



Article

Realistic Boundary Conditions in Testing Diaphyseal Femoral Implants

Levita Valeriant ^{1,2}, Tobias Barth ^{2,3}, Matthias Münch ^{2,3}, Arndt-Peter Schulz ^{2,3} and Nicholas Bishop ^{1,*}

¹ Department of Medical Technology, Faculty of Life Sciences, Hamburg University of Applied Sciences (HAW Hamburg), 21033 Hamburg, Germany; levitagraciella.valeriant@haw-hamburg.de

² Centre for Clinical Research, BG Klinikum Hamburg, 21033 Hamburg, Germany; t.barth@bgk-hamburg.de (T.B.); m.muench@bgk-hamburg.de (M.M.); a.schulz@bgk-hamburg.de (A.-P.S.)

³ Fraunhofer IMTE, Fraunhofer Research Institution for Individualized and Cell-Based Medical Engineering, 23538 Lübeck, Germany

* Correspondence: nicholas.bishop@haw-hamburg.de

Abstract

Background: Considerable variation exists in the proximal and distal fixture designs used in biomechanical testing of diaphyseal femoral implants. In most of them a load is applied to the femoral head, but muscle forces are rarely represented, so bending of the diaphysis is unphysiologically high. **Methods:** This study investigated twelve testing configurations reported in the literature using finite element and analytical models of the femoral diaphysis and a lateral fixation plate. The models differed in proximal and distal displacement constraints, femoral orientation, and load application. Internal force distributions, as well as bone and implant stresses and bone strains, were calculated for each configuration. **Results:** Common experimental conditions, in which the load is applied to the femoral head, generally produced high bending moments, resulting in, what we interpret to be, unrealistically high bone and implant stresses and bone strains. In contrast, axial loading along the femoral shaft generated substantially lower bending, which we propose to be more physiological, particularly in two such cases with reduced rotational constraint. **Conclusions:** Their constant bending distribution should improve reproducibility and their implementation should provide a practical approach to biomechanical testing of diaphyseal femoral implants.

Keywords: biomechanics; test setup; femur; femoral diaphysis; biomechanical analysis and modeling; implant testing



Academic Editor: Frederic Marin

Received: 30 June 2026

Revised: 11 August 2026

Accepted: 24 August 2026

Published: 28 August 2026

Copyright: © 2026 by the authors.

Licensee MDPI, Basel, Switzerland.

This article is an open access article distributed under the terms and conditions of the [Creative Commons Attribution \(CC BY\)](https://creativecommons.org/licenses/by/4.0/) license.

1. Introduction

Diaphyseal femoral fractures are among the most common injuries in orthopaedic practice, with reported incidences ranging from approximately 10 to 21 cases per 100,000 individuals annually [1–3]. These fractures typically result from high-energy trauma, such as motor vehicle accidents, particularly in younger people. However, in elderly patients, they are more commonly associated with low-energy trauma, such as slips, trips, or falls [4,5]. When not appropriately treated, femoral shaft fractures can lead to disability. The literature states that costs of femoral shaft fractures are between approximately €24,000–€25,000 per patient [6,7].

Intramedullary nailing and plate osteosynthesis are the standard treatments in most cases. The development and evaluation of orthopaedic implants and their interfaces with the bone require preclinical testing to ensure their safety, durability, and effectiveness.

Biomechanical testing plays a role in this process, providing insights into how implants interact with the body under physiological loading [8].

One of the most critical activities considered in biomechanical femoral testing is gait, as it is an unavoidable and repetitive daily activity for patients as they convalesce [9]. Many studies suggest that during walking, muscle activity plays an important role in limiting bending moments in the femur [10–12]. Taylor et al. [11] used a finite element model of the proximal femur with gait joint loads to demonstrate a much-reduced bending deflection when the main muscle group forces were included. They demonstrated that this low bending deflection was similar to that measured from X-rays for one-legged stance *in vivo*.

Considerable variation is found between the distal and proximal fixture designs employed in experimental testing of implanted femurs, which makes their interpretation and comparison difficult [13]. Inacio et al. [14] emphasised that the mechanical performance of such constructs is highly sensitive to the boundary conditions imposed by the test fixtures, with increasing motion constraint leading to increased stiffness of the construct.

This study investigates which boundary conditions are suitable for biomechanical experiments representing one-legged stance during walking loading of the femoral diaphysis. Based on the arguments presented above, we assume that low bending of the bone is most realistic. A simplified finite element model of the femur was used to determine reaction forces, which were then used to analytically calculate internal forces and moments along the bone diaphysis and in a simulated lateral fracture fixation plate. Bending stresses and strains were then derived.

