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Reducing femoral peri-implant fracture risk through optimized plate length and screw configuration – a biomechanical study

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Abstract

Background Locked plating of femur fractures is associated with secondary peri-implant fractures which may be a result of stress concentrations at the proximal plate end region. The aim of this study was to investigate whether the strength of healed femoral bone-locking-compression-plate constructs can be increased by modifying the screw configurations and plate length to minimize the risks of peri-implant femur fractures.

Methods The detached shaft of a variable angle condylar locking compression plate (VA-LCP Condylar Plate; Johnson & Johnson MedTech) was fixed to the proximal two-third of twenty-four intact artificial femurs in four different configurations ($n=6$) distinguished by either using a short plate with cortical or locking screws whereby the most proximal screw was inserted in the femoral shaft 50 mm below the lesser trochanter, or using a long plate with either cortical or locking screws whereby the most proximal screw was positioned in the femoral neck. Simulating a situation after fracture healing, constructs were cyclically tested under progressively increased loading until catastrophic failure.

Results Long plates fixed with a cortical screws demonstrated the highest failure load (1091 N ± 142 N) which was significantly higher compared to long plates fixed with locking screws (888 N ± 80 N), short plates fixed with cortical screws (471 N ± 42 N), and short plates fixed with locking screws (450 N ± 19 N). In addition, whereas the locking screw construct with a long plate was associated with a significantly higher failure load compared to both short plate constructs, there were no significant differences between the latter two. The failure modes were predominantly characterized by neck screw pull-out in both long plate constructs and peri-implant bone fractures at the most proximal screw in both constructs with short plates. None of the specimens exhibited a femoral neck fracture.

Conclusion The findings of this study performed on synthetic bones indicate that from a biomechanical perspective long plates that extend into the femoral neck sustained higher failure loads compared to short plates. In addition, long plates fixed with a cortical neck screw further enhanced the construct strength and reduced the risk of peri-implant fractures compared to the use of a locking neck screw. Therefore, this study supports the use of long locking plates combined with use of cortical neck screws, particularly in high-risk patients, such as those with severe osteoporosis.

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Keywords Secondary fracture, Peri-implant fracture, Distal femur fracture, Locking compression plate, Femur plate, Plate osteosynthesis, Neck screw, Biomechanics

Introduction

Peri-implant fractures can occur after treatment of femur fractures with locking plates. Osteoporotic patients older than 50 years are particularly affected with an incidence reaching 7% after the treatment of distal femur fractures [1, 2]. Peri-implant fractures usually require another surgical intervention associated with increased mortality and morbidity in this older patient group [3–5]. The reason for these fractures is the difference in stiffness between the fragile bone and the much stiffer bone-plate construct [1, 3, 6, 7]. This difference creates high stress zones under bending that lead to fractures at the most proximal screw [1, 3, 6, 7].

The risk of peri-implant fractures likely persists until metal removal [1, 6]. Removing the metal would eliminate the stress zones caused by differences in stiffness. However, this could lead to a secondary fracture through an empty screw hole [1], and a second surgical intervention is associated with perioperative complications particularly in this age group.

Two previous studies showed that for synthetic bone cylinders the resistance against bending forces can be increased if a cortical screw is used at the end of the plate instead of the usual locking screw [6, 8]. The cortical screw may allow a slightly higher flexibility and, thereby, a smoother stiffness transition. Moreover, a cortical screw with its free screw angle of 15° can be positioned bicortically even when the plate is not correctly aligned, thus reducing the risk of a peri-implant fracture even more [9]. However, two further biomechanical studies on synthetic bone cylinders that investigated variations of end screws could not report a significant difference regarding the use of cortical versus locking end screws [10, 11]. All these studies used synthetic bone cylinders and further research using anatomical bone models and clinical studies is still needed. Other biomechanical studies addressing this issue observed predominantly failure modes other than peri-implant fractures (plate bending, screw pull-out, other mechanical failures), or no catastrophic failures [12–19], limiting the relevance of these findings on peri-implant fracture risk.

Alternative to plates ending in the shaft region, long plates with the most proximal screw extending into the femoral neck can be used. In this case, the whole femur would be stabilized, reducing stiffness differences and stress. However, empirical evidence is lacking for such constructs and the related analysis requires use of anatomical bone models.

