

# Effect of Controlled Simulated Cam Morphology on Proximal Femoral Stress Distribution: A Finite-Element Study

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## Research Article

**Keywords:** Femur, Finite Element Analysis, Cam deformity, Osteoporotic bone, Simulation

**Posted Date:** September 29th, 2026

**DOI:** <https://doi.org/10.21203/rs.3.rs-11176469/v1>

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**Additional Declarations:** The authors declare no competing interests.

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## **Effect of Controlled Simulated Cam Morphology on Proximal Femoral Stress Distribution: A Finite-Element Study**

**Running title: Effect of Controlled Simulated Cam Morphology on Stress in the Proximal Femur**

### **ABSTRACT**

#### **Background**

Cam-type femoral impingement has been associated with altered hip biomechanics. However, the effect of increasing cam morphology on stress distribution within the proximal femur remains incompletely understood. Finite-element analysis may provide a preliminary method for investigating this relationship.

**Methods:** A three-dimensional finite element analysis model was developed from computed tomography data of a single morphologically normal proximal femur. Three models were created with controlled geometric modification to condition simulating low bone density based on alpha angles: 40° (no deformity), 60° (moderate), and 80° (severe). Simulations were conducted under two loading conditions: sit-to-stand and trochanteric lateral forces, with physiological and supraphysiological loads.

**Results:** Increasing the simulated  $\alpha$  angle was associated with changes in the magnitude and distribution of calculated stress and deformation in the proximal femur. The magnitude of these changes varied according to the loading condition and anatomical region examined (neck or intertrochanteric). The greatest values occurred in the 80° model when testing the sit-to-stand scenario, increasing both stress concentration (+207.39% in local von Mises equivalent stress [LE] at the trochanteric area) and stress distribution (up 367% in femoral neck and 487% in intertrochanteric), compared to the 40°, our reference model.

**Conclusions:** In this single-femur computational model, the simulated cam-like geometry acted as a stress amplifier tested for low bone density conditions. These findings should be regarded as exploratory and hypothesis-generating. External validation is required for any clinical relevance in real bone setting.

**Keywords:** Femur; Finite Element Analysis; Cam deformity; Osteoporotic bone; Simulation.

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## INTRODUCTION

Femoroacetabular impingement (FAI) is associated with morphological abnormalities of the femoral head and acetabulum. Cam-type morphology is primarily defined by the alpha angle, whereas pincer-type morphology is characterized by acetabular overcoverage. The coexistence of both features defines mixed-type impingement [1-3].

While classically linked to osteoarthritis, FAI is also associated with traumatic injuries. It is a risk factor for posterior hip dislocation due to a cantilever effect [4-6] and is linked to femoral neck [7], intertrochanteric [8], and stress fractures [9, 10]. The role in proximal femoral fracture risk remains controversial [11], and no prior studies, in our literature review, have evaluated its influence on proximal femur stress distribution. Thus, the mechanical consequences of cam morphology within the proximal femur remain uncertain. The clinical associations do not establish, with certainty, that cam morphology increases fracture risk. These findings may depend on bone quality, cortical thickness, cancellous architecture, loading direction, and individual femoral anatomy.

Finite element analysis (FEA) has been widely used in orthopedic research to simulate mechanical scenarios *in silico* [12]. This technique enables the development of virtual three-dimensional geometric models subjected to simulated loading conditions, allowing detailed assessment of stress distribution, deformation patterns and biomechanical interactions.

Due to the lack of biomechanical studies linking FAI to stress in the proximal femur, we performed a finite-element analysis to investigate: (1) whether cam deformity alters proximal femur stress concentration, and (2) how progressively greater simulated cam-like geometry influence stress distribution in different areas (femoral neck and intertrochanteric region).