2. Methods

2.1. Finite Element (FE) Model

Various boundary conditions were applied to the proximal and distal ends of a tubular finite element model of the femoral diaphysis and a fixation plate model (ANSYS 2024 R1, Ansys Inc., Canonsburg, PA, USA) and the reaction forces and moments were sampled at the distal end. The analysis focused on a frontal plane representation of the femur, whereby horizontal translation ($d_{horizontal}$) and rotation within the plane (r) were applied proximally and distally (Figure 1) to represent various published testing conditions [15–21]. Due to the uniaxial nature of most testing machines, a “vertical” force was applied to the proximal end of the femur and vertical motion ($d_{vertical}$) was fixed distally. Two load application points were investigated: (i) the femoral head (applied at an angle 10° to the diaphyseal axis) and (ii) the femoral notch, along the diaphyseal axis, according to common experimentally applied load configurations: femoral head: [15–19]; proximal notch: [20,21]. Twelve combinations of boundary conditions were tested (Figure 1). The models differ in proximal and distal displacement constraint configurations, femoral orientation, and load application.

The femoral diaphysis was modelled as a 400 mm cortical bone tube [22], with an outside diameter of 30 mm [23], a wall thickness of 5 mm [24], an elastic modulus of 17 GPa and a Poisson’s ratio of 0.3. A force of 2000 N was applied either to a rigid lever arm 50 mm from the bone axis [25], to represent the medial offset of the femoral head (Models 1 to 7), or along the axis of the bone (Models 8 to 12). This load represents the peak hip joint contact force measured *in vivo* during one-legged stance (mean $2.7 \times$ body weight for a 75 kg patient [26]). The direction of the hip contact force in the frontal plane during dynamic activities, such as slow walking, normal walking, fast walking, ascending and descending stairs, sitting down, standing up, and knee bends, ranges between 12° and 16° . During one-legged stance, the angle is approximately 7° [26]. The 10° angle was selected to represent a midpoint in this range.

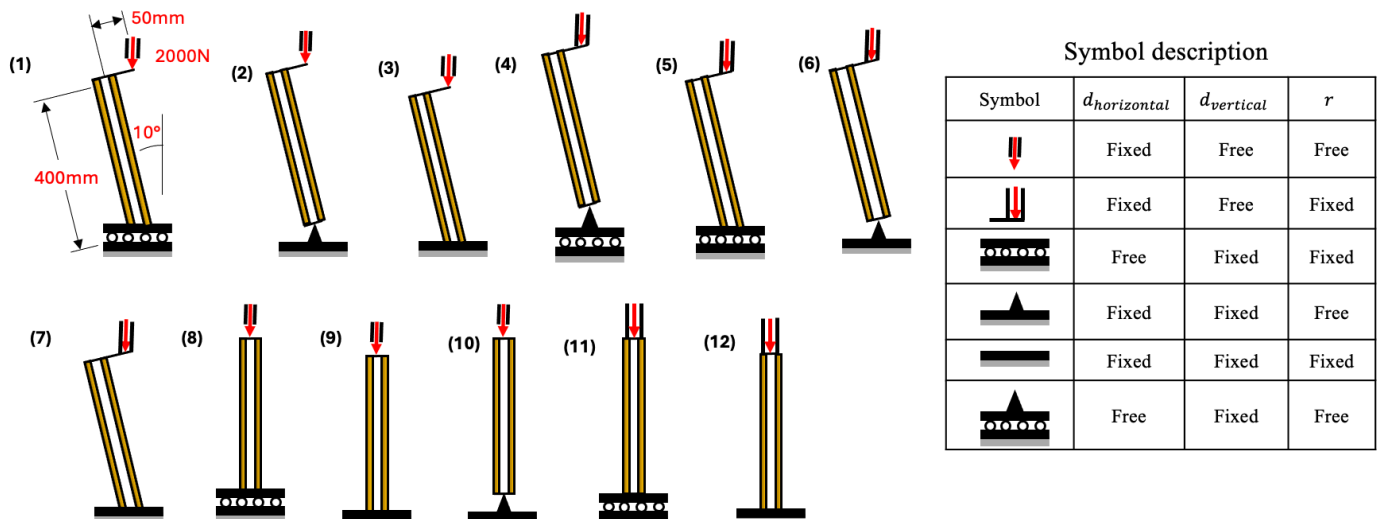


Figure 1. The twelve combinations of boundary conditions representing test conditions reported in the literature and investigated in this study. $d_{horizontal}$ and $d_{vertical}$ are respective displacements in the plane shown and r is the rotation around an axis out of the plane.

A second finite element model consisting of the lateral fixation plate alone was created. The plate was represented as a 400 mm long beam with a rectangular cross-section width of 12 mm and a thickness of 6 mm. The elastic modulus assigned to the plate was that of titanium alloy ($E = 110$ GPa). The same twelve loading and boundary condition configurations applied to the tubular femoral model were applied to the plate model.

