

Finite Element Analysis of Different Internal Fixation Configurations for Delbet Type II Femoral Neck Fractures in Preadolescent Children: A Parametric Biomechanical Study

Xinxiao Chen

Weifang People's Hospital, Shandong Second Medical University

Yang Gao

The Third Bethune Hospital of Jilin University

Changpeng Cao

The Third Bethune Hospital of Jilin University

Haiyang Xing

The Second Bethune Hospital of Jilin University

Faxin Cao

Rizhao People's Hospital

Zhipeng Du

The 988th Hospital of the Joint Logistics Support Force of the Chinese People's Liberation Army

Guanning Huang

The Third Bethune Hospital of Jilin University

Gang Wang

w_g@jlu.edu.cn

The Third Bethune Hospital of Jilin University

Research Article

Keywords: pediatric femoral neck fracture, Delbet Type II, internal fixation, finite element analysis, biomechanic

Posted Date: October 21st, 2026

DOI: <https://doi.org/10.21203/rs.3.rs-11156482/v1>

License: © ⓘ This work is licensed under a Creative Commons Attribution 4.0 International License.

[Read Full License](#)

Additional Declarations: No competing interests reported.

Finite Element Analysis of Different Internal Fixation Configurations for Delbet Type II Femoral Neck Fractures in Preadolescent Children: A Parametric Biomechanical Study

Authors:

Xinxiao Chen¹, Yang Gao², Changpeng Cao², Haiyang Xing³, Faxin Cao⁴, Zhipeng Du⁵, Guanning Huang² & Gang Wang^{2*}

Affiliations:

¹ Weifang People's Hospital, Shandong Second Medical University, Weifang, Shandong 261000, China.

² The Third Bethune Hospital of Jilin University, Changchun, Jilin 130033, China.

³ The Second Bethune Hospital of Jilin University, Changchun, Jilin 130021, China.

⁴ Rizhao People's Hospital, Rizhao, Shandong 276827, China.

⁵ The 988th Hospital of the Joint Logistics Support Force of the Chinese People's Liberation Army, Zhengzhou, Henan 450007, China.

email:

Xinxiao Chen¹: cxx19980313@163.com

Yang Gao²: ayjlzhm1012@qq.com

Changpeng Cao²: caocp17808072260@163.com

Haiyang Xing³: xinghaiyang@jlu.edu.cn

Faxin Cao⁴: xfc17862817752@163.com

Zhipeng Du⁵: 1264960325@qq.com

Guanning Huang²: hgn940314@163.com

Gang Wang^{2*}: w_g@jlu.edu.cn

Abstract

Pediatric femoral neck fracture (PFNF) are rare but serious injuries with high complication rates. Delbet Type II fractures are the most common subtype. Optimal internal fixation remains controversial regarding screw number, diameter, and hybrid constructs. Finite element analysis (FEA) offers a non-destructive method to compare fixation biomechanics. To biomechanically evaluate nine internal fixation configurations for Delbet Type II PFNF, quantifying the relative influence of screw diameter versus screw number, evaluating a hybrid screw–Kirschner wire construct, and providing quantitative evidence to inform clinical decision-making. This study established a three-dimensional finite element model of a Delbet Type II PFNF based on CT data from the femur of a healthy 12-year-old male (50 kg). Nine configurations were compared: six screw constructs (2×4.5 mm through 3×6.5 mm), crossed K-wires, a hybrid screw–K-wire construct, and a proximal femoral locking plate (PFLP). Loading scenarios included double-leg stance (500 N), single-leg stance (1,000 N), and walking (1,500 N with 5 N·m torque). Peak stress, peak displacement, fracture gap relative displacement, fracture surface peak stress and construct stiffness were evaluated. Under walking load, the locking plate (Group I) showed the lowest gap displacement (0.174 mm) and highest stiffness (8,620.7 N/mm). Among screws, Group H (3×6.5 mm) had the lowest implant stress (181.58 MPa) and gap displacement (0.215 mm). The hybrid construct (Group D) reduced gap displacement by 28% versus the screw-only equivalent (Group B). K-wire fixation (Group A) produced implant stress of 1,330.9 MPa, exceeding titanium alloy yield strength. Groups C, F, G, H, and I maintained gap displacement below 0.5 mm under all loads. Screw diameter exerts biomechanical influence comparable to screw number in pediatric Delbet Type II fractures. Three large-diameter screws provide optimal stability when neck space permits; two large-diameter screws or a hybrid construct are viable alternatives. Crossed K-wires should be reserved for young, compliant patients with protected weight-bearing.

Keywords: pediatric femoral neck fracture; Delbet Type II; internal fixation; finite element analysis; biomechanic

Introduction

Pediatric femoral neck fractures (PFNFs) constitute a rare but clinically consequential injury, accounting for less than 1% of all pediatric fractures and approximately 0.3% to 0.5% of all hospital admissions for pediatric skeletal trauma^{1,2}. Despite their relative infrequency, these injuries carry disproportionately high morbidity. The peak incidence occurs between 10 and 13 years of age, with a reported male-to-female ratio of approximately 2.5:1^{2,3}. Most cases result from high-energy trauma such as motor vehicle accidents or falls from height, and associated ipsilateral injuries to the hip, pelvis, or femoral shaft are common⁴⁻⁶. The overall complication rate for PFNFs ranges from 20% to 50%, encompassing avascular necrosis (AVN), nonunion, coxa vara deformity, and premature physeal closure^{1,2,7-10}. Among the Delbet classification subtypes^{11,12}, Type II (transcervical) fractures are the most prevalent, constituting 40.3% of cases in a meta-analysis by Moon and Mehlman encompassing 360 patients¹³, with other series reporting frequencies of 40% to 50%. The transcervical location of Delbet Type II fractures places the fracture line directly across the tenuous epiphyseal blood supply, rendering this subtype particularly vulnerable to vascular compromise¹⁴.

The complication burden is particularly severe for Delbet Type II fractures. A 2023 meta-analysis by Xin et al. encompassing 1,185 cases reported an overall AVN rate of 22% across all PFNF subtypes¹⁵. For Delbet Type II specifically, AVN rates range from 25% to 35%^{13,15,16}, with individual series reporting 26.5%¹⁶, 28%¹³, and 32%¹⁵. Xin et al. further demonstrated that Delbet Type I and II fractures carry an odds ratio of 3.02 ($p < 0.001$) for AVN compared with Delbet Type III and IV fractures¹⁵. Given that AVN in skeletally immature patients can lead to femoral head collapse, progressive deformity, and debilitating secondary osteoarthritis, these data underscore the imperative for optimizing internal fixation strategies in this specific fracture subtype.

The choice of fixation construct for PFNFs remains controversial¹⁷. Treatment options include smooth Kirschner wires, cannulated cancellous screws, and pediatric proximal femoral locking plates^{18,19}. Each modality presents distinct biomechanical and biological trade-offs. Smooth K-wires are often employed in younger children or when anatomical constraints limit screw placement; There is also debate regarding the appropriate age for transphyseal fixation with screws. Transphyseal fixation with screws has been deemed acceptable in patients ≥ 10 years of age¹, while other studies indicate that screws should only be passed through the growth plate in patients aged 15 years or older, as this minimizes the risk of limb length discrepancy caused by premature growth arrest^{20,21}. The narrow dimensions of the pediatric femoral neck frequently restrict the surgeon's ability to place multiple large-diameter screws without violating cortical margins or compromising the blood supply²². Biomechanical studies have traditionally favored three-screw constructs for enhanced stability; however, clinical data reveal a counterintuitive paradox. Wang et al.²³ reported in a multicenter study of 153 cases that 2-screw fixation was associated with an AVN rate of 21.4% compared with

44.8% for 3-screw fixation ($P=0.027$). The same study identified that screw diameter exceeding 16.5% of femoral neck width increased AVN risk ($RR=1.455$, $P=0.004$), and that a bipolar diaphyseal angle percentage (BDAP%) less than 51.6% was similarly associated with elevated AVN risk²³. Two-screw constructs also offer operative advantages including reduced surgical time, decreased blood loss, and diminished vascular disruption to the lateral ascending cervical vessels²⁴⁻²⁶. Conversely, locking plates provide maximal initial stability^{27,28} but cannot generate interfragmentary compression and may restrict the micromotion thought to be beneficial for fracture healing^{12,19}. These clinical and biomechanical tensions highlight the need for systematic evidence to guide implant selection.

FEA offers a controlled, reproducible platform for evaluating internal fixation biomechanics, particularly in the pediatric population where cadaveric specimen availability is limited and ethical constraints restrict in vivo experimentation. FEA permits parametric variation of implant diameter, quantity, and configuration under standardized loading conditions, enabling quantitative comparison of construct behavior without the confounding variables of specimen age, bone density, or anatomical variation inherent to cadaveric studies. The methodology is especially advantageous for investigating pediatric injuries, as developing bone exhibits material properties and structural geometries distinct from adult tissue. Precedent studies have established the feasibility of this approach: Tucker et al.²⁹ conducted parametric FEA of proximal femur fracture fixation, and Yin et al.³⁰ performed pediatric-specific FEA comparing screw-only, screw-plus-K-wire, and plate fixation for Delbet Type II fractures. However, no prior study has systematically compared the isolated and combined biomechanical effects of screw diameter, screw number, and hybrid screw-K-wire configurations within a single unified modeling framework for this fracture subtype.

This study represents the first systematic parametric biomechanical evaluation of screw diameter versus screw number versus hybrid screw-K-wire configurations specifically for Delbet Type II PFNFs. By quantifying the relative biomechanical weights of implant diameter and quantity, and by validating the clinical hybrid construct concept, this investigation aims to provide evidence-based guidance for internal fixation selection in this challenging pediatric injury.

The objectives of this FEA were: (1) to biomechanically compare nine internal fixation configurations for Delbet Type II PFNFs under physiologically grounded loading conditions; (2) to quantify the relative influence of screw diameter versus screw number on construct stability; (3) to evaluate the biomechanical efficacy of a hybrid cannulated screw-K-wire construct as a modification for space-limited femoral necks; and (4) to provide quantitative evidence to inform clinical decision-making regarding implant selection.

Materials and Methods

Ethics Statement and Subject Selection

This study was approved by the Institutional Ethics Committee, and written informed consent was obtained from the patient's guardians. All CT data were fully de-identified before transfer to the research team. The geometric model was reconstructed from the left femur of a healthy 12-year-old male (height 150 cm, body weight 50 kg) with no history of skeletal disease or prior lower-limb trauma. The age of 12 years was selected because the incidence of pediatric femoral neck fractures peaks between 10 and 13 years, corresponding to the peripubertal growth spurt when proximal femoral bone mineral density is transiently reduced^{31,32}. Furthermore, the 12-year-old skeleton retains an open proximal femoral physis, a critical feature influencing both fracture pattern and fixation strategy in this age group^{22,33}. A single preadolescent model was therefore employed, with the understanding that inter-individual geometric variation represents an acknowledged limitation addressed in Section 2.10.

Model Construction and Fracture Definition

CT scanning of the left femur was performed with the subject supine; images were exported in DICOM format. A 3D reconstruction workflow was executed using Mimics 21.0 (Materialise, Leuven, Belgium) for segmentation, Geomagic 2017 (3D Systems, Morrisville, NC, USA) for surface smoothing, SolidWorks 2021 (Dassault Systemes, Velizy-Villacoublay, France) for solid modeling, and Ansys 2023 R1 (Ansys, Canonsburg, PA, USA) for finite element analysis^{19,34}. Cortical and cancellous compartments were segmented by threshold-based mask differentiation followed by manual verification^{35,36}. The proximal femoral physis was identified as a distinct 1.5–2.0 mm transitional zone between metaphyseal cancellous and epiphyseal subchondral bone³⁵⁻³⁷.

The fracture was defined as a Delbet Type II transcervical fracture with a Pauwels angle of 70 degrees, measured between the fracture line and the horizontal plane on the anteroposterior projection²⁹. A Pauwels angle of 70 degrees was selected to represent a worst-case unstable scenario, as angles exceeding 50 degrees generate high shear forces at the fracture interface³². The fracture line was generated by intersecting the intact femoral solid with an oblique cutting plane oriented through the center of the femoral neck (Fig. 1). A local Cartesian coordinate system was established at the femoral head center: the X-axis oriented medial–lateral, the Y-axis anterior–posterior, and the Z-axis superior–inferior. The cutting plane was inclined 70 degrees relative to the horizontal (X–Y) plane, rotated about the Y-axis to simulate the Pauwels shear orientation^{29,30}.

After virtual fracture creation, the proximal and distal fragments were re-approximated to simulate anatomic reduction with zero residual gap. Nine fixation constructs were then modeled (Fig. 2):

- **Group A:** Four 2.5 mm crossed K-wires, transphyseal, tips 5 mm subchondral.
- **Group B:** Two 4.5 mm cannulated screws (CSs), parallel, non-transphyseal.
- **Group C:** Three 4.5 mm CSs, inverted-triangle, non-transphyseal.
- **Group D:** Two 4.5 mm CSs plus two 2.0 mm K-wires; CSs non-transphyseal, K-wires transphyseal stopping 5 mm subchondral.
- **Group E:** Two 5.5 mm CSs, parallel (same arrangement as B).
- **Group F:** Three 5.5 mm CSs, inverted-triangle (same arrangement as C).
- **Group G:** Two 6.5 mm CSs, parallel (same arrangement as B).
- **Group H:** Three 6.5 mm CSs, inverted-triangle (same arrangement as C).
- **Group I:** Proximal femoral locking plate; proximal screws non-transphyseal, distal locking screws bicortical.

All implants were Ti-6Al-7Nb alloy. Trajectories were planned in the coronal and sagittal planes to avoid cortical perforation and maximize subchondral purchase^{34,37}.

D-Group Pre-Experimental Configuration Selection

Prior to finalizing the nine-group comparison, a pre-experimental screening determined the optimal K-wire positioning for the hybrid D-group construct. Three candidates were compared against the B-group baseline (two 4.5 mm screws alone):

- **D1:** Two 2.0mm K-wires placed anterior or posterior to the screw pair, parallel to the screws, transphyseal.
- **D2:** One 2.0mm K-wire placed posterosuperior to the screws, parallel to the screws, transphyseal, forming an inverted triangle with the screw pair.
- **D3:** Two K-wire centered within the each of the two 4.5 mm CSs, transphyseal, creating a crossed internal arrangement.

All candidates were subjected to identical loading (1,500 N plus 5 N·m torsion). D1 demonstrated the most favorable profile: lower peak screw stress, lower femoral displacement, and lower fracture gap displacement than the B-group baseline. D3 produced unacceptable stress concentration at the screw-K-wire junction. Compared with D1, D2 indicated load transfer inefficiency and did not demonstrate any significant biomechanical advantages over Group B. D1 was therefore selected as the definitive D-group construct.

Material Properties

All materials were modeled as homogeneous, isotropic, and linear elastic. Properties were assigned from pediatric-specific literature (Table 1). Cortical bone modulus (16,000 MPa, $\nu = 0.30$) was drawn from pediatric bending tests and ultrasound studies^{38,39}. Femoral head and neck cancellous moduli (800 MPa and 600 MPa, respectively) were adapted from pediatric proximal femoral trabecular bone data^{36,40}. The physis was assigned 6 MPa with $\nu = 0.49$ ^{35,36}; although layered properties

exist (reserve zone ~11 MPa, proliferative ~3.4 MPa, hypertrophic ~3.3 MPa) ⁴¹, a single representative modulus was adopted for computational tractability. All metallic implants were Ti-6Al-7Nb (110,000 MPa, $\nu = 0.33$) ⁴².

