

Biomechanical Evaluation of the Percutaneous Compression Plating System for Hip Fractures

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Objective: To investigate the biomechanical properties of the percutaneous compression plating system for intertrochanteric hip fractures.

Design: A biomechanical laboratory investigation on human cadaveric upper femora was conducted.

Setting: Biomechanical laboratory.

Patients: Sixteen femora from cadavers of patients aged 60 to 85 years.

Intervention: An intertrochanteric fracture was performed along the intertrochanteric line and fixed with the percutaneous compression plating system.

Main Outcome Measurements: Postfixation and postcycling bending and torsional stiffnesses, and load to failure at 0, 7, and 25° of adduction.

Result: The normalized postfixation bending and torsional

stiffnesses in the neutral position were 65% and 59%, respectively, and higher with adduction. After percutaneous compression plating fixation, bending stiffness increased to 72% following cyclic loading ($p < 0.05$). It was comparable to that of similar fixation constructs in the literature, whereas the torsional stiffness was somewhat higher. The mode of failure was distal femoral neck sagging, indicating good femoral head fixation. No varus displacement and cutouts were observed.

Conclusions: Our biomechanical data indicate that the percutaneous compression plating provides adequate bending stiffness and torsional stability. With sliding capability this fixation leads to controlled fracture impaction, which is further enhanced by cyclic loading.

Key Words: Fracture fixation, Intertrochanteric fracture, Biomechanical testing, Hip, Minimally invasive.

A significant increase in hip fractures has been reported worldwide (1,23,27) and is recognized as an orthopaedic epidemic (21,38); almost half are intertrochanteric fractures (16). Despite advanced treatment of intertrochanteric fractures, many complications prevail (3,15,36), and postoperative morbidity and mortality have remained high (3,5,24,28,30). Many attempts to improve the surgical outcomes have resulted in primarily biomechanical, rather than biologic, modifications of the procedure, including cementing techniques (4,10,11), osteotomies (9,13,14,33), and fixation designs and devices (2,8,17,26,31).

In the current study, we evaluate the biomechanical behavior of the percutaneous compression plating system (PCCP), a new device for intertrochanteric fracture fixation (19,20). The combination of percutaneous fixation to minimize operative trauma and small-diameter drilling

to prevent additional bone damage in the remaining lateral trochanteric wall without exposing the fracture site are perceived advantages. Biomechanically, the PCCP system is a double-axis fixation with a sliding capability that could theoretically facilitate rotational stability and so provide controlled fracture impaction.

The objectives of the current study were to evaluate the biomechanical characteristics of the PCCP system for use in hip fractures. Although this study was performed before the introduction of the PCCP fixation into orthopaedic practice, no reports were published on its biomechanical properties that could allow comparison with other devices (6,10).

MATERIALS AND METHODS

Specimens

Sixteen femora were harvested from cadavers of patients aged 60 to 85 years who had not been afflicted with diseases that might affect bone quality. After the soft tissues were stripped off, bone density of the proximal femurs was determined from anteroposterior radiography using the Singh index (35), and only grade 3 bones were included in the study. The bones were stored at -18°C and were thawed overnight at room temperature before testing. The distal femoral condyles were removed, and

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The devices that are the subject of this manuscript are FDA approved.

the femoral shaft was anchored by six screws into the holding device of the electromechanical servo-controlled testing machine (Instron Universal Testing Instrument; floor model TT-D, Instron, United Kingdom; Fig. 1). The biomechanical tests were conducted at room temperature with bones kept moist at all times.

Fractures and Fixation

After the determination of intact stiffnesses (see below), an intertrochanteric fracture was created by drilling multiple 3.2-mm holes along the intertrochanteric line in a circumferential manner, followed by a blow from a plastic mallet. This fracturing technique resulted in a very reproducible standard spike-edged fracture line that resembled clinical fractures and was comparable with similar published fracture models (37). The fracture was then reduced under direct vision and fixed with the

