J Orthop Sci
DOI 10.1007/s00776-015-0735-3

ORIGINAL ARTICLE

Periprosthetic femoral fracture fixation: a biomechanical comparison between proximal locking screws and cables

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Received: 18 December 2014 / Accepted: 28 April 2015

Abstract
Background The incidence of periprosthetic femoral fractures (PFF) around a stable stem is increasing. The aim of this biomechanical study was to examine how three different methods of fixation, for Vancouver type B1 PFF, alter the stiffness and strain of a construct under various configurations, in order to gain a better insight into the optimal fixation method.
Methods Three different combinations of proximal screws and Dall–Miles cables were used: (A) proximal unicortical locking screws alone; (B) proximal cables and unicortical locking screws; (C) proximal cable alone, each in combination with distal bicortical locking screws, to fix a stainless steel locking compression plate to five synthetic femora with simulated Vancouver type B1 PFFs. In one synthetic femora, there was a 10-mm fracture gap, in order to simulate a comminuted injury. The other four femora had no fracture gap, to simulate a stable injury. An axial load was applied to the constructs at varying degrees of adduction, and the overall construct stiffness and surface strain were measured.
Results With regards to stiffness, in both the gap and no gap models, method of fixation A was the stiffest form of fixation. The inclusion of the fracture gap reduced the stiffness of the construct quite considerably for all methods of fixation. The strain across both the femur and the plate was considerably less for method of fixation C, compared to A and B, at the locations considered in this study.
Conclusion This study highlights that the inclusion of cables appears to damage the screw fixations and does not aid in construct stability. Furthermore, the degree of fracture reduction affects the whole construct stability and the bending behaviour of the fixation.

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Introduction

The worldwide incidence of intra-operative and post-operative periprosthetic femoral fractures (PFFs) around hip prostheses is between 0.1 and 6 % of all total hip arthroplasties (THAs) [1]. This incidence is likely to increase as more THA are performed. It is estimated that currently more than 800,000 THAs are carried out every year worldwide [2].

There is no single universal classification system used in the management of PFF, although the Vancouver classification is most commonly used in current orthopaedic practice [4–6]. In this system, fractures are subdivided into Types A, B and C. Type B refers to fractures within the stem length, and is subclassified into B1 when the implant is stable, B2 when the implant is unstable and B3 when the bone stock is inadequate.

The application of locking compression plates (LCPs) in the treatment of PFFs is a relatively new concept, having been introduced in 2000, with the first results reported in 2003 [7]. LCPs have an advantage over conventional plates that can be applied at a distance over the periosteum, because the screws lock onto the plate, preserving the periosteal blood supply and thus enhancing fracture healing. In addition, the fixation of the locked screw to the plate provides improved angular stability to ensure secure anchoring in osteoporotic bone. Based on these comparative advantages, LCPs have become the preferred method of treatment for Vancouver B1 (VB1) PFFs [8], with clinical data reporting overall union rates of 91 % [9].

The efficiency of LCPs has been questioned by various researchers, since incorrect surgical application of the plate may produce a very rigid fixation, which prevents bone healing by secondary intention [10–13]. Some authors have advocated less-rigid fixation around the fracture site [10], or the use of cables proximally instead of screws. However, a combination of proximal screws and cables may appeal in the case of unstable fractures. Fracture stability and loading configuration have been thought to play a crucial role in the performance of any type of fracture fixation device [14, 15], yet to the best of our knowledge, no study has investigated these effects in PFF fixations.

Furthermore, no optimal fixation method has yet been established, and there has been limited biomechanical or clinical data comparing various methods of screw and cable attachment [16, 17]. Little is known about the clinical and biomechanical differences between cable and screw fixation using locking plates. Therefore, the aim of this biomechanical study is to compare the performance of (A) proximal unicortical locking screw fixation alone, versus (B) proximal cable and unicortical screw fixation, versus (C) proximal cable fixation alone in stable and unstable VB1 PFF fixation under three different loading conditions. Overall construct stiffness and strain on the locking plate were compared between the cases to gain better insight into the optimal fixation method.