Mesh Generation and Convergence Analysis

Solid models were imported into Ansys Workbench and discretized using tetrahedral elements. Mesh sensitivity analysis confirmed convergence prior to full analysis. Mesh sizes were reduced stepwise decrements, and peak von Mises stress in the proximal femoral neck was recorded at each level. Convergence was defined as <5% change in peak stress between successive refinements. The convergence procedure ensured that reported values were not discretization artifacts. Mesh sizes were determined by stress gradients: femoral shaft and metaphysis at 3.0 mm; K-wires at 2.0 mm; CSs and locking screws at 1.5 mm; locking plate at 2.0 mm (Fig. 3). Element and node counts for each group are shown in Table 2.

Contact Definitions

Assume anatomical reduction of the fracture, with the fracture surfaces in contact. Refer to relevant literature to define the contact relationships between various materials (Table 3). The contact relationships between the femoral cortical bone and cancellous bone, the plate and locking screws, the locking screws and the femur, and the threaded portion of CSs and the femur are all defined as “bound”; The contact relationship between the plate and the femur is friction, with a friction coefficient of 0.2; the contact relationship between the fracture surfaces is friction, with a friction coefficient of 0.46 ²⁹; the contact relationship between the K-wire and the femur is friction, with a friction coefficient of 0.1 ³⁰; and the contact relationship between the smooth shaft of CSs and the femur is friction, with a friction coefficient of 0.3 ⁴³.

Boundary Conditions

The distal femur was fully constrained to simulate distal limb fixation. All nodes on the distal femoral cutting plane (120 mm distal to the lesser trochanter) were fixed in all six degrees of freedom (translations and rotations in X, Y, and Z). This boundary condition approximates the mechanical constraint of the intact distal limb during single-limb loading and is consistent with published proximal femoral finite element studies ^{30,44}. No constraint was applied to the femoral head articular surface.

Loading Conditions

Four loading scenarios were applied sequentially to simulate progressively demanding physiological conditions (Table 4, Fig. 4). All forces were applied as static nodal loads distributed over the center of the femoral head weight-bearing zone.

Load magnitudes were established as peak physiological estimates derived from the subject's body weight ($BW = 50 \text{ kg} \times 9.81 \text{ m/s}^2 = 490.5 \text{ N}$)⁴⁵⁻⁴⁷. The double-leg stance load (500 N, $\sim 1.0 \times BW$) represents the vertical hip contact force during quiet standing, with approximately 50% of body weight transmitted through each hip^{48,49}. The single-leg stance load (1,000 N, $\sim 2.0 \times BW$) represents static single-limb support^{44,48}. The walking load (1,500 N, $\sim 3.1 \times BW$) represents the upper range of pediatric hip contact forces during gait ($2.5\text{--}3.5 \times BW$ for children of comparable mass)^{48,50}. This load was decomposed into a vector with 10° adduction in the coronal plane (hip abductor pull) and 9° posterior inclination in the sagittal plane (late stance-phase extension)^{50,51}.

A torsional load of 5 N·m was applied along the femoral long axis concurrently with each compressive load. This torque represents a simplified static estimate of muscle-induced twisting during gait. While in vivo telemetric data report torsional moments of 1.72–1.91% BW·m for walking (scaling to $\sim 7\text{--}14 \text{ N}\cdot\text{m}$ for a 50-kg subject)⁴⁸, the 5 N·m value was adopted as a conservative, reproducible baseline that maintains consistency with the computed results dataset. Components were applied in the established local coordinate system: positive X = lateral, positive Y = posterior, positive Z = superior.

Observation Metrics

The following output parameters were extracted for each construct under each loading scenario:

1. **Peak von Mises stress** (MPa): Maximum equivalent stress in femoral bone and each implant component.
2. **Peak displacement** (mm): Maximum resultant nodal displacement magnitude in femoral bone and each implant component.
3. **Fracture gap relative displacement** (mm): The maximum nodal displacement vector difference between proximal and distal fracture surfaces, extracted using the MAX and Probe functions in Ansys.
4. **Fracture surface peak stress** (MPa): Maximum equivalent stress in fracture surface.
5. **Construct stiffness** (N/mm): Defined as applied force divided by fracture gap displacement ($\text{Stiffness} = F / \Delta \text{gap}$), providing a normalized index of fixation rigidity.

Study Assumptions and Limitations

The following assumptions were explicitly adopted:

1. **Anatomic reduction**: The fracture was reduced with perfect fragment opposition and zero residual gap. In clinical practice, imperfect reduction and interposed soft tissue may alter contact mechanics.
2. **Material linearity and homogeneity**: All materials were modeled as homogeneous, isotropic, and linear elastic. Real bone is anisotropic,

heterogeneous, and viscoelastic, with ongoing mineralization gradients not captured by single-value moduli.

3. **Static analysis:** Loading was applied statically; dynamic effects (impact, fatigue, creep) were not modeled. Scenarios represent peak instantaneous forces rather than full gait cycles.
4. **Absence of soft tissues:** Muscles, ligaments, joint capsule, and periosteum were excluded. These structures provide substantial in vivo load sharing not represented here.
5. **Single-subject representation:** One 12-year-old model was used; inter-individual variation in geometry, bone mineral density, and physal maturity among children aged 10–13 years is not captured.
6. **Simplified torsion:** The 5 N·m torque is a static estimate; actual muscle-induced torsion during gait is time-varying and directionally complex.

These assumptions are standard in comparative orthopedic finite element studies and were applied consistently across all nine groups, preserving inter-group comparative validity while acknowledging the gap between computational idealization and clinical reality^{30,34,37}.

Results

Implant Peak Stress

Peak von Mises stress in each fixation implant was recorded under the three loading conditions. Stress distributions are presented in Fig. 5-7, and numerical values are summarised in Table 5.

Under the walking load (1500 N), the peak stresses for Groups A and B were 1330.9 MPa and 524.65 MPa, respectively, both exceeding the yield stress of Ti-6Al-7Nb (450 MPa). The remaining groups recorded values below this threshold: Group D (381.7 MPa), Group E (345.37 MPa), Group C (341.27 MPa), Group G (292.59 MPa), Group I (283.42 MPa), Group F (246.83 MPa), and Group H (181.58 MPa).

Under the single-leg stance load (1000 N), the values ranked from highest to lowest as follows: Group A (479.34 MPa), Group B (269.99 MPa), Group D (228.52 MPa), Group F (212.91 MPa), Group C (195.95 MPa), Group E (181.93 MPa), Group I (132.59 MPa), Group G (123.62 MPa), and Group H (87.42 MPa).

Under the double-leg stance load (500 N), the values were: Group A (263.09 MPa), Group B (173.89 MPa), Group D (163.94 MPa), Group C (157.71 MPa), Group E (128.84 MPa), Group F (111.78 MPa), Group G (73.962 MPa), Group I (69.013 MPa), and Group H (44.573 MPa).

In all groups, peak implant stress was located in the vicinity of the fracture line, at the implant–femoral neck cortex interface. For Group D (2 CSs + 2 K-wires), the reported value corresponds to the screw component.

Implant Peak Displacement

Peak implant nodal displacement was recorded under the three loading conditions (Fig. 8, Fig. 9, Fig. 10, Table 7).

Under the walking load (1500 N), the values ranked as follows: Group A (6.522 mm), Group B (3.808 mm), Group D (3.692 mm), Group E (3.677 mm), Group C (3.614 mm), Group F (3.538 mm), Group H (3.514 mm), Group G (3.443 mm), and Group I (3.07 mm).

Under the single-leg stance load (1000 N), the values were: Group A (2.375 mm), Group B (1.21 mm), Group D (1.158 mm), Group C (1.152 mm), Group E (1.141 mm), Group H (1.117 mm), Group F (1.104 mm), Group G (1.095 mm), and Group I (0.843 mm).

Under the double-leg stance load (500 N), the values ranked as follows: Group A (1.242 mm), Group C (0.58 mm), Group B (0.578 mm), Group D (0.567 mm), Group H (0.565 mm), Group E (0.559 mm), Group G (0.549 mm), Group F (0.543 mm), and Group I (0.429 mm).

Peak implant displacement was located at the proximal portion of the implant, near the entry point into the femoral head.

Femoral Peak Stress

Peak femoral stress was recorded under the three loading conditions (Fig. 11, Fig. 12, Fig. 13, Table 6).

Under the walking load (1500 N), the values ranked as follows: Group A (163.76 MPa), Group G (108.4 MPa), Group C (103.12 MPa), Group H (103.09 MPa), Group E (102.64 MPa), Group F (100.29 MPa), Group D (100.13 MPa), Group I (97.064 MPa), and Group B (95.95 MPa).

Under the single-leg stance load (1000 N), the values were: Group A (81.187 MPa), Group F (63.521 MPa), Group G (57.202 MPa), Group E (48.15 MPa), Group B (44.253 MPa), Group I (43.892 MPa), Group D (38.36 MPa), Group C (32.692 MPa), and Group H (31.432 MPa).

Under the double-leg stance load (500 N), the values ranked as follows: Group A (42.93 MPa), Group B (41.877 MPa), Group F (35.417 MPa), Group D (34.129 MPa), Group G (33.243 MPa), Group C (28.76 MPa), Group I (28.018 MPa), Group E (27.668 MPa), and Group H (25.034 MPa).

Peak femoral stress was located at the medial inferior aspect of the femoral neck, adjacent to the calcar region.

Femoral Peak Displacement

Peak femoral nodal displacement was recorded under the three loading conditions (Fig. 14, Fig. 15, Fig. 16, Table 7).

Under the walking load (1500 N), the values ranked as follows: Group A (7.187 mm), Group B (5.821 mm), Group E (5.569 mm), Group D (5.37 mm), Group G (5.168 mm), Group C (5.155 mm), Group F (5.129 mm), Group H (5.059 mm), and Group I (4.491 mm).

Under the single-leg stance load (1000 N), the values were: Group A (2.619 mm), Group B (2.004 mm), Group D (1.844 mm), Group E (1.827 mm), Group G (1.728 mm), Group C (1.701 mm), Group H (1.673 mm), Group F (1.67 mm), and Group I (1.308 mm).

Under the double-leg stance load (500 N), the values ranked as follows: Group A (1.377 mm), Group B (1.006 mm), Group D (0.96 mm), Group E (0.931 mm), Group G (0.91 mm), Group C (0.886 mm), Group H (0.879 mm), Group F (0.863 mm), and Group I (0.703 mm).

Peak femoral displacement was located at the femoral head.

Fracture Gap Relative Displacement

Fracture gap relative displacement was calculated as the maximum nodal displacement vector difference between the proximal and distal fracture surfaces, extracted using the MAX and Probe functions in the post-processing environment (Fig. 17, Table 8).

Under the walking load (1500 N), the values ranked as follows: Group A (1.232 mm), Group B (0.708 mm), Group D (0.511 mm), Group E (0.499 mm), Group G (0.281 mm), Group F (0.264 mm), Group C (0.226 mm), Group H (0.215 mm), and Group I (0.174 mm).

Under the single-leg stance load (1000 N), the values were: Group A (0.591 mm), Group B (0.332 mm), Group D (0.246 mm), Group E (0.21 mm), Group G (0.155 mm), Group F (0.121 mm), Group C (0.114 mm), Group H (0.111 mm), and Group I (0.083 mm).

Under the double-leg stance load (500 N), the values ranked as follows: Group A (0.329 mm), Group B (0.154 mm), Group D (0.131 mm), Group E (0.098 mm), Group G (0.084 mm), Group C (0.07 mm), Group F (0.067 mm), Group H (0.064 mm), and Group I (0.055 mm).

Peak fracture gap displacement was located at the posterosuperior margin of the fracture interface, and decreased sequentially from posterosuperior to anteroinferior.

Fracture Surface Peak Stress

Peak fracture surface stress was recorded under the three loading conditions (Fig. 18, Table 9).

Under the walking load (1500 N), the values ranked as follows: Group A (160.89 MPa), Group B (92.007 MPa), Group D (62.81 MPa), Group G (62.13 MPa), Group E (61.148 MPa), Group C (59.438 MPa), Group I (54.689 MPa), Group F (52.771 MPa), and Group H (45.999 MPa).

Under the single-leg stance load (1000 N), the values were: Group A (72.731 MPa), Group B (36.234 MPa), Group G (34.897 MPa), Group E (31.652 MPa), Group D (31.215 MPa), Group F (28.789 MPa), Group C (26.052 MPa), Group I (24.369 MPa), and Group H (22.383 MPa).

Under the double-leg stance load (500 N), the values ranked as follows: Group A (41.19 MPa), Group E (26.008 MPa), Group F (25.737 MPa), Group G (20.552 MPa), Group B (18.324 MPa), Group D (16.447 MPa), Group C (14.834 MPa), Group I (12.65 MPa), and Group H (11.866 MPa).

Construct Stiffness

Construct stiffness was calculated as the ratio of applied force to fracture gap relative displacement ($\text{Stiffness} = \text{Applied Force} / \text{Fracture Gap Displacement}$) for each group under each loading condition (Table 10).

Under the walking load (1500 N), the values ranked as follows: Group I (8,620.7 N/mm), Group H (6,976.7 N/mm), Group C (6,637.2 N/mm), Group F (5,681.8 N/mm), Group G (5,338.1 N/mm), Group E (3,006.0 N/mm), Group D (2,935.4 N/mm), Group B (2,118.6 N/mm), and Group A (1,217.5 N/mm).

Under the single-leg stance load (1000 N), the values were: Group I (12,048.2 N/mm), Group H (9,009.0 N/mm), Group C (8,772.0 N/mm), Group F (8,264.5 N/mm), Group G (6,451.6 N/mm), Group E (4,761.9 N/mm), Group D (4,065.0 N/mm), Group B (3,012.0 N/mm), and Group A (1,692.0 N/mm).

Under the double-leg stance load (500 N), the values ranked as follows: Group I (9,090.9 N/mm), Group H (7,812.5 N/mm), Group F (7,462.7 N/mm), Group C (7,142.9 N/mm), Group G (5,952.4 N/mm), Group E (5,102.0 N/mm), Group D (3,816.8 N/mm), Group B (3,246.8 N/mm), and Group A (1,519.8 N/mm).

Summary Statement

As this study was based on a single finite element model for each configuration, no statistical tests were performed. All reported values represent the output of individual finite element analyses and are presented as measured without inferential statistical comparison.

Discussion

This finite element analysis evaluated nine fixation constructs for Delbet Type II PFNF. Three principal findings emerged. First, larger-diameter screws and three CSs constructs generally reduced implant stress and fracture gap displacement compared with smaller-diameter or two CSs alternatives. Second, adding two supplementary K-wires to a two-screw construct (Group D) improved mechanical stability relative to screws-only fixation of equivalent diameter (Group B), suggesting a potential role for hybrid fixation. Third, constructs with walking-state gap displacement exceeding 0.5 mm (Groups A, B, D, and E) may require protected weight-bearing to optimise the mechanical environment for healing. These findings are consistent with the broader literature and support the hypothesis that construct selection should be individualised based on fracture stability, femoral neck anatomy, and anticipated compliance.

The von Mises stress distribution within implants is a critical predictor of mechanical failure in pediatric patients who may experience high cycle counts during healing. The results indicate that K-wire fixation (Group A) generated stresses substantially exceeding the yield strength of stainless steel under both single-leg stance and walking loads, suggesting a high probability of plastic deformation or fatigue failure. The two-screw 4.5 mm construct (Group B) also exceeded yield stress (450MPa⁵²) during simulated walking, raising concern regarding construct durability when compliance with activity restriction is variable. All remaining groups maintained stresses below the yield limit, consistent with expected mechanical performance in adolescent bone²⁹. These findings support the hypothesis that protected weight-bearing is essential when K-wire or small-diameter dual-screw constructs are selected. Clinical studies have reported hardware failure as a recognised complication in noncompliant patients or those with unstable fracture patterns¹³. For Groups A and B, a spica cast or strict partial-weight-bearing regimen may be warranted until radiographic healing is evident.