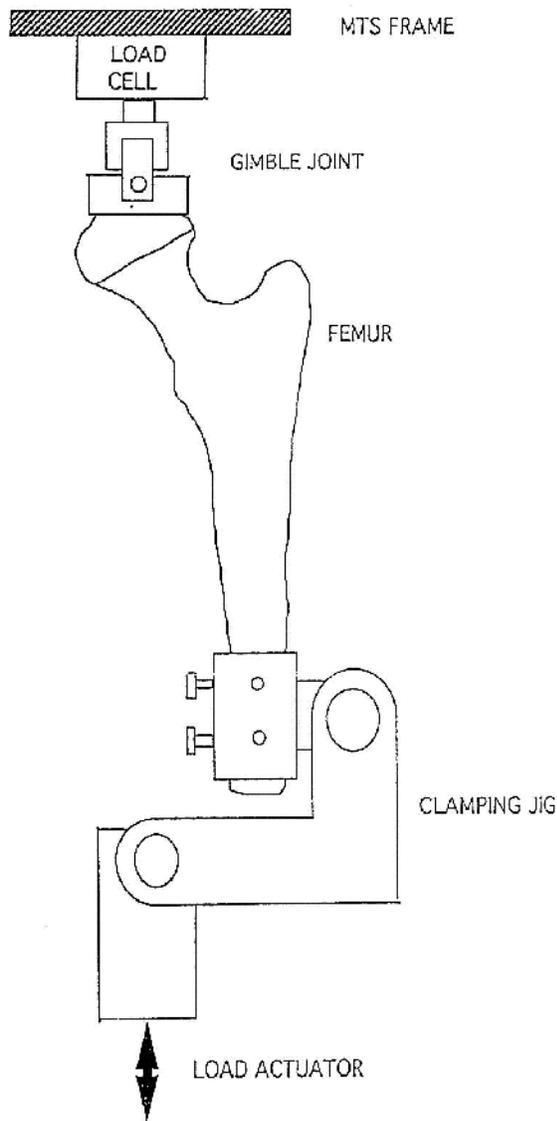


FIG. 1. Scheme of bending and fatigue device.

PCCP device (Fig. 2). The fracture was compressed with the two telescoping screws. The fixed femora were examined radiologically to ascertain the implant's position and fracture reduction.

Biomechanical Tests

The experimental protocol followed that of Swiontkowski et al. (37), but we used only one constant rate of loading-to-failure instead of varying levels of loading-to-failure.

Bending stiffness was determined in femora anchored in the holding device installed in the testing machine keeping the femoral neck in the coronal plane. Cyclic compression loads of 100 N were applied 5 times to settle each specimen in the holding device, and then load was applied by ramp displacement at a rate of 1 mm/min up to a maximal load of 890 N. All femora were tested in 0° and 7° of adduction (37), and 5 were also tested in 25° to allow comparison with published data (6,10).

Torsional stiffness was measured under maintained compression (37) in a specifically built holding device where the femoral head was grasped by two wings to ensure firm grip without use of penetrating spikes that damage and weaken the head (Fig. 3). The device was attached to the testing machine with the femoral neck in vertical orientation and in the center of revolution (Fig. 3).

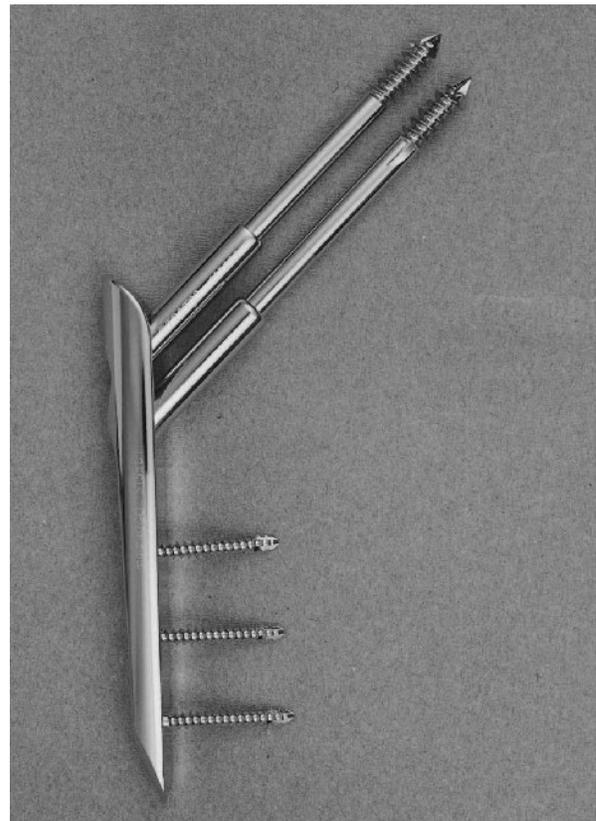


FIG. 2. Percutaneous compression plating (PCCP) device for hip fractures is comprised of a plate, two telescoping neck screws, and three shaft screws.