In both models the distal reaction forces (F_{1x} , F_{1y} , M_{1z}) (Figure 2) were sampled for each of the 12 constraint sets in Figure 1.

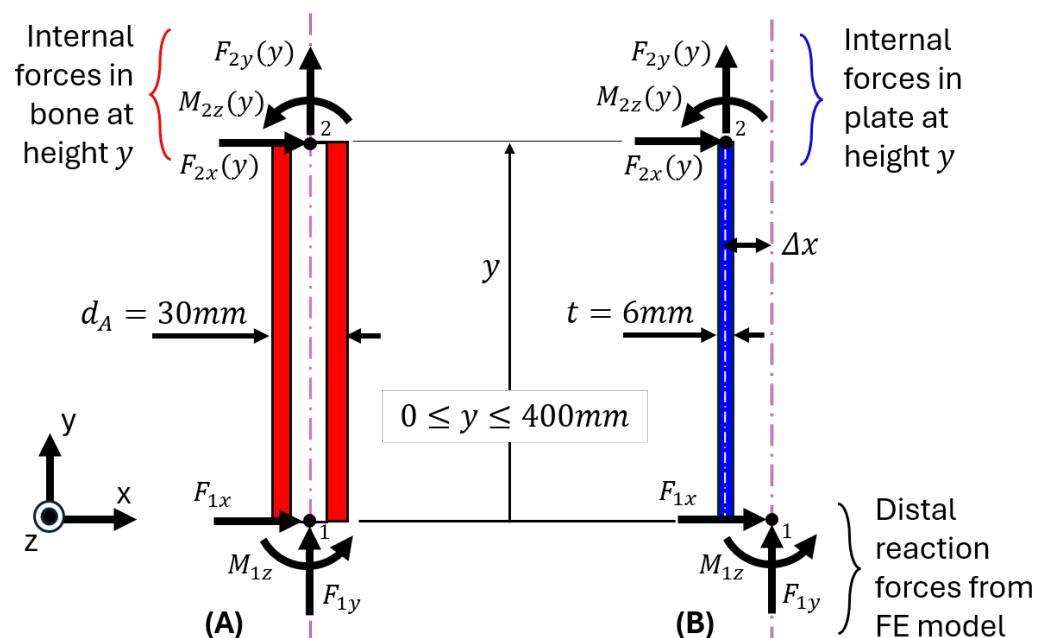


Figure 2. Free body diagrams for the calculation of the internal force distribution $F_{2x}(y)$, $F_{2y}(y)$, $M_{2z}(y)$ from distal reaction forces F_{1x} , F_{1y} , M_{1z} calculated with finite element models: (A) for the bone diaphysis alone ($\Delta x = 0$ mm); (B) for the lateral plate alone ($\Delta x = 18$ mm), where Δx is the distance from the bone axis (dot-dash, magenta).

2.2. Internal Forces, Stresses and Strains

The distal boundary reaction forces F_{1x} , F_{1y} , M_{1z} from the finite element models were used to calculate the internal force distributions, where

$$F_{2x}(y) = -F_{1x} \quad (1)$$

$$F_{2y}(y) = -F_{1y} \quad (2)$$

$$M_{2z}(y) = -M_{1z} - F_{1x} \cdot y - F_{1y} \cdot \Delta x \quad (3)$$

in the bone model and the plate model (Figure 2A,B, respectively). These distributions were determined by assuming static equilibrium, resulting in Equations (1)–(3). Internal forces were calculated along the bone axis ($\Delta x = 0$) and along the plate axis ($\Delta x = 18$ mm). Simple bending theory was applied to find the maximum bending stresses at the surfaces of the bone and implant, from the bending moment M_{2z} , as well as the maximum strain at the surface of the bone (Appendix A).

3. Results

The internal shear forces $F_{2x}(y)$ and axial forces $F_{2y}(y)$ along the bone were constant with position y for each configuration (not presented). Shear forces were less than 20% of the joint load in Models 1–5 (−370 N to −248 N), whereas Models 6–12 exhibited negligible shear forces (0 N to 2 N) along the bone. The axial forces were dominated by compressive loading with magnitudes similar to the applied joint load of 2000 N.

For configurations in which the load was applied at the femoral head, bending moments $M_{2z}(y)$ varied linearly along the bone with magnitudes of −139 to 70 Nm (Figure 3, red lines) and along the plate with larger magnitudes, ranging from −174 to 69 Nm (Figure 3, blue lines). In contrast, load application along the femoral shaft axis (Models 8–12) resulted in no bending moments along the bone. Models 6 and 7 (with heavily constrained boundaries) also produced negligible bending moments along the femur, despite femoral head loading. For the plate, however, moments were generated for the plate under axial loading for some of the less constraining boundary conditions (Models 8–10), which were constant along the length of the plate for Models 8 and 10.

