td>
<td>135</td>
<td>0.225</td>
</tr>
<tr>
<td>Titanium alloy (fixation implant)</td>
<td>4.62</td>
<td>96000</td>
<td>0.36</td>
</tr>
</tbody>
</table>

The geometry of FE models corresponded to the definition of femur neck fractures, including the subcapital, transcervical, basicervical [12], and vertical fractures [7]. The neck fractures were simulated in 3D CAD software of SolidWorks®. Then, the 3D models of implant and femur were imported to the ANSYS® software (Ansys 19.0, Ansys, Inc., Canonsburg, PA, USA) for placing the FNS® in the optimal position and subsequently establishing the FE model by remeshing (Fig. 1). For FE analysis, the principles of model construction were uniform, as follows: (1) The plate with one hole made the contact with the femoral diaphysis. (2) The trajectory of the screws was chosen based on the locking hole of the plate so that they protruded over 2 mm on the opposite side. (3) The contact between plate and screw was simulated as the bonding with virtual mechanical rigid links to mimic the locking head screw (LHS) mechanism. (4) The bolt (screw) was inserted through the femoral head center or center-inferior at less than 10 mm in any direction from the outer boundary of the femoral head in concordance with the well-accepted technique of the manufacturer's instructions [13].

**Material properties, boundary conditions, and stress analysis of fixation constructs**

The material properties for the synthetic femur were assigned according to the manufacturer's specification for the fourth-generation Sawbones (Table 1). We set the Young's modulus of the cortical bone at 7,200 MPa with a Poisson’s ratio (γ) of 0.350, and set it for the cancellous bone at 135 MPa with Poisson ratio (γ) of 0.225. The density of the cortical bone was 1.5g/cm³, and that of the cancellous bone was 0.2g/cm [14, 15]. All the metal of the implants was assumed to have the elastic, isotropic and homogeneous properties of titanium alloy in this study. The Young's modulus of the titanium alloy was set at 96,000 MPa with a Poisson's ratio (γ) of 0.36 and the density of the implant was 4.62g/cm [14].

The distal end of the femur model was completely fixed, and the loads of 1950 N, equivalent to tripling the body weight of the subject (65 kg), were applied to the center of the femoral head [16]. To mimic the normally physiologic alignment of lower limbs in the standing position, each assembly model was abducted 7° in the vertical ground (Fig. 1). The force vector was set laterally at an angle of 3° and posteriorly at 15°, because the femoral neck was slightly anteverted in relation to the position of the femoral condyles in the horizontal or transverse plane [17]. The 3D shear stress on the X axis was 98.57
N, 1947.3 N on the Y axis, and 26.4 N on the Z axis. We assumed that the implant was in direct contact with the bone (Table 1). According to the well-established and approved test contact setup method described in previous studies, a binding contact was formed between the internal fixation screw and the femur (Table 2) [18]. We could not evaluate the torsional results in these models. We assumed that the implant had a direct contact with the bone and did the analysis using commercial finite-element software of Ansys® with von Mises stress (VMS) in megapascal (MPa), fracture displacement of the implant relative to the bone (as a measure of relative fixation strength).

<table>
<thead>
<tr>
<th>Contact pair</th>
<th>Contact type</th>
<th>Friction coefficient</th>
<th>Elements</th>
</tr>
</thead>
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<tr>
<td>Cortical-cancellous bone in all</td>
<td>Bonded</td>
<td>4824293</td>
<td>3363112</td>
</tr>
<tr>
<td>Fracture surface in all</td>
<td>No separation</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Cortical bone-FNS¹ in all</td>
<td>No separation</td>
<td>4931251</td>
<td>3434440</td>
</tr>
<tr>
<td>Cancellous bone-FNS in all</td>
<td>No separation</td>
<td>4927640</td>
<td>3415588</td>
</tr>
<tr>
<td>Screw-plate hole in all</td>
<td>Bonded</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Plate-bolt &amp; antirotation screw in FNS</td>
<td>Bonded</td>
<td>6794680</td>
<td>4704320</td>
</tr>
<tr>
<td>Lag screw-plate with barrel in DHS²</td>
<td>No separation</td>
<td>4923734</td>
<td>3425442</td>
</tr>
</tbody>
</table>

### Results

According to the displacement of the assembly model, the maximum displacement occurs at the upper part of the femoral head, as shown in Fig. 2. The displacements of the proximal femur were 9.28 mm for the no-fracture model, 9.61 mm for the subcapital fracture, 9.77 mm for the transcervical fracture, 9.86 mm for the basicervical fracture, and 9.87 mm for the vertical fracture. The VMS distributions on bone were assessed and are shown in Fig. 3. Compared with the no-fracture model, the subcapital and transcervical fracture had a similar distribution of VMS, which was the medial exit point of the screw through the plate (Fig. 4). The vertical fracture was the lateral insertion point of the plate. However, for the basicervical fracture, the max point of VMS was different from that in the other models and was located in the posteromedial side of the fracture site (Fig. 5).