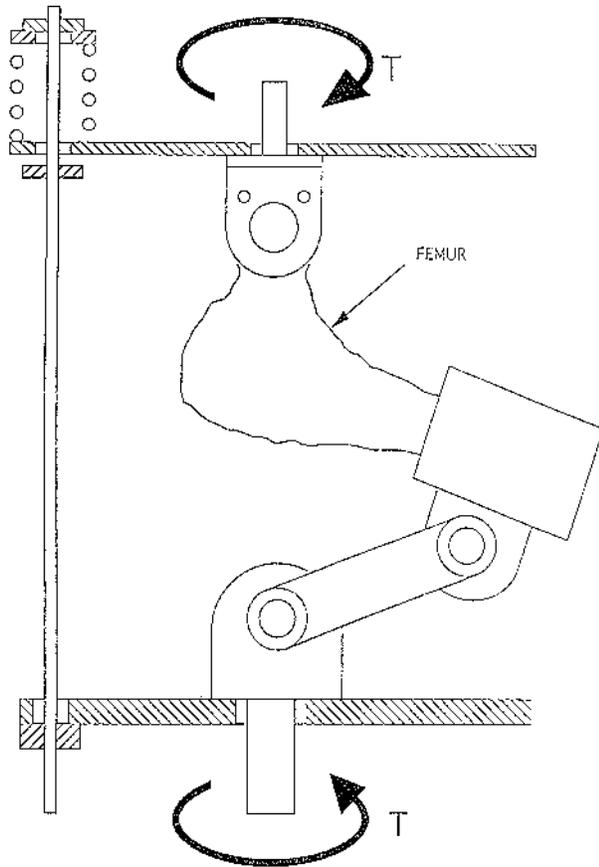


FIG. 3. Scheme of torsion compression device.

Torsional stiffness was determined with the bone loaded at compression of 890 N, and torsional moment was applied at a rate of $0.032^\circ/\text{second}$ up to 8.8 newton-meters.

The mode of failure was established by visual inspection. Load-to-failure was determined with specimens loaded at a continuous rate of 1 mm/min. Load versus deformation data were recorded, and failure was defined as the point where the constant increase in deformation resulted in leveling off or sudden decrease in the applied load (32), and not as the load at which gross destruction occurs. Eight specimens were tested at 0° , 8 at 7° , and 5 at 25° of adduction.

The load-to-failure was also converted to the maximal moment acting on the proximal femur at the intertrochanteric line. From the measurements of each femur's head diameter, neck length, and neck-shaft angle, the moment was calculated by multiplying the applied load value by the perpendicular distance from the load's line of action to the intersection of the neck center line and the intertrochanteric line.

Experimental Protocol

Initially, bones were tested for bending stiffness at adduction of 0° (neutral), 7° , and 25° , and for torsional

stiffness at 0° . Then a fracture was generated and fixed with the PCCP system, as detailed above. Postfixation bending stiffness measurements were repeated in the same fashion and positions, followed by torsional stiffness determination. Then two stages of cyclic bending loads were applied in 0° and 7° of adduction to assess the affect of cyclical loading on the stiffness of the PCCP fixed bones. First, 1,000 cycles from 90 N to 365 N, at a rate of 1 cycle/second, and then 500 cycles from 334 N to 1,000 N at a rate of 1 cycle/second. Postcycling bending and torsional stiffness were then tested again.

After the last stiffness testing, the specimens were loaded to failure as detailed above. The experiment was concluded with visual examination of the mode of failure.

Data Analysis

Load-deflection curves from 0 to 687 N (corresponding to a body weight of 70 kg) were obtained from each specimen in each test. From them bending stiffness was recorded as the slope of the linear portion of the curve and the maximum load-to-failure as the load value at which significant deformation occurred at diminishing load increments. Torsional stiffness was similarly determined from the torsion test. For comparison with other studies, the bending and torsion stiffnesses of each specimen were normalized by division of its postfixation and postcycling measurements by the corresponding values of the intact bone.