When the femoral stress of children exceeds 100 MPa, bone resorption will outpace bone formation, ultimately resulting in the failure of the internal fixation^{53,54}. The locking plate construct (Group I) produced the lowest femoral displacement and gap motion, reflecting superior stiffness. However, this rigidity may predispose to stress shielding adjacent to the plate–bone interface, with potential consequences for remodelling¹⁵. Conversely, Group A (K-wire) generated the highest femoral stress concentrations, which may increase the risk of localised bone resorption if sustained micromotion exceeds the threshold for callus preservation⁵². An optimal mechanical balance appears to be achieved by constructs maintaining fracture gap motion below the critical threshold for bone resorption while distributing stress sufficiently to avoid excessive shielding. Groups C (three 4.5 mm screws), F (three 5.5 mm screws), and H (three 6.5 mm screws) exhibited moderate femoral stress levels combined with gap displacements under 0.5 mm across all loading conditions. These constructs may represent a favourable compromise between mechanical stability and biological preservation, although this

requires prospective clinical–biomechanical correlation. In addition, the femoral peak stresses (Groups A–I) were all approached or exceeded the reference threshold of 100 MPa under walking loading, therefore, we recommend avoiding walking and high-intensity functional exercises until the fracture healing.

Fracture healing is fundamentally dependent on the mechanical environment within the interfragmentary gap. Perren’s strain theory posits that interfragmentary strain determines the differentiation pathway of repair tissue, with low strain favouring direct bone formation and excessive strain promoting fibrous nonunion⁵⁵. Animal studies have demonstrated that gap displacements below 0.5 mm are conducive to callus formation, whereas motion exceeding 1.0 mm may lead to bone resorption and persistent nonunion^{56,57}. These thresholds have been established in experiment and are widely applied in fixation assessment⁵⁶.

In this analysis, five groups (C, F, G, H, and I) maintained gap displacement below 0.5 mm under all loading conditions, suggesting a mechanically favourable environment for secondary healing without prolonged activity restriction. However, four groups (A, B, D, and E) demonstrated walking-state gap displacements exceeding or approaching 0.5 mm, with Group A (1.232mm) exceeding the threshold associated with bone resorption. For these constructs, protected weight-bearing may be necessary to maintain interfragmentary strain within the optimal range. The clinical significance is underscored by the documented association between excessive fracture site motion and nonunion in PFNFs, a complication occurring in 5 %–30 % of cases depending on displacement and treatment modality^{13,58}.

A central question is whether screw diameter or screw number exerts greater influence on mechanical stability. The present results suggest that this relationship is conditional, and construct selection must integrate biomechanical performance with biological considerations of vascular preservation. When comparing constructs of comparable total implant volume, larger-diameter screws appeared to confer modest advantages. Group G (two 6.5 mm screws) and Group F (three 5.5 mm screws) demonstrated similar stiffness (5,338.1 N/mm versus 5,681.8 N/mm) and walking-state gap displacement (0.281 mm versus 0.264 mm), with Group F producing only marginally superior parameters. However, Group F exhibited lower peak implant stress (246.83 MPa versus 292.59 MPa), suggesting that load distribution across three screws may reduce individual implant overload. In contrast, when comparing smaller-diameter alternatives, Group C (three 4.5 mm screws) markedly outperformed Group E (two 5.5 mm screws) in stiffness (6,637.2 N/mm versus 3,006.0 N/mm) and gap control (0.226 mm versus 0.499 mm). These comparisons indicate that the relative importance of diameter versus number depends on absolute implant size and the available cross-sectional area of the femoral neck.

This biomechanical analysis must be interpreted alongside clinical evidence regarding avascular necrosis. Wang et al.²³ reported a paradoxical finding in 153 pediatric patients: two-screw fixation was associated with a lower AVN rate (21.4

525 %) than three-screw fixation (44.8 %; $P = 0.027$). The authors attributed this to
526 increased vascular trauma from a third screw, which disrupts the tenuous blood
527 supply to the proximal femoral epiphysis—particularly the lateral retinacular and
528 intraosseous cervical vessels⁵⁸. Wang et al.²³ further identified that screws
529 exceeding 16.5 % of femoral neck width independently increased AVN risk ($RR =$
530 1.455 ; $P = 0.004$), underscoring the mechanical–biological trade-off in large-
531 diameter fixation. Resolution of this paradox requires individualisation. In femoral
532 necks with adequate diameter and length, three screws may be preferred for
533 stability, provided screw size is constrained to avoid excessive vascular violation. In
534 narrow necks, or when the fracture extends proximally toward the physis, two
535 larger-diameter screws with protected weight-bearing may be preferable. The
536 present data support the hypothesis that two large-diameter screws can achieve
537 acceptable stability under controlled loading, although this strategy demands
538 greater patient compliance with activity restriction.

539 The hybrid construct of two 4.5 mm screws and two supplementary K-wires
540 (Group D) warrants specific consideration. Relative to the screws-only equivalent
541 (Group B), the K-wire reduced walking-state gap displacement by approximately 28
542 % and peak CSs stress by approximately 27 %. This improvement is attributable to
543 the K-wire acting as a tension-band element against varus displacement—a loading
544 mode particularly relevant during single-leg stance. The clinical indication for
545 hybrid fixation arises when femoral neck anatomy is insufficient for three screws, or
546 when the proximal fracture line extends into the subcapital region, precluding safe
547 placement of a third screw distal to the physis. In such cases, transphyseal K-wire
548 placement—while not without risk—may be biomechanically preferable to an
549 inadequately spaced third screw or a two-screw construct with excessive
550 micromotion. These results are consistent with Kenmegne et al.⁵⁹, who reported
551 satisfactory outcomes with hybrid fixation in complex pediatric proximal femur
552 injuries where screw-only constructs were anatomically precluded.

553 The mechanical trends identified are consistent with published finite element
554 analyses. Yin et al.³⁰ modelled an 8-year-old child under 600 N single-leg load and
555 reported interfragmentary displacements of 1.64 mm for CSs, 1.18 mm for screws
556 with K-wire, and 1.11 mm for a lateral plate. Although absolute magnitudes in this
557 analysis differ due to the higher walking load (1,500 N, approximately $3.1 \times$ body
558 weight), the relative construct ranking is preserved—hybrid superior to screws-
559 only, and three screws constructs superior to two screws fixation. This consistency
560 supports the hypothesis that the mechanical hierarchy is robust rather than
561 artefactual. Tucker et al.²⁹ demonstrated parametrically that increasing implant
562 diameter reduces interfragmentary displacement nonlinearly, with diminishing
563 returns beyond a threshold determined by bone geometry. The present findings are
564 consistent with this relationship, and the systematic variation of implant diameter
565 and number within a single anatomical model follows the approach described by
566 Tucker et al.²⁹.

Integration of biomechanical findings with clinical evidence permits formulation of a provisional decision algorithm. The following hierarchy is proposed as a guide, acknowledging that individual factors—including age, bone quality, fracture pattern, and compliance—must modify these recommendations. First, for severely displaced, unstable fractures with compromised posteromedial cortical integrity, or in patients with neuromuscular conditions precluding protected weight-bearing, a proximal femoral locking plate (Group I) provides the greatest mechanical stability. Second, for patients with adequate femoral neck diameter, three large-diameter screws (Group H) offer an optimal balance of stability and minimally invasive fixation. Third, when neck anatomy is limited or the fracture line extends proximally, hybrid fixation (Group D) may improve mechanical performance relative to a two-screw construct alone, accepting the biological compromise of transphyseal hardware. Fourth, in young, compliant patients with minimally displaced fractures and strong posteromedial support, crossed K-wires with spica cast immobilisation (Group A) remain biomechanically acceptable provided strict non-weight-bearing is maintained. Fifth, when choosing between three small versus two large screws, the decision should integrate biomechanical findings with the AVN data of Wang et al.²³: in wide femoral necks where screw diameter can remain below 16.5 % of neck width, three screws may be preferable; in narrow necks, two larger screws with stringent postoperative protection may represent the safer biological compromise.

Limitations

This analysis is subject to several methodological limitations. First, a single geometric model representing a 12-year-old child was employed. Pediatric proximal femoral anatomy varies dynamically across the age spectrum, with substantial changes in neck-shaft angle, anteversion, physeal width, cortical thickness, and cancellous density⁵⁷. The findings may not generalise to younger children with more cartilaginous skeletons or adolescents approaching physeal closure. Second, static loading conditions do not capture the dynamic nature of physiological gait. The bimodal force curve of walking, with distinct loading and push-off peaks, as well as viscoelastic behaviour of bone and soft tissues, were not represented¹⁶. The use of a single resultant force vector and simplified 5 N·m torque neglects distributed muscle contributions to hip joint loading. Third, bone was modelled as isotropic and linear elastic. Cortical and cancellous bone exhibit pronounced anisotropy, and the physis is highly anisotropic with longitudinal stiffness approximately ten-fold greater than transverse stiffness^{21,44}. This simplification may have altered stress distributions near the growth plate, particularly for transphyseal implant paths. Fourth, no soft tissue structures were included. The hip joint capsule, ligamentum teres, periarticular musculature, and retinacular vessels contribute to both mechanical stability and biological perfusion^{3,58}. Their exclusion means the model may underestimate stabilising effects of intact soft tissues and cannot address the vascular mechanisms underlying AVN. Fifth, no patient-specific factors were

incorporated. Bone quality, fracture comminution, reduction accuracy, and posterior cortical defects—all of which influence in vivo mechanical behaviour—were held constant. Sixth, material parameters were derived from limited published data and may not accurately represent bone mineralisation in this age group²¹. These uncertainties underscore that computational findings should complement, rather than replace, clinical judgement.

Future Directions

Several avenues for future research may address these limitations. Multi-subject statistical shape modelling would enable generation of age-specific geometric variants, permitting population-level analysis of developmental morphological changes on construct performance. Dynamic simulations with time-varying gait loads and viscoelastic properties would more accurately replicate the cyclic mechanical environment during healing. Advanced constitutive models—including anisotropic and porous representations—would improve stress prediction fidelity near the physis and implant–bone interface.

Prospective, multi-center clinical studies with concurrent biomechanical correlation are needed to evaluate the thresholds proposed and determine whether computational differences translate into measurable differences in union rates, AVN incidence, and functional outcomes. Sensitivity analyses for material properties, loading magnitudes, and fracture gap geometry would quantify the robustness of findings to parameter uncertainty. Finally, musculoskeletal modelling approaches, such as those described by Heller et al., would permit subject-specific estimation of muscle force contributions to implant loading during rehabilitation¹⁶. These advances would collectively bridge the gap between computational biomechanics and individualised clinical decision-making in pediatric orthopedic trauma.

Conclusions

For Delbet Type II PFNFs, screw diameter exerts biomechanical influence comparable to screw number. When femoral neck diameter permits, three large-diameter screws (Group H) offer an optimal balance of stability and minimally invasive fixation. When neck space is constrained, two large-diameter screws (Groups E or G) represent a biomechanically acceptable alternative; the comparable stiffness and gap displacement between 2×6.5 mm and 3×5.5 mm constructs (5,338 vs. 5,682 N/mm; 0.281 vs. 0.264 mm) suggest that increasing diameter may compensate for reduced screw number.

when femoral necks anatomy is limited or the proximal fracture line extends into the subcapital region, the hybrid construct (Group D) improves stability over screw-only alternatives of equivalent diameter, which warranted specific consideration. Crossed K-wire fixation (Group A) is biomechanically inferior to

screw and plate constructs, and should be reserved for young, compliant patients with minimal displacement, managed with delayed weight-bearing and spica cast protection. PFLP (Group I) provide the highest construct stiffness and are recommended for severely displaced, unstable fractures where maximum immediate stability is required.

The selection among these options should integrate biomechanical performance with patient age, compliance, fracture displacement, and neck morphology.

Data Availability Statement

All data generated or analyzed during this study are included in this manuscript.

References

- 1 Patterson, J. T., Tangtiphaiboon, J. & Pandya, N. K. Management of Pediatric Femoral Neck Fracture. *Journal of the American Academy of Orthopaedic Surgeons*. **26**, 411-419 (2018). <https://doi.org/10.5435/jaaos-d-16-00362>
- 2 Boardman, M. J., Herman, M. J., Buck, B. & Pizzutillo, P. D. Hip Fractures in Children. *JAAOS - Journal of the American Academy of Orthopaedic Surgeons*. **17**, 162-173 (2009).
- 3 Spence, D. *et al.* Osteonecrosis After Femoral Neck Fractures in Children and Adolescents: Analysis of Risk Factors. *Journal of Pediatric Orthopaedics*. **36**, 111-116 (2016). <https://doi.org/10.1097/bpo.0000000000000424>
- 4 Leung, P. C. & Lam, S. F. Long-term follow-up of children with femoral neck fractures. *The Journal of Bone and Joint Surgery. British volume*. **68-B**, 537-540 <https://doi.org/10.1302/0301-620x.68b4.3733826>
- 5 Sanghavi, S., Patwardhan, S., Shyam, A., Nagda, T. & Naik, P. Nonunion in Pediatric Femoral Neck Fractures. *J Bone Joint Surg Am*. **102**, 1000-1010 (2020). <https://doi.org/10.2106/jbjs.19.01117>
- 6 Wang, W. T. *et al.* Risk factors for the development of avascular necrosis after femoral neck fractures in children: a review of 239 cases. *Bone Joint J*. **101-b**, 1160-1167 (2019). <https://doi.org/10.1302/0301-620x.101b9.Bjj-2019-0275.R1>
- 7 Bimmel, R., Bakker, A., Bosma, B. & Michielsen, J. Paediatric hip fractures: a systematic review of incidence, treatment options and complications. *Acta Orthop Belg*. **76**, 7-13
- 8 Yeranossian, M., Horneff, J. G., Baldwin, K. & Hosalkar, H. S. Factors affecting the outcome of fractures of the femoral neck in children and adolescents. *The Bone & Joint Journal*. **95-B**, 135-142 (2013). <https://doi.org/10.1302/0301-620x.95b1.30161>
- 9 Li, H., Zhao, L., Huang, L. & Kuo, K. N. Delayed Slipped Capital Femoral Epiphysis After Treatment of Femoral Neck Fracture in Children. *Clinical*

687 *orthopaedics and related research*. **473**, 2712-2717
688 <https://doi.org:10.1007/s11999-014-4067-y>

689 10 Kuo, F.-C., Kuo, S.-J. & Ko, J.-Y. Overgrowth of the femoral neck after hip
690 fractures in children. *Journal of orthopaedic surgery and research*. **11**, 50
691 <https://doi.org:10.1186/s13018-016-0387-9>

692 11 Sankar, W. N. & Mehlman, C. T. The Community Orthopaedic Surgeon Taking
693 Trauma Call: Pediatric Femoral Neck Fracture Pearls and Pitfalls. *J Orthop*
694 *Trauma*. **33 Suppl 8**, S22-s26 (2019).
695 <https://doi.org:10.1097/bot.0000000000001541>

696 12 Pinto, D. A. & Aroojis, A. Fractures of the Proximal Femur in Childhood: A
697 Review. *Indian Journal of Orthopaedics*. **55**, 23-34 (2020).
698 <https://doi.org:10.1007/s43465-020-00259-4>

699 13 Moon, E. S. & Mehlman, C. T. Risk Factors for Avascular Necrosis After
700 Femoral Neck Fractures in Children: 25 Cincinnati Cases and Meta-analysis of
701 360 Cases. *Journal of Orthopaedic Trauma*. **20**, 323-329 (2006).