## METHODS

### Virtual anatomical model

A three-dimensional model was generated from a computed tomography (CT) scan. Imaging was performed using a 16-slice CT scanner (Emotion, Siemens<sup>TM</sup>, Germany) with a resolution of  $512 \times 512$  and a slice thickness of 1.0 mm. The femur used in this study was obtained on an anonymized image database [13], from a previous study. The specimen was selected based on the availability of an anatomically complete femur, hemi pelvis, and adequate image quality for three-dimensional

reconstruction. The database did not provide sufficient clinical information to determine whether the subject had osteoporosis, osteopenia, previous fracture, or other conditions affecting bone quality. Therefore, the selected femur cannot be considered representative of a defined age, sex, or bone-health population.

The selected images were independently reviewed by three orthopaedic surgeons. The  $\alpha$  angle and lateral center-edge angle were measured on the original CT images to assess whether the femur showed radiographic features of cam or pincer morphology, and avoid bias from preexisting deformities. As the independent measures showed no abnormalities, images were considered adequate for this study and the creation of the controlled geometric modification. The left hemipelvis and left proximal femur were segmented from digital imaging and communications in medicine files and converted into standard tessellation language (STL) format.

### **Creation of virtual models and cam-type deformities**

Three-dimensional models of the bone structures were generated from the STL files using Rhinoceros™ 6 (Robert McNeel & Associates, United States). FEA simulations were performed using SimLab™ (HyperWorks, United States) with the OptiStruct solver.

Three femoral models, with simulated increases in the alpha angle, were created: one with an angle of 40° (reference model) and two with simulated cam-type deformities at the head-neck junction, with alpha angles of 60° and 80°, values deliberately chosen to represent moderate and severe deformities, respectively.

These modifications were intended to isolate the mechanical effect of increasing the measured geometric parameter. They were not derived from serial patient imaging and do not reproduce the full biological development or anatomical variability of cam morphology. In particular, the present modelling procedure does not establish whether the additional geometry in a clinical cam deformity consists primarily of cortical bone, cancellous bone, or a combination of both.

The cam deformity was created with its apex at the 1:30 clock position, extending from 12 to 3 o'clock, corresponding to the region most commonly associated with cam-type FAI [14, 15] (Fig. S1). The original femoral head radius was preserved, the neck-shaft angle was unchanged, and the cortical shell thickness was preserved.

The acetabular morphology was not modified, resulting in models aiming to simulate a cam-

type FAI (Fig. 1a, 1b and 1c), After geometric modification, the  $\alpha$  angle was remeasured on each model to confirm the target values of 40°, 60°, and 80°.

### **Discretization, material properties, and FEA**

The data were segmented to reconstruct the external geometry of the femur. The cortical bone and cancellous bone, were represented according to the segmentation and geometric-processing procedures described below. The internal bone structure was simplified for computational modelling and should not be interpreted as a complete representation of the subject-specific cortical thickness or trabecular architecture.

Models were imported into SimLab™ for FEA. Cortical and trabecular regions were meshed using second-order tetrahedral elements. The 40°, 60°, and 80° models contained 131,230 elements and 1,236,581 nodes; 142,365 elements and 1,325,489 nodes; and 149,658 elements and 1,398,476 nodes, respectively (Fig. S2). This difference in the number of elements and nodes is due to the geometric modification associated with the simulated deformity.

The cortical shell was generated by applying a uniform geometric offset to the external femoral surface, and the remaining internal volume was assigned as cancellous bone. The reference model (40°) matched established biomechanical data [16]. Material properties intended to approximate low-density bone were assigned. [17] were assigned using a Poisson's ratio of 0.3 for both tissues, with a Young's modulus of 13.7 GPa for cortical bone and 1.37 GPa for trabecular bone. The material-property values were adopted from previously published data [17] of vertebral bone and were used as approximate values because femoral properties are not available in the literature or the FEA software. These may not fully represent the mechanical properties of femoral cortical and cancellous bone. This limitation was considered when interpreting absolute stress values.