Two-way analysis of variance with repeated measures (equivalent to *t* test for dependent samples) tested the differences of normalized bending stiffness with classification by stage (postfixation or postcycling) and by angle of adduction. Normalized bending stiffnesses were compared using the Tukey-Kramer Honestly Significant Difference test for comparison of group means, and *p* values ≤ 0.05 were considered statistically significant. Data are presented as mean \pm SD.

RESULTS

Fourteen femora were included in the experiment, as one was used just for setup, and another was excluded because of malfixation demonstrated on radiography. There were no failures or damages of the fixation plates or screws during the tests, and all of the twenty-eight telescoping neck screws, except one, maintained sliding ability until the end of the testing.

Bending and Torsional Stiffness

The values of the bending stiffness are presented in Table 1. Intact stiffness was higher by 63% in adduction of 7° from that in the neutral position ($p = 0.0018$), and by 54% ($p = 0.04$) at 25° of adduction.

Immediately postfixation, the bending stiffness decreased to 65% of the corresponding intact bending stiffness (i.e., the normalized value was 65%) in the neutral

TABLE 1. Bending stiffness and normalized bending stiffness in three anatomic positions at different stages of PCCP evaluation

	Neutral		7° adduction		25° adduction	
	n	Mean ± SD	n	Mean ± SD	n	Mean ± SD
Intact (MN/m)	14	0.790 ± 0.291	8	1.285 ± 0.338	5	1.216 ± 0.547
Postfixation (MN/m)	14	0.497 ± 0.164	8	0.741 ± 0.275	5	1.021 ± 0.441
Postcycling (MN/m)	9	0.684 ± 0.226	7	0.816 ± 0.260		
Postfixation, normalized	14	0.650 ± 0.121	8	0.585 ± 0.154	8	0.855 ± 0.145
Postcycling, normalized	9	0.872 ± 0.203	7	0.621 ± 0.126		

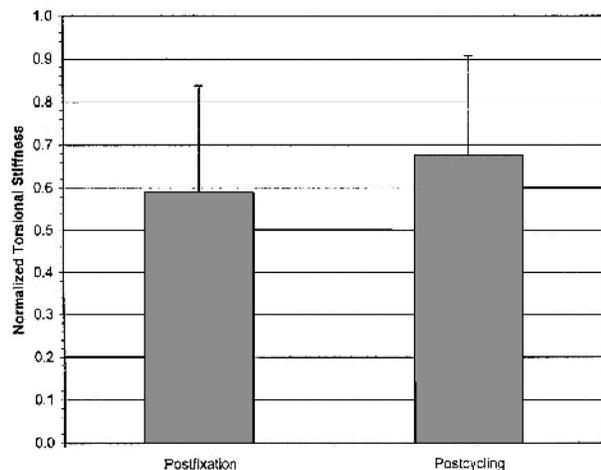
position, to 59% in 7 degrees of adduction, and to 85% at 25° of adduction (Table 1). The differences between the values at 25 degrees and those at 0° or 7° were significant ($p = 0.013$ and $p = 0.003$, respectively). The absolute postfixation bending stiffness increased with the degree of adduction (Table 1).

After cycling, bending stiffness in the neutral position, was 0.684 ± 0.226 MN/m, corresponding to a normalized postcycling stiffness of 87%. Compared with the postfixation normalized value of 65%, this presents a significant improvement ($p < 0.05$). At 7° of adduction, the postcycling bending stiffness was 0.816 ± 0.260 MN/m, a 19% increase from the neutral position, and was statistically not different from the postfixation stiffness.

The postfixation torsional stiffness was 109.3 ± 45.4 newton-meters/rad, compared with intact torsional stiffness of 195.8 ± 59.2 newton-meters/rad ($p = 0.002$), representing a normalized value of 59% (Fig. 4). The increase in torsional stiffness after cyclic loading was not statistically significant.