702 14 Song, K. S. Displaced fracture of the femoral neck in children. *The Journal of*
703 *Bone and Joint Surgery. British volume*. **92-B**, 1148-1151 (2010).
704 <https://doi.org:10.1302/0301-620x.92b8.24482>

705 15 Xin, P., Li, Z., Pei, S., Shi, Q. & Xiao, L. The incidence and risk factors for
706 femoral head necrosis after femoral neck fracture in pediatric patients: a
707 systematic review and meta-analysis. *Journal of Orthopaedic Surgery and*
708 *Research*. **18** (2023). <https://doi.org:10.1186/s13018-023-03502-4>

709 16 Akar, M. S. Can Formation of Avascular Necrosis Really Be Prevented in
710 Delbet Type 2 Femoral Neck Fractures? *Turkish Journal of Trauma and*
711 *Emergency Surgery* (2020). <https://doi.org:10.14744/tjtes.2020.33568>

712 17 Akkari, M., Santili, C., Akel, E. & Angelim, R. Femoral neck fracture in children:
713 treatment and complications. *Revista da Associação Médica Brasileira*. **61**, 5-7
714 (2015). <https://doi.org:10.1590/1806-9282.61.01.005>

715 18 Eberl, R., Singer, G., Ferlic, P., Weinberg, A. M. & Hoellwarth, M. E. Post-
716 traumatic coxa vara in children following screw fixation of the femoral neck.
717 *Acta Orthopaedica*. **81**, 442-445 (2010).
718 <https://doi.org:10.3109/17453674.2010.501744>

719 19 Chen, S. Y. et al. Comparison of pediatric hip locking compression plate and
720 cannulated screws in the treatment of Delbet type II and III femoral neck
721 fractures in children [in Chinese]. *Zhonghua Xiao Er Wai Ke Za Zhi*. **37**, 898-
722 902 (2016).

723 20 Palocaren, T. Femoral Neck Fractures in Children: A Review. *Indian Journal of*
724 *Orthopaedics*. **52**, 501-506 (2018).
725 https://doi.org:10.4103/ortho.ijortho_404_17

726 21 Pandey, R. A. & John, B. Current controversies in management of fracture
727 neck femur in children: A review. *Journal of Clinical Orthopaedics and*
728 *Trauma*. **11**, S799-S806 (2020). <https://doi.org:10.1016/j.jcot.2020.05.029>

729 22 Gao, L. Y. et al. Biomechanical analysis of modified cannulated screw system
730 for Pauwels type III femoral neck fractures in children [in Chinese]. *Zhongguo*
731 *Xian Dai Yi Sheng*. **59**, 77-81 (2021).

- 732 23 Wang, W. *et al.* Effect of the Number, Size, and Location of Cannulated Screws
733 on the Incidence of Avascular Necrosis of the Femoral Head in Pediatric
734 Femoral Neck Fractures: A Review of 153 Cases. *Journal of Pediatric*
735 *Orthopaedics*. **42**, 149-157 (2022).
736 <https://doi.org/10.1097/bpo.0000000000002018>
- 737 24 Wang, W. *et al.* Do the Number, Size, and Position of Partially Threaded
738 Screws Affect the Radiological Healing of Surgically Treated Displaced
739 Femoral Neck Fractures? A Review of 136 Children. *Medicina*. **58** (2022).
740 <https://doi.org/10.3390/medicina58091153>
- 741 25 Pavone, V. *et al.* Surgical treatment with cannulated screws for pediatric
742 femoral neck fractures: A case series. *Injury*. **50**, S40-S44 (2019).
743 <https://doi.org/10.1016/j.injury.2019.01.043>
- 744 26 Shrader, M. W., Jacofsky, D. J., Stans, A. A., Shaughnessy, W. J. & Haidukewych,
745 G. J. Femoral Neck Fractures in Pediatric Patients. *Clinical Orthopaedics and*
746 *Related Research*. **454**, 169-173 (2007).
747 <https://doi.org/10.1097/01.blo.0000238794.82466.3d>
- 748 27 Gurnea, T. P., Nielsen, R. C. & Althausen, P. L. Use of Proximal Humerus
749 Locking Plates for Fixation of Pediatric Femoral Neck Fractures: Technical
750 Trick. *Journal of Orthopaedic Trauma*. **34**, E312-E315 (2020).
751 <https://doi.org/10.1097/bot.0000000000001765>
- 752 28 Duffy, S., Gelfer, Y., Trompeter, A., Clarke, A. & Monsell, F. The clinical
753 features, management options and complications of paediatric femoral
754 fractures. *European Journal of Orthopaedic Surgery & Traumatology*. **31**, 883-
755 892 (2021). <https://doi.org/10.1007/s00590-021-02933-1>
- 756 29 Tucker, S. M., Wee, H., Fox, E., Reid, J. S. & Lewis, G. S. Parametric Finite
757 Element Analysis of Intramedullary Nail Fixation of Proximal Femur
758 Fractures. *J Orthop Res*. **37**, 2358-2366 <https://doi.org/10.1002/jor.24401>
- 759 30 Yin, W., Ouyang, Z., Zhou, J., Yang, B. H. & W, Y. K. Finite element analysis of
760 different internal fixation methods for Delbet type II femoral neck fractures
761 in children [in Chinese]. *Zhongguo Zu Zhi Gong Cheng Yan Jiu*. **27**, 3476-3480
762 (2023).
- 763 31 Shi, F. *et al.* Experimental study on the effect of different diameters of physeal
764 penetrating injuries on epiphyseal development [in Chinese]. *Zhonghua Xiao*
765 *Er Wai Ke Za Zhi*. 53-55 (2002).
- 766 32 Chen, J. C., Chen, S. Y. & Pan, Y. C. Study on the effect of screw insertion depth
767 on the treatment of Delbet type II and III femoral neck fractures in children
768 [in Chinese]. *Lin Chuang Xiao Er Wai Ke Za Zhi*. **19**, 573-578 (2020).
- 769 33 Ding, Q., Liu, Y. F., Su, Y. & Wang, X. H. Kirschner wire and cannulated screw
770 internal fixation for Delbet type II and III femoral neck fractures in children
771 [in Chinese]. *Zhong Yi Zheng Gu*. **32**, 69-71 (2020).
- 772 34 Shen, M., Wang, C., Chen, H., Rui, Y.-f. & Zhao, S. An update on the Pauwels
773 classification. *Journal of Orthopaedic Surgery and Research*. **11**, 161 (2016).
774 <https://doi.org/10.1186/s13018-016-0498-3>
- 775 35 Yadav, P., Shefelbine, S. J. & Gutierrez-Farewik, E. M. Effect of growth plate
776 geometry and growth direction on prediction of proximal femoral

777 morphology. *J Biomech.* **49**, 1613-1619 (2016).
778 <https://doi.org:10.1016/j.jbiomech.2016.03.039>
779 36 Carriero, A., Jonkers, I. & Shefelbine, S. J. Mechanobiological prediction of
780 proximal femoral deformities in children with cerebral palsy. *Computer*
781 *Methods in Biomechanics and Biomedical Engineering.* **14**, 253-262 (2011).
782 <https://doi.org:10.1080/10255841003682505>
783 37 Weng, P.-W. *et al.* The effects of tibia profile, distraction angle, and knee load
784 on wedge instability and hinge fracture: A finite element study. *Med Eng*
785 *Phys.* **42**, 48-54 <https://doi.org:10.1016/j.medengphy.2017.01.007>
786 38 Semaan, M., Karam, E., Baron, C. & Pithioux, M. Estimation of the elastic
787 modulus of child cortical bone specimens via microindentation. *Connective*
788 *Tissue Research.* **60**, 399-405 (2019).
789 <https://doi.org:10.1080/03008207.2019.1570170>
790 39 Berteau, J.-P. *et al.* In vitro ultrasonic and mechanic characterization of the
791 modulus of elasticity of children cortical bone. *Ultrasonics.* **54**, 1270-1276
792 (2014). <https://doi.org:10.1016/j.ultras.2013.09.014>
793 40 Kainz, H. *et al.* The influence of maximum isometric muscle force scaling on
794 estimated muscle forces from musculoskeletal models of children with
795 cerebral palsy. *Gait & Posture.* **65**, 213-220 (2018).
796 <https://doi.org:10.1016/j.gaitpost.2018.07.172>
797 41 Sergerie, K., Lacoursière, M.-O., Lévesque, M. & Villemure, I. Mechanical
798 properties of the porcine growth plate and its three zones from unconfined
799 compression tests. *J Biomech.* **42**, 510-516 (2009).
800 <https://doi.org:10.1016/j.jbiomech.2008.11.026>
801 42 Kobayashi, E., Wang, T. J., Doi, H., Yoneyama, T. & Hamanaka, H. Mechanical
802 properties and corrosion resistance of Ti-6Al-7Nb alloy dental castings. *J*
803 *Mater Sci Mater Med.* **9**, 567-574 <https://doi.org:10.1023/a:1008909408948>
804 43 Sommers, M. B., Fitzpatrick, D. C., Madey, S. M., Vande Zanderschulp, C. &
805 Bottlang, M. A surrogate long-bone model with osteoporotic material
806 properties for biomechanical testing of fracture implants. *J Biomech.* **40**,
807 3297-3304 <https://doi.org:10.1016/j.jbiomech.2007.04.024>
808 44 Wang, G., Tang, Y., Wu, X. & Yang, H. Finite element analysis of a new plate for
809 Pauwels type III femoral neck fractures. *J Int Med Res.* **48**,
810 300060520903669 <https://doi.org:10.1177/0300060520903669>
811 45 Cui, Y. *et al.* Characterization of novel intramedullary nailing method for
812 treating femoral shaft fracture through finite element analysis. *Exp Ther Med.*
813 **20**, 748-753 <https://doi.org:10.3892/etm.2020.8763>
814 46 Wang, J. *et al.* Comparative finite element analysis of three implants fixing
815 stable and unstable subtrochanteric femoral fractures: Proximal Femoral
816 Nail Antirotation (PFNA), Proximal Femoral Locking Plate (PFLP), and
817 Reverse Less Invasive Stabilization System (LISS). *Orthop Traumatol Surg*
818 *Res.* **106**, 95-101 <https://doi.org:10.1016/j.otsr.2019.04.027>
819 47 Huang, B.-X., Zhan, S.-Z., Yang, M. & Zhang, D.-Y. Biomechanical effects of
820 internal fixation with self-lock compression anti-rotation blade for Pauwels

type III femoral neck fractures: a comparative finite element analysis. *BMC Musculoskelet Disord.* **24**, 292 <https://doi.org/10.1186/s12891-023-06386-x>

48 Bergmann, G. *et al.* Hip contact forces and gait patterns from routine activities. *J Biomech.* **34**, 859-871 [https://doi.org/10.1016/s0021-9290\(01\)00040-9](https://doi.org/10.1016/s0021-9290(01)00040-9)

49 Mu, J.-X., Xiang, S.-Y., Ma, Q.-Y. & Gu, H.-L. Selection of internal fixation method for femoral intertrochanteric fractures using a finite element method. *World J Clin Cases.* **9**, 6343-6356 <https://doi.org/10.12998/wjcc.v9.i22.6343>

50 Heller, M. O. *et al.* Musculo-skeletal loading conditions at the hip during walking and stair climbing. *J Biomech.* **34**, 883-893 [https://doi.org/10.1016/s0021-9290\(01\)00039-2](https://doi.org/10.1016/s0021-9290(01)00039-2)

51 Nie, S.-B. *et al.* Medial support nail and proximal femoral nail antirotation in the treatment of reverse obliquity inter-trochanteric fractures (Arbeitsgemeinschaft für Osteosynthesfragen/Orthopedic Trauma Association 31-A3.1): a finite-element analysis. *Chinese Medical Journal.* **133**, 2682-2687 (2020). <https://doi.org/10.1097/cm9.0000000000001031>

52 Vajgel, A. *et al.* Comparative Finite Element Analysis of the Biomechanical Stability of 2.0 Fixation Plates in Atrophic Mandibular Fractures. *Journal of Oral and Maxillofacial Surgery.* **71**, 335-342 (2013). <https://doi.org/10.1016/j.joms.2012.09.019>

53 Meng, Y. *et al.* A finite element model of a six-year-old child for simulating pedestrian accidents. *Accident Analysis & Prevention.* **98**, 206-213 (2017). <https://doi.org/10.1016/j.aap.2016.10.002>

54 Wang, H. L., Li, H. Y., He, L. J. & Cui, S. H. Establishment of finite element model of femur in 6-year-old children and simulation analysis of femoral neck injury [in Chinese]. *Tianjin Ke Ji Da Xue Xue Bao.* **27**, 60-63 (2012). <https://doi.org/10.13364/j.issn.1672-6510.2012.03.008>

55 Perren, S. M. Evolution of the internal fixation of long bone fractures. The scientific basis of biological internal fixation: choosing a new balance between stability and biology. *J Bone Joint Surg Br.* **84**, 1093-1110 (2002). <https://doi.org/10.1302/0301-620x.84b8.13752>

56 Glatt, V., Evans, C. H. & Tetsworth, K. A Concert between Biology and Biomechanics: The Influence of the Mechanical Environment on Bone Healing. *Front Physiol.* **7**, 678 (2016). <https://doi.org/10.3389/fphys.2016.00678>

57 Kenwright, J. & Gardner, T. Mechanical influences on tibial fracture healing. *Clinical orthopaedics and related research*, S179-190 <https://doi.org/10.1097/00003086-199810001-00019>

58 Chung, S. The arterial supply of the developing proximal end of the human femur. *JBJS.* **58**, 961-970 (1976).

59 Kenmegne, G. R. *et al.* Postoperative clinical outcome and complications of combined cannulated cancellous screw with Kirschner wire in adolescent femoral neck fractures. *Frontiers in Pediatrics.* **11** (2023). <https://doi.org/10.3389/fped.2023.1169581>

866

867 **Authors' Contributions**

868 X.C. : Conceptualization, Methodology, Writing - Original Draft. Y.G. : Software. C.C.
869 and H.X. : Validation, Formal analysis. F.C. and Z.D. : Data Curation, Visualization.
870 G.H. : Writing - Review & Editing. G.W. : Supervision.

871 **Funding**

872 This study was not supported by any fundings.

873 **Ethics Approval**

874 This study was approved by the Institutional Ethics Committee at Weifang People's
875 Hospital. All CT data were fully de-identified before analysis in accordance with the
876 Declaration of Helsinki. Written informed consent was obtained from the patient's
877 guardians for the use of de-identified imaging data in research.

878 **Competing Interests**

879 The authors declare that they have no competing interests.

880 **Clinical Trial Number**

881 Clinical trial number: not applicable.

882 **Table Legends**

883 **Table 1. Material properties assigned in the finite element models**

Material	Elastic Modulus (MPa)	Poisson's Ratio	Source
Femoral cortical bone	16,000	0.30	Semaan et al. 2019 [38]; Berteau et al. 2014 [39]
Femoral head cancellous bone	800	0.30	Carriero et al. 2011 [36]; Kainz et al. 2020 [40]
Femoral neck cancellous bone	600	0.30	Carriero et al. 2011 [36]; Kainz et al. 2020 [40]

Material	Elastic Modulus (MPa)	Poisson's Ratio	Source
Proximal femoral physis	6	0.49	Yadav et al. 2016 [35] Carriero et al. 2011 [36];
Ti-6Al-7Nb alloy	110,000	0.33	Kobayashi et al. 1998 [42]

Table 2. Finite element mesh statistics for each fixation construct model.

