



Should Plate-Bone Gap be Preserved in Far-Cortical Locking Technique? A Biomechanical Study

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Abstract

Purpose The use of a locking plate eliminates excessive pressure on bone for anatomical reduction and thus preserves the periosteal blood supply, which is important for fracture healing. The far-cortical locking technique with a semi-rigid locking screw reduces structural stiffness and parallel motion, allowing uniform callus formation at the fracture site. Although previous studies demonstrated the superior clinical and biomechanical outcomes of semi-rigid locking screws over rigid ones, it is unclear whether a gap between the plate and bone should be preserved in the far-cortical locking technique.

Methods The present study conducted finite element analyses with mechanical calibration to clarify the influence of a plate-bone gap on the biomechanical performance of the far-cortical locking technique. A simulated mid-shaft fracture model was fixed using a locking plate and six semi-rigid locking screws. The plate-bone distance was 0 to 2 mm and the axial compressive load was 500 N.

Results Gliding guidance at the plate-bone interface enhanced parallel intersegmental motion but reduced intersegmental movement, which is a mechanical stimulant for callus formation, and may increase pressure on the bone. Screw stresses increased with increasing plate-bone gap distance.

Conclusion For the far-cortical locking technique, the results suggest a minor plate-bone gap should be preserved. Engagement between the plate and bone should be avoided both before and after the application of mechanical load.

Keywords Far-cortical locking · Semi-rigid locking screw · Gap · Stiffness · Parallel motion · Finite element analysis · Biomechanics

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1 Introduction

The application of locking plates has greatly changed surgical techniques for orthopedic trauma [1–3]. For a compression plate, anatomical reduction with tight binding between the bone and plate is necessary to achieve sufficient fixation stability. The use of a locking plate creates a structure with great mechanical strength and thus preserves the periosteal blood supply [3], which is a key factor in bone healing. Clinical outcomes for compression plates depend on primary (direct) bone healing, whereas those for locking plates depend on secondary bone healing [1, 3, 4]. However, several studies found no differences in the stiffness of fixation structure between locking and compression plates [5, 6], caused the delayed or non-union which may need extra surgical treatments, implant fractures, alignment loss [7–11]. The high stiffness of a fixation structure that uses a locking plate reduces the efficiency of bone healing. To reduce structural stiffness, the far-cortical locking technique, where fixation is done only in the far-cortex region using a semi-rigid locking screw, has been proposed [12–15]. The parallel movement at the fracture site in the far-cortical locking technique removes the problem of uneven callus formation caused by conventional locking plates and screws [15, 16].

Nevertheless, the plate-bone gap remains an issue for the locking plate technique. A biomechanical study found that an excessive plate-bone gap jeopardizes the safety of a fixation structure [17]. When a semi-rigid locking screw is used, the performance of the fixation structure becomes more complicated and difficult to anticipate. Besides structural stiffness, the influence of the plate-bone gap on parallel motion at the fracture site should be examined. The present study investigates whether the plate-bone gap should be preserved in dynamic fixation using a semi-rigid screw from a biomechanical perspective. Structural stiffness, parallel motion at the fracture site, and screw stress are discussed.

2 Material and Methods

2.1 Virtual Model Construction

A cylindrical diaphyseal mid-shaft fracture model was constructed with a hollow cortex layer (outer diameter: 30 mm; cortex thickness: 7.0 mm; Young's modulus: 15,100 MPa; Poisson's ratio: 0.3) filled with solid to simulate cancellous bone (Young's modulus: 100 MPa; Poisson's ratio: 0.3). The total length of the cylindrical model was 230 mm, with a fracture gap of 10 mm in the middle of the shaft [12, 18].