Failure

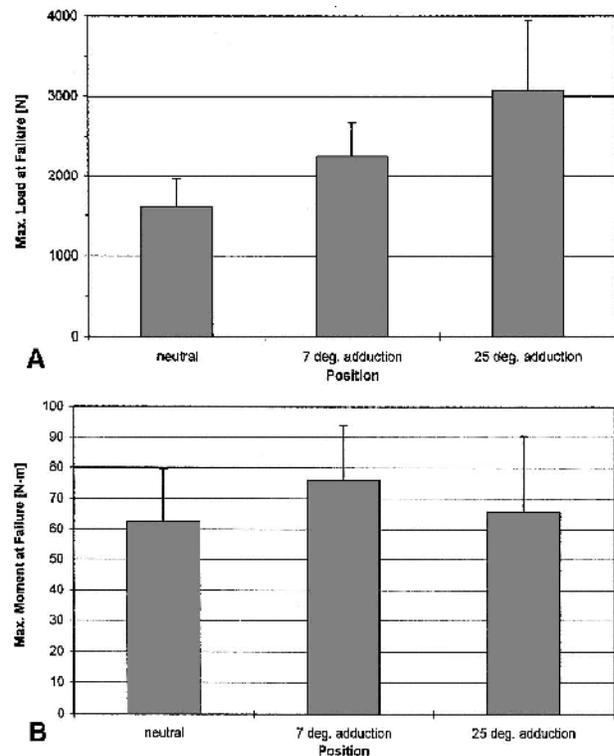
The load-to-failure was highly dependent on the angle of adduction, with the maximum at 25°, the position of a single-leg stance (29). There it was $6,164 \pm 347$ N when clamped, and 3066 ± 884 N when the load was applied free-hinged. These results correspond to 9 and 4.5 times

**FIG. 4.** PCCP normalized torsional stiffness immediately after fracture fixation ("postfixation") and after cyclical loading ("postcycling").

the body weight, respectively. The load-to-failure at 7° of adduction was $2,248 \pm 421$ N and neutral position $1,645 \pm 360$ N (Fig. 5A).

The calculation of maximal moment accounts primarily for the degree of eccentricity of load with regard to the center of fracture fixation in each adduction position. Each load produces some impaction and bending at the fracture site, and the maximal moment at failure separates out the bending moment from the combined loading. For loading at 0° the maximal moment was 62.4 ± 17.1 newton-meters ($n = 8$), whereas for 7° of adduction it was 76.1 ± 17.8 newton-meters ($n = 8$) and for 25° it was 65.5 ± 24.8 newton-meters ($n = 3$; Fig. 5B). These values were statistically not different.

The failure at 0° and 7° of adduction was characterized by slow bending into varus, whereas at 25° a milder distal migration of the femoral neck into the intertrochanteric region occurred.

**FIG. 5. A:** Maximum load to failure in three anatomic positions. **B:** Maximum moment of failure accounting for the moment arm.

DISCUSSION

This study demonstrated the biomechanical characteristics of the PCCP system for hip fracture fixation to be similar to those of similar devices, but they are further improved with cyclic loading, and the torsional stiffness of the double-axis fixation is better than single-axis fixation. Our post-PCCP fixation normalized bending stiffness of $64.3 \pm 15.2\%$ is statistically not different (by *t* test) from the $76.3 \pm 21.8\%$ reported by Swiontkowski et al. for seven commercial multiple pin-and-screw femoral neck fracture fixation constructs that had been tested under a similar compression range (37). Furthermore, sufficient statistical power (>80%) exists to infer that the normalized bending stiffness of the PCCP is indeed equal (within one standard deviation) to those of the seven fixation devices.

Immediately after fracture fixation, the torsional stiffness decreased to about 60% of the intact torsional stiffness, well within the range reported by Swiontkowski et al. (37) for five multiple-axis hip fracture fixation constructs. The PCCP, a double-axis fixation device, provides higher resistance against torsion, like the above multiple-axis fixations, compared with the Dynamic Hip Screw (Synthes, Paoli, PA, U.S.A.), which is a single-axis fixation. Our results contradict those of Blair et al. (6), who reported no significant differences in torsional stiffness among the following groups: three cancellous screw fixation, sliding hip screw, and sliding screw with an additional cancellous screw, in a basicervical fracture model. This contradiction may be explained by the presence of a sideplate in the PCCP, which locks the fixation screws to become a closed fixation system with better torsional resistance, and also perhaps by the nature of the fracture.