The maximum bending stresses in bone and plate and bone surface strain magnitudes are summarised in Figure 3. Magnitudes were greatest for Models 1 to 5, with maxima for Model 4 of 70.3 MPa, 2449 MPa and 4137 $\mu\epsilon$ for bone stress, plate stress and bone strain, respectively. Stress and strain magnitudes are proportional to bending moments and were up to an order of magnitude lower for the models with low bending (Models 6–12). Stresses in the plate were minimal for Models 11 and 12, with heavily constrained rotation boundary conditions, but were moderate and constant along the whole length for the less constrained axially loaded Models 8 and 10 (highlighted in grey, Figure 3).

Model	Bending Moment Distribution in the bone			Bending Moment Distribution in the plate			Max bone stress magnitude [MPa]	Max plate stress magnitude [MPa]	Max bone microstrain [$\mu\epsilon$]	High stress:	Constant low stress:	No stress:
1							51.3	1888	3018	X		
2							51.8	1904	3045	X		
3							51.2	1861	3012	X		
4							70.3	2449	4137	X		
5							38.2	992	2248	X		
6							5.3	542	310			
7							5.4	28	316		X	
8							5.1	528	300	X		
9							5.1	528	300			
10							5.1	528	300	X		
11							5.1	28	300		X	
12							5.1	28	300		X	

Figure 3. Top rows: Internal moment distributions for the twelve sets of boundary conditions: upper (red) for the bone model; lower (blue) for the plate model. The conditions are grouped into those generating moments in the bone (**left**) and those generating negligible moments in the bone (**right**). Bottom rows: Maximum magnitudes of bone and plate bending stresses and corresponding bone strains on the surface of the bone. Models 8 and 10, highlighted in grey, represent what are proposed to be sensible axial loading conditions.

4. Discussion

4.1. Study Aim

In this study femoral boundary configurations were assessed for the magnitudes of the bending moments generated for gait loading, whereby it was assumed here that low bending moments in the bone are physiologically realistic.

4.2. Evidence of Low Bending Due to Muscles

Several studies have suggested that the femur is primarily subjected to compressive loading during walking, with bending moments reduced by muscle forces. For example, Duda, Schneider and Chao [12] demonstrated that muscle activity significantly reduces bending moments acting on the femoral shaft during gait. Similarly, Taylor et al. [11] showed that physiological loading conditions in the femur lead to a predominantly compressive stress distribution in the diaphysis, rather than bending. Earlier studies by Pauwels [10] also demonstrated that muscles and ligaments help limit bending stresses in the femur during the stance phase of gait. Munih, Kralj and Bajd [27] also reported that muscle forces substantially reduce the overall bending moments generated by joint forces in the lower limb in one-legged stance.

4.3. Common Experimental Representation

Based on this evidence we assume that physiologically realistic loading of the femur during gait should involve relatively low bending moment magnitudes. However, it is very difficult to simulate realistic muscle forces experimentally, necessitating the attachment of cables to the bone [28,29]. The effects of realistic muscle loading are therefore commonly neglected, with only the joint force applied, using a uniaxial mechanical testing machine in combination with the various boundary conditions tested in this study.

4.4. Bending Results

Our results indicate that such femoral head loading configurations are generally related to bending moments with high peak magnitudes, which also tend to vary in magnitude along the femoral shaft. We therefore propose that such commonly employed loading conditions lead to non-physiologically high stresses which deviate from patterns described in the literature. In contrast, Models 8 and 10, with axial load application along the femoral shaft, produced moderate and more constant bending moments in the plate and may better reflect the *in vivo* loading expected during walking.

4.5. Further Arguments Against Configurations Generating Bending

For configurations generating bone bending (Models 1–5) it is observed that the stress and strain magnitudes calculated are unphysiological, since they approach the material strengths of bone and titanium, as well as approaching functional bone strain magnitudes. The strength of cortical bone could be assumed to be roughly 100 MPa [30] and for titanium it is an order of magnitude greater, at a maximum of 1000 MPa (Ti-6Al-4V: 795–875 MPa [31]). For models generating bending, maximum bone stresses ranged from 38.2 to 70.3 MPa, and 992 to 2449 MPa for the implant, which would suggest unrealistically low, or non-existent, safety factors for the implant under gait loading. These models also resulted in bone strains of up to 4137 $\mu\epsilon$, which exceed those reported for functional loading of bones (~2000 to 3000 $\mu\epsilon$ in various animals [32]). In contrast, axial loading produced substantially lower stress and strain levels.