Cortical and cancellous bone were assigned uniform material properties within their respective regions. This was a simplifying assumption made to permit a controlled comparison among the simulated geometries. The model therefore does not reproduce regional differences in density, anisotropy, cortical thickness, or elastic properties among the femoral head, neck, trochanter, and shaft. In addition, subject-specific bone mineral density was not available, so the model was created with values from literature [17]. It should not be interpreted as a validated representation of an osteoporotic femur.

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## Boundary conditions (BCs) and applied loads

The magnitude of the applied loads was based on values previously obtained from direct measurements in instrumented hip implants taken during activities of daily living [18]. The supraphysiological condition was generated by multiplying the physiological force magnitudes by two to examine the model response under an increased-load scenario.

Two BCs were applied: sit-to-stand movement and an idealized lateral loading condition, which we created based on the common trauma mechanism of a sideways fall, associated with trochanteric fractures in older adults [21-22]. The condition was inspired by the clinical mechanism of a sideways fall but was not intended to reproduce the complete kinematics or impact dynamics of an actual fall.

The pubis, sacroiliac region, and ischium were fixed in translation along the X, Y, and Z axes in both scenarios. Rotational constraints were also applied. In the lateral-loading condition, an additional constraint was applied at the distal femoral shaft. For the sit-to-stand condition, the femur was positioned at 90° of flexion, 10° of adduction, and 10° of internal rotation according to the prescribed model configuration. (Fig. S2). The lateral loading scenario kept the femur at zero (neutral) flexion, 10° of adduction, and 10° of internal rotation.

The loading and boundary conditions represented idealized mechanical scenarios rather than complete simulations of physiological movement or actual sideways falls. The prescribed constraints were used to compare the relative response of the different simulated geometries under identical computational conditions.

### *Condition 1 (BC 1): sit-to-stand movement*

The first condition simulated the sit-to-stand movement, an activity associated with high loads on the proximal femur in biomechanical studies [18, 19].

Vector loads were applied at the greater trochanter to represent the approximate insertions of the gluteus minimus (GMin), gluteus medius (GMed), and short external rotators (SERs), along with a flexion force applied at the distal femur (F). These vectors were intended to simulate the action of the primary muscle groups acting on the hip (Fig. 2). The following loads were applied: physiological loading – GMin 123.4 N, GMed 156.5 N, SERs 37 N, F 203.7 N; and supraphysiological loading – GMin 246.8 N, GMed 313 N, SERs 74 N, F 407.4 N.

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### *Condition 2 (BC 2): lateral loading condition*

The second condition simulated a lateral load at the greater trochanter. In this condition, an additional fixation point was applied at the distal femoral shaft. A load of 3000 N was applied on the lateral surface of the greater trochanter, with the force vector aligned along the axis of the femoral neck (Fig. 3).

### **Study variables**

The primary outcomes were total displacement (TD), total von Mises stress (TE), and local von Mises stress in the femoral-neck (LE1) and intertrochanteric regions (LE2). For each model and loading condition, the maximum value was recorded. Results were reported as absolute values and as percentage differences relative to the 40° reference model. Heatmaps were used for graphical comparison across geometric configurations, loading conditions, and anatomical regions.

## **RESULTS**

Across the simulated models, increasing the  $\alpha$  angle was associated with changes in calculated equivalent stress and deformation under the prescribed loading conditions. The direction and magnitude of these changes differed according to the loading condition and anatomical region. Because the analysis involved one femur and one model realization for each geometric condition, no statistical inference or population-level estimate can be made.

Table 1 presents the absolute values for all model and loading conditions. Calculated stress and deformation generally increased as the simulated  $\alpha$  angle increased, although the magnitude and pattern of change differed according to the loading condition and anatomical region.

In the sit-to-stand simulation (BC 1), percentage analysis demonstrated progressive increase in the three variables (TD, TE, local von Mises), as the deformity grew and with the transition from physiological to supraphysiological loading. The highest local von Mises values were observed in the intertrochanteric region (LE2) in the 80° model under supraphysiological loading (+207.39%) and, in the same group, for TD (+324.71%) compared with the 40° model.