Group	Nodes	Elements
A	96,884	55,991
B	107,017	62,683
C	120,205	70,040
D	114,323	66,464
E	109,388	64,059
F	121,794	71,118
G	110,163	64,377
H	122,403	71,215
I	116,168	67,229

Table 3. Contact definitions between model components

Contact Pair	Contact Type	Friction Coefficient
Cortical bone–Cancellous bone	Bonded	—
Plate–Femur	Frictional	0.2
Fracture surfaces (proximal–distal)	Frictional	0.46
K-wire–Femur	Frictional	0.1
Screw thread–Bone	Bonded	—
the smooth shaft of CSs–Bone	Frictional	0.3

Table 4. Loading conditions

Loading Scenario	Load Magnitude	Direction	Physiological Basis
Double-leg stance	500 N (~1.0×BW)	Vertical downward (-Z)	Shared standing load per hip [48]
Single-leg stance	1,000 N (~2.0×BW)	Vertical downward (-Z)	Static single-limb support [48]
Walking	1,500 N (~3.1×BW)	10° adduction (coronal), 9° extension (sagittal)	Peak hip contact force, late stance phase [48,50]
Torsion (Walking)	5 N·m	About femoral long axis (+/-Z)	Simplified muscle-induced twisting [48]

BW = body weight. Subject mass = 50 kg; BW = 50 × 9.81 = 490.5 N. The 1,500 N walking load represents the upper physiologic range for pediatric gait.

Table 5. Peak von Mises stress (MPa) recorded in implant components across all fixation constructs and loading conditions. The Ti-6Al-7Nb yield stress of 450 MPa is provided as a reference threshold. For Group D, the reported value corresponds to the screw component.

Group	Double-leg Stance (500 N)	Single-leg Stance (1,000 N)	Walking (1,500 N)
A	263.09	479.34	1,330.90
B	173.89	269.99	524.65
C	157.71	195.95	341.27
D	163.94	228.52	381.70
E	128.84	181.93	345.37
F	111.78	212.91	246.83
G	73.96	123.62	292.59
H	44.57	87.42	181.58
I	69.01	132.59	283.42

Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

Table 6. Peak von Mises stress (MPa) recorded in the femoral bone across all fixation constructs and loading conditions. A cortical bone reference stress of 100 MPa is provided as a comparative threshold.

Group	Double-leg Stance (500 N)	Single-leg Stance (1,000 N)	Walking (1,500 N)
A	42.93	81.19	163.76
B	41.88	44.25	95.95
C	28.76	32.69	103.12
D	34.13	38.36	100.13
E	27.67	48.15	102.64
F	35.42	63.52	100.29
G	33.24	57.20	108.40
H	25.03	31.43	103.09
I	28.02	43.89	97.06

Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

Table 7. Peak displacement (mm) of implant components and femoral bone across all fixation constructs and loading conditions. Implant displacement represents the maximum total deformation within the fixation hardware; femoral displacement represents the maximum total deformation of the entire femoral model. For Group D, the reported value of Implant displacement corresponds to the screw component.

Group	Implant Displacement (mm)			Femoral Displacement (mm)		
	Double-leg	Single-leg	Walking	Double-leg	Single-leg	Walking
A	1.242	2.375	6.522	1.377	2.619	7.187
B	0.578	1.210	3.808	1.006	2.004	5.821
C	0.580	1.152	3.614	0.886	1.701	5.155
D	0.567	1.158	3.692	0.960	1.844	5.370
E	0.559	1.141	3.677	0.931	1.827	5.569
F	0.543	1.104	3.538	0.863	1.670	5.129
G	0.549	1.095	3.443	0.910	1.728	5.168
H	0.565	1.117	3.514	0.879	1.673	5.059
I	0.429	0.843	3.070	0.703	1.308	4.491

Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

Table 8. Relative displacement (mm) across the fracture gap for all fixation constructs under three loading conditions. Values represent the maximum relative displacement between opposing fracture surfaces under double-leg stance (500 N), single-leg stance (1,000 N), and walking (1,500 N). The fracture gap healing threshold (0.5 mm) and bone resorption threshold (1.0 mm) are provided as clinical reference values.

Group	Double-leg Stance (500 N)	Single-leg Stance (1,000 N)	Walking (1,500 N)
A	0.329	0.591	1.232
B	0.154	0.332	0.708
C	0.070	0.114	0.226
D	0.131	0.246	0.511
E	0.098	0.210	0.499
F	0.067	0.121	0.264
G	0.084	0.155	0.281
H	0.064	0.111	0.215
I	0.055	0.083	0.174

Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

Table 9. Peak von Mises stress (MPa) on the fracture surface for all fixation constructs under three loading conditions. Values represent the maximum stress magnitude observed on the fracture plane itself, reflecting the mechanical environment at the bone healing interface under double-leg stance (500 N), single-leg stance (1,000 N), and walking (1,500 N) loading.

Group	Double-leg Stance (500 N)	Single-leg Stance (1,000 N)	Walking (1,500 N)
A	41.19	72.73	160.89
B	18.32	36.23	92.01
C	14.83	26.05	59.44

D	16.45	31.22	62.81
E	26.01	31.65	61.15
F	25.74	28.79	52.77
G	20.55	34.90	62.13
H	11.87	22.38	46.00
I	12.65	24.37	54.69

Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

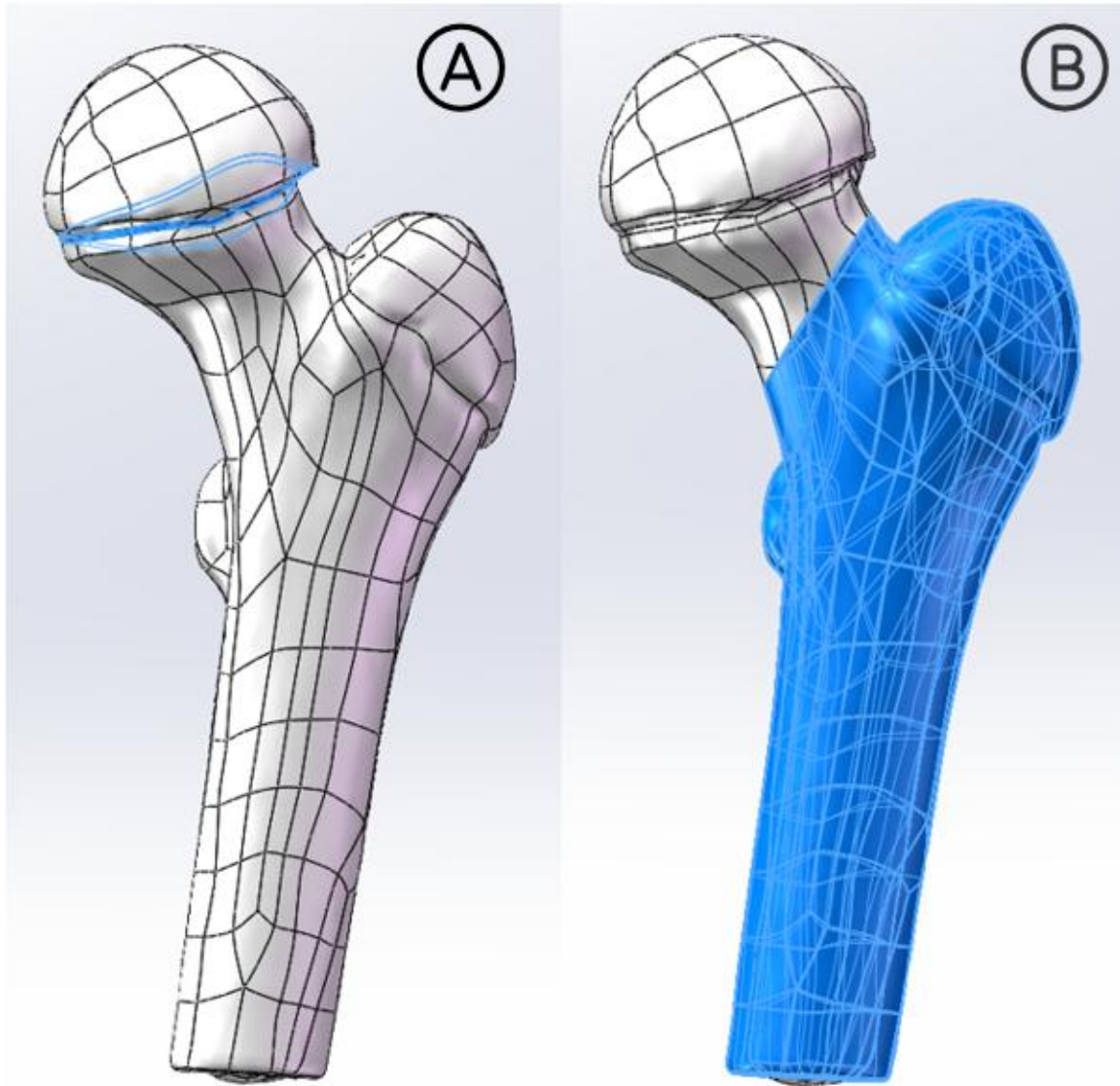
Table 10. Construct stiffness (N/mm) of each fixation construct under three loading conditions. Higher values indicate greater mechanical stability of the construct.

Group	Double-leg Stance (N/mm)	Single-leg Stance (N/mm)	Walking (N/mm)
A	1,519.8	1,692.0	1,217.5
B	3,246.8	3,012.0	2,118.6
C	7,142.9	8,772.0	6,637.2
D	3,816.8	4,065.0	2,935.4
E	5,102.0	4,761.9	3,006.0
F	7,462.7	8,264.5	5,681.8
G	5,952.4	6,451.6	5,338.1
H	7,812.5	9,009.0	6,976.7
I	9,090.9	12,048.2	8,620.7

Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

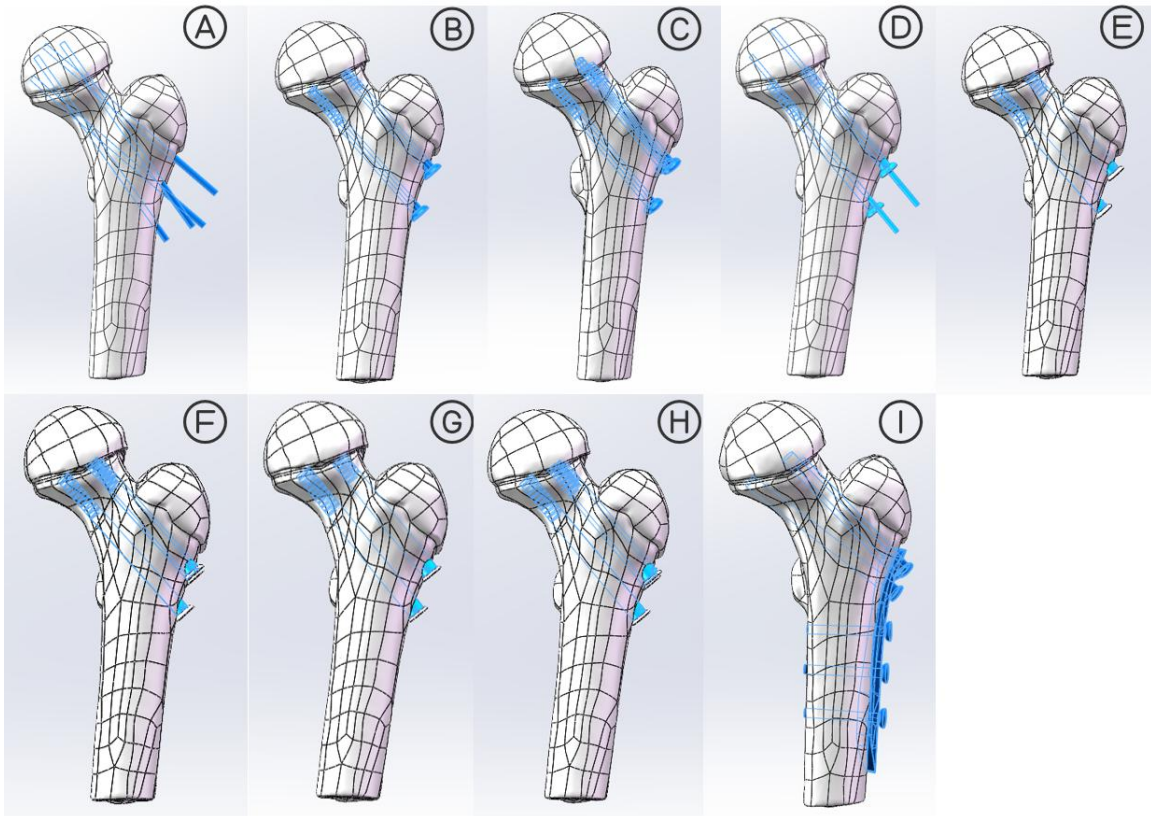
Figure Legends

Fig. 1. Three-dimensional geometric models of the pediatric proximal femur with a simulated Delbet Type II femoral neck fracture



(A) Intact proximal femur model showing anatomical landmarks including the femoral head, neck, greater and lesser trochanters, and proximal femoral physis. (B) Transcervical fracture line oriented obliquely across the femoral neck, consistent with Delbet Type II classification.

Fig. 2. Diagram of fracture and internal fixation assembly mode.



Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

Fig. 3. The mesh distribution appearance diagram.