A simple bone plate (length: 212 mm; width: 19 mm; thickness: 7 mm) construct was built with 12 screw holes

(screw hole span: 16 mm). A commercial 42-mm semi-rigid locking screw (DDTL screw, A Plus Biotechnology Co., Ltd., Taiwan) was reconstructed and the screw threads were removed (Fig. 1). Three screws were inserted into the 1st, 3rd, and 5th screw holes, respectively, on the bone plate to fix the upper shaft, and three screws were inserted into the 8th, 10th, and 12th screw holes, respectively, to fix the lower shaft. The material properties of a titanium alloy were assigned to all metallic components (Young's modulus: 110,000 MPa; Poisson's ratio: 0.3). Screws were fixed on the shaft in the far-cortex region. A 5.0-mm tunnel was retained for dynamic motion of the screw shaft (3.4 mm) at near-cortex regions (Fig. 2).

Three gap distances between the bone and the bone plate, namely 0, 1, and 2 mm, were used (Fig. 3). The corresponding models are denoted as G00, G01, and G02, respectively.

2.2 Finite Element Modeling

The three models were meshed with tetrahedral elements in the commercial software ANSYS Workbench 17.0 (ANSYS Inc., Canonsburg, PA, USA). A convergence test was performed with an axial compressive load of 500 N applied to the structure. The von Mises stress on the bone plate/screws and the axial displacement in the interfracture region were taken into convergence concern. The final numbers of element numbers were 165,330 for the G00 model, 166,102 for the G01 model, and 165,983 for the G02 model. Element sizes were 1.5 mm (globally), 1.3 mm (bony construct), and 0.85 mm (metallic implants).

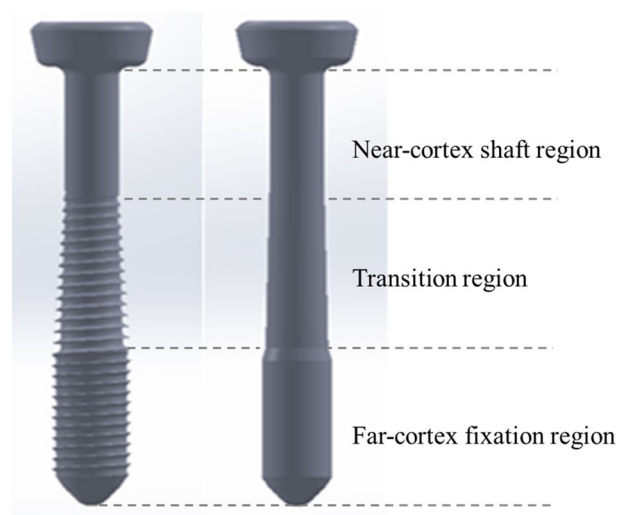


Fig. 1 Model of commercial dynamic locking screw (left). The threaded region was simplified (right) for finite element analyses