In the lateral load simulation (BC 2), percentage analysis likewise demonstrated more critical conditions in the 80° model. The highest local von Mises values were observed in the femoral neck (LE1;

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+132.54%) and for TD (+47.47%) relative to reference (40°) model. Table S1 shows the percentage variation of all variables across the different experimental scenarios.

In the sit-to-stand simulation (BC 1), comparative heatmap analysis demonstrated an increase in the relative local stress values. In the femoral neck, the highest value was observed in the 80° model under supraphysiological loading, with an overall increase of 367% compared with the baseline 40° model (Table S2).

In the intertrochanteric region (LE2), the 80° model showed a peak overall increase of 487% relative to the 40° model. The 60° model also showed higher calculated local stress values than the 40° reference model, with increases exceeding 200% under the relevant loading condition. (Table 2).

Fig. 4 illustrates peak stress concentration across the different sit-to-stand simulations (BC 1). Stress concentration was observed near the simulated cam-like region, with additional high-stress regions identified in the femoral neck and intertrochanteric/metaphyseal areas.

In the lateral load simulation (BC 2, with supraphysiological loading), cross-analysis of local stresses demonstrated a progressive percentage increase in both evaluated regions. The highest values were observed in the femoral neck in the 80° model, with a +132% increase relative to the baseline 40° model (Table S3).

In the intertrochanteric region, peak stress was also observed in the 80° model, with an overall increase of +87% compared with the baseline 40° model (Table 3).

Fig. 5 illustrates peak stress concentration across the different lateral load simulations (BC 2). A tendency toward stress concentration in the posterior cortical region of the femoral neck was observed, with progressive intensification as deformity severity increased.

## DISCUSSION

The principal finding of this exploratory analysis was that controlled geometric increases in the simulated  $\alpha$  angle altered the calculated stress and deformation patterns in a single femoral model. The results suggest that proximal femoral geometry may influence local load transfer under prescribed loading conditions.

This trend was observed in both global measurements—total displacement (TD) and global von Mises stress (TE)—and localized analyses of the femoral neck (LE1) and intertrochanteric region (LE2). The magnitude of these changes differed between the two loading conditions and was generally greater

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under BC1.

Under the sit-to-stand condition, increasing the simulated  $\alpha$  angle was associated with higher calculated values of total displacement, global von Mises stress, and local von Mises stress in the femoral neck and intertrochanteric regions. Greater displacement under the prescribed loading conditions was consistent with increased calculated deformability in the models with larger simulated  $\alpha$  angles. This finding may reflect a reduction in apparent structural stiffness; however, stiffness was not directly determined from a force–displacement analysis.

The regional stress patterns differed according to the geometric and loading conditions. Higher calculated stress was observed in both the femoral neck and intertrochanteric regions, although the region with the highest value varied between model configurations. These differences may reflect altered load transfer through the proximal femur. That is a possible interpretation not proven mechanism, as the change in the force-transmission axis was not directly measured.

Under the idealized lateral trochanteric loading condition, the 80° model showed the highest calculated local stress values, including in the posterior cortical region of the femoral neck. Because this condition used prescribed static constraints and did not reproduce the complete kinematics or impact dynamics of an actual sideways fall, the results should be interpreted as a comparison of model responses under controlled lateral loading rather than as a validated simulation of a clinical fall-related fracture.

These findings are particularly relevant given the ongoing controversy regarding the relationship between FAI and traumatic hip injuries. Previous studies suggest that FAI, especially cam and mixed morphologies, may be associated with traumatic posterior dislocation of the hip, possibly due to a cantilever effect caused by impingement between the femoral head-neck junction and the acetabular rim during flexion and internal rotation [4, 6, 7].