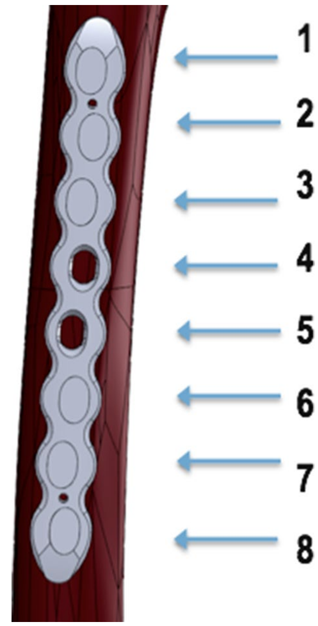

Fig. 1 The Stryker eight-hole locking plate showing the numbering of the holes

Materials and methods

Five synthetic sawbone femora (Sawbone 4th Generation, Pacific Research Laboratories Inc, USA) were secured in stainless steel modules using polymethylmethacrylate (PMMA) cement. After preparation of the proximal femur, an Exeter Stem (Stryker SA, Montreux, Switzerland) was implanted using PMMA cement. A VB1 fracture was simulated by a transverse osteotomy 10 mm distal to the tip of the stem. The VB1 fracture was stabilised by a single surgeon using an eight-hole LCP (Stryker SA, Montreux, Switzerland), Fig. 1. The following three methods of fixation were used.

Method A

The LCP was fixed with three proximal unicortical screws (holes 1–3), and three bicortical screws distally (holes 6–8) (Fig. 2a).

Method B

The LCP was fixed with three proximal unicortical screws (holes 1–3), three proximal Dall–Miles cables (holes 1–3) and three bicortical screws distally (holes 6–8) (Fig. 2b).

Method C

The LCP was fixed with three Dall–Miles cable (holes 1–3) and three bicortical screws distally (holes 6–8) (Fig. 2c).

The Dall–Miles cables were applied by resting the cable over a custom cable slot in the screw hole of the plate, and tensioned using a custom tensioner to apply a tension of 150 N.

Four of the five specimens (Specimen 1–4) were fixed as if anatomically reduced, referred to here as the ‘no gap model’ and also representing a stable fracture pattern. To simulate the worst-case scenario, where there is no bony contact across the fracture, a gap of 10 mm was left across the fracture site in a single specimen (Specimen 5), referred to as the ‘gap model’. This represents an unstable fracture pattern.

Eight uniaxial strain gauges (ST 1-8) (GFLA-3-50, Tokyo Sokki Kenkyujo, Tokyo, Japan) were attached using cyanoacrylate adhesive to the gap model (Fig. 3). Strain was only measured in the gap model because it was anticipated that the values here would be considerably larger than in the no gap model, and therefore could be more reliably measured. All gauges were positioned so that their axes were aligned with the longitudinal axis of the femur.