The max VMS of the fracture site was 67.01 MPa for the subcapital fracture, 68.56 MPa for the transcervical fracture, 344.54 MPa for the basicervical fracture, and 130.59 MPa for the vertical fracture (Fig. 6). For the stress distribution on the FNS® implant, there were some differences based on the fracture morphologies. The max VMS of the implant was 840.34 MPa for the subcapital fracture, 637.37 MPa for the transcervical fracture, 464.07 MPa for the basicervical fracture, and 421.01 MPa for the vertical fracture (Fig. 6). For the stress distribution on the implant, the max points of VMS were the bolt
around fracture site in all models and were located in the junction site between the fracture site and the barrel of the plate (Fig. 7). There were two kinds of stress distribution of the bolt according to the fracture morphologies. The max point of the subcapital and transcervical fractures was the upper junction site, like that in the no-fracture model, and was the lower junction site for the basicervical and vertical fractures (Fig. 8).

Considering the max VMS distributions on the assembly models, the max VMS of the implant corresponded to the value of the entire fixation construct; so the FNS® mainly served for load bearing, because the stress value of the fracture site was small except for the basicervical fracture. In terms of the load-bearing role, the implant’s VMS was the highest in the subcapital fracture and lowest in the vertical fracture. For the basicervical and vertical fractures, the stress distribution between the implant and fracture sites differed significantly; the basicervical fracture had higher VMS in the bone, implant, and fracture sites (Fig. 9).

**Discussion**

Although controversy remains regarding optimal fixation techniques and constructs, strategies for achieving optimal stability are crucial to minimizing the complications and sequelae in the management of high-energy femur neck fractures. We conducted the FE analysis to assess the structural-mechanical stability of FNS® in the non-osteoporotic femur neck fractures. This computational analysis enabled us to arrive at several interesting findings:

(1) The max VMS of FNS® corresponded to the value of the entire fixation construct and mainly functioned as the load-bearing implant. (2) For the subcapital and transcervical fractures, the stress distribution mainly concentrated on the implant; so, the proximal osseous anchoring of the bolt might be important for maintaining the rotational and angular stability. (3) For the stress distribution on the fracture site, the max VMS point was located in the posteroinferior side of the fracture site, just as in the basicervical and vertical fractures. The VMS of the basicervical fracture was significantly larger than that of vertical fracture. (4) Compared with the no-fracture model, the VMS distribution of the vertical fracture was most similar in the fracture site and implant.

Although various implants exist for the operative fixation of femoral neck fractures, the use of FNS® has been increased because of its biomechanical advantages with minimally invasiveness [8]. Thus, the surgical complications, including the cut-out, nonunion, and femoral head necrosis, inevitably occurred [19]. To our best knowledge, there was no clinical report of FNS® complications according to the fracture morphologies or Pauwels angle, although several studies have reported the comparative results of FNS® and MCS [19–21]. Thus, considering these limitations of clinical case studies, we aimed to investigate the biomechanical behaviors of FNS® based on the traditional classification of femur neck fractures. For the VMS distribution of fracture sites, the vertical fracture was similar to the no-fracture model but was much different from the basicervical fracture. The stress distribution of the fracture site was notably increased in the basicervical fracture. Thus, if the FNS® fixation is considered for a basicervical fracture.
the related factors of the variant types [22–24] and reduction adequacy, which are anatomically cortical contact, posteroinferior comminution, and fracture gap should be verified intraoperatively before the definitive fixation.

Because the vertically oriented fracture may contribute to the high failure rate in Pauwels’ grade III fractures in healthy young patient, the DHS with or without a derotational screw has been regarded as a superior fixation construct [4, 25]. Before this investigation to identify the optimal indications, we anticipated that FNS® may not be appropriate for vertical fractures, because it is a smaller implant of plate and lag screw (bolt) than is DHS. Although we did not analyze the direct comparison between FNS® and DHS, our results demonstrated that the stress value of the fixation construct was concentrated on the implant and not the fracture site. The max VMS value of the implant was not significantly different but was much different in the fracture site. Thus, despite these prejudices, we think the vertical fracture might be more suitable for FNS® fixation than is the basicervical fracture. Furthermore, considering that 96% of vertical neck fractures had major comminution, which was mostly located inferiorly and posteriorly [26], the FNS® for vertical fractures might be a superior implant based on the results of this study.