Some studies also associate cam morphology with proximal femoral fractures [8, 9]. It was also previously suggested, that limited internal rotation combined with torsional trauma creates a cantilever effect, increasing proximal femur stress [6, 7, 9]. This is consistent with our sit-to-stand simulation, showing progressive stress increases with larger artificially created alpha angles. Similarly to Yang et al. [7], higher loads and increasing deformity in our model consistently elevated femoral neck stress.

The present results cannot be directly applied to clinical scenarios. They support the hypothesis that changes in proximal femoral morphology may alter calculated stress and load-transfer patterns, but

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this hypothesis requires validation using multiple patient-derived models and experimental or clinical data.

Despite this inherent limitation, the present results suggest that the simulated cam deformity does not produce a single or uniform effect, but rather alters stress distribution in various areas of the proximal femur, increasing measured stress in both the femoral neck and intertrochanteric region, depending on the magnitude of the created deformity and also the forces involved. The evaluation of multiple loading scenarios is therefore a strength of this study.

Liu et al. [23], using FEA models with varying alpha angles, demonstrated that cartilage contact pressures may show minimal differences during gait but increase substantially during activities involving larger hip flexion, such as sit-to-stand movement, especially for alpha angles  $\geq 80^\circ$  [18, 23]. In the present study, BC 1 was designed to replicate this functional transition, incorporating muscle force vectors to reduce bias in periarticular load distribution — a common limitation in previous studies that did not include muscle forces. Our results demonstrate that the effects of simulated cam deformity are not limited to the articular surface, but can also extend to different areas at the proximal femur.

Other FEA studies demonstrate how FAI alters hip biomechanics. Ng et al. [24] showed that cam deformity with varus neck-shaft angles elevates acetabular stress during squatting, while Hellwig et al. [25] linked anatomical variations to joint degeneration via repetitive overloading. A systematic review confirmed increased stress in cam-type FAI, particularly in the anterosuperior quadrant at higher ranges of hip motion [26]. Our study examines whether a similar relationship may also be present within the proximal femur, as progressive alpha angles ( $40^\circ$ ,  $60^\circ$ ,  $80^\circ$ ), in our model using simplified low-density bone material properties, increase stress concentration in different regions of the proximal femur.

These findings provide a related, but not directly equivalent, computational perspective. Simulating an increased alpha angle, raises not only the joint and cartilage loads described in previous simulations but also the stress concentration in the proximal femur (acting as a supporting structure), especially in the femoral neck and intertrochanteric regions. These are regions clinically relevant to various fracture patterns, although the FEA does not predict clinical outcomes.

Our study has important limitations. First, it used one femoral anatomy and therefore cannot establish population-level generalizability, it overlooks variations in bone density related to sex, ethnicity, comorbidities, or prior injury. Second, it relies on a computational simulation with inherent simplifications, omitting structures like the joint capsule and labrum. Also, local finite-element stress

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peaks may be amplified by mesh density, sharp geometric transitions, contact definitions, or rigid constraints. Third, cortical and cancellous bone were represented as homogeneous, isotropic materials, without subject-specific bone mineral density or regional variation in cortical thickness, anisotropy, or trabecular architecture. Consequently, the assigned properties should not be interpreted as a validated representation of osteoporosis. Fourth, the model focuses exclusively on femoral cam deformity with a fixed pelvis, excluding pincer deformities or spinopelvic interactions. Also, the loading and boundary conditions were idealized. The sit-to-stand and lateral trochanteric conditions were used to compare the geometries under controlled computational conditions but did not reproduce the full dynamics of physiological movement as there are dynamic variations in the muscle tension in different moments during motion, which our static simulations fail to detect. Finally, FEA requires validation with bone models. The absence of experimental validation and formal mesh-convergence testing, limits confidence in the absolute stress values and local peaks, particularly near geometric transitions and constrained regions.