4.6. How to Load Realistically

These arguments suggest that common loading configurations, in which the joint load is applied to the femoral head, with no muscle forces simulated, cause unrealistically high bending loads. A force applied along the bone axis (Models 8–12) reduces bending and may be more realistic by better representing the effects of muscles. Low bending moments were also observed for two of the models with femoral head loading (Models 6 and 7), demonstrating that the magnitude of bending is due not only to the location of load application, but also the boundary conditions. In these two cases the proximal rotation

was constrained. But restriction of rotation of the femoral head is not easily achieved experimentally. Similarly, it is noted that the axially loaded configurations with both ends fixed in rotation (Models 11 and 12) eliminated bending in the plate, resulting in, what we interpret as, unrealistically minimal stressing! Thus, over-constraint of the axial model should also be avoided, leaving Models 8 and 10 as practical candidates.

4.7. Constant and Predictable Loading Along the Length of the Bone

In contrast to configurations generating high bending, an attribute of axial loading may be that it produces more constant loading along the length of the bone (Figure 3). This is beneficial for experimental testing because it reduces the sensitivity of the results to the exact positioning of the osteosynthesis plate along the femur. In fact, a constant, known moment generated by four-point bending is proposed as a testing method for such implants in the ASTM standard [33]. When bending moments vary along the bone (Figure 3), small variations in plate placement can significantly affect stressing of the implant system. The mounted position of the implant is rarely described in detail, thus reducing reproducibility and comparability between studies. It is noted that Model 9, with a fixed distal end, resulted in such a linearly varying bending distribution along the plate (Figure 3), despite our proposal of axial loading, and may not be the best axial configuration.

4.8. Complicated Current Configurations

Femoral head loading configurations often require the femur to be positioned at a specific tilt angle in order to represent the hip joint force direction in a vertical loading machine. In experimental setups, especially when using cadaveric bones, reproducing this exact orientation can be challenging. Small variations in this angle may occur between experiments or between different studies, which can influence the resulting bending moments along the femur. As a result, test setups based on femoral head loading may be less reproducible and more difficult to compare across studies. Instead, axial loading (Models 8–12) may provide a more predictable moment magnitude along the bone, due to a constant moment arm between load and implant axis, that can be easily determined.

4.9. Limitations

One may question why the loading distributions were not determined directly from the finite element models. Our arguments are based on bending moments in the femur and these can only be determined using the analytical model. The elastic continuum FE model was used to determine the reaction forces, since some of the boundary conditions make the system analytically indeterminate. All other parameters were derived analytically, for simplicity, and demonstrate the different magnitudes and effects of bending regardless of absolute accuracy. Indeed, it is noted that the absolute stress and strain magnitudes calculated should be interpreted cautiously because real variations in femoral geometry, cortical thickness, curvature, and subject-specific characteristics are not represented in our model.

Implant stresses were evaluated using a plate-only model. This represents a “worst case”, with no load sharing between the bone fragments, so that plate stressing would be maximised. This model allows plate loading to be assessed at any position along the lateral femoral shaft and compared between the different sets of boundary conditions. The plate would generally be mounted on the bone experimentally to test the biomechanical behaviour of the interface [8].

Forces acting out of the frontal plane were not analysed, partly because they are rarely included in such experimental test setups. The anterior component of the joint load has been measured to be ~10% of overall joint force [7] but is proposed to be neutralised in

the diaphysis by proximal external rotator muscle groups, based on the results of Taylor et al. [11].

Bending is predominantly discussed here, in preference to the axial and shear forces, because bending stresses tend to be orders of magnitude greater than others. It should be acknowledged that muscle activity is unlikely to completely neutralise bending of the skeleton, and stresses may be greater than those generated by the axial force loading proposed here [12,34]. A physiological combination of bending and axial loading occurs in lateral plates due to the axial force acting at a distance from the bone axis. It should be noted that this proposed axial loading mode would eliminate any bending of an intramedullary nail, which would be loaded under axial compression only. But this may be realistic for gait loading, in which the horizontal locking screws and their interfaces are most susceptible to failure by shearing.

Finally, it should be acknowledged that not only one-legged stance is relevant to the preclinical testing of implants. But it does represent the most common high-load activity that a convalescing patient may undertake. And there seems to be little evidence that other activities are specifically represented by any of the commonly employed loading conditions, in which high bending is induced by loading of the femoral head.

5. Conclusions

In conclusion, we suggest that common testing configurations for diaphyseal femoral implants under one-legged stance, in which the joint load is intuitively applied to the femoral head, without simulating muscle forces, causes unrealistically high femoral bending. It also generates implant stressing that is unpredictable and sensitive to implant positioning. We further suggest that loading applied down the axis of the bone, with limited rotation boundary constraints (for example, Model 10, with lateral translation fixed at both ends, or Model 8, with lateral translation fixed at the proximal end and rotation fixed distally, highlighted in grey in Figure 3) offers more realistic reduced femoral loading, which is constant and predictable and is also simpler to set up experimentally. We propose that the consistent use of the axially loaded experimental setups represented by models 8 and 10 to test femoral diaphyseal implants could improve comparability between studies, improve their reproducibility, and better represent physiological loading.