Therefore, the aim of this study was to investigate on synthetic specimens whether the strength of femoral bone-locking-compression-plate constructs can be increased by modifying screw configurations and plate lengths to minimize the risks of peri-implant femur fractures after healing.

Methods

Specimens

Twenty-four synthetic proximal femora (LD 2220, Synbone AG, Zizers, Switzerland) were used. The bone model with low cortical density and soft cancellous bone was selected to reflect the most severely affected osteoporotic patient group. To simulate secondary fractures after fracture healing, a fracture gap was omitted. Four variations of locking compression plate constructs ($n = 6$) were tested on proximal femora: short plate with locking screws (SL), short plate with cortical screws (SC), long plate with locking screws (LL), long plate with cortical screws (LC). In short plates, the uppermost screw was positioned 50 mm below the lesser trochanter, while in long plates, it was placed within the femoral neck. Each plate length was used in one group receiving 5.0 mm locking screws and another group receiving 4.5 mm cortical screws (Fig. 1).

Surgical technique

All 24 specimens were instrumented by the same surgeon according to the manufacturer's guidelines using stainless steel (316 L) Variable Angle Locking Compression Condylar Plates (VA-LCP 4.5/5.0) and corresponding screws by the same manufacturer (Johnson & Johnson MedTech, Zuchwil, Switzerland).

For specimens with short plates, the plate was positioned on the lateral side of the femur such that the most proximal screw was located 50 mm below the lesser trochanter. Afterwards, the three further distal plate holes were occupied. For specimens with long plates and neck screws, the plates were bent and positioned to ensure proper alignment with the bone and maintain a screw angulation of no more than 15° (plate-neck screw angle of 75°–90°). Following plate positioning, the 90 mm long neck screw was inserted under fluoroscopic guidance to ensure centered neck screw positioning in anteroposterior and mediolateral directions. Afterwards, the four remaining shaft screws were positioned on the same height as the screws in the short plate groups. All shaft screws were 44 mm long ensuring bicortical fixation. The 5.0 mm locking screws were tightened as specified by the manufacturer. For the 4.5 mm cortical screws, the ideal screw torque of 80% of the previously determined stripping torque was applied [20]. All femora were cut