Despite these limitations, this study has several strengths. We did not identify a previous study that evaluated simultaneously the effect of cam deformity simulations on stress patterns across the neck and trochanteric regions, they focused on the articular surface. Plus, the progressive analysis of three  $\alpha$ -angle configurations allowed an exploratory assessment of whether the response varied progressively with the magnitude of geometric modification. Additionally, the inclusion of two BCs and different load levels enhanced functional scenarios, and both the sit-to-stand and lateral-loading conditions were inspired in clinically relevant fracture mechanisms, even though were not complete simulations of real-life situations. Further, the model included force vectors intended to approximate the actions of selected periarticular muscles, which were lacking in previous FEA studies. Also, we aimed to, at least in part, address bone density with both cortical and trabecular bone.

The findings support the hypothesis that changes in head-neck geometry may alter calculated proximal femoral load transfer. Future studies should include multiple patient-derived femora with and without cam morphology, separate analyses of mild and severe deformity, subject-specific bone mineral density, regional and anisotropic material properties, mesh-convergence assessment, clinically representative loading conditions, and comparison with experimental strain measurements or clinical fracture patterns

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## CONCLUSION

In a single-femur finite-element model, controlled increases in the simulated  $\alpha$  angle were associated with changes in calculated proximal femoral stress and deformation under prescribed loading conditions. The findings suggest that head-neck geometry may influence local load transfer. However, due to the artificially modified geometry, a single bone with simplified properties, idealized loading and boundary conditions, the results should be regarded as exploratory and hypothesis-generating to guide additional research.

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## Ethical standards

All procedures were conducted in accordance with the ethical standards of the institutional research committee. Informed consent was waived due to the non-interventional design of the study and the use of an anonymized image database

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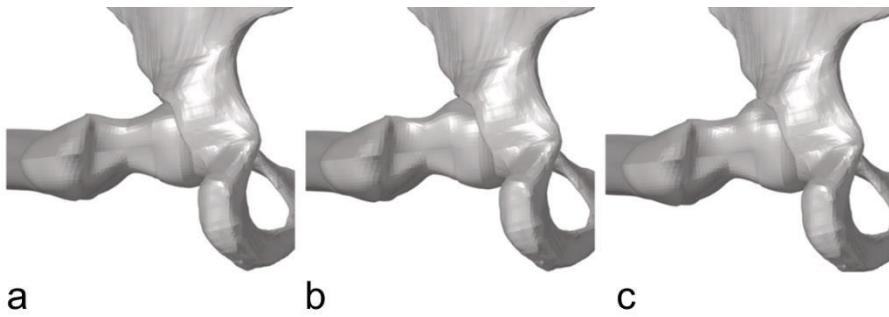
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## FIGURE LEGENDS

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**Fig. 1** Models generated according to simulated alpha angle/degree of deformity. (a) Alpha angle of 40° (reference model, no deformity). (b) Alpha angle of 60° (moderate deformity). (c) Alpha angle of 80° (severe deformity)



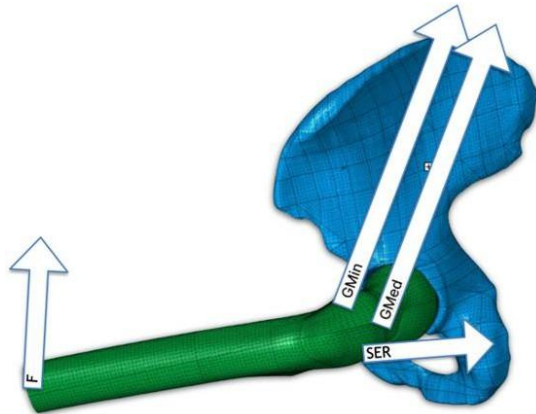
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Descriptive caption: Three-dimensional computational representations to simulated alpha angle and degree of deformity. The first image illustrates 40 degrees alpha angle (no deformity/referene), the second illustrates 60 degrees alpha angle (moderate deformity), and the third illustrates 80 degrees alpha angle (severe deformity).