Author Contributions: Conceptualization, T.B. and A.-P.S.; methodology, T.B., M.M. and N.B.; formal analysis, L.V., T.B., M.M. and N.B.; investigation, L.V. and N.B.; writing—original draft preparation, L.V. and N.B.; writing—review and editing, T.B., M.M. and N.B.; visualization, L.V. and N.B.; supervision, N.B. All authors have read and agreed to the published version of the manuscript.

Funding: This work has been supported by a grant from the German Federal Ministry of Research, Technology and Space (BMFTR) as part of the funding line “KMU-innovativ: Medizintechnik” under the contract number 13GW0199 (project SmartFix). Furthermore, the project was funded by the European Union under the Horizon Europe grant agreement No. 101136376 (project SmILE), and from the Swiss State Secretariat for Education, Research and Innovation (SERI). Views and opinions expressed are however those of the authors only and do not necessarily reflect those of the European Union nor European Health and Digital Executive Agency (HaDEA). Neither the European Union nor HaDEA can be held responsible for them.

Institutional Review Board Statement: Not applicable.

Informed Consent Statement: Not applicable.

Data Availability Statement: The raw data supporting the conclusions of this article will be made available by the authors on request.

Conflicts of Interest: The authors declare no conflicts of interest.

Appendix A

Appendix A.1. Bending Stresses

Simple beam theory allows the maximum bending stress σ to be derived from the internal bending moment M , the maximum distance c of the outside surface from the neutral plane, and the second moment of area I around this plane, according to the following:

$$\sigma = M \cdot c / I$$

where for the bone tube

$$I = \frac{\pi}{4} \cdot (R_{\text{outside}}^4 - R_{\text{inside}}^4) \text{ and } c = R_{\text{outside}}$$

and for the plate

$$I = \frac{b \cdot d^3}{12} \text{ and } c = d/2$$

for the respective cross-sectional geometries given below:

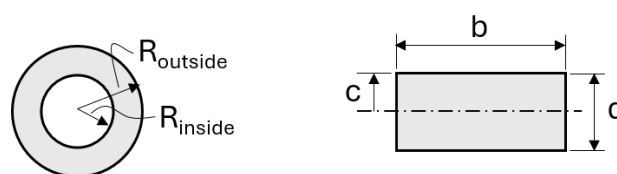


Figure A1. Cross-sectional geometries of the bone (left) and plate (right).

Appendix A.2. Surface Strains

Strains ε are assumed to be linearly proportional to stresses σ , according to the elastic modulus of the material E , where

$$\varepsilon = \sigma / E$$

References

1. Weiss, R.J.; Montgomery, S.M.; Al Dabbagh, Z.; Jansson, K.-Å. National Data of 6409 Swedish Inpatients with Femoral Shaft Fractures: Stable Incidence between 1998 and 2004. *Injury* **2009**, *40*, 304–308. [\[CrossRef\]](#) [\[PubMed\]](#)
2. Lundin, N.; Huttunen, T.T.; Enocson, A.; Marcano, A.I.; Felländer-Tsai, L.; Berg, H.E. Epidemiology and Mortality of Pelvic and Femur Fractures—A Nationwide Register Study of 417,840 Fractures in Sweden across 16 Years: Diverging Trends for Potentially Lethal Fractures. *Acta Orthop.* **2021**, *92*, 323–328. [\[CrossRef\]](#) [\[PubMed\]](#)
3. Enninghorst, N.; McDougall, D.; Evans, J.A.; Sisak, K.; Balogh, Z.J. Population-Based Epidemiology of Femur Shaft Fractures. *J. Trauma Acute Care Surg.* **2013**, *74*, 1516–1520. [\[CrossRef\]](#) [\[PubMed\]](#)
4. Denisiuk, M.; Afsari, A. Femoral Shaft Fractures. In *StatPearls*; StatPearls Publishing: Treasure Island, FL, USA, 2023.
5. Zhu, Y.; Xing, X.; Liu, S.; Chen, W.; Zhang, X.; Zhang, Y. Epidemiology of Low-Energy Wrist, Hip, and Spine Fractures in Chinese Populations 50 Years or Older: A National Population-Based Survey. *Medicine* **2020**, *99*, e18531. [\[CrossRef\]](#) [\[PubMed\]](#)
6. Cnossen, J.D.; Panneman, M.J.M.; Polinder, S.; Verhofstad, M.H.J.; Van Lieshout, E.M.M. Trends in Incidence, Health Care Consumption, and Medical and Productivity Costs of Femoral Shaft Fractures in the Netherlands between 2005 and 2019. *Injury* **2023**, *54*, 111140. [\[CrossRef\]](#) [\[PubMed\]](#)
7. Bonafede, M.; Espindle, D.; Bower, A.G. The Direct and Indirect Costs of Long Bone Fractures in a Working Age US Population. *J. Med. Econ.* **2013**, *16*, 169–178. [\[CrossRef\]](#) [\[PubMed\]](#)
8. Schorler, H.; Wendlandt, R.; Jürgens, C.; Schulz, A.-P.; Kaddick, C.; Capanni, F. Bone Plate-Screw Constructs for Osteosynthesis—Recommendations for Standardized Mechanical Torsion and Bending Tests. *Biomed. Eng. Biomed. Tech.* **2017**, *63*, 719–727. [\[CrossRef\]](#) [\[PubMed\]](#)
9. Bergmann, G.; Graichen, F.; Rohlmann, A.; Bender, A.; Heinlein, B.; Duda, G.N.; Heller, M.O.; Morlock, M.M. Realistic Loads for Testing Hip Implants. *Bio-Med. Mater. Eng.* **2010**, *20*, 65–75. [\[CrossRef\]](#) [\[PubMed\]](#)
10. Pauwels, F. *Biomechanics of the Locomotor Apparatus*, 1st ed.; Springer: Heidelberg, Germany, 1980; ISBN 978-3-642-67140-1.
11. Taylor, M.E.; Tanner, K.E.; Freeman, M.A.R.; Yettram, A.L. Stress and Strain Distribution within the Intact Femur: Compression or Bending? *Med. Eng. Phys.* **1996**, *18*, 122–131. [\[CrossRef\]](#) [\[PubMed\]](#)