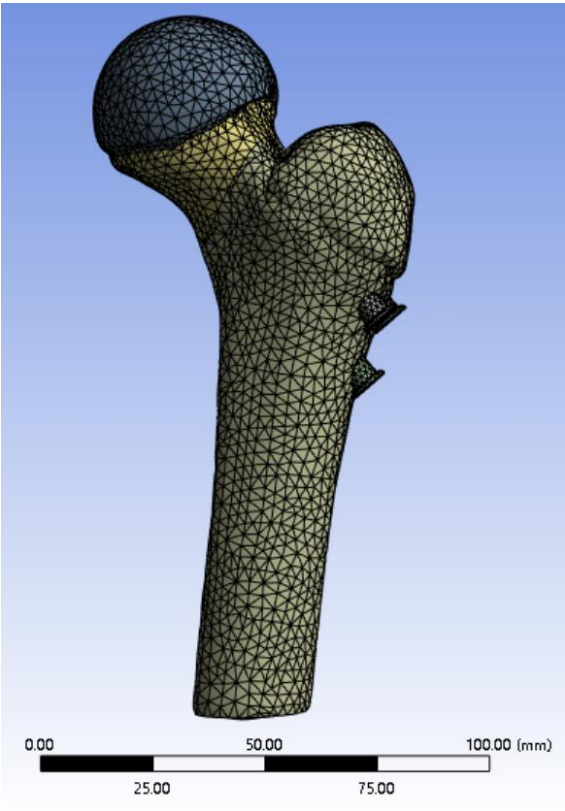
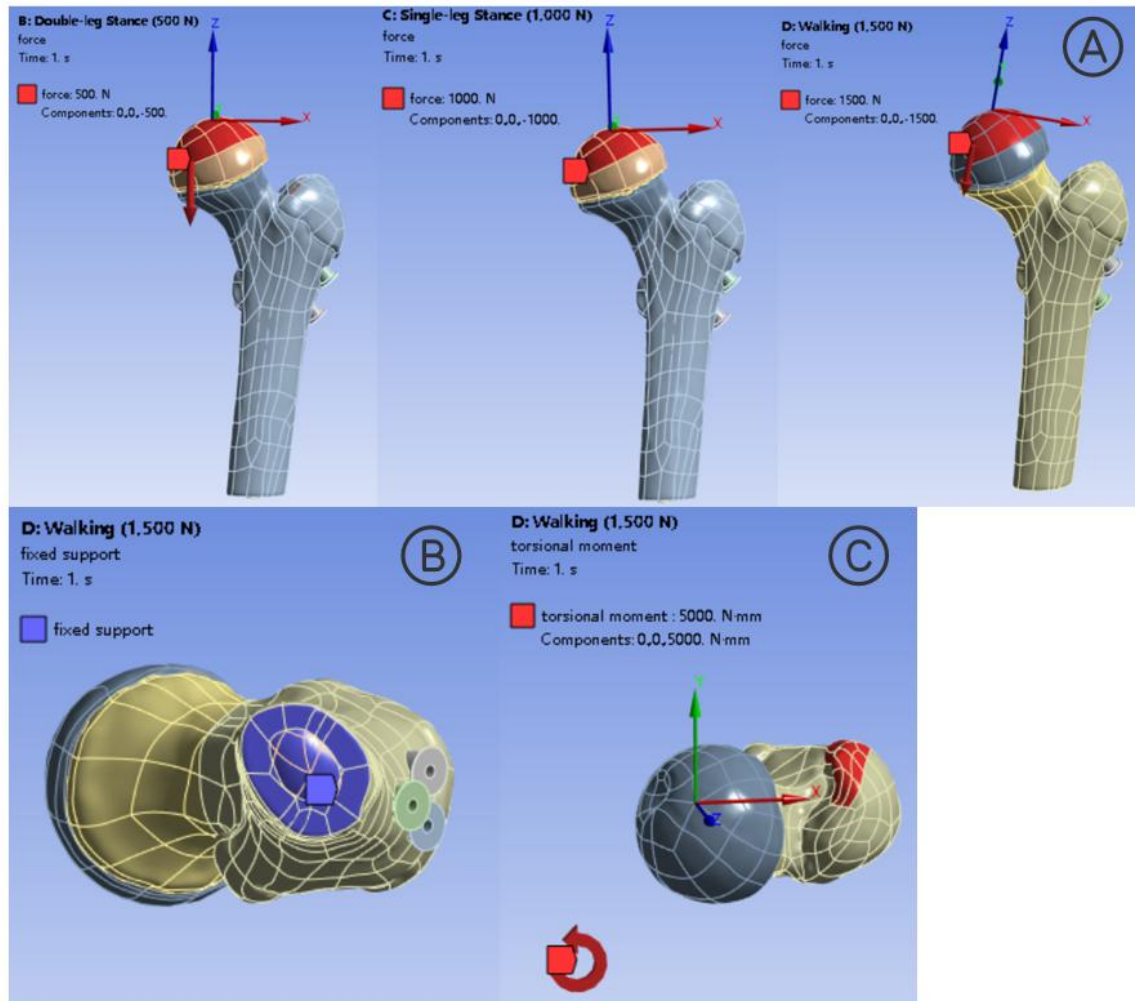
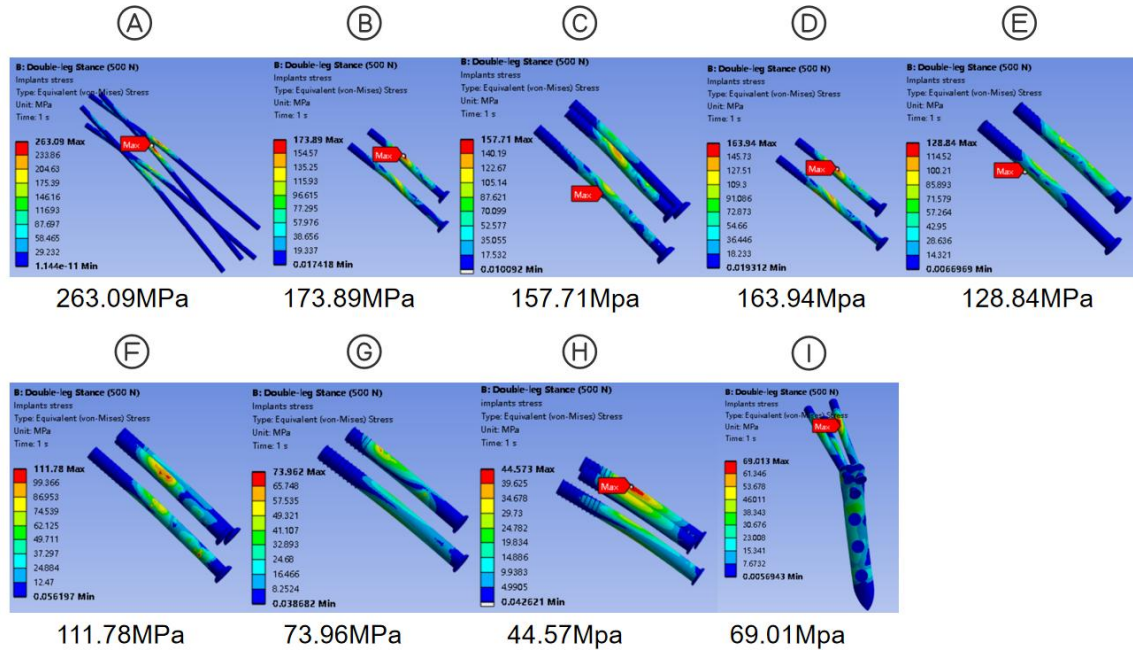


Fig. 4. Boundary conditions and loading configurations applied in the finite element model.



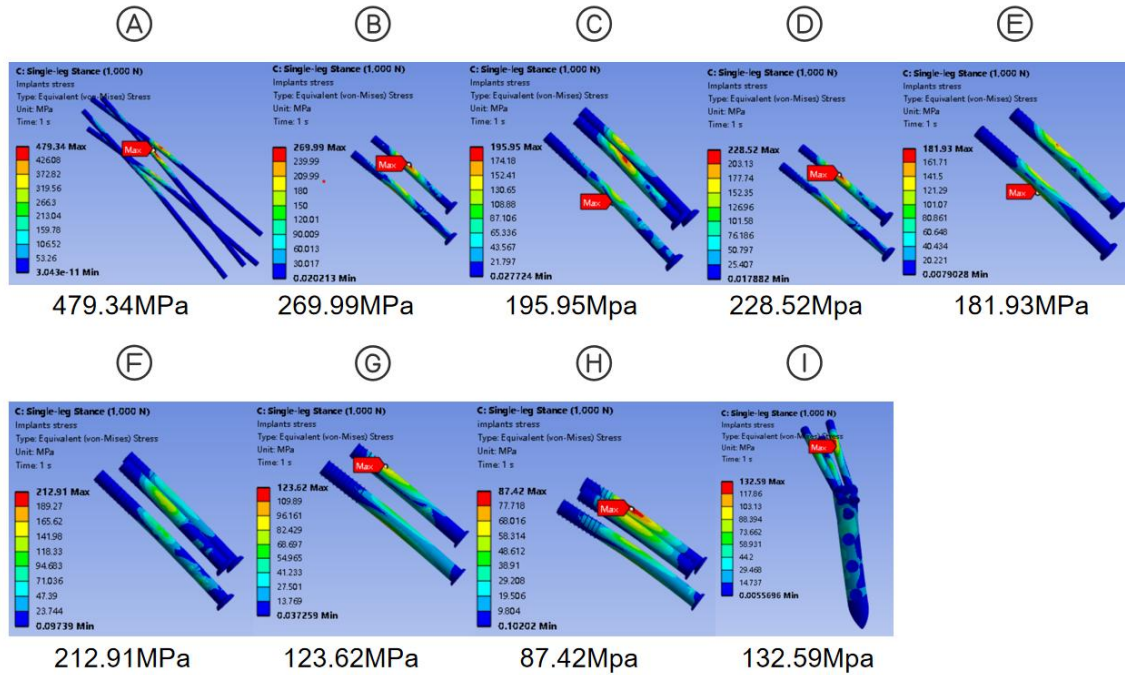
(A) Joint reaction forces were applied at the femoral head center via a coupled reference point simulating double-leg stance (500 N, vertical), single-leg stance (1,000 N, vertical), and walking (1,500 N, 10° coronal adduction, 9° sagittal extension). (B) The distal femoral shaft was fully constrained in all six degrees of freedom at the distal metaphyseal transection plane. (C) A torsional moment of 5 N·m about the femoral long axis was applied during walking simulations.

Fig. 5. Implant von Mises stress distribution cloud maps under double-leg stance loading (500 N, $\sim 1.0 \times BW$). The figure displays the implant stress distribution for each fixation construct (Groups A–I). Stress concentrations are visible in the vicinity of the fracture line. The Ti-6Al-7Nb alloy yield stress of 450 MPa is indicated as a threshold reference. Under this low-magnitude loading, all constructs remained below the material yield limit.



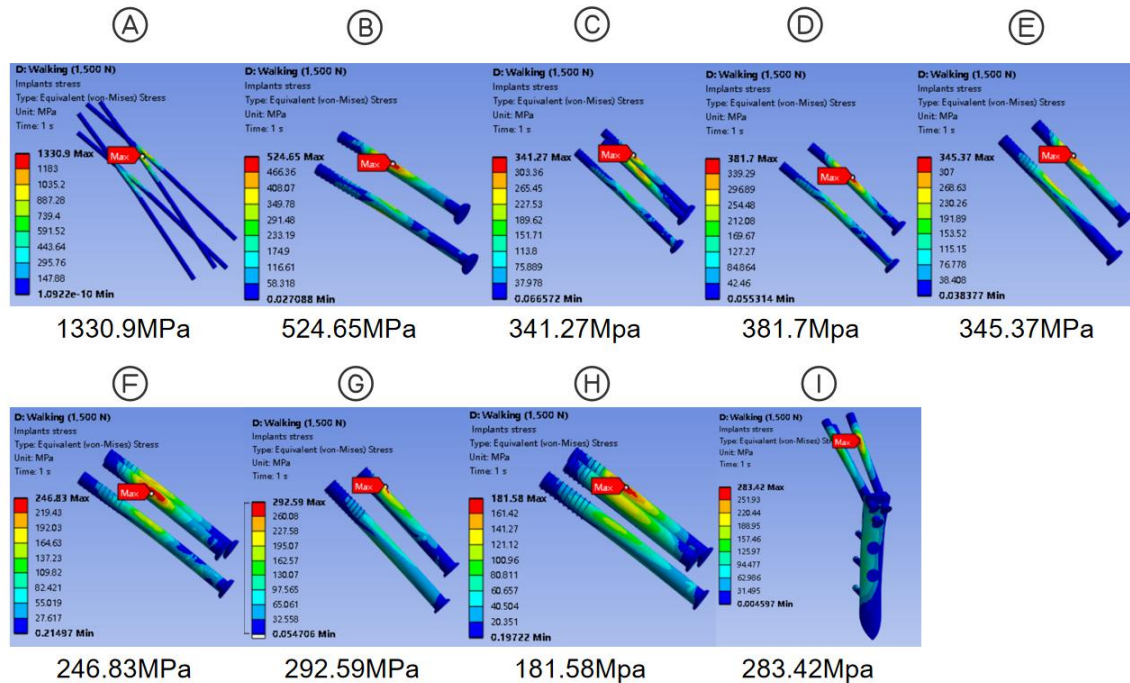
Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

Fig. 6. Implant von Mises stress distribution cloud maps under single-leg stance loading (1,000 N, $\sim 2.0 \times BW$). The figure displays the implant stress distribution for each fixation construct (Groups A–I). Under this intermediate loading, Group A (crossed K-wires) exhibited peak stress exceeding the Ti-6Al-7Nb yield stress of 450 MPa, while all screw and plate constructs remained below this limit. Stress concentrations were most pronounced in the vicinity of the fracture line.



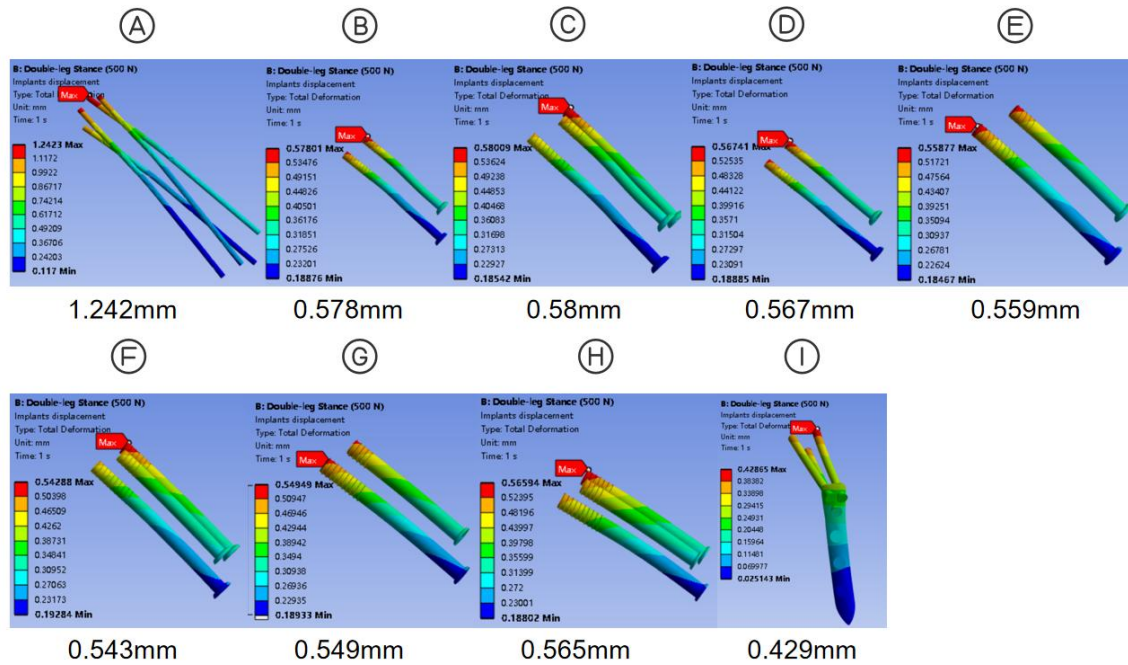
Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

Fig. 7. Implant von Mises stress distribution cloud maps under walking loading (1,500 N, $\sim 3.1 \times BW$ with 10° adduction, 9° posterior inclination, 5 N·m torsion). The figure displays the implant stress distribution for each fixation construct (Groups A–I). Under this peak physiological loading, Group A exhibited severe stress exceeding the 450 MPa Ti-6Al-7Nb yield threshold. Groups B also exceeded the yield limit. The remaining groups recorded values below this threshold, with Group H showing the lowest peak implant stress.



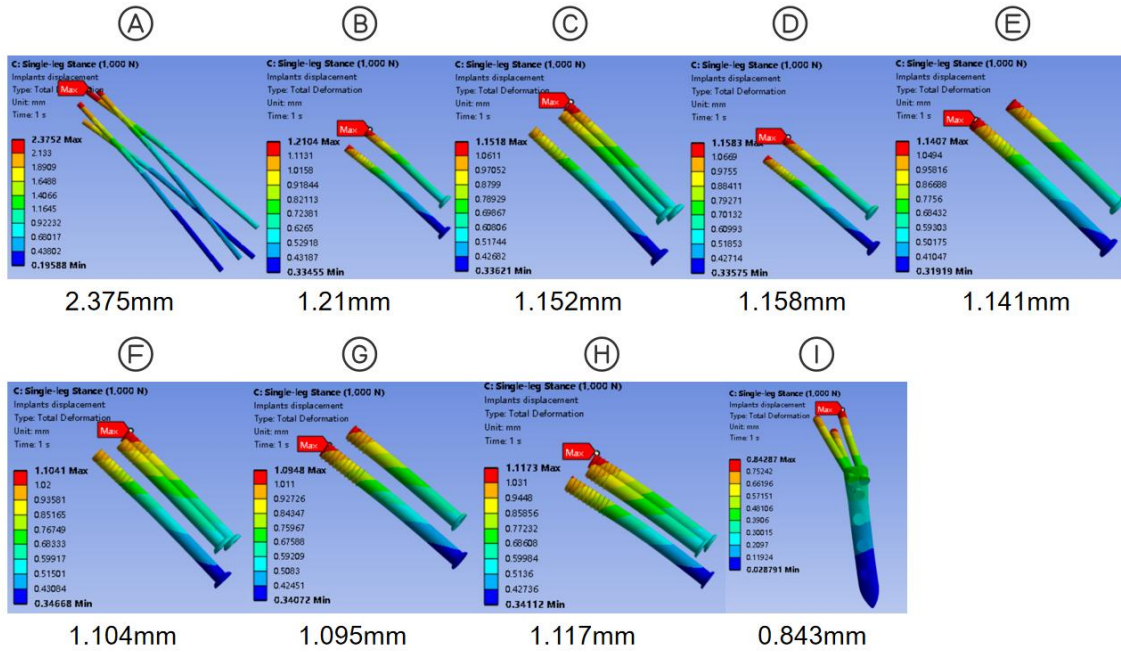
Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

Fig. 8. Implant displacement distribution cloud maps under double-leg stance loading (500 N, $\sim 1.0 \times BW$). The figure displays the total displacement magnitude within the implant components for each fixation construct (Groups A–I). Displacement values represent the magnitude of deformation relative to the initial implant position. Group A exhibited the greatest implant displacement (1.242 mm), while Group I showed the least (0.429 mm). K-wire constructs demonstrated notably greater deformation than screw or plate-based fixation.