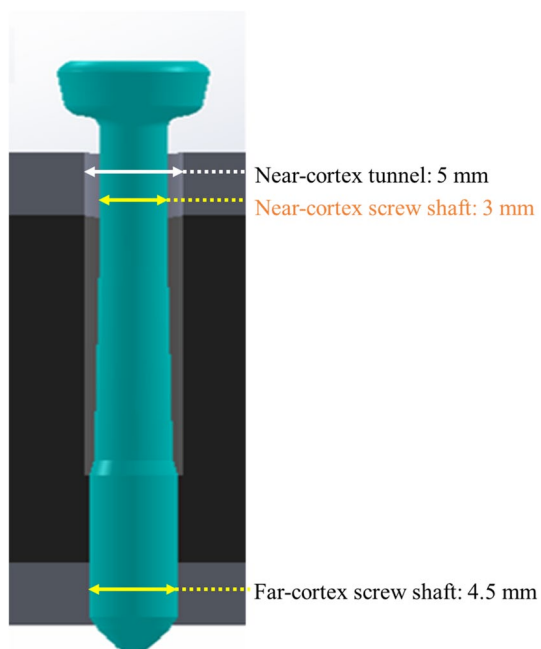
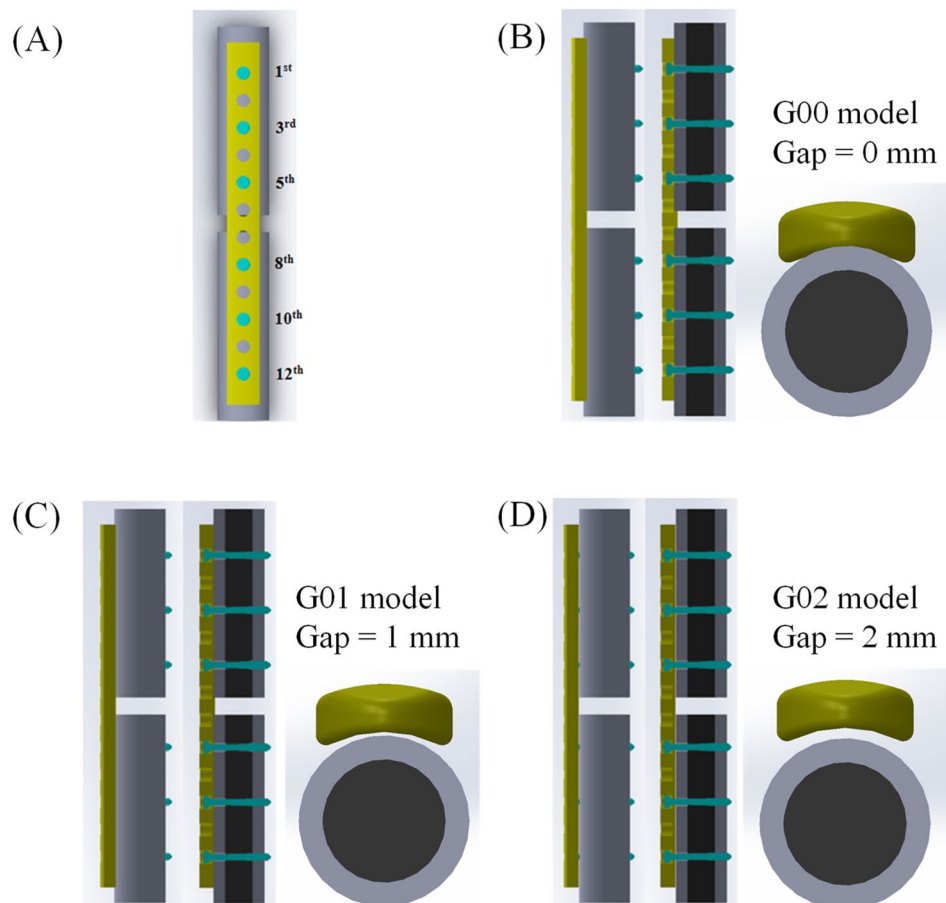


Fig. 2 Illustration of screw insertion. The tip of the screw is fixed at the far cortex of the bony shaft and clearance is retained in the tunnel region

Fig. 3 **A** Configuration of locking plate and semi-rigid locking screws. The side views and cross-sectional views of **B** G00, **C** G01, and **D** G02 models represent the correlation of bone and implants



2.3 Boundary and Loading Condition

The contact between the bone plate and the bony shaft was set to be frictionless. A coefficient of friction of 0.5 was set for the fracture gap in case of contact after load application. The screw heads were set to be bonded with the bone plate. For the screw body, the far-cortex region was bonded with the bony shaft. The screw shaft at near-cortex region was frictionless, and the transition region had a coefficient of friction of 0.5 against the bony tunnel.

A compressive axial load was applied to the top of the bony shaft. The initial load was 100 N; the load was increased in steps of 100 N until it reached 500 N. The bony shaft model was fixed at the bottom of the lower fragment. In the finite element analyses, the stresses on the implants were recorded and the axial displacement of the load-exertion surface on the model was used to calculate the structural stiffness. Interfragmentary motion was evaluated by recording the fracture site displacements on near- and far-cortical sides, calculated as tilting angle to clarify if ideal parallel motion can be preserved under different fixation settings.