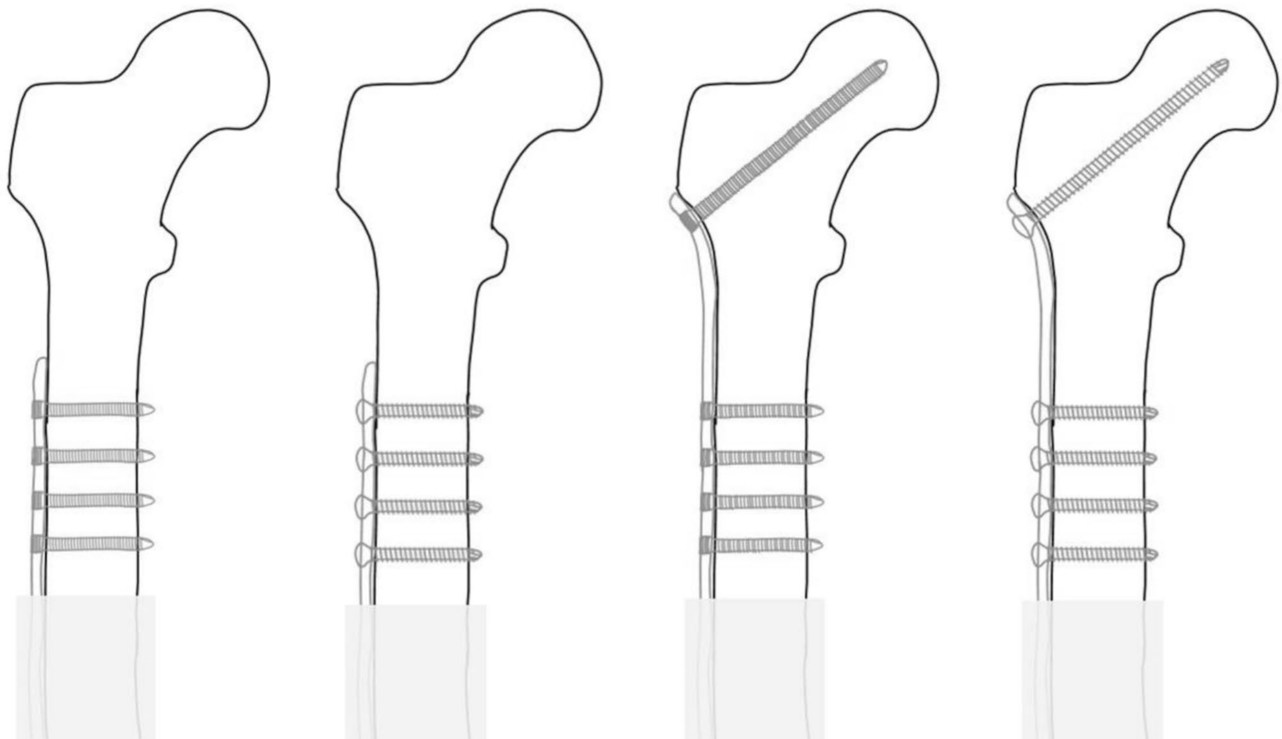

Fig. 1 Four variations of locking compression plate constructs tested in the current study (from left to right): short plate with locking screws (SL), short plate with cortical screws (SC), long plate with locking screws (LL), long plate with cortical screws (LC)

to a length of 29 cm and the plates were shortened as needed at the distal end. The plate and femoral shaft were embedded together in a 65 mm high and 48 mm wide polymethylmethacrylate (PMMA; SCS-Beracryl D28, Suter Kunststoffe AG, Fraubrunnen, Switzerland) base.

Biomechanical testing

Biomechanical testing was performed on a servohydraulic testing machine (858 Mini Bionix®II, MTS Systems Corp., Eden Prairie, USA) equipped with a 5 kN load cell (MCS 10, HBM, Darmstadt, Germany). The femora were positioned in 17° adduction and 11° flexion to mimic the main vector of the hip joint reaction force during human gait [21]. The femoral head was loaded in compression along the machine axis via a spherically shaped, greased PMMA cup attached to the machine actuator and the load cell (Fig. 2). The distal end was connected to the machine frame via a cardan joint to prevent transmission of torques and shear forces. The loading protocol was adapted from previous studies [22–25] and comprised an initial quasi-static ramp in compression from 50 N to 100 N at 5 N/s, followed by cyclic loading at 2 Hz with a double-peaked physiological compression profile of each cycle. Keeping the valley load of each cycle at a constant level of 50 N throughout the whole test, the peak load, starting at 100 N, was progressively increased cycle by cycle at a rate of 0.1 N/cycle until the stop criterion of 40 mm displacement was fulfilled, which was found in previously conducted pilot study sufficient to provoke catastrophic failures in the constructs.

Data acquisition & evaluation

Machine data in terms of axial displacement and axial load were acquired at a rate of 200 Hz throughout the tests. Axial stiffness of the bone-implant construct was derived from the ascending slope of the load-displacement curve within a linear region of the initial quasi-static ramp. Based on the cyclic tests, the numbers of cycles until 2 mm, 3 mm, and 5 mm displacement, reached with respect to the machine displacement at the beginning of the cyclic test, were evaluated together with the corresponding loads, considered in loaded condition and defined as failure loads according to these three criteria. Furthermore, the numbers of cycles until onset of catastrophic failure were determined from machine data together with the corresponding catastrophic failure load. Finally, the modes of catastrophic failure were analyzed macroscopically.

Statistical evaluation among the parameters of interest was performed using Python 3.12.3 (Python Software Foundation, Delaware, USA). The normal distribution of the data within each group was screened and proved using the Shapiro-Wilk test. Explorative outcome measures are reported in terms of mean value and standard