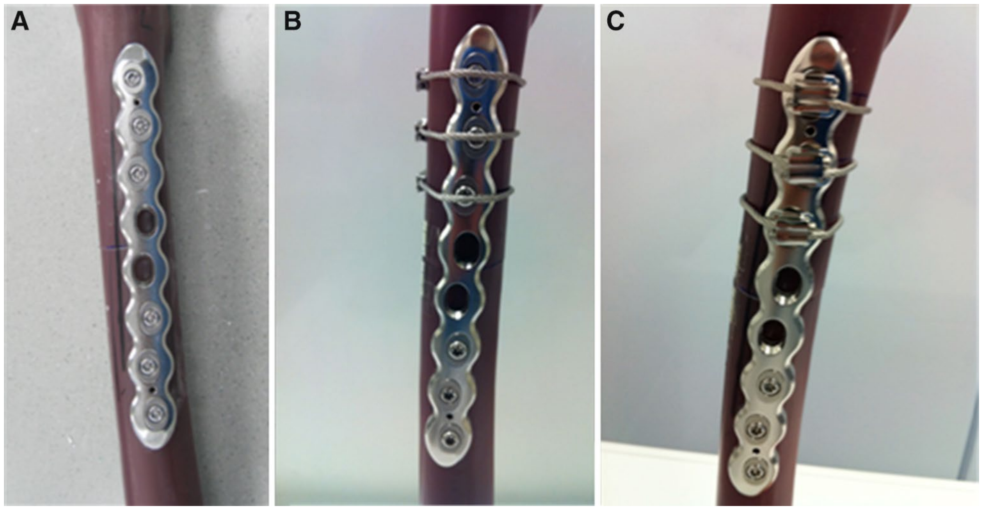

Fig. 2 Method of fixation **a** Stryker eight-hole LCP fixed with three proximal unicortical screws (holes 1–3), and three bicortical screws distally (holes 6–8). Method of fixation **b** Stryker eight-hole LCP fixed with three proximal unicortical screws (holes 1–3), three proximal Dall–Miles cables (holes 1–3) and three bicortical screws distally (holes 6–8). Methods of fixation **c** Stryker eight-hole LCP fixed with three Dall–Miles cable (holes 1–3) and three bicortical screws distally (holes 6–8)

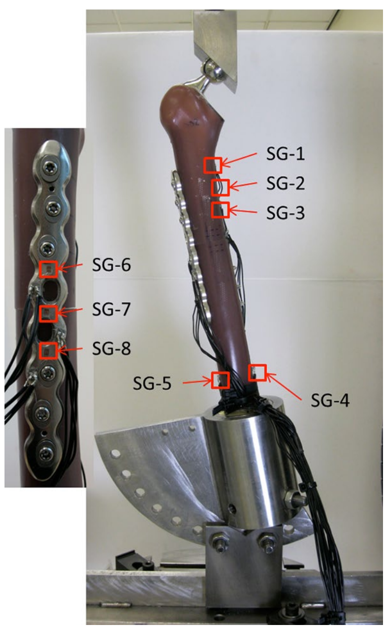

Fig. 3 Position of the strain gauges 1–8 on S5

Five gauges with a gauge length of 3 mm were attached to the surface of the femur. Four gauges were positioned on the medial length of the femur, at 0, 40, 80, and 200 mm distal to the lesser trochanter, with an additional gauge positioned on the lateral side of the femur, 200 mm distal to the lesser trochanter. This arrangement was used to capture the strain magnitude in the proximal and distal bony fragment. Three strain gauges with a gauge length of 0.3 mm were then positioned on the eight-hole stainless steel LCP, one above hole 4, one above hole 5 and one above hole 6. These gauges were positioned in order to capture the strain around the empty screw holes and below the most proximal screw, where there is the highest risk of fixation failure.

A material testing machine (Instron 8874, Instron, Norwood, MA, USA) was used to load the femoral constructs. An axial load was applied via a hemispherical cup to the femoral head stem, at varying degrees of adduction, 0°, 10° and 20° in the no gap model, and 10° in the gap model, this being the most anatomical position. Using displacement control, a vertical load was applied at the femoral head at 5 mm/min, to a maximum subclinical level of 500 N, in order to keep the specimens within the linear elastic range and avoid damage. The constructs were preconditioned over 20 loading cycles and were then loaded six more times, during which the appropriate measurements, i.e., stiffness and strain, were taken.

From the original, no load position, the load was applied and the construct reached its maximum deformation, i.e., bending as a result of axial loading, at 500 N. Construct stiffness was calculated as the slope of the linear part of the load–displacement data that was obtained from the material testing machine, i.e., the difference between the final loading (500 N) and initial loading (0 N) was divided by the difference between the corresponding displacements. Strain values were measured based on the signals from strain gauges that were amplified, recorded and converted to microstrain at 500 N using LabVIEW (National Instruments, TX, USA). Both stiffness and strain values were recorded over the six repeated loading cycles and averaged in each case.

Results

All planned tests were successfully completed and there was no incomplete data. There were no observed failures of the specimens and at no point was fixation lost. A comparison of the mean stiffness of the constructs using the three fixation methods in the no gap model and the stiffness of the construct using three fixation methods in the gap model, are presented in Fig. 4. The recorded strain across the plate and bone in the fracture gap model for all three methods of fixation is demonstrated in (Fig. 5a, b).

With regards to the stiffness, the most striking finding was that in the no gap model, method of fixation A was clearly the stiffest form of fixation at all loading angles. Furthermore, the difference in stiffness between method B and C in the no gap model was very small, as is indicated by the relatively small standard deviations. However, in the gap model, the differences in stiffness, measured at 10°, between all methods of fixation were very low.