Fig. 4 Configuration of the mechanical test



2.4 Model Calibration

The results from the simulated models were calibrated using a practical mechanical test with identical parameters for the screws (DDTL screw, A Plus Biotechnology Co., Ltd.), bone plate (ABS Locking Plate, A Plus Biotechnology Co., Ltd.), and surrogate bony structure. Metallic implants were fixed onto composite sawbones [1, 12–15] (Sawbones #3403; Pacific Research Laboratories, Vashon, Washington) with a length of 110 mm for both the upper and lower shafts. A 10-mm osteotomy gap was preserved [12]. Plates were fixed at 2 mm from the bone surface (Fig. 4). The 42-mm screws were secured onto the bone plate to a torque of 4 Nm using a torque limiter. A static axial compression test was conducted with a 0.1 mm/s load rate until a 500-N load was reached. Mechanical test was conducted one-time on a single sample, and the load–displacement correlation and the calculated axial stiffness from the mechanical test were used to calibrate the aforementioned simulated models.

3 Results

3.1 Model Calibration

The load–displacement results from the mechanical tests (obtained under identical geometrical and loading conditions) were applied for calibrating the G02 model in the finite element analysis. The load–displacement curve for the mechanical test (average of three sets of mechanical test data) and that for the G02 model are plotted in Fig. 5.

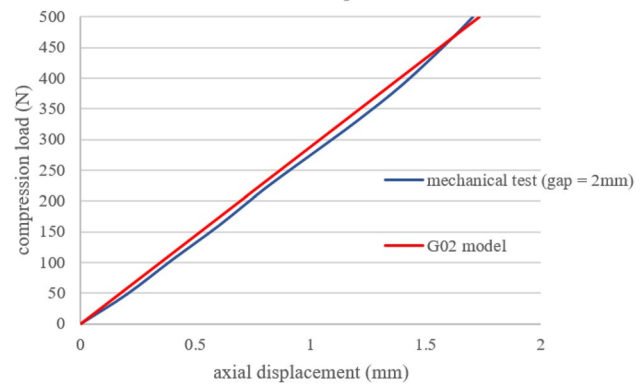


Fig. 5 Load–displacement curves for mechanical test and finite element analysis of G02 model. The two curves are similar

Table 1 Structural stiffness of simulated finite element models

Model	Stiffness (N/mm)
G00 (gap=0 mm)	381.7
G01 (gap=1 mm)	315.3
G02 (gap=2 mm)	288.8

Table 2 Interfragmentary motion at simulated fracture sites

Model	Interfragmentary motion (mm)	
	Near-cortex	Far-cortex
G00 (gap=0 mm)	1.00	1.62
G01 (gap=1 mm)	1.17	2.00
G02 (gap=2 mm)	1.29	2.17

The structural stiffness was 293.3 ± 11.35 N/mm (range: 284.15–306.02 N/mm) for the mechanical test specimens and 288.8 N/mm for the G02 model. These values are similar, indicating that the finite element analysis results are reliable.

3.2 Structural Stiffness

The simulation results show that the structural stiffness decreases with increasing plate-bone gap distance. The structural stiffness was largest for the G00 model (381.7 N/mm); those for the G01 and G02 models were 17.3% and 24.3% lower, respectively (Table 1).

3.3 Interfragmentary Motion

Table 2 shows the interfragmentary motion in the near- and far-cortex regions. Movements increased with increasing plate-bone gap distance. The largest movements were observed for the G02 model. The calculated tilting angle,

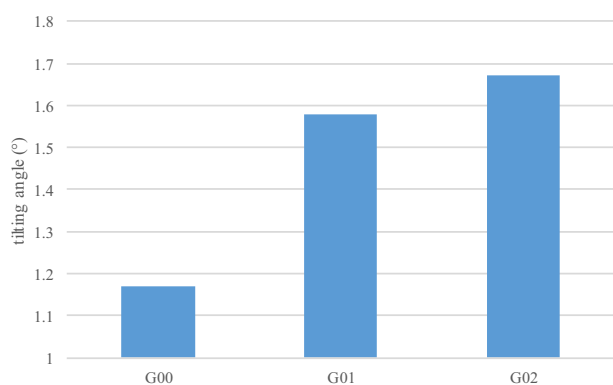


Fig. 6 Comparison of tilting angles at the fracture site to evaluate parallel motion between the fractured segments

Table 3 Comparisons of screw stresses

Screw location	von Mises stress (MPa)		
	G00 model (gap=0 mm)	G01 model (gap=1 mm)	G02 model (gap=2 mm)
1st hole	448.8	438.3	454.9
3rd hole	453.3	497.0	510.9
5th hole	479.0	542.3	558.0
8th hole	482.5	562.7	583.7
10th hole	461.0	475.1	492.7
12th hole	466.1	459.3	470.9

used to evaluate parallel motion, increased with the elevation of bone plate. The tilting angles were respectively 1.17° , 1.58° , and 1.76° for the G00, G01, and G02 models (Fig. 6).