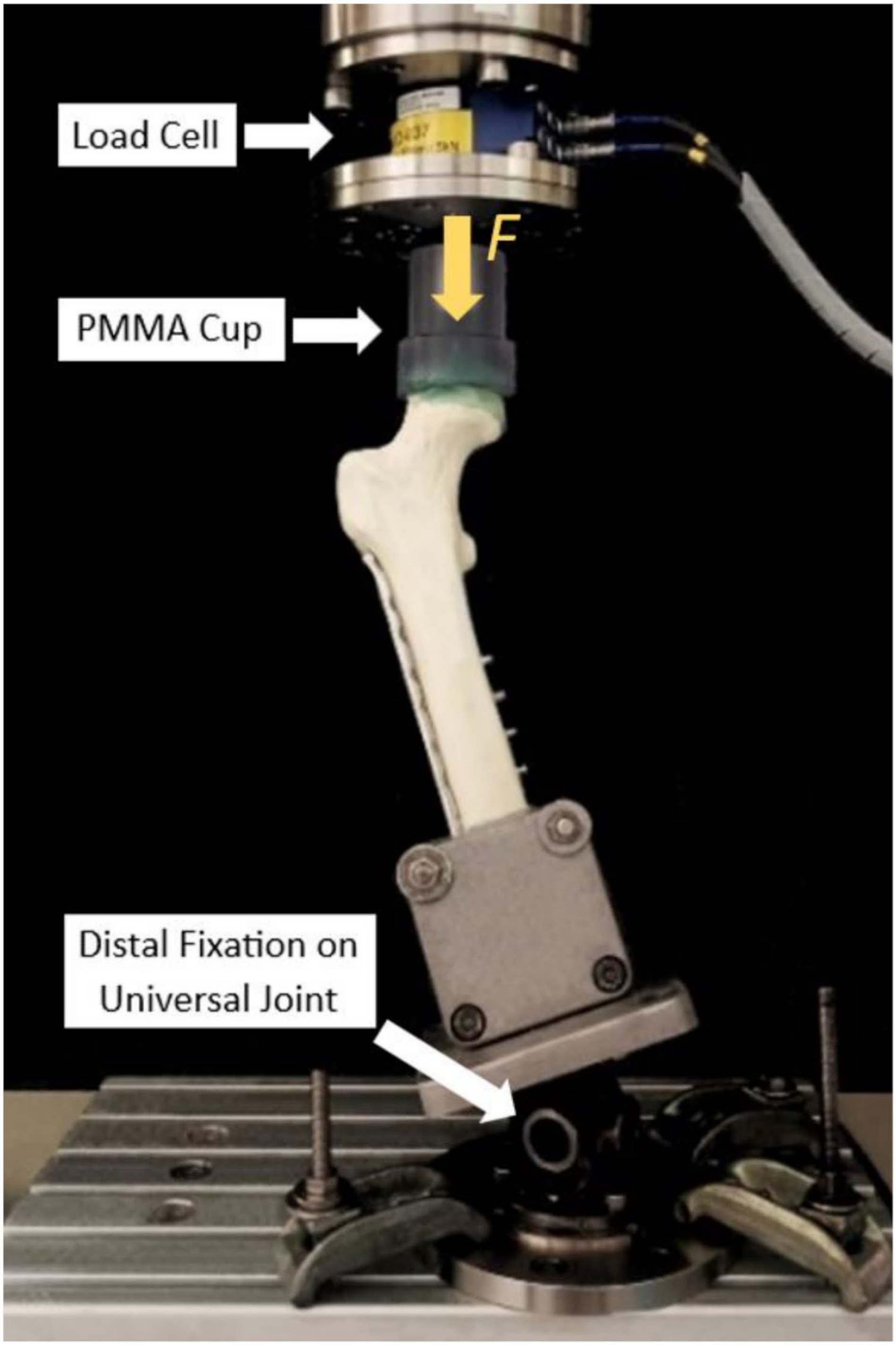

Fig. 2 Experimental setup with a specimen instrumented with a long plate and cortical screws, and mounted for biomechanical testing. Vertical arrow denotes loading (F) direction

deviation (SD). A General Linear Model (GLM) Repeated Measures (RM) test was conducted to explore the differences between the groups for cycles to 2 mm, 3 mm, and 5 mm machine displacement, and to corroborate within each group the machine displacement progression over time. Significant differences in catastrophic failure loads between the four instrumentation techniques were identified using One-Way Analysis of Variance (ANOVA) tests and Tukey's Honest Significant Difference (HSD) test with Šidák post-hoc p-value correction for multiple comparisons. The general significance level was set at 0.05 for all statistical tests.

