Structure-Mechanical Analysis of Various Fixation Constructs for Basicervical Fractures of the Proximal Femur and clinical implications: Finite Element Analysis

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Article

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Abstract

Objective: This present study was conducted to determine the structural-mechanical stability of various fixation constructs through finite element (FE) analysis following simulation of a basicervical fracture and to introduce the clinical implications.

Materials and Methods: We simulated fracture models by using a right synthetic femur (SAWBONES®). We imported the implant models into ANSYS® for placement in an optimal position. Five assembly models were constructed: (1) multiple cancellous screws (MCS), (2) FNS (femoral neck system®), (3) dynamic hip screw (DHS), (4) DHS with anti-rotation 7.0 screw (DHS + screw), and PFNA-II (Proximal Femoral Nail Antirotation-II®). The femur model's distal end was completely fixed and 7° abducted. We set the force vector at a 3° angle laterally and 15° posteriorly from the vertical ground. Analysis was done using Ansys® software with von Mises stress (VMS) in megapascals (MPa) and displacement (mm).

Results: The displacements of the proximal femur were 10.25 mm for MCS, 9.66 mm for DHS, 9.44 mm for DHS + screw, 9.86 mm for FNS, and 9.31 mm for PFNA-II. The maximum implant VMS was 148.94 MPa for MCS, 414.66 MPa for DHS, 385.59 MPa for DSH + screw, 464.07 MPa for FNS, and 505.07 MPa for PFNA-II. The maximum VMS at the fracture site was 621.13 MPa for MCS, 464.14 MPa for DHS, 64.51 MPa for DHS + screw, 344.54 MPa for FNS, and 647.49 MPa for PFNA-II. The maximum VMS at the fracture site was in the superior area with the high point around the posterior screw in the MCS, anterosuperior corner in the DHS, the posteroinferior site of the FNS, and posterosuperior site around the entry point in the PFNA-II. In the DHS + screw, the stresses were distributed evenly and disappeared at the maximum VMS fracture site.

Conclusion: Based on the fracture site and implant's stress distribution, the model receiving the optimal load was a DHS + screw construct, and the FNS implant could be applied to anatomically reduced fractures without comminution. Considering the high-stress concentration around the entry point, a PFNA-II fixation has a high probability of head-neck fragment rotational instability.

Introduction

It is crucial to determine the optimal treatment methods and to minimize complications of hip fractures in orthopedic trauma, especially in the geriatric field. Basicervical fractures are unique among hip fractures occurring at the base of the neck of the proximal femur. Biomechanical studies¹⁻³ suggest that these fractures may best be treated similarly to intertrochanteric fractures, with a DHS rather than with multiple cancellous screws (MCS). Although the surgical results are said to be similar to those of intertrochanteric fractures, the surgical fixation of basicervical fracture has been associated with high rates of failure because of their inherent propensity to collapse and to have rotational instability.⁴,⁵ Concerning the guide to choose the optimal implant for basicervical fracture, Dekhne et al.⁶ recently noted that basicervical fracture repairs with DHS and a cephalomedullary nail (CMN) produce similar failure and reoperation
rates, based on a systematic review. However, in recent decades, there has been considerable controversy about this being a highly unstable variant when managed with CMN fixation.\textsuperscript{4,7–9}

Finite element (FE) analysis is a computational method that has received wide acceptance in orthopedic research.\textsuperscript{10–12} FE analysis is the preferred method for solving numerical problems encountered in orthopedic biomechanics by treating the stress and strain analysis of bones and load-bearing implants as structure-mechanical issues. Compared to gross biomechanical investigation, research studies via computer simulation provide additional insight into mechanical stability under similar conditions. Additionally, this method allows for a 360° free rotation with magnification in any plane to analyze the stress distribution at the fracture or implant site. Therefore, we used the FE method to confirm the mechanical stability of various fixation constructs after simulating a basicervical fracture of the proximal femur and introduce the clinical implications.