3.4 Stress on Semi-rigid Locking Screws

The von Mises stresses on the semi-rigid screws are listed in Table 3. The two center screws (at the 5th and 8th screw holes) had the greatest stress for all models. The largest stresses at the screws adjacent to the fracture gap were found for the G02 model (558.0 and 583.7 MPa), followed by the G01 (542.3 and 562.7 MPa) and G00 (479.0 and 482.5 MPa) models.

4 Discussion

The periosteal blood supply preserves active bone healing [3]. The use of a locking plate preserves this important mechanism, allowing effective fracture healing [1–3]. The far-cortical locking technique, which replicates the biomechanical behavior and biological healing response of external fixator by internal fixation [19], increases mechanical stimulation and dramatically enhances secondary bone

healing [12–15]. However, it is unclear whether the clearance spared for the periosteal blood supply affects the efficiency of the far-cortical locking technique. The present study conducted biomechanical evaluations to determine whether the locking plate should be flush with the bone or a plate-bone gap should be preserved. The comprehensive comparisons of structural stiffness, interfragmentary motion, and screw stress provide surgeons with information to optimize the far-cortical locking technique.

When a locking plate is used to reconstruct fracture segments, sufficient interfragmentary motion is required for secondary bone healing [20]. This can be most directly achieved by reducing mechanical strength or structural stiffness. A previous biomechanical study measured stiffness for various gap distances between the locking plate and the bone; stiffness similar to that achieved with a compression plate was found when the locking plate was flush with the bone [16]. When the gap distance was too large, structural safety was jeopardized. Biomechanical and clinical studies suggest a maximum plate-bone gap of 1 to 2 mm [12, 14, 15]. With the far-cortical locking technique, the stiffness of the fixation structure is reduced by enlarged screw deformation within the screw hole tunnel [12–15]. Compared to the structural stiffness of the G02 model with a 2-mm plate-bone gap, those of the G01 and G00 models were 9.2% and 32.1% higher, respectively. The performance of the model with a locking plate flush to the bone is similar to that found in a biomechanical study for a conventional locking screw. When the plate is flush to the bone, there is no clearance for radial plate-bone approaching after a mechanical load is applied, so the bending deformation of the semi-rigid screws is limited.

Parallel motion between the fractured segment is important for adequate callus formation to achieve uniform healing in the near- and far-cortex regions [12–16, 21]. A previous animal study compared the efficiency of fracture healing between rigid and semi-rigid locking screws and found that the far-cortical locking technique with semi-rigid screws enhanced the quality of callus formation and decreased the recurrence of fracture after plate removal [13]. In the present study, the tilting angle between the fractured segments was 1.67° (compared to complete parallel motion) for the G02 model. When the gap was reduced to 1-mm (G01 model), the tilting angle of the fractured segments decreased to 1.60° (– 5.4%). Without a gap (G00 model), the tilting angle further decreased to 1.17° (– 29.9%). This obvious difference was caused by the effect of gliding guidance when the plate is engaged with bone under an axial compressive load. Because a thick locking plate model (7 mm) was utilized in the present study, the plate was strong enough to prevent large deformation; the fractured segments were thus forced to glide along the interior surface of the plate, decreasing the tilting angle. Although this seems to be ideal, the insufficient radial clearance at the plate-bone interface

will increase contact pressure at the interface and thus suppress the periosteal blood supply, which may lead to bone loss even though the plate does not initially strongly press on the bone. In addition, if the plate is pre-pressed to the bone when the semi-rigid screws are secured, the pressure-induced resistance at the plate-bone interface may interfere with gliding guidance and thus result in uneven interfragmentary motion between the near and far cortex.