Table 1 Numbers of cycles to 2 mm, 3 mm and 5 mm of displacement shown in terms of mean value and standard deviation for the four groups: short plate with locking screws (SL), short plate with cortical screws (SC), long plate with locking screws (LL), long plate with cortical screws (LC)

	SL	SC	LL	LC
Cycle at 2 mm displacement	1698 ± 213	1767 ± 347	2784 ± 499	2910 ± 467
Cycle at 3 mm displacement	2793 ± 274	2859 ± 468	4423 ± 636	4586 ± 641
Cycle at 5 mm displacement	4224 ± 121	4431 ± 542	7259 ± 742	7547 ± 849

Results

The numbers of cycles to 2 mm, 3 mm, and 5 mm machine transducer displacement are summarized in Table 1. The differences between the groups for cycles to 2 mm, 3 mm, and 5 mm machine displacement, and the machine displacement progression over time within each group were all significant ($P < 0.001$). The two groups using locking screws (SL and LL) showed only non-significantly lower cycle counts ($P \geq 0.383$) before reaching these displacement thresholds compared to their respective groups with cortical screws.

The two groups with long plates reached approximately two-fold loads at onset of catastrophic failure compared to short plates (Fig. 3). Long plates with locking screws (LL) reached significantly lower loads at onset of catastrophic failure (888 N ± 80 N) compared to long plates with cortical screws (LC), which reached the highest loads (1091 N ± 142 N) among the four groups. Short plates with locking screws (SL) withstood the lowest loads (450 N ± 19 N), followed by short plates with cortical screws (SC; 471 N ± 42 N). All pairwise comparisons between the four groups showed significant differences ($P \leq 0.036$), except the comparison of the two screw types for short plates (SL and SC; $P = 0.999$). The corresponding numbers of cycles until catastrophic failure were 4,356 ± 256 for SL; 4,637 ± 443 for SC; 9,330 ± 1,147 for LL; and 11,760 ± 1,660 for LC.

All specimens experienced catastrophic failure with a bone fracture before reaching the test stop criterion. All constructs with short plates exhibited failure due to peri-implant fractures of the bone at the most proximal