12. Duda, G.N.; Schneider, E.; Chao, E.Y.S. Internal Forces and Moments in the Femur during Walking. *J. Biomech.* **1997**, *30*, 933–941. [[CrossRef](#)] [[PubMed](#)]
13. Schorler, H.; Capanni, F.; Gaashan, M.; Wendlandt, R.; Jürgens, C.; Schulz, A.-P. Bone Plates for Osteosynthesis—a Systematic Review of Test Methods and Parameters for Biomechanical Testing. *Biomed. Eng. Biomed. Tech.* **2016**, *62*, 235–243. [[CrossRef](#)] [[PubMed](#)]
14. Inacio, J.V.; Schwarzenberg, P.; Yoon, R.S.; Kantzos, A.; Malige, A.; Nwachuku, C.O.; Dailey, H.L. Boundary Conditions Matter—Impact of Test Setup on Inferred Construct Mechanics in Plated Distal Femur Osteotomies. *J. Biomech. Eng.* **2022**, *144*, 081009. [[CrossRef](#)] [[PubMed](#)]
15. Wisanuyotin, T.; Paholpak, P.; Sirichativapee, W.; Sirichativapee, W.; Kosuwon, W. Effect of Bone Cement Augmentation with Different Configurations of the Dual Locking Plate for Femoral Allograft Fixation: Finite Element Analysis and Biomechanical Study. *J. Orthop. Surg. Res.* **2023**, *18*, 405. [[CrossRef](#)] [[PubMed](#)]
16. Hoffmann, M.F.; Burgers, T.A.; Mason, J.J.; Williams, B.O.; Sietsema, D.L.; Jones, C.B. Biomechanical Evaluation of Fracture Fixation Constructs Using a Variable-Angle Locked Periprosthetic Femur Plate System. *Injury* **2014**, *45*, 1035–1041. [[CrossRef](#)] [[PubMed](#)]
17. Fulkerson, E.; Koval, K.; Preston, C.F.; Iesaka, K.; Kummer, F.J.; Egol, K.A. Fixation of Periprosthetic Femoral Shaft Fractures Associated With Cemented Femoral Stems: A Biomechanical Comparison of Locked Plating and Conventional Cable Plates. *J. Orthop. Trauma* **2006**, *20*, 89–93. [[CrossRef](#)] [[PubMed](#)]
18. O’Connell, R.S.; Owen, J.R.; Hansen, E.J.; Bashir, A.S.; Wayne, J.S.; Satpathy, J.; Kates, S.L. Biomechanical Evaluation of Osteoporotic Proximal Periprosthetic Femur Fractures With Proximal Bicortical Fixation and Allograft Struts. *J. Orthop. Trauma* **2018**, *32*, 508. [[CrossRef](#)] [[PubMed](#)]
19. Wähnert, D.; Schröder, R.; Schulze, M.; Westerhoff, P.; Raschke, M.; Stange, R. Biomechanical Comparison of Two Angular Stable Plate Constructions for Periprosthetic Femur Fracture Fixation. *Int. Orthop.* **2013**, *38*, 47–53. [[CrossRef](#)] [[PubMed](#)]
20. Pierret, M.; Favreau, H.; Bonnomet, F.; Elizondo Moreno, D.; Bahlouli, N.; Martz, P.; Ehlinger, M. Comparison of Five Methods for Locked-Plate Fixation of Complex Diaphyseal Fractures. *Orthop. Traumatol. Surg. Res.* **2022**, *108*, 103400. [[CrossRef](#)] [[PubMed](#)]
21. Windolf, M.; Klos, K.; Wähnert, D.; van der Pol, B.; Radtke, R.; Schwieger, K.; Jakob, R.P. Biomechanical Investigation of an Alternative Concept to Angular Stable Plating Using Conventional Fixation Hardware. *BMC Musculoskelet. Disord.* **2010**, *11*, 95. [[CrossRef](#)] [[PubMed](#)]
22. Polguy, M.; Bliźniewska, K.; Jędrzejewski, K.; Majos, A.; Topol, M. Morphological Study of Linea Aspera Variations—Proposal of Classification and Sexual Dimorphism. *Folia Morphol.* **2013**, *72*, 72–77. [[CrossRef](#)] [[PubMed](#)]