Stress on the screw can be a safety indicator. The mechanical calibration conducted in the present study did not reach forces that could break the screw or plastically damage the structure. Therefore, the reported screw stress is an estimation and not direct evidence of screw safety. According to the simulation results, the two center screws (at the 5th and 8th screw holes) adjacent to the fracture site had the largest stresses for all models (558.0 and 583.7 MPa, 542.3 and 562.7 MPa, and 479.0 and 482.5 MPa for the G02, G01, and G00 models, respectively). The relatively low stresses for the G00 model resulted from insufficient screw deformation because of the lack of radial clearance at the plate-bone interface. The stresses on the other screws (away from the fracture site) were lower.

Mechanical calibration was conducted for the finite element model. Nevertheless, some limitations of the present study should be noted. First, only the performance of the G02 model was calibrated due to limited resources (available specimens for mechanical tests), the evaluation of biomechanical influences basing on the calibrated model and simple variations from the model should be considerable. Second, the load applied in the mechanical calibration test did not reach the magnitude required for implant failure, so the results of the finite element analysis cannot be used to infer component safety. In addition, the present study focused on parallel motion at the fracture site. The load was applied and calibration was performed only for evaluating the initial stiffness, as described in the study by Bottlang et al. [12]. Third, the current study evaluated the influence of plate-bone gap on implant stresses and interfragmentary motion by computational simulation. The fatigue durability and safety were not validated and should be based on practical biomechanical test. Finally, the configuration of the testing specimens was not as precise as that of the three-dimensional models in the simulation. Nevertheless, the results of the present study can be used as a reference by clinicians and in future studies.

5 Conclusion

The results of biomechanical evaluations suggest that for the far-cortical locking technique, the locking plate should be placed as near to the bone as possible to ensure parallel motion between fractured segments, which promotes

uniform callus formation. In the present study, a 1-mm plate-bone gap was found to be preferable than a 2-mm gap in view of implant safety (lower screw stress) and capability in maintaining parallel interfragmentary motion. Although the slightly engaged plate-bone interface may induce gliding guidance for parallel interfragmentary motion, pressure loading on the bone surface should be avoided to preserve the periosteal blood supply for greater efficiency of fracture healing.

Author Contributions All authors contributed to the study conception and design. Material preparation, data collection and analysis were performed by S-PW and C-CL. The first draft of the manuscript was written by C-CL and all authors commented on previous versions of the manuscript. All authors read and approved the final manuscript.

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Declarations

Conflict of interest The authors have no relevant financial or non-financial interest to disclose.

Ethical Approval None to declare.

Consent to Participate None to declare.

Consent to Publish None to declare.

References

1. Claes, L. (2011). Biomechanical principles and mechanobiologic aspects of flexible and locked plating. *Journal of Orthopaedic Trauma*, 25(S1), S4–S7.
2. Frigg, R., Appenzeller, A., Christensen, R., Frenk, A., Gilbert, S., & Schavan, R. (2001). The development of the distal femur Less Invasive Stabilization System (LISS). *Injury*, 32(S3), SC24–SC31.
3. Perren, S. M. (2002). Evolution of the internal fixation of long bone fractures. The scientific basis of biological internal fixation: choosing a new balance between stability and biology. *The Journal of Bone and Joint Surgery. British*, 84, 1093–1110.
4. Claes, L. E., Heigele, C. A., Neidlinger-Wilke, C., Kaspar, D., Seidl, W., Margevicius, K. J., et al. (1998). Effects of mechanical factors on the fracture healing process. *Clinical Orthopaedics and Related Research*, 355(Suppl), S132–S147.
5. Fitzpatrick, D. C., Doornink, J., Madey, S. M., & Bottlang, M. (2009). Relative stability of conventional and locked plating fixation in a model of the osteoporotic femoral diaphysis. *Clinical Biomechanics (Bristol, Avon)*, 24, 203–209.
6. Stoffel, K., Booth, G., Rohrl, S. M., & Kuster, M. (2007). A comparison of conventional versus locking plates in intraarticular calcaneus fractures: A biomechanical study in human cadavers. *Clinical Biomechanics (Bristol, Avon)*, 22, 100–105.
7. Button, G., Wolinsky, P., & Hak, D. (2004). Failure of less invasive stabilization system plates in the distal femur: A report of four cases. *Journal of Orthopaedic Trauma*, 18, 565–570.
8. Henderson, C. E., Bottlang, M., Marsh, J. L., Fitzpatrick, D. C., & Madey, S. M. (2008). Does locked plating of periprosthetic supracondylar femur fractures promote bone healing by callus