The inclusion of the fracture gap reduced the stiffness of the construct quite considerably at 10° adduction for all methods of fixation. These results suggest that the fracture-gap and no gap models behave differently, and the results from the gap model are not simply an amplification of those of the no gap model.

Strain was only measured in the gap model, loaded at 10° adduction. The strain recorded on both the femur and the plate was very similar for fixation methods A and B. However, the strain was considerably lower on the plate and the femur for method of fixation C, compared to A and B.

Discussion

The aim of this biomechanical study was to examine how the three different methods of fixation for stable and unstable PFFs around a stable stem alter the stiffness and strain of the fixation.

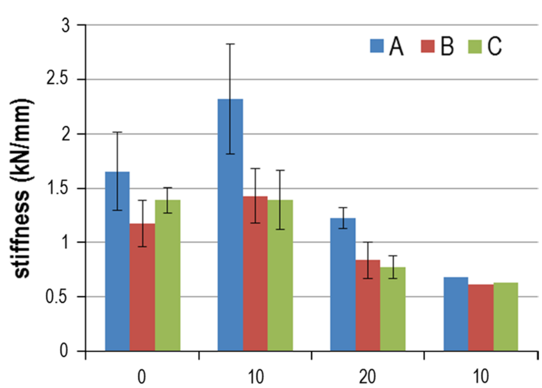

Fig. 4 The stiffness of the no gap model loaded at three different adduction angles; the means are shown (*error bars* = ±SD), and the stiffness of the gap model loaded at 10°

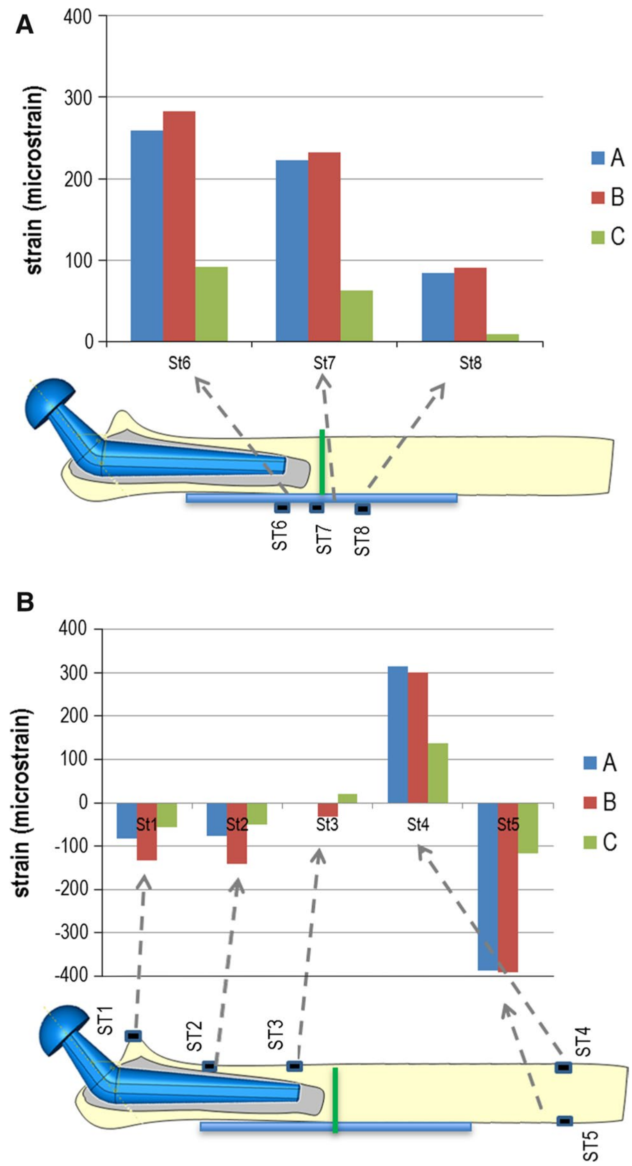

Fig. 5 **a** The recorded strain on the plate for specimen 5. **b** The recorded strain on the bone for specimen 5

The most clear finding from this study was that in the no gap model, simulating a stable injury, method of fixation A was considerably stiffer than methods B and C, when loaded at 0°, 10° and 20°. This suggests that when using proximal unicortical screws, the addition of cables most likely results in the screws being pushed into the bone as the cables are applied, and subsequently the screws loose their fixation onto the bone. The results therefore imply that the unicortical screws become obsolete in the fixation when used in combination with cables. Results from the gap model go some way to reinforcing this theory. Conventional surgical wisdom would suggest that fixation method

B would be stiffer than method A. However, the stiffness of the two is almost identical, implying that while the addition of cables increases the overall stiffness, that advantage is lost in equal measure through changes in screw fixation.