For the subcapital fracture, the stress distribution mainly concentrated on the implant, and the max points of VMS were the bolt around the fracture site in the junction between the fracture site and the barrel of the plate. In order to arrive at this result, one should assume that the anchoring between the proximal bolt and the cancellous bone of the femoral head is theoretically maximized. In the personal communication between orthopedic trauma surgeons, we found that fixation failure of FNS® was not uncommon, although the critical factors could not be analyzed. However, this FE analysis seems to show that the proximal osseous anchoring of the bolt might be essential for maintaining the rotational and angular stability. The subcapital fracture might be cautiously applied, because there was a short working length in the femoral head fragment. Adding an anti-rotational screw to the FNS® might increase the proximal anchoring and angular stability; so further research on this topic will be needed in the future (Fig. 10).

Despite interesting findings, this computational simulation study has several fundamental limitations. First, our fracture models were very simplified for simulating the perfect reduction without gap and comminution between fragments. Second, our results had descriptive characteristics, because our models were simulated from a normal non-osteoporotic femur without considering the bone quality. Third, the fracture impact by controlled sliding of the lag screw/blade could not be simulated, because of technical difficulties. Our results just showed the initial strength of the fixation construct. Nevertheless, our computational analysis could be assessed on structural-mechanical strength and the VMS distribution of the fracture site and the implant under the same conditions. Although the implant should be chosen in terms of the extent of displacement, fracture configuration, physiological age, bone quality, and other factors, our results might be able to directly suggest technical relevance to maximize the structural strength of the FNS® fixation construct for femur neck fractures.

Conclusions
Considering the stress distribution of fracture sites and implants, the FNS® fixation construct might be appropriate for transcervical and vertical fractures. For a basicervical fracture, an FNS® implant might be applied in the anatomically reduced fracture without gap and comminution. When treating the subcapital fracture, there are two factors to consider: (1) maximizing the osseous anchorage between the proximal bolt and cancellous bone of the femoral head. (2) verifying preoperatively the working length of the bolt in the femoral head fragment.

Abbreviations

CAD  computer-aided design
FNS  femur neck system
FE   finite-element
MPa  megapascal
VMS  von Mises stress

Declarations

Acknowledgements

This work was not supported by research fund.

Author contributions

H.S.S., D.H.K., and S.W.K. planned the experiments. S.L.J. and G.H. J. performed processing of the model analysis. G.H.J. wrote original draft preparation. H.S.S., D.H.K., and S.W.K. reviewed and edited the writing. G.H.J., D.H.K. and H.S.S. performed the data analysis. All authors have read and agreed to the published version of the manuscript.

Competing interests

The authors declare no competing interests.

Availability of data and materials

The datasets analyzed during the current study are available from the corresponding author on reasonable request.

References


Figures

Figure 1

The neck fractures were simulated in 3D CAD software of SolidWorks 2019®. Then, the 3D models of the implant and femur were imported to ANSYS® software (Ansys 19.0, Ansys Inc., Canonsburg, PA, USA) for placing the FNS® in the optimal position.
Figure 2

According to the displacement of the assembly model, the maximum displacement occurs at the upper part of the femoral head. The displacement was the largest in the basicervical and vertical fractures.

Von mises stress of bone

Figure 3
The VMS distributions on bone. Compared with the no-fracture model, the subcapital and transcervical fractures had a similar distribution of VMS, which was the medial exit point of the screw through the plate. However, for the basicervical and vertical fractures, the max point of VMS was different from that in other models and was located in the medial side of the fracture site.

**Figure 4**

Compared with the no-fracture model, the subcapital and transcervical fractures had the similar distribution of VMS, which was the medial exit point of the screw through the plate.
Figure 5

The stress distribution of the fracture site was notably increased in the basicervical fracture, for which the max point of VMS was different from that in other models and was located in the posteroinferior area of the fracture site
The max VMS of the fracture site was 67.01 MPa for the subcapital fracture, 68.56 MPa for the transcervical fracture, 344.54 MPa for the basicervical fracture, and 130.59 MPa for the vertical fracture.
For the stress distribution on the implant, the max points of VMS were the bolt around the fracture site in all models; it was located in the junction between the fracture site and the barrel of the plate.
There were two kinds of stress distribution of the bolt according to the fracture morphologies. The max point of the subcapital and transcervical fractures was the upper junction site, as in the no-fracture model, and was the lower junction site for the basicervical and vertical fractures.
In terms of the load-bearing role, the implant’s VMS was the highest in the subcapital fracture and lowest in the vertical fracture. Comparing the basicervical and vertical fractures, the stress distribution between the implant and fracture sites differed significantly, and the basicervical fracture had higher VMS in the bone, implant, and fracture sites.
A 54-year-old man sustained a femur neck fracture caused by falls from a 2-m height. (A, B) Plain radiographs and the intraoperative fluoroscopic image showed the subcapital fracture. (C) The FNS c anti-rotational screw was applied to increase the anchoring and stability of the femoral head fragment.