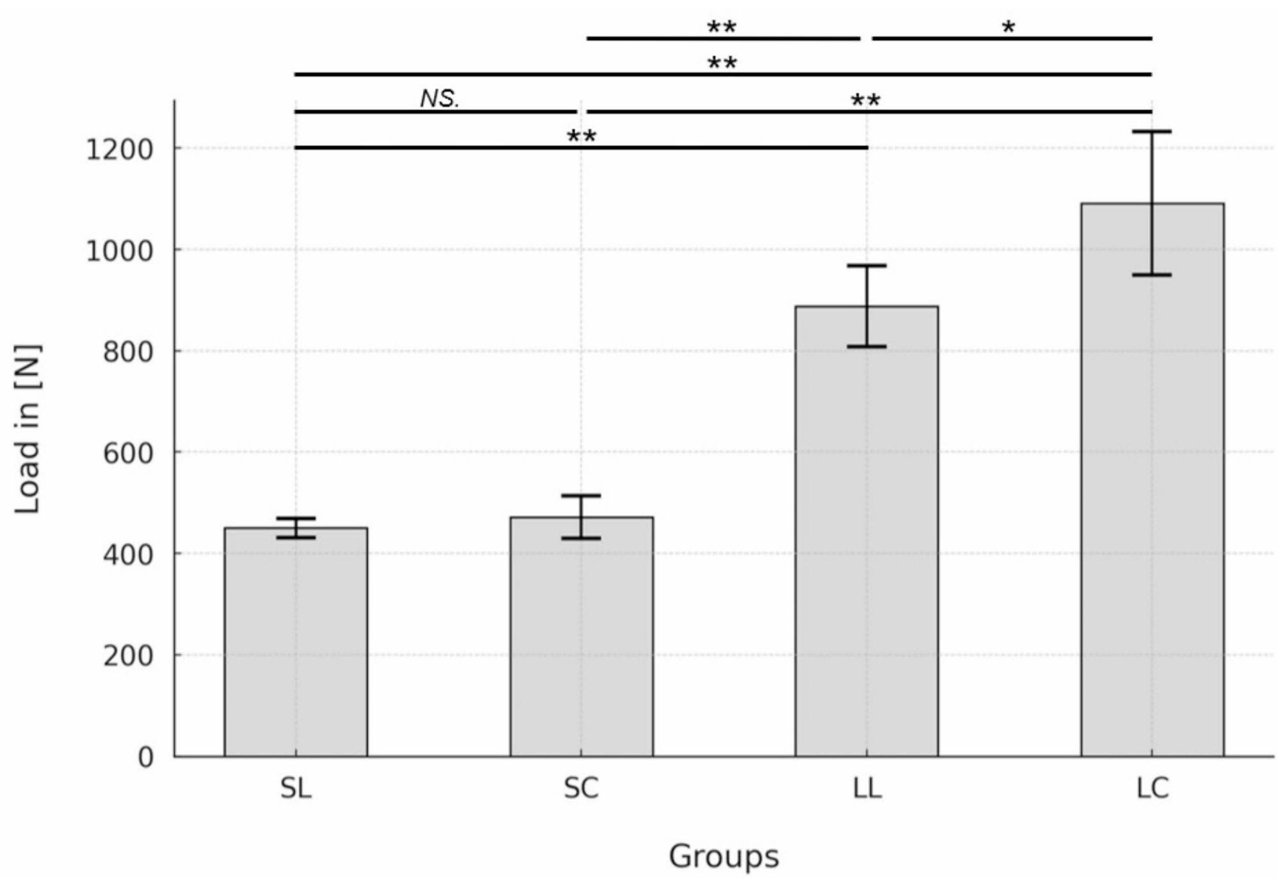

Fig. 3 Loads at onset of catastrophic failure of each group presented in terms of mean value and standard deviation (short plate with locking screws (SL), short plate with cortical screws (SC), long plate with locking screws (LL), long plate with cortical screws (LC)). Asterisks denote statistical significance: ** P < 0.001, * P < 0.05, NS: not significant