Unicortical screws are easier to apply and are an overall less invasive method of fixation, with less soft tissue stripping than cables; they therefore provide a more desirable form of fixation. In addition, cables may produce fritting corrosion against the plate, may loosen and brake with the time, and also could strangulate the periosteal blood supply. Considering stability of fixation only, these results indicate that proximal unicortical screws provide the optimal method of fixation when managing VB1 PPF in both a stable and an unstable injury.

With regards to strain, this was only measured in the gap model loaded at 10°, the most anatomical loading position. Common reported sites of fixation failure are the plate in the region close to the fracture and also the distal femur [9–11]. Method of fixation C had considerably lower strains recorded across all three strain gauges on the plate, compared to methods A and B, which were very similar. Furthermore, method of fixation C also recorded lower strain readings on the femur compared to A and B. While these results may suggest that proximal cables alone place less strain across all areas of the fixation, it must be noted that there might have been a high strain region on the plate in fixation C that our strain gauges did not capture. Therefore, care must be taken in the interpretation of these results.

Increasing the fracture gap reduced the stiffness of the construct quite considerably when loading at 10° for the three methods of fixation. Confirming that when unstable injuries are managed in clinical practice and absolute stability is not achieved by fixation with a LCP, the stiffness of the construct is decreased. It may therefore be wise to use additional fixations such as anterior allografts, or to at least strongly recommend partial weight bearing for patients with unstable injuries until union is achieved [14, 15, 18, 19]. If the plate is considered inadequate for achieving safe fixation, then a long stem revision is recommended to address both bone healing and stem survival [20, 21].

A key limitation of the study was the number of specimens tested. The fracture-gap experiments were undertaken on a single specimen, as a feasibility study to evaluate whether the gap made a major difference in the behaviour. The results indicated that this was the case, and the behaviour was altered, rather than just being a multiple of the no gap model. Therefore, further tests should be undertaken on a larger group to enable statistical evaluation. Additionally, cyclic loading should also be undertaken to assess the effects of cable and screw loosening over time.

A limitation of all biomechanical studies, including this study, is the difficulty in translating results seen in the laboratory to potential results in clinical practice. For example, there are multiple patient and external factors that influence not only the choice of fixation, but also the way the fixation is applied. These include the age of the patient, underlying comorbidities, additional injuries, the skill of the surgeon and the equipment available, to name but a few. Additionally, it is difficult in a laboratory to create the forces that a fixation in a patient would undergo, and the present study only considered axial loading in one plane. Common reasons for failure of fixation in clinical practice are the forces that result from a patient falling, and this has not been taken into account in this study. However, even in the standardized model used here, there were marked differences seen between the fixation methods, indicating that their performance in vivo would be different and that the stability of the construct should be considered in the management of PPF.

It is technically challenging for a surgeon to manage PFF around a stable stem. The results of this study provide two important biomechanical outcomes. First, the inclusion of cables appears to damage the screw fixations and does not aid in the construct stability in locking plates. Second, that the degree of fracture reduction affects the overall stability of the construct. Not only was the construct less stable where a fracture gap was present, but the results showed different trends, indicating that the whole bending behaviour changed when there was a fracture gap present.

Conflict of interest There are no conflicts of interest.