Maintenance of torsional stiffness is very important for differentiating fracture impaction from controlled fracture impaction. We refer to fracture compression as the surgical maneuver performed during surgery and define fracture impaction as the postsurgical weight-bearing compression provided by the sliding capability of the fixation device. Controlled fracture impaction occurs when the fixation device contributes to torsional stability in addition to sliding capability. It is of utmost importance because it maintains reduction throughout fracture healing by controlling fracture instability while complying with dynamic events across the fracture such as cyclic loading and fracture remodeling. In contrast, we define fracture collapse as fracture impaction where reduction is not maintained or when additional fracturing occurs, such as fracture of the trochanteric lateral wall.

After fracture fixation and cyclic loading, the bending stiffness improved, most significantly seen in the normalized bending stiffness in the neutral position and to a lesser degree in adduction. Torsional stiffness behaved similarly. These improvements could be the result of controlled fracture impaction that provides extra stability and strength of fixation and emphasize the importance of

having a sliding mechanism in any hip fracture fixation device.

When the femoral shafts are mounted at different angles of adduction their biomechanical characteristics differ. The intact bending stiffness in adduction was significantly higher from that of the neutral position. This is because of the shorter moment arm from the applied load to the center of fracture fixation in adduction. Immediately postfixation this difference remained, but at lower stiffness values, which were most pronounced in the normalized value of the largest adduction.

The maximal load-to-failure doubled at 25° of adduction, from 3,066 N to 6,164 N, when the gimbal joint was eliminated. In the setup of Blair et al. (6), using a base of neck model, a load-to-failure of 2,880 N for the sliding hip screw, 2,903 N for the sliding hip screw and cancellous screw, and 1,736 N for multiple cancellous screws were found.

Bone quality and the implant's biomechanical characteristics determine the holding power of the fixation, and they combine at the bone implant unit (12). Clark et al. found bone quality to be the single most important factor for the stability of the bone implant unit (12). Choueka et al. (10) concluded that the interaction between the design and materials of the fixation device with the surrounding bone is the factor that determines biomechanical behavior in trochanteric fractures. They also showed that increased projected area of the dome plunger within the femoral head (area of the device through a plane perpendicular to the applied load) increased the overall strength of fixation and improved sliding ability. Since the surgeon cannot change the quality of bone at time of hip fracture presentation, efforts must be directed at the improvement of the implant's characteristics and its interaction with the surrounding bone.

Three segments along the fixation device deserve particular attention: the proximal fixation within the femoral head, the midsegment, and the distal fixation at the lateral trochanteric wall or intramedullary area. The ability to maintain reduction in these three parts is the key to unimpaired fracture healing. Furthermore, controlled fracture impaction should be provided by the fixation system to adjust for bone resorption because the more unstable a fracture is, the more ability to control impaction is required. Initially, the fracture's anatomy, the number of fragments, and the remaining bone stock determine its stability. Appropriate reduction can lower this instability, and fracture compression will further enhance stability, as demonstrated in our tests. However, bone damage at the fracture site during or after surgery could produce secondary fracture instability.

The proximal segment of the implant unit should provide load-bearing capacity, whereas controlling the caudal shift and varus tilt primarily ensures stability (11). This load-bearing capacity is reflected in bending stiffness, load-to-failure, and torsional stability. Bending stiffness is a function not only of implant material and design (10), but also of sliding ability that impacts the

fracture, as demonstrated here. The load-to-failure capacity is greatly affected by the mode of fixation in the femoral head (11). Increased projected area of fixation in the femoral head increases the strength of fixation and improves sliding ability (10). As for stability, it was demonstrated that in uncemented Dynamic Hip Screw (Synthes) fixation of unstable intertrochanteric fractures, cyclic loading caused irreversible caudal shift and varus tilt of the femoral head, while movement in other directions was reversible and small; cementing the fixation significantly reduced these irreversible movements (11).

Swiontkowski et al. (37) found that the Neufeld and the Dynamic Hip Screw (Synthes) provide little, if any, torsional stability. This was attributed to the single-axis fixation of the Dynamic Hip Screw (Synthes) and to the absence of the head part, which enables fracture site compression in the Neufeld pin. The torsional stability of the PCCP is similar to that of other multiple-axis fixations, providing higher resistance against torsion (37). Hence, improvement of the projected area of fixation in the femoral head, enhancement of torsional stability, and adequate bending stiffness are needed to perfect the implant's proximal fixation segment.