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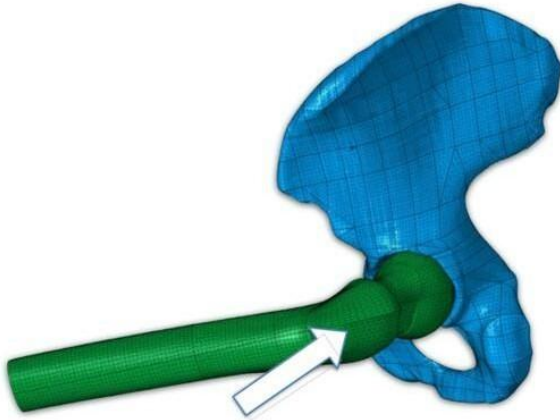
**Fig. 2** Force vectors applied in condition 1 (sit-to-stand)

F, flexion vector; GMin, gluteus minimus force; GMed, gluteus medius force; SER, short external rotator force.



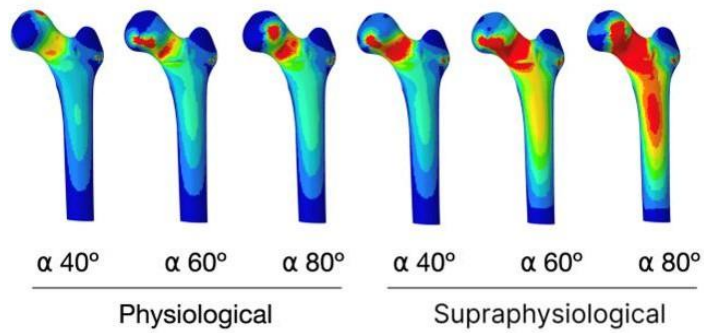
Descriptive caption: Force vectors applied in condition one, sit-to-stand. The diagram shows arrows representing muscle and reaction forces acting on the meshed pelvic and femoral models. F indicates flexion vector, GMin indicates gluteus minimus force, GMed indicates gluteus medius force, and SER indicates short external rotator force.

**Fig. 3** Force vector applied in condition 2 (lateral force at the greater trochanter)



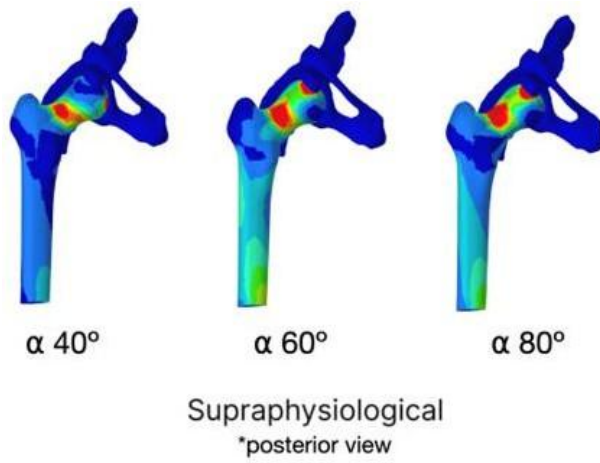
Descriptive caption: Force vector applied in condition two, lateral force. The diagram shows a single white arrow pointing directly at the greater trochanter region of the meshed femoral model.

**Fig. 4** Stress distribution at peak loading during condition 1 (sit-to-stand)



Descriptive caption: Stress distribution at peak loading during condition one, sit-to-stand. The image displays six proximal femur models showing colored maps of mechanical stress. The first three models represent physiological loads with alpha angles of 40 degrees, 60 degrees, and 80 degrees. The next three models represent supraphysiologic loads with alpha angles of 40 degrees, 60 degrees, and 80 degrees. Red and orange areas indicating higher stress concentrations increase progressively with larger alpha angles and higher loads.