23. Pierre, M.A.; Zurakowski, D.; Nazarian, A.; Hauser-Kara, D.A.; Snyder, B.D. Assessment of the Bilateral Asymmetry of Human Femurs Based on Physical, Densitometric, and Structural Rigidity Characteristics. *J. Biomech.* **2010**, *43*, 2228–2236. [[CrossRef](#)] [[PubMed](#)]
24. Arnone, J.C.; Crist, B.D.; Ward, C.V.; El-Gizawy, A.S.; Pashuck, T.; Della Rocca, G.J. Variability of Human Femoral Geometry and Its Implications on Nail Design. *Injury* **2021**, *52*, 109–116. [[CrossRef](#)] [[PubMed](#)]
25. Unnanuntana, A.; Toogood, P.; Hart, D.; Cooperman, D.; Grant, R.E. Evaluation of Proximal Femoral Geometry Using Digital Photographs. *J. Orthop. Res.* **2010**, *28*, 1399–1404. [[CrossRef](#)] [[PubMed](#)]
26. Bergmann, G.; Deuretzbacher, G.; Heller, M.; Graichen, F.; Rohlmann, A.; Strauss, J.; Duda, G.N. Hip Contact Forces and Gait Patterns from Routine Activities. *J. Biomech.* **2001**, *34*, 859–871. [[CrossRef](#)] [[PubMed](#)]
27. Munih, M.; Kralj, A.; Bajd, T. Bending Moments in Lower Extremity Bones for Two Standing Postures. *J. Biomed. Eng.* **1992**, *14*, 293–302. [[CrossRef](#)] [[PubMed](#)]
28. Chang, P.B.; Williams, B.J.; Bhalla, K.S.B.; Belknap, T.W.; Santner, T.J.; Notz, W.I.; Bartel, D.L. Design and Analysis of Robust Total Joint Replacements: Finite Element Model Experiments with Environmental Variables. *J. Biomech. Eng.* **2001**, *123*, 239–246. [[CrossRef](#)] [[PubMed](#)]
29. Park, Y.; Albert, C.; Yoon, Y.-S.; Fernlund, G.; Frei, H.; Oxland, T.R. The Effect of Abductor Muscle and Anterior-Posterior Hip Contact Load Simulation on the in-Vitro Primary Stability of a Cementless Hip Stem. *J. Orthop. Surg. Res.* **2010**, *5*, 40. [[CrossRef](#)] [[PubMed](#)]
30. Morgan, E.F.; Unnikrisnan, G.U.; Hussein, A.I. Bone Mechanical Properties in Healthy and Diseased States. *Annu. Rev. Biomed. Eng.* **2018**, *20*, 119–143. [[CrossRef](#)] [[PubMed](#)]
31. Niinomi, M. Mechanical Properties of Biomedical Titanium Alloys. *Mater. Sci. Eng. A* **1998**, *243*, 231–236. [[CrossRef](#)]
32. Rubin, C.T. Skeletal Strain and the Functional Significance of Bone Architecture. *Calcif. Tissue Int.* **1984**, *36*, S11–S18. [[CrossRef](#)] [[PubMed](#)]
33. ASTM F382-24; Standard Specification and Test Method for Metallic Bone Plates. ASTM International: West Conshohocken, PA, USA, 2024.
34. Cordey, J.; Borgeaud, M.; Frankle, M.; Harder, Y.; Martinet, O. Loading Model for the Human Femur Taking the Tension Band Effect of the Ilio-Tibial Tract into Account. *Injury* **1999**, *30*, SA26–SA30. [[CrossRef](#)] [[PubMed](#)]

Disclaimer/Publisher’s Note: The statements, opinions and data contained in all publications are solely those of the individual author(s) and contributor(s) and not of MDPI and/or the editor(s). MDPI and/or the editor(s) disclaim responsibility for any injury to people or property resulting from any ideas, methods, instructions or products referred to in the content.