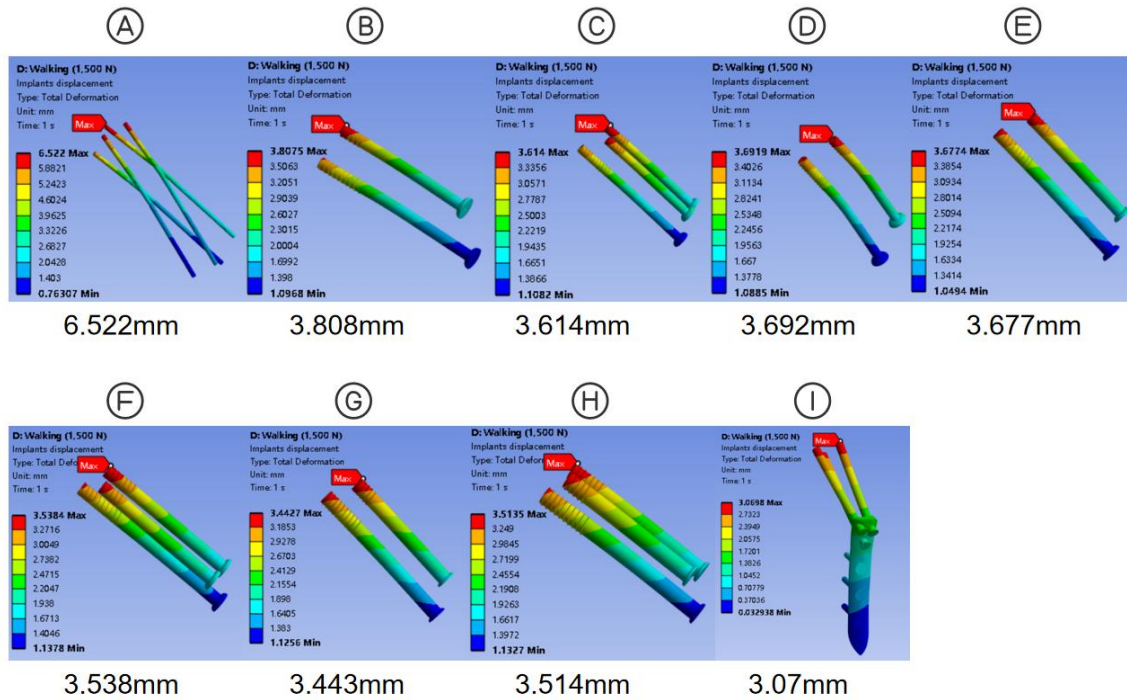
Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

Fig. 9. Implant displacement distribution cloud maps under single-leg stance loading (1,000 N, $\sim 2.0 \times BW$). The figure displays the total displacement magnitude within the implant components for each fixation construct (Groups A–I). Group A exhibited the greatest implant displacement (2.375 mm), consistent with the lower stiffness of smooth K-wire fixation. Groups B–H showed intermediate displacement values (1.095–1.210 mm), while Group I (proximal femoral locking plate) demonstrated the smallest displacement (0.843 mm). The displacement pattern reflects the relative bending stiffness of each construct.



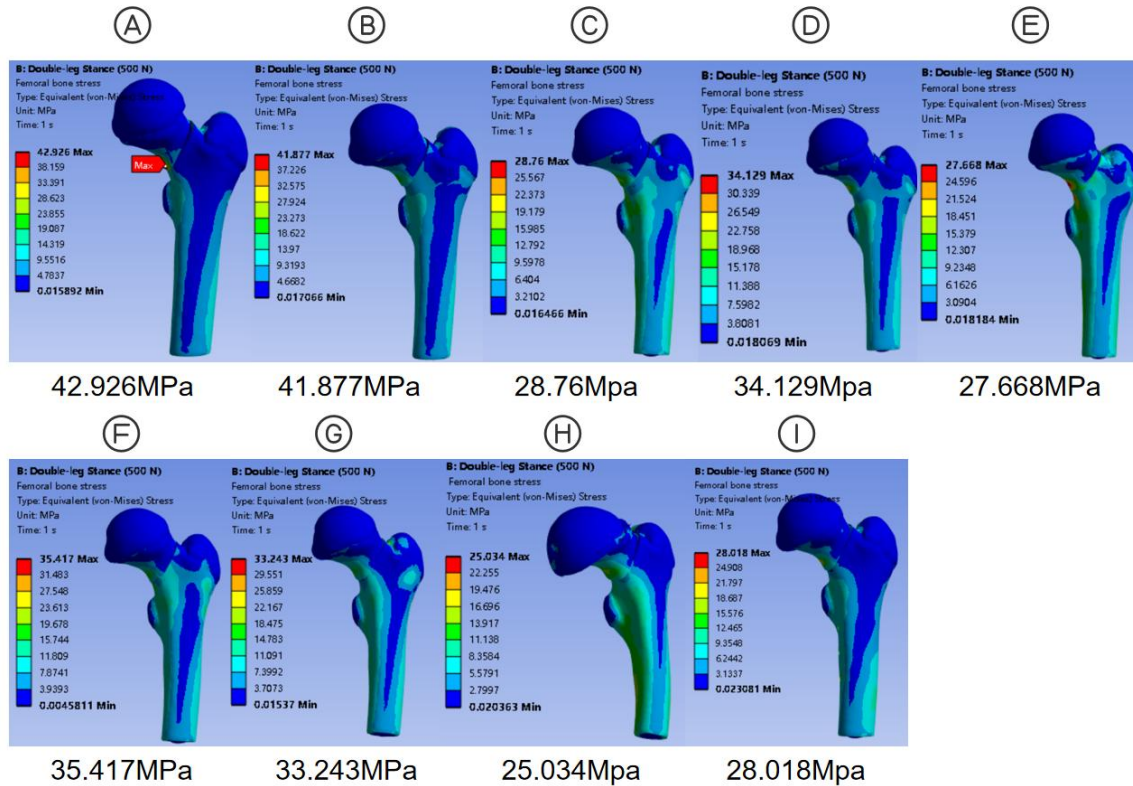
Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

Fig. 10. Implant displacement distribution cloud maps under walking loading (1,500 N, $\sim 3.1 \times BW$ with 10° adduction, 9° posterior inclination, 5 N·m torsion). The figure displays the total displacement magnitude within the implant components for each fixation construct (Groups A–I). Under peak physiological loading, Group A showed substantially greater implant displacement (6.522 mm) than all other constructs. Group I maintained the lowest displacement (3.070 mm). Three-screw inverted-triangle configurations (Groups C, F, H) demonstrated intermediate displacement values, generally lower than their two-screw counterparts.



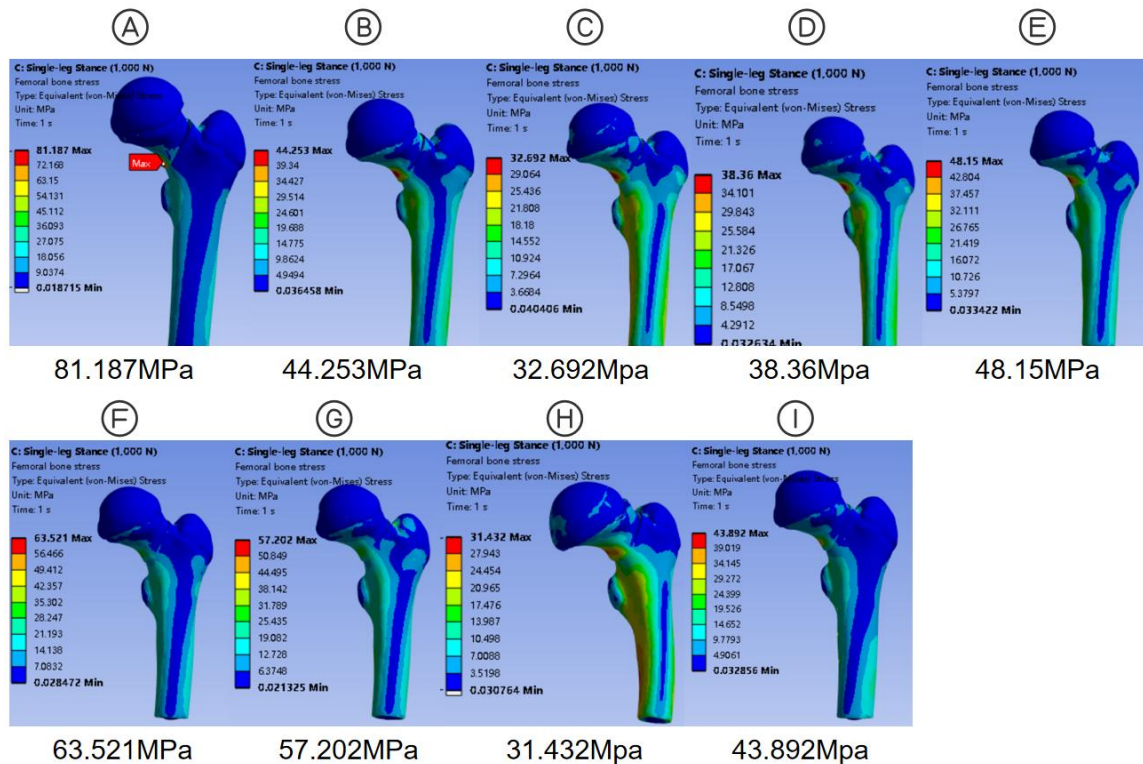
Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

Fig. 11. Femoral bone von Mises stress distribution cloud maps under double-leg stance loading (500 N, $\sim 1.0 \times BW$). The figure displays the stress distribution within the femoral cortical and cancellous bone for each fixation construct (Groups A–I). The femoral yield stress of 100 MPa is indicated as a threshold reference. Stress concentrations were observed in the calcar region, lateral femoral neck cortex, and peritrochanteric areas. Group A exhibited the highest femoral stress (42.93 MPa), while Group H showed the lowest (25.03 MPa). All constructs remained well below the 100 MPa reference threshold under this low-magnitude loading.



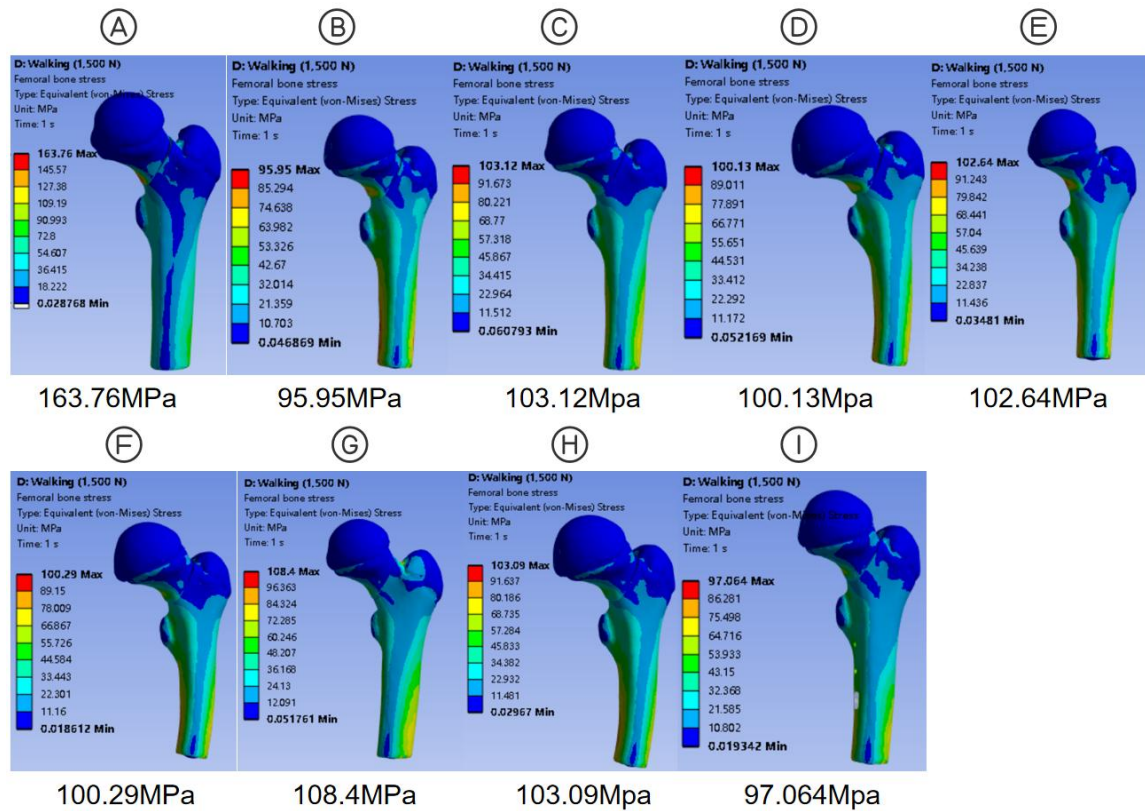
Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

Fig. 12. Femoral bone von Mises stress distribution cloud maps under single-leg stance loading (1,000 N, $\sim 2.0 \times BW$). The figure displays the stress distribution within the femoral cortical and cancellous bone for each fixation construct (Groups A–I). Group A exhibited the highest femoral stress (81.19 MPa), while Group H showed the lowest (31.43 MPa). All screw and plate constructs remained below the 100 MPa reference threshold. Stress distribution patterns showed medial column compression and lateral neck tension characteristic of femoral neck loading biomechanics.



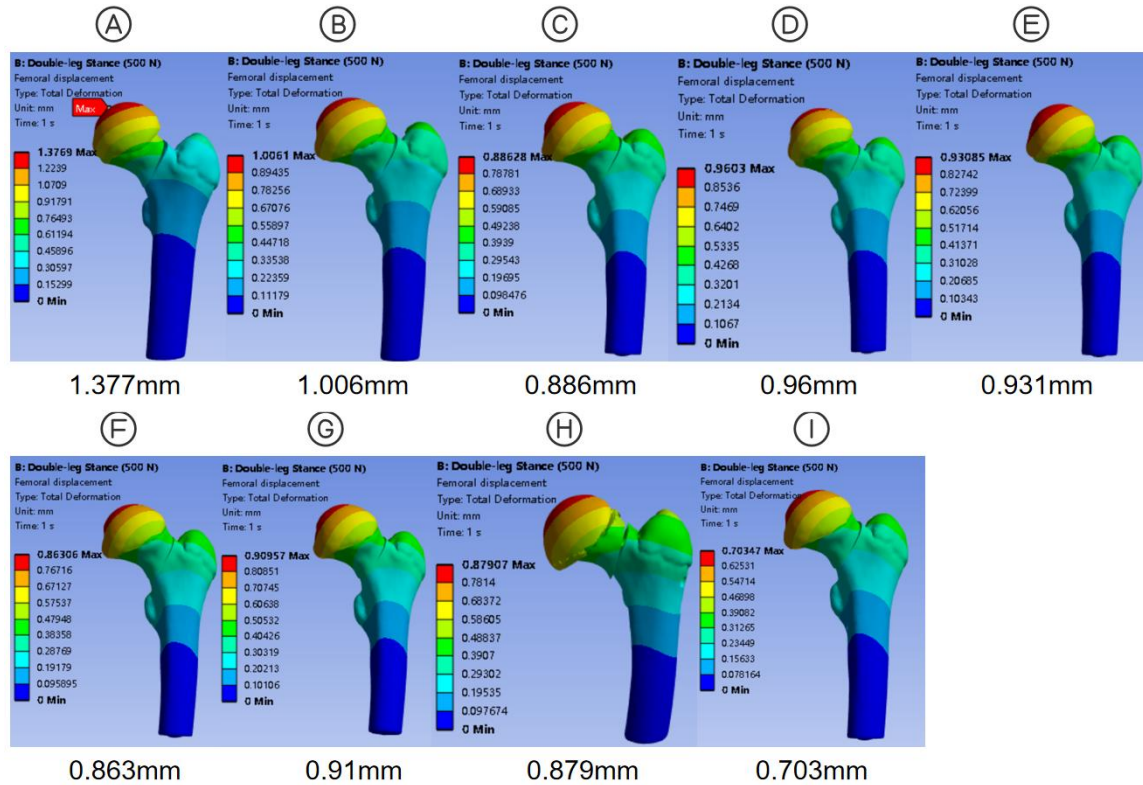
Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

Fig. 13. Femoral bone von Mises stress distribution cloud maps under walking loading (1,500 N, $\sim 3.1 \times BW$ with 10° adduction, 9° posterior inclination, 5 N·m torsion). The figure displays the stress distribution within the femoral cortical and cancellous bone for each fixation construct (Groups A–I). The peak stress in the femur of Group A was 163.76 MPa, significantly exceeding the reference threshold of 100 MPa. The peak stresses in the other groups were also approached or slightly exceeded this threshold.