**Fig. 5** Stress distribution at peak loading during condition 2 (lateral force at the greater trochanter)



Descriptive caption: Stress distribution at peak loading during condition two, lateral force at the greater trochanter. The image displays a posterior view of three hip joint models under supraphysiologic loads, corresponding to alpha angles of 40 degrees, 60 degrees, and 80 degrees. Colored maps show mechanical stress on the femur and pelvis, with red and orange areas indicating high stress concentration at the femoral neck junction across all three angles.

**TABLES**

**Table 1 Absolute values obtained in the study models**

| Stress distribution across the study models (absolute) |           |               |       |        |                    |         |        |
|--------------------------------------------------------|-----------|---------------|-------|--------|--------------------|---------|--------|
| Simulation                                             | Variable  | Physiological |       |        | Supraphysiological |         |        |
|                                                        |           | 40°           | 60°   | 80°    | 40°                | 60°     | 80°    |
|                                                        |           |               |       |        |                    |         |        |
| Sit-to-stand                                           | TE (MPa)  | 843.24        | 920.5 | 1078.5 | 1373.3             | 1840.7  | 2892.4 |
|                                                        | LE1 (MPa) | 102.05        | 132.5 | 168.24 | 183.98             | 261.57  | 477.24 |
|                                                        | LE2 (MPa) | 43.84         | 66.26 | 91.67  | 83.75              | 149.09  | 237.42 |
|                                                        | TD (mm)   | 0.64          | 0.69  | 1.29   | 0.85               | 1.39    | 3.61   |
| Lateral Force                                          | TE (MPa)  |               |       |        | 1346.40            | 1777.31 | 2083.8 |
|                                                        | LE1 (MPa) |               |       |        | 98.54              | 142.19  | 229.13 |
|                                                        | LE2 (MPa) |               |       |        | 29.14              | 38.74   | 54.64  |
|                                                        | TD (mm)   |               |       |        | 0.99               | 1.12    | 1.46   |

LE1, local von Mises equivalent stress in the femoral neck; LE2, local von Mises equivalent stress in the intertrochanteric region; TD, total displacement; TE, total von Mises equivalent stress

**Table 2 Heatmap of stress variables located in the intertrochanteric region of the femur in the sit-to-stand simulation**

| Alpha angle |                    | % Heatmap – Sit-to-stand – LE2 (intertrochanteric) |      |      |                    |      |     |
|-------------|--------------------|----------------------------------------------------|------|------|--------------------|------|-----|
| 40°         | Physiological      | X                                                  | -34  | -52  | -48                | -71  | -83 |
| 60°         |                    | 51.1                                               | X    | -28  | -21                | -56  | -74 |
| 80°         |                    | 109                                                | 38.3 | X    | 9.5                | -39  | -64 |
| 40°         | Supraphysiological | 91                                                 | 26.4 | -8.6 | X                  | -44  | -68 |
| 60°         |                    | 240                                                | 125  | 62.6 | 78                 | X    | -42 |
| 80°         |                    | 487                                                | 289  | 181  | 207                | 72.7 | X   |
|             |                    | Physiological                                      |      |      | Supraphysiological |      |     |
|             |                    | 40°                                                | 60°  | 80°  | 40°                | 60°  | 80° |

LE2, local von Mises equivalent stress in the intertrochanteric region

**Table 3 Heatmap of stress variables located in the intertrochanteric region of the femur in the lateral force at the greater trochanter simulation**

| Alpha angle | % Heatmap – Sideways fall – LE2 (intertrochanteric) |                    |     |     |
|-------------|-----------------------------------------------------|--------------------|-----|-----|
| 40°         | Supraphysiological                                  | X                  | -25 | -47 |
| 60°         |                                                     | 32.9               | X   | -29 |
| 80°         |                                                     | 87.5               | 41  | X   |
|             |                                                     | Supraphysiological |     |     |
|             |                                                     | 40°                | 60° | 80° |

LE2, local von Mises equivalent stress in the intertrochanteric region