References

1. Berry DJ. Epidemiology: hip and knee. Orthop Clin North Am. 1999;30(2):183–90.
2. Tsiridis E, Pavlou G, Venkatesh R, Bobak P. Gie, G periprosthetic femoral fractures around hip arthroplasty: current concepts in their management. Hip Int. 2009;19(2):75–86.
3. Kingdom, N.J.R.U., 2009/2010.
4. Brady OH, Garbuz DS, Masri BA, Duncan CP. Classification of the hip. Orthop Clin North Am. 1999;30(2):215–20.
5. Duncan CP, Masri BA. Fractures of the femur after hip replacement. Instr Course Lect. 1995;44:293–304.
6. Masri BA, Meek RM, Duncan CP. Periprosthetic fractures evaluation and treatment. Clin Orthop Relat Res. 2004;420:80–95.
7. Gautier E, Sommer C. Guidelines for the clinical application of the LCP. Injury. 2003;34(Suppl 2):B63–76.
8. Chakravarthy J, Bansal R, Cooper J. Locking plate osteosynthesis for Vancouver Type B1 and Type C periprosthetic fractures of femur: a report on 12 patients. Injury. 2007;38(6):725–33.
9. Graham SM, Moazen M, Leonidou A, Tsiridis E. Locking plate fixation for Vancouver B1 periprosthetic femoral fractures: a critical analysis of 135 cases. J Orth Sci. 2013;18(3):426–36.
10. Moazen M, Jones AC, Leonidou A, Jin Z, Wilcox RK, Tsiridis E. Rigid versus flexible plate fixation for periprosthetic femoral fracture-computer modelling of a clinical case. Med Eng Phys. 2012;34:1041–8.
11. Buttaro MA, Farfalli G, Paredes Nunez M, Comba F, Piccaluga F. Locking compression plate fixation of Vancouver type-B1 periprosthetic femoral fractures. J Bone Joint Surg Am. 2007;89(9):1964–9.

12. Bottlang M, Doornink J, Lujan TJ, Fitzpatrick DC, Marsh JL, von Augat P, Rechenberg B, Lesser M, Madey SM. Effects of construct stiffness on healing of fractures stabilized with locking plates. J Bone Joint Surg Am. 2010;92(Suppl 2):12–22.
13. Lujan TL, Henderson CE, Madey SM, Fitzpatrick DC, Marsh JL, Bottlang M. Locked plating of distal femur fracture leads to inconsistent and asymmetric callus formation. J Orthop Trauma. 2010;24:156–62.
14. Augat P, Merk J, Ignatius A, Margevicius K, Bauer G, Rosenbaum D, Claes L. Early, full weight bearing with flexible fixation delays fracture healing. Clin Orthp Relat Res. 1996;328:194–202.
15. Moazen M, Mak JH, Etchels LW, Jin Z, Wilcox RK, Jones AC, Tsiridis E. The effect of fracture stability on the performance of locking plate fixation in periprosthetic femoral fractures. J Arthroplast. 2013;28(9):1589–95.
16. Moazen M, Jones AC, Jin Z, Wilcox RK, Tsiridis E. Periprosthetic fracture fixation of the femur following total hip arthroplasty: a review of biomechanical testing. Clin Biomech. 2011;26:13–22.
17. Shah S, Kim SYR, Dubov A, Schemitsch EH, Bougherara H, Zdero R. The biomechanics of plate fixation of periprosthetic femoral fractures near the tip of a total hip implant: cables, screws, or both? Proc Inst Mech Eng Part H. 2011;225:845–56.
18. Haddad FS, Duncan CP, Berry DJ, Lewallen DG, Gross AE, Chandler HP. Periprosthetic femoral fractures around well-fixed implants: use of cortical onlay allografts with or without a plate. J Bone Joint Surg [Am]. 2002;84:945–50.
19. Sariyilmaz K, Dikici F, Dikmen G, Bozdag E, Sunbuloglu E, Bekler B, Yazicioglu O. The effect of strut allograft and its position on Vancouver type B1 periprosthetic femoral fractures: a biomechanical study. J Arthroplast. 2014;29(7):1485–90.
20. Moazen M, Mak JH, Etchels LW, Jin Z, Wilcox RK, Jones AC, Tsiridis E. Periprosthetic femoral fracture–a biomechanical comparison between Vancouver type B1 and B2 fixation methods. J Arthroplast. 2014;29(3):495–500.
21. Neumann D, Thaler C, Born U. Management of Vancouver B2 and B3 femoral periprosthetic fractures using a modular cementless stem without allografting. Int Orthop. 2012;36:1045–50.