- formation? Two cases with opposite outcomes. *Iowa Orthopaedic Journal*, 28, 73–76.
9. Henderson, C. E., Kuhl, L. L., Fitzpatrick, D. C., & Marsh, J. L. (2011). Locking plates for distal femur fractures: Is there a problem with fracture 272 healing? *Journal of Orthopaedic Trauma*, 25(Suppl 1), S8–S14.
 10. Lujan, T. J., Henderson, C. E., Madey, S. M., Fitzpatrick, D. C., Marsh, J. L., & Bottlang, M. (2010). Locked plating of distal femur fractures leads to inconsistent and asymmetric callus formation. *Journal of Orthopaedic Trauma*, 24, 156–162.
 11. Vallier, H. A., Hennessey, T. A., Sontich, J. K., & Patterson, B. M. (2006). Failure of LCP condylar plate fixation in the distal part of the femur. A report of six cases. *Journal of Bone and Joint Surgery. American Volume*, 88(4), 846–853.
 12. Bottlang, M., Doornink, J., Fitzpatrick, D. C., & Madey, S. M. (2009). Far cortical locking can reduce stiffness of locked plating constructs while retaining construct strength. *Journal of Bone and Joint Surgery. American Volume*, 91, 1985–1994.
 13. Bottlang, M., Lesser, M., Koerber, J., Doornink, J., von Rechenberg, B., Augat, P., et al. (2010). Far cortical locking can improve healing of fractures stabilized with locking plates. *Journal of Bone and Joint Surgery. American Volume*, 92, 1652–1660.
 14. Döbele, S., Gardner, M., Schröter, S., Höntzsch, D., Stöckle, U., & Freude, T. (2014). DLS 5.0—The biomechanical effects of dynamic locking screws. *PLoS ONE*, 9, e91933.
 15. Doornink, J., Fitzpatrick, D. C., Madey, S. M., & Bottlang, M. (2011). Far cortical locking enables flexible fixation with periarticular locking plates. *Journal of Orthopaedic Trauma*, 25(Suppl 1), S29–S34.
 16. Adams, J. D., Jr., Tanner, S. L., & Jeray, K. J. (2015). Far cortical locking screws in distal femur fractures. *Orthopedics*, 38(3), e153–e156.
 17. Ahmad, M., Nanda, R., Bajwa, A. S., Candal-Couto, J., Green, S., & Hui, A. C. (2007). Biomechanical testing of the locking compression plate: When does the distance between bone and implant significantly reduce construct stability? *Injury*, 38, 358–364.
 18. Moazen, M., Mak, J. H., Jones, A. C., Jin, Z., Wilcox, R. K., & Tsiridis, E. (2013). Evaluation of a new approach for modelling the screw-bone interface in a locking plate fixation: A corroboration study. *Proceedings of the Institution of Mechanical Engineers. Part H*, 227(7), 746–756.
 19. Bottlang, M., & Feist, F. (2011). Biomechanics of far cortical locking. *Journal of Orthopaedic Trauma*, 25(Suppl 1), S21–S28.
 20. Egol, K. A., Kubiak, E. N., Fulkerson, E., Kummer, F. J., & Koval, K. J. (2004). Biomechanics of locked plates and screws. *Journal of Orthopaedic Trauma*, 18(8), 488–493.
 21. Ries, Z., Hansen, K., Bottlang, M., Madey, S., Fitzpatrick, D., & Marsh, J. L. (2013). Healing results of periprosthetic distal femur fractures treated with far cortical locking technology: A preliminary retrospective study. *Iowa Orthopaedic Journal*, 33, 7–11.