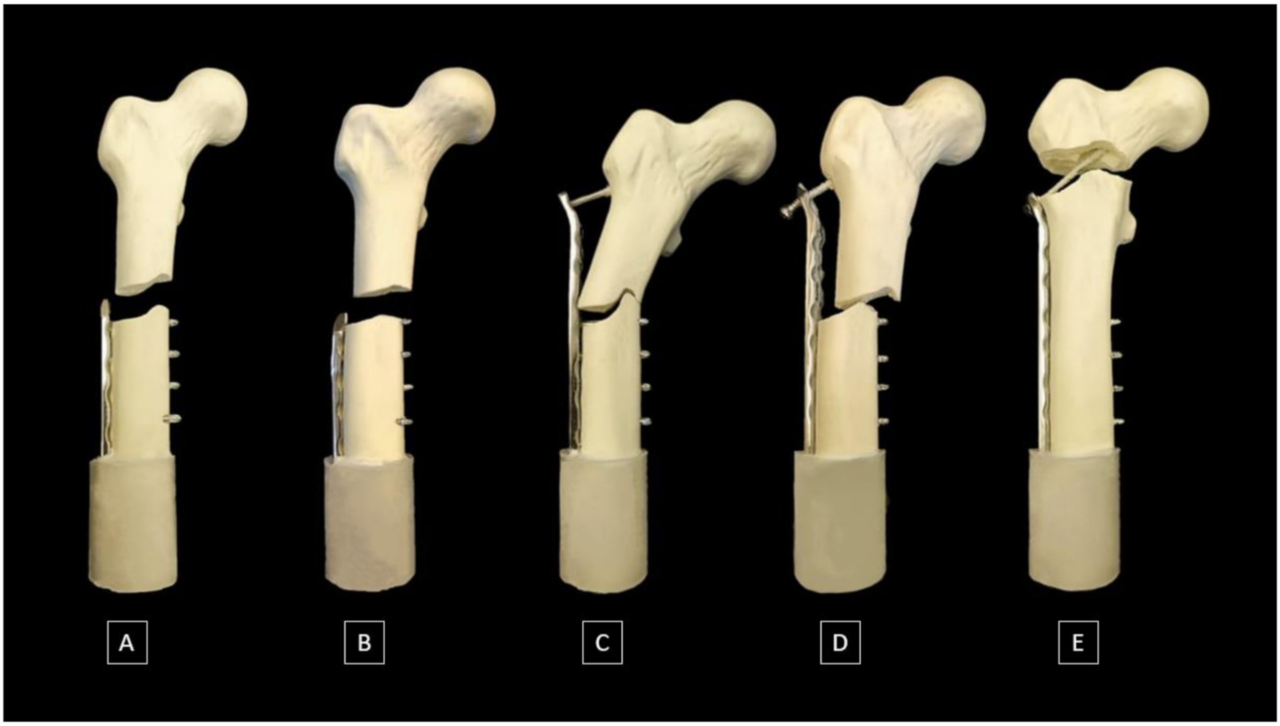

Fig. 4 Photographs of the different failure mechanisms. From left to right: all specimens (6/6) in SL group exhibited a peri-implant fracture (**A**), all specimens (6/6) in the SC group also exhibited a peri-implant fracture (**B**), all specimens (6/6) in the LL group exhibited pull-out of the neck screw and fracture at the most proximal shaft screw (**C**), 4/6 specimens in the LC group exhibited pull-out of the neck screw and fracture at the most proximal shaft screw (**D**), 2/6 specimens in the LC group exhibited a peri-implant fracture at the neck screw (**E**)

screw. These fractures were characterized as slightly oblique spiral ones, initiating precisely at the center of the screw beneath the plate and extending slightly proximally to the opposite medial cortex. Notably, for the constructs with short plates and cortical screws the fractures were observed to be slightly more oblique compared to those seen for the constructs with short plates and locking screws. Ten out of twelve specimens with long plates failed due to pull-out of the neck screw, followed by a fracture at the most proximal shaft screw (Fig. 4). It could be observed visually that no neck screw bending happened during the pull-out, this occurred later when the fracture dislocated due to axial compression of these specimens. In contrast, the other two specimens with long plates and cortical screws failed due to peri-implant intertrochanteric fractures. Notably, these two specimens exhibited the highest failure loads. None of the tested specimens exhibited a femoral neck fracture.

Discussion

This study compared the mechanical characteristics of two screw configurations and two different plate lengths in Variable Angle Locking Compression Condylar Plates (VA-LCP 4.5/5.0) to identify designs reducing risk of secondary peri-implant fractures in synthetic proximal femora. Our results demonstrated that long plate constructs with the most proximal screw inserted in the femoral neck resisted higher loads until failure, and therefore provided better protection of the bone than short plates terminating at the shaft. This is a logical consequence of the mechanical construct enforcement by protecting a longer distance up to the femoral head. Under loading, whereas the bone transmits a higher portion of the forces to the implant via the neck screw construct, it does not exhibit load sharing in this region in the short plate constructs.

Short plates all failed due to peri-implant fractures at the most proximal screw. We observed only non-significant advantages in terms of loads and numbers of cycles until reaching displacements of 2 mm, 3 mm, and 5 mm for the short plates with cortical versus locking screws. However, short plates with proximal cortical end screws demonstrated less oblique fracture lines compared to proximal locking end screws, suggesting a different bone behavior during the development of peri-implant shaft fractures. For cylinders with fracture gaps, previous studies showed either no difference [10, 11] or superiority of constructs with proximal cortical end screws [6, 8]. However, cylinders do not reflect the anatomical shape of human bones and fracture gaps recapitulate a transient stage of a few weeks after osteosynthesis, when peri-implant fractures are rare [26]. This limits the relevance of cylinders with fracture gaps for investigating the risk of peri-implant fractures. An underpowered clinical study showed four peri-implant fractures of distal femora for proximal locking end screws and none for proximal cortical end screws among 88 patients, however, these numbers were too low to yield conclusive results [1].

The cortical configuration with long plates in the current study demonstrated the highest loads and cycle counts before reaching catastrophic failure. Failures eventually occurred, predominantly due to the pull-out of the neck screw, followed by fractures at the most proximal shaft screw. The performance variance between cortical and locking neck screws may be due to differences in the pull-out forces. This could be attributed to the greater thread depth of cortical screws (0.65 mm for a 4.5 mm screw using a 3.2 mm drill bit for pre-drilling), compared to locking screws (0.35 mm for a 5.0 mm screw with a 4.3 mm drill bit for pre-drilling). The deeper cortical screw threading could provide better anchorage especially in osteoporotic conditions with a thin cortex [27–29]. In good-quality bones, the likelihood of screw pull-out is lower for both 4.5 mm cortical screws and 5.0 mm locking screws [30, 31]. Therefore, the failure mechanism could change in constructs with less pull-out risk to fractures above the proximal neck screw, a fracture pattern seen in two specimens of the current study (Fig. 4). Another factor is the use of variable angle locking screws to achieve a center-center position of the neck screw. Previous studies have shown that these screws tend to disengage from the locking mechanism earlier than perpendicular fixed-angle locking screws [32–37]. Therefore, classic perpendicular fixed-angle locking screws in the neck could probably offer greater stability but require perfect, precise plate pre-bending, which is technically more demanding. In addition, none of the specimens in the present study suffered a femoral neck fracture. The neck screw may provide additional stability in the femoral neck and could not only prevent peri-implant fractures, but also femoral neck fractures in high risk patients. However, further studies are needed to confirm this hypothesis.