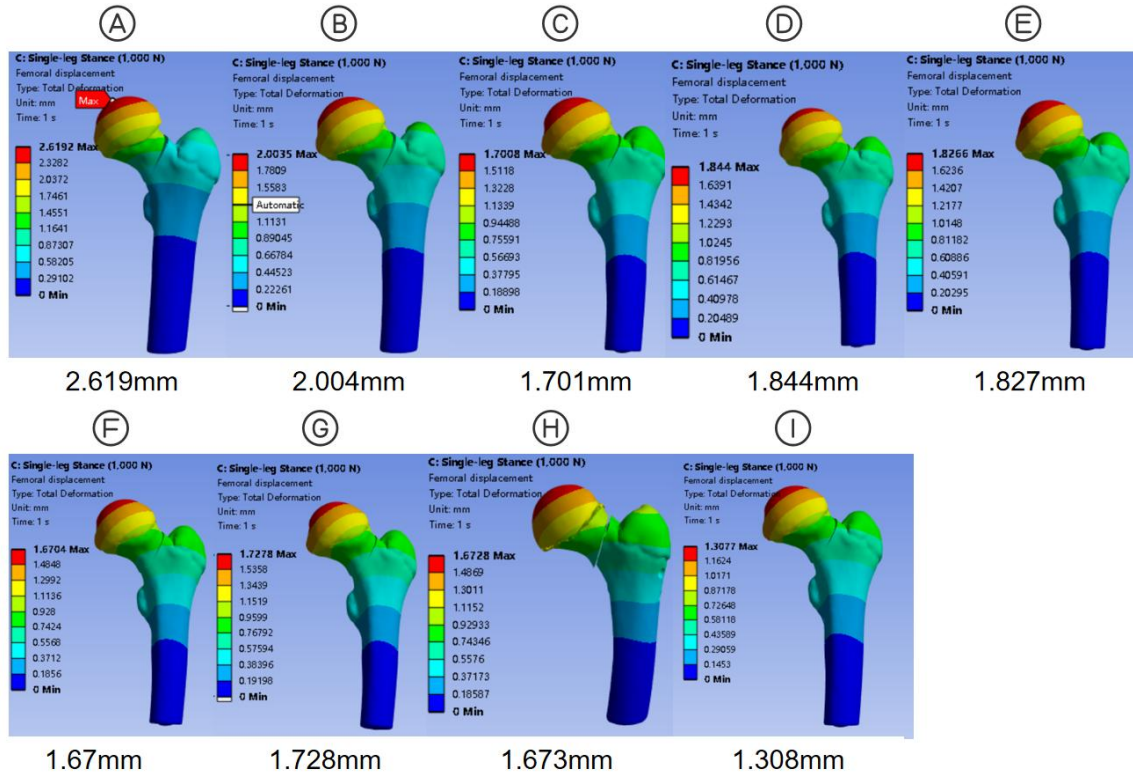
Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

Fig. 14. Femoral displacement distribution cloud maps under double-leg stance loading (500 N, $\sim 1.0 \times BW$). The figure displays the total displacement magnitude of the femoral bone for each fixation construct (Groups A–I). Group A exhibited the greatest femoral displacement (1.377 mm), while Group I showed the least (0.703 mm). Three-screw constructs (Groups C, F, H) and the locking plate (Group I) demonstrated superior resistance to femoral deformation compared with two-screw and K-wire constructs.



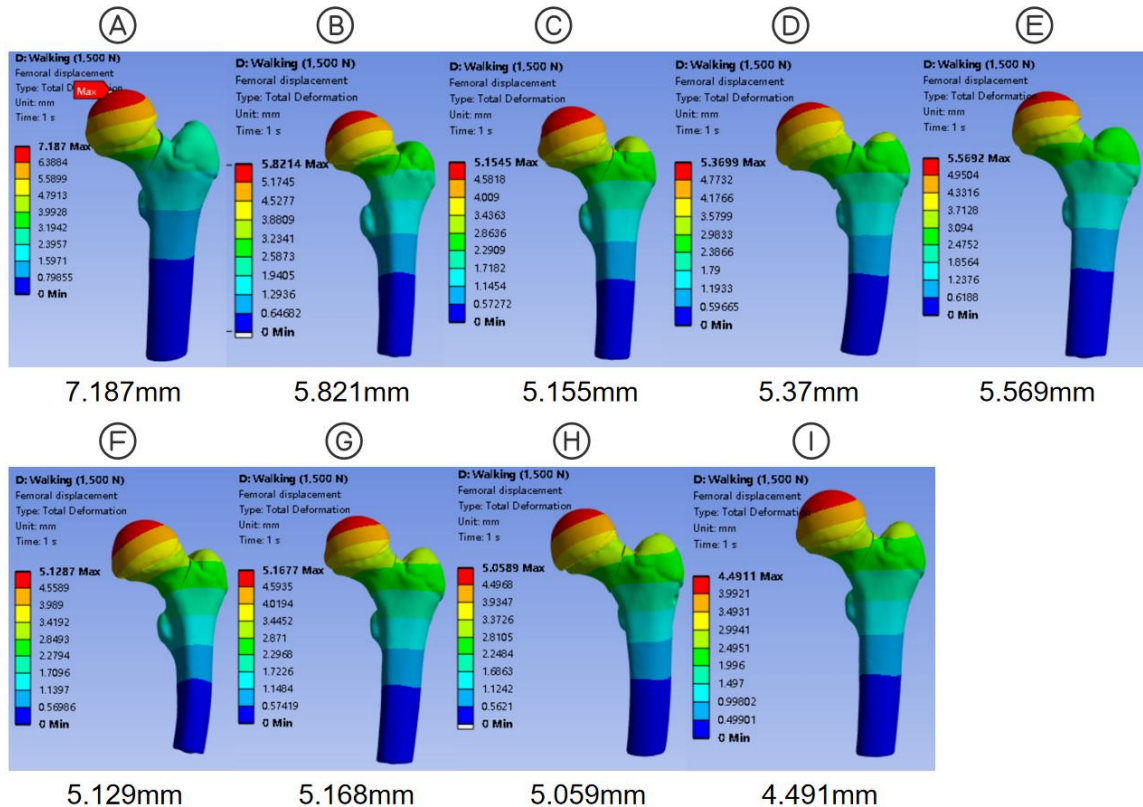
Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

Fig. 15. Femoral displacement distribution cloud maps under single-leg stance loading (1,000 N, $\sim 2.0 \times BW$). The figure displays the total displacement magnitude of the femoral bone for each fixation construct (Groups A–I). Group A exhibited the greatest femoral displacement (2.619 mm), while Group I demonstrated the least (1.308 mm). The three-screw inverted-triangle configurations (Groups C, F, H) provided progressively greater stability with increasing screw diameter. The PFLP (Group I) distributed load across a broader bone–implant interface, resulting in the smallest overall femoral deformation.



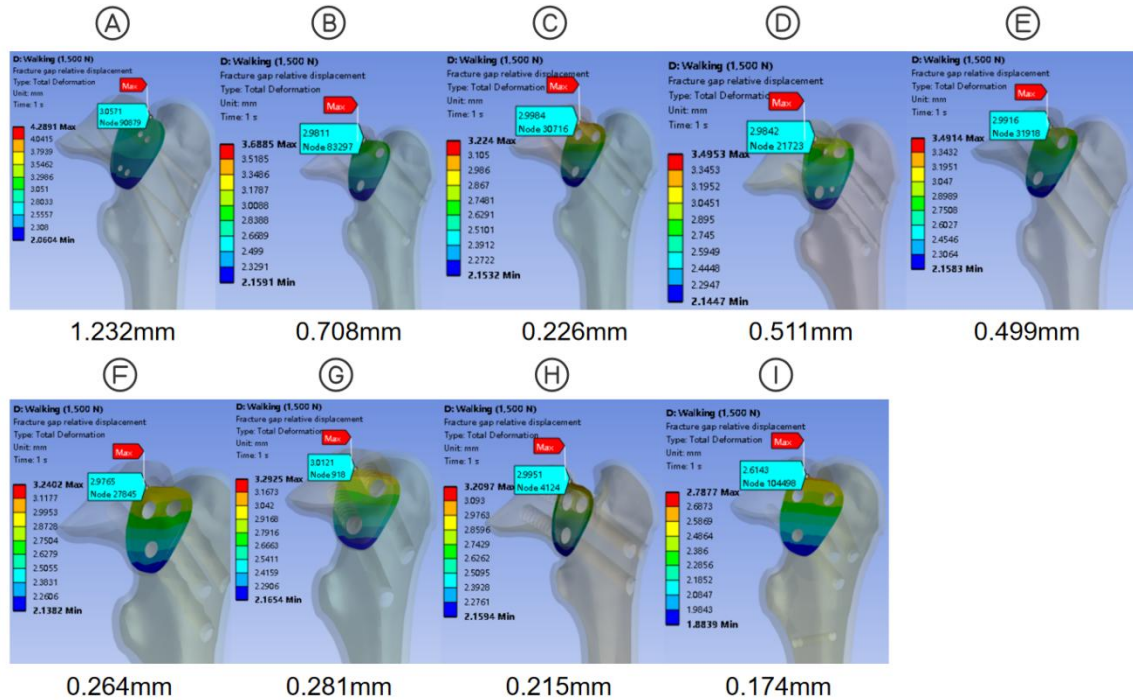
Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

Fig. 16. Femoral displacement distribution cloud maps under walking loading (1,500 N, $\sim 3.1 \times BW$ with 10° adduction, 9° posterior inclination, 5 N-m torsion). The figure displays the total displacement magnitude of the femoral bone for each fixation construct (Groups A–I). Group A exhibited substantially greater femoral displacement (7.187 mm) than all other constructs. Group I maintained the lowest femoral displacement (4.491 mm). Among cannulated screw constructs, three-screw configurations (Groups C, F, H) consistently outperformed their two-screw counterparts (Groups B, E, G) in resisting femoral deformation under combined multiplanar loading.



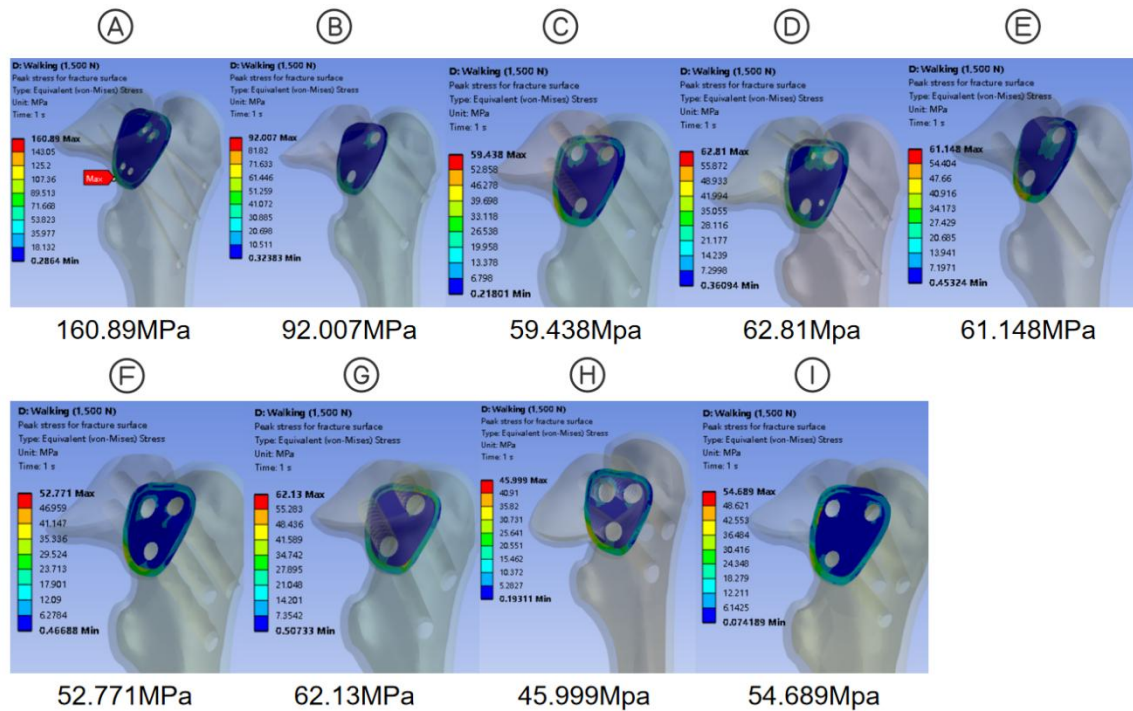
Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

Fig. 17. Fracture gap relative displacement distribution cloud maps under walking loading (1,500 N, $\sim 3.1 \times BW$ with 10° adduction, 9° posterior inclination, 5 N·m torsion). The figure displays the relative displacement magnitude across the fracture interface for each fixation construct (Groups A–I). The fracture gap healing threshold is 0.5 mm and bone resorption threshold is 1.0 mm. Group A (1.232 mm) exceeded the resorption threshold, indicating unacceptable micromotion for fracture union. Groups B, D, and E exceeded or approached the healing threshold. Groups C, F, G, H, and I maintained fracture gap displacement below 0.5 mm, favorable for fracture healing.



Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.

Fig. 18. Peak stress and stress distribution cloud maps for fracture surface under walking loading (1,500 N, $\sim 3.1 \times BW$ with 10° adduction, 9° posterior inclination, 5 N·m torsion). The figure displays the stress distribution of fracture surface for each fixation construct (Groups A – I). Group A (160.89MPa) significantly exceeded the reference threshold of 100 MPa. Groups B (92.01MPa) approached the threshold. The peak stresses in the other groups were all well below this threshold, and Group H showed the lowest (46.00 MPa).



Groups A–I represent the following fixation constructs: Group A = four 2.5 mm crossed Kirschner wires (K-wires); Group B = two 4.5 mm CSs; Group C = three 4.5 mm CSs in inverted-triangle configuration; Group D = two 4.5 mm CSs combined with two 2.0 mm K-wires (hybrid construct); Group E = two 5.5 mm CSs; Group F = three 5.5 mm CSs in inverted-triangle configuration; Group G = two 6.5 mm CSs; Group H = three 6.5 mm CSs in inverted-triangle configuration; Group I = PFLP.