**Materials And Methods**

**Development of the FE model**

We did not need Institutional Review Board approval since we used a commercially available computer-aided design (CAD) three-dimensional (3D) image file of a standard fourth-generation composite right femur bone model (SAWBONES®\textsuperscript{19}, Vashon, WA, USA). We modeled full-scale 3D implants for basicervical fracture fixation using 3D CAD software from SolidWorks 2019\textsuperscript{20} (Dassault Systems SolidWorks Co, MA, USA), commercially available CMN (PFNA-II\textsuperscript{21}, DePuy Synthes, Co, West Chest, PA, USA), dynamic hip screws (DHS), the Femur Neck System\textsuperscript{22} (FNS, DePuy Synthes), and 7.0-mm cannulated screws. We imported both the 3D femur and implants into SolidWorks Software\textsuperscript{23} for further polishing and meshed them using 1.0-mm tetrahedral mesh (Table 1).

<table>
<thead>
<tr>
<th>Assembly units</th>
<th>Mesh size</th>
<th>Node</th>
<th>Elements</th>
</tr>
</thead>
<tbody>
<tr>
<td>FNS\textsuperscript{1}</td>
<td>1mm</td>
<td>4824293</td>
<td>3363112</td>
</tr>
<tr>
<td>DHS\textsuperscript{2}</td>
<td>1mm</td>
<td>4931251</td>
<td>3434440</td>
</tr>
<tr>
<td>DHS + ARS\textsuperscript{3}</td>
<td>1mm</td>
<td>4927640</td>
<td>3415588</td>
</tr>
<tr>
<td>MCS\textsuperscript{4}</td>
<td>1mm</td>
<td>6794680</td>
<td>4704320</td>
</tr>
<tr>
<td>PFNA-II\textsuperscript{5}</td>
<td>1mm2</td>
<td>4923734</td>
<td>3425442</td>
</tr>
</tbody>
</table>

\textsuperscript{1}Femur Neck System\textsuperscript{®} (FNS, DePuy Synthes), \textsuperscript{2}Dynamic Hip Screw (DHS, DePuy Synthes), \textsuperscript{3}Dynamic Hips Screw with 7.0 mm cannulated screw, \textsuperscript{4}Multiple Screw Fixation with 7.0mm cannulated screw (MCS), \textsuperscript{5}Proximal Femoral Nail Antirotation II\textsuperscript{®} (PFNA-II, DePuy Synthes)
The geometry of FE models corresponded to the definition of basicervical fracture\textsuperscript{3,8,13}, and the fracture models were simulated in 3D CAD software of SolidWorks 2019\textsuperscript{®}. Then, the 3D models of implant and femur were imported to ANSYS\textsuperscript{®} software (Ansys 19.0, Ansys Inc., Canonsburg, PA, USA) for placing the implant in the optimal position and subsequently establishing the FE model by remeshing. We used the following principles of model construction for FE analysis: 1) Based on the fixed angle of locking hole, we determined the trajectory of the screws. 2) The contact between plate and screw was designed to simulate the locking head screw (LHS) mechanism. 3) For purchasing the opposite cortex, all screws were protruded over 2mm except lag screw. Five assembly models were constructed: (1) a multiple screw fixation (MCS) model, (2) an FNS\textsuperscript{®} fixation (FNS) model, (3) a dynamic hip screw fixation (DHS) model, (4) a DHS with 7.0 screw fixation (DHS + screw) model, and (5) a PFNA-II\textsuperscript{®} fixation (PFNA-II) model.