The concept with use of long plates in the current study was meant to achieve spanning from the distal femur condyle to the proximal femur, and therefore to protect the whole femur. A similar approach has been described by Ma et al. as a plate-on-plate technique in a retrospective study, where 11 patients were successfully treated with this method [38]. This is a therapeutic method for treating peri-implant fractures by maintaining the original plate osteosynthesis and adding a second overlapping plate that spans the entire femur. The plate-on-plate technique addresses complications post-occurrence. In contrast, the present study focuses on preventing these peri-implant fractures before they occur using extended plates in the initial surgical intervention. Additionally, due to the lack of biomechanical studies on the plate-on-plate technique, there are uncertainties about the outcomes when combining these implants as described.

The current work has some limitations: Synthetic bone models do not fully capture the complex properties of human bones, such as differences in density and microarchitecture [39], and bone remodeling around the metal following osteosynthesis. This is an important limitation, since bone remodelling could differ between various end screw types, affecting the stress zones and fracture risk. Additionally, synthetic bones do not represent the interpatient variability in bone geometry, which could lead to different load distributions and fracture patterns. Nevertheless, we chose synthetic bones because they show significantly less variability under all types of loading (< 10%) than human cadaveric bones, thus increasing statistical power [40]. Furthermore, the pull-out behavior in synthetic bones is comparable to that of real bones [41]. We used a model with low cortical density and soft cancellous bone to represent the osteoporotic patient risk group and to ensure occurrence of a peri-implant fracture with no other failure mechanisms. This resulted in rather low failure loads (238 N to 290 N). We also only compared femoral plate osteosyntheses that extend into the femoral neck with those that end at the femoral shaft 50 mm below the lesser trochanter. No other neck screw angles or other plate lengths were tested, such as those ending further distally on the shaft. Furthermore, we did not include a control group with intact non-instrumented femora and our number of specimens per group was relatively low. Additionally, the study design omitted a fracture gap to simulate a healed state, thereby neglecting the immediate postoperative period. This precludes a direct comparison of construct failure risks to the direct post-operative phase. A disadvantage of using long plates with neck screws is the technically more demanding plate bending and positioning, which prolongs the surgical intervention. Additionally, the use of a long plate is more invasive and results in a larger surgical wound.

Future research should expand to real bone specimens and explore not only different implant configurations but also the effects of varying patient-specific factors such as bone quality, bone geometry and comorbid conditions. Moreover, introducing loading conditions that more closely simulate fall situations could provide deeper insights into the performance of these implants in vivo. Further investigation of the stress zones above the end screw at the shaft in short plate constructs could provide more information on the mechanism of these peri-implant fractures and the advantages and disadvantages of different screw variations at the plate end. Exploring various neck screw angles or multiple neck screws in long plate constructs could potentially reduce the observed pull-out effect.

Conclusion

From a biomechanical perspective, long plates that extend into the femoral neck sustained higher failure loads compared to short plates under cyclic loading in synthetic femora. In addition, long plates fixed with a cortical neck screw further enhanced the construct strength and reduced the risk of peri-implant fractures compared to the use of a locking neck screw. Therefore, this study supports the use of long locking plates combined with the use of cortical neck screws, particularly in high-risk patients, such as those with severe osteoporosis.

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Author contributions
Conceptualization KS; methodology LL, IZ, LP; formal analysis LP; data curation LP; writing—original draft preparation LP; writing—review and editing KS, BG, IZ; visualization LP; supervision KS, BG. All authors have read and agreed to this version of the manuscript.

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Data availability
No datasets were generated or analysed during the current study.

Declarations

Ethics approval and consent to participate
Not applicable, this study does not report on or involve the use of any animal or human data or tissue.

Consent for publication
Not applicable, this manuscript does not contain data from any individual person.

Competing interests
The authors declare no competing interests.

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