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

We assigned the material properties of the synthetic femur in accordance with the manufacturer’s specification for fourth generation Sawbones (Table 2). The Young's modulus of cortical bone was set to 7,200 MPa with a Poisson's ratio (\( \nu \)) of 0.350. The Young's modulus (135 MPa) and Poisson's ratio (\( \nu \)) were set to 0.225 for cancellous bone. The density of cortical bone was 1.5 g/cm\textsuperscript{3} and cancellous bone, 0.2 g/cm\textsuperscript{3}.\textsuperscript{14,15} This study assumed that all metals in the implants had homogeneous, isotropic, and elastic properties of titanium alloy. The Young's modulus of titanium alloy was set at 96,000 MPa with a Poisson's ratio (\( \nu \)) of 0.36 and an implant density of 4.62 g/cm.\textsuperscript{14} The distal end of femur model was completely fixed, and we applied a 1950 N load (i.e., triple subject's body weight of 65 kg), to the center of the femoral head.\textsuperscript{16} To mimic normal physiologic lower limb alignment in a standing position, each assembly model was abducted 7° from the vertical ground (Fig. 1). The force vectors were set at an angle of 3° laterally and 15° posteriorly because the femoral neck was slightly anteverted in relation to the position of the femoral condyles in the horizontal or transverse planes (Fig. 1). The 3D shear stress was 98.57 N on the X-axis, 1947.3 N on the Y-axis, and 26.4 N on the Z-axis. We could not evaluate torsional results in these models. We assumed that the implant was in direct contact with the bone (Table 3).

According to the well-established and approved test contact setup method described in previous studies, binding contact was formed between the internal fixation screw and the femur (Table 3).\textsuperscript{17} We did finite element analysis using commercially available Ansys\textsuperscript{®} software and VMS measured in megapascals and fracture displacement of the implant relative to the bone (as a measure of relative fixation strength).
### Table 2
Material properties of bone and implant

<table>
<thead>
<tr>
<th>Materials</th>
<th>Density ((g/cm^3))</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>

### Table 3
Contact Details of assembly unit

<table>
<thead>
<tr>
<th>Contact pair</th>
<th>Contact type</th>
<th>Friction coefficient</th>
<th>Elements</th>
</tr>
</thead>
<tbody>
<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></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Cortical bone-FNS(^1) in all</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Cancellous bone-FNS in all</td>
<td></td>
<td></td>
<td></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 DHS(^2)</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

\(^1\)Femur Neck System® (FNS, DePuy Synthes), \(^2\)Dynamic Hip Screw (DHS, DePuy Synthes)

### Results
The assembly model displacement indicates the maximum displacement occurs at the upper part of the femoral head, as shown in Fig. 2 (total deformation). The proximal femur displacements were 10.25 mm for the MCS model, 9.66 mm for DHS model, 9.44 mm for DHS + screw model, 9.86 mm for FNS model, and 9.31 mm for PFNA-II model (Fig. 3). The VMS distributions for each assembly were assessed and are shown in Fig. 4. Stress distribution areas in all assembly models had maximum VMS values at the fracture site except for the DHS + screw model, which was at the inner side of the inferior most screw (Fig. 4).

The maximum VMS at the fracture site was 621.13 MPa for the MCS model, 464.14 MPa for the DHS model, 64.51 MPa for the DHS + screw model, 344.54 MPa for the FNS model, and 647.49 MPa for the
PFNA-II model (Fig. 5). The maximum VMS area at the fracture sites varied greatly, depending upon the type of fixation implant (Fig. 6). In the MCS model, fracture site stresses appeared to be concentrated at the superior area, with high VMS points around the posterior screw cortex. In the DHS model, fracture site stresses were concentrated in the superior area, with the maximum VMS stress at the anterosuperior corner. Stresses in the DHS + screw model were evenly distributed, and the maximum VMS of the anterior corner disappeared. In the FNS model, the max area was in the posteroinferior site and posterior superior site around the entry portal (trochanteric fossa) of head-neck fragment, in the PFNA-II model.

The maximum VMS areas in the implants were generally found around the fracture site. In the MCS model, the maximum area was in the inferior screw around the fracture site and in the area between the lag screw and the barrel of the DHS model (Fig. 7). The maximum VMS in the implants was 148.94 MPa for MCS model, 414.66 MPa for DHS model, 385.59 MPa for DHS + screw model, 464.07 MPa for FNS model, and 505.07 MPa for PFNA-II model (Fig. 8). The fixation strength of assembly models and the loading patterns of implant had a significant difference. (Fig. 9). In terms of load bearing around the fracture site, the DHS + screw / FNS model was lower, and MCS / PFNA model, higher than other models. Considering the load-sharing aspect of implant, the MCS model had the lowest value, but the FNS / PFNA-II models had a higher load-sharing point. Compared with the DHS + screw model, the FNS model had significantly increased stress distribution around the bone and fracture site. However, the implant did not have much difference. Comparing DHS + screw and the PFNA models, the stress distribution between the implant and fracture sites differed significantly, and the PFNA model had higher VMS in all bone, implant, and fracture sites.

**Discussion**

Basicervical fractures of the proximal femur are controversial type. Anatomically and biomechanically, they have been considered to be either intracapsular or extracapsular fractures. Concerning the surgical management of basicervical fracture, although internal fixation might be associated with increased failure and reoperation rates, related factors could not be clearly introduced because of variant fracture types, several available implants, and inconsistencies in fixation technique and in enrolling patients. Because of these clinical case study limitations, we applied FE analysis to investigate the biomechanical behavior of assembly models and stress distribution in the fracture sites. This computational analysis enabled us to detect several interesting findings; 1) The DHS + screw and PFNA-II models had the least displacement, and MCS models had the most displacement. 2) The stress distribution pattern and VMS around the fracture sites differed significantly, depending upon the implants used. The MCS and DHS models had maximum VMS in the fracture site's superior aspect. However, the PFNA-II model had maximum VMS at the head-neck fragment's entry point, around the trochanteric fossa. 3) As compared to the single screw DHS model, the two integrated FNS and DHS + screw model had uniform stress distribution around the fracture site, and the maximal VMS disappeared in the superior aspect. 4) The stress distribution of bone and fracture site was similar in the MCS and PFNA-II models, but this was much different in the implant.
As VMS distribution for implants is an indicator of the metal yield and can be explained by the mechanics of load-sharing\textsuperscript{19,20}, our results demonstrate that PFNA-II had the highest stress values and the MCS model had the lowest. Although an MCS fixation for basicervical fractures is seldom used, our FEA results showed that the MCS model has the highest displacement and lowest load-sharing. The maximum VMS in the MCS model was placed in the superior aspect of the fracture site and inferior screw, and this leads a higher possibility of implant failure. Compared with the DHS + screw model as the extramedullary fixation, the FNS model had some difference in fracture site stress distribution. Even if there was not much difference between the implant stress values, the fracture site distributions gave us meaningful patterns. By adding an anti-rotational screw to the DHS model (DHS + screw), the maximum VMS disappears on superior aspect of fracture site, similar to the FNS model. When the stress values at the fracture site were compared with the DHS + screw model (64.51 MPa) and the FNS model (344.54 MPa), there were many differences (Fig. 5). Therefore, an FNS for basicervical fractures should be used for load-sharing implant, and cortical contact should be achieved without comminution to prevent collapse because the VMS values at the fracture site are high. Considering that the variant type of basicervical fracture have been recently reported\textsuperscript{18,21}, the FNS fixation for these types might not be appropriate based on the results of this study.

Although there is no consensus on whether extramedullary or intramedullary fixation is the best treatment for intertrochanteric fracture, CMN has been regarded currently as the effective implant because it provides a biomechanical advantage over DHS, particularly for unstable fracture patterns or under the signs of comminution.\textsuperscript{22–24} Thus, the indications for CMN have been broadened greatly and led to increased CMN use for basicervical fracture patterns.\textsuperscript{11} Although fractures of two fragments is usually associated with fewer complications, it has been well known that basicervical fractures fixed with CMN had an increased rate of cut-out mechanical failure.\textsuperscript{8,25,26} For preventing lag screw mechanical failure, helical blades and two integrated screw types could be used because they had higher rotational stability than lag screws.\textsuperscript{27,28} The results of this study show that the PFNA-II model had the least displacement and greatest implant VMS, and it was found that the PFNA-II model played a very good role in load-sharing. Interestingly, however, regarding the stress distribution in the fracture site, the maximum VMS was concentrated at the entry portal around trochanteric fossa of head-neck fragment (Fig. 10). Therefore, given the high risk of reverse wedge effect following CMN fixation for the basicervical fracture, this may not reduce neck collapse or mechanical cut-out failure despite the increased rotational stability of the helical blade.\textsuperscript{29}

In CMN fixation for trochanter fractures, clinical complications include the rotation of femoral head and the cut-out phenomenon of the lag screw, which is the most common cause of failure and remains one of the major clinical challenges.\textsuperscript{30} Despite remarkable research efforts in evolving concepts of postoperative stability for trochanter fractures, detailed mechanics of how these complications arise have not been fully documented. Throughout our result of fracture site stress distribution, we could identify the rotational movement caused by the axial load-induced deformation of the PFNA-II model, which was suppressed by the proximal osseous support around the entry point, especially in the anterior and posterior parts of the
head-neck fragment. Thus, surgical technique and implant design modifications to increase the osseous support might help increase the rotational instability and prevent the reverse wedge effect.

Despite these interesting findings, this computational simulation study has several limitations. First, we simulated a highly simplified model in a normal non-osteoporotic femur not indicated for CMN fixation. Second, our results had rather descriptive characters because the basicervical fracture was perfectly reduced without gap. Third, the fracture impact caused by lag screw/blade-controlled sliding could not be considered because of technical difficulties. Nevertheless, our computational simulation can evaluate the structure-mechanical strength, VMS distribution in the fracture site, and the load sharing-/bearing degree under the same conditions. Although implant choice should be determined according to the degree of displacement, fracture type, physiological age, and bone quality, our results may directly suggest the technical relevance of maximizing structural strength and minimizing stress concentration around the fracture site.

**Conclusion**

Considering the stress distribution of the fracture site and implant, the optimal load-bearing implant was a DHS + screw fixation construct, and an FNS implant might be applied to anatomically reduce noncomminuted basicervical fractures. When performing CMN fixation, high stress concentrations around the entry point in the head-neck fragment should be minimized by increasing the osseous support around proximal end of the CMN.

**Declarations**

**Acknowledgements**

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**Author contributions**

J.W.K, C.W.O, and G.H.J. planned the experiments. S.L.J. and G.H. J. performed processing of the model analysis. G.H.J. wrote original draft preparation. J.W.K, C.W.O, and B.S.K. reviewed and edited the writing. J.W.K, B.S.K, and G.H.J. 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.

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

References


Figures

Figure 1
The distal end of femur model was completely fixed, and 1950 N loads were applied to the center of the femoral head. Each assembly model was adducted 7° in the vertical ground to mimic the normally physiologic alignment of the lower limbs in the standing position.

Figure 2

According to the displacement of assembly model, the maximum displacement occurs at the upper part of the femoral head.
Figure 3

The displacement of the proximal femur was the largest in the MCS model and smallest in the DHS + screw model.
Figure 4

The VMS distributions for each assembly were assessed, and the maximal area of all models was in the fracture site except the DHS + screw model.
Figure 5

The VMS distributions of the fracture sites were the largest in the PFNA-II and smallest in the DHS + screw models.
Figure 6

The fracture site's maximum VMS area had many differences, depending upon the fixation constructs.
Figure 7

The maximum VMS areas of implant were generally located around the fracture site. In the MCS model, the maximum area was at the inferior screw around the fracture site as well as the area between lag screw/blade and barrel in the DHS model/PFNA-II model.

Max von Mises stress of implants

Equivalent Stress(MPa)

Figure 8

The VMS distribution of implant was largest in the PFNA-II and smallest in the MCS model, which does not have a load-sharing function.
Figure 9

Regarding the fixation strength of assembly models and loading patterns of implant, there was a significant difference among them. In the load-bearing aspect around fracture site, the DHS+screw and FNS model was lower than others and higher in the MCS and PFNA model.
The PFNA-II model had the least displacement and the largest VMS of implant, and we found that it played a major role as a load-sharing implant. However, in the stress distribution of fracture site, the maximum VMS was concentrated at the entry portal around trochanteric fossa of the head-neck fragment.