If the integrity of the proximal bone implant unit is not maintained after fracture fixation, subsequent damage to bone trabeculae can initiate secondary fracture instability. Such reported failures were the superior migration of the screw (11) or a cutout through the femoral head (10). If, however, the integrity of the proximal bone implant unit is maintained, a different mode of failure was observed, namely, sagging of the inferior neck in five cases and separation of the sideplate in one (10). The mode of failure of the PCCP was that of inferior neck sagging into the intertrochanteric space, indicating sound proximal bone implant integrity.

The midsegment of the fixation device bridges the fracture site while connecting proximal and distal bone fragments. It is where the device's mechanical quality of bending stiffness meets the biologic state of fracture stability, determining together the biomechanical nature of the segmental bone-implant unit. Interaction of these factors is further subjected to dynamic events of cyclic loading and fracture site remodeling by resorption and formation. Bending stiffness, sliding capability, and the number of axial fixation points are the main characteristics of the fixation device's midsegment. Impaired sliding capability could turn a load-sharing device into a rigid load-bearing device that may result in femoral head penetration (22).

The distal segment of the fixation device must provide stable anchorage in the proximal femur. Drilling or reaming could initiate secondary instability. Subtrochanteric fractures produced intraoperatively and postoperatively in pertrochanteric fractures were reported after Gamma Nail fixation (Howmedica, Rutherford, NJ, U.S.A.) (8,31) and after femoral neck fracture fixation by a variable-length cannulated screw system (7) or nailing (34). Intraoperative fractures of the femoral lateral trochanter-

ic cortex during Gamma Nail and Dynamic Hip Screw (Synthes) fixations were also reported (25). Depending on the size and geometry of the drill (18), drilling the proximal lateral femur in the presence of intertrochanteric fracture may jeopardize the area of the greater trochanter and cause fracture and displacement of its lateral wall. In unstable fractures break of the lateral wall of the greater trochanter and its consequential superior migration prevent impaction of the distal buttress by the proximal femoral neck fragment, an important element for fracture compression. Increased instability and possible varus collapse compromise fracture healing. Therefore, a small-diameter drill should be used, especially when a delicate lateral wall is present (19,20).

When hip fractures are fixed with pins and screws without a lateral plate, lateral anchorage is provided as long as fracture compression is maintained. However, after fracture line resorption and loosening of compression, the screw head can regress, and instability with possible loss of reduction could result. Clark et al. (12) reported cannulated lag screws that were forced back out of the drilling hole during loading, and Blair et al. demonstrated failure of multiple cancellous screw fixation resulting from toggling of the screws at their points of insertion in the lateral cortex (6).

The weakness of this study stems from the use of cadaver bones, where the same specimens participated in multiple biomechanical tests. Results should be evaluated with this reservation in mind.

CONCLUSION

Our data indicate that the PCCP provides adequate bending stiffness and torsional stability, which with sliding capability produce controlled fracture impaction. Cyclic loading enhances the performance.

REFERENCES

1. Anderson GH, Raymakers R, Gregg PJ. The incidence of proximal femoral fractures in an English county. *J Bone Joint Surg Br* 1993; 75:441-444.
2. Babst R, Martinet O, Renner N, et al. Die DHS-Abstützplatte für die Versorgung der instabilen proximalen Femurfrakturen. *Helv Chir Acta* 1992;59:521-525.
3. Bannister GC, Gibson AG, Ackroyd CE, et al. The fixation and prognosis of trochanteric fractures: a randomized prospective controlled trial. *Clin Orthop* 1990;254:242-246.
4. Bartucci EJ, Gonzalez MH, Cooperman DR, et al. The effect of adjunctive methylmethacrylate on failures of fixation and function in patients with intertrochanteric fractures and osteoporosis. *J Bone Joint Surg Am* 1985;67:1094-1107.
5. Bendo JA, Weiner LS, Strauss E, et al. Collapse of intertrochanteric hip fractures fixed with sliding screws. *Orthop Rev* 1994; (suppl):30-37.
6. Blair B, Koval KJ, Kummer F, et al. Basicervical fractures of the proximal femur: a biomechanical study of three internal fixation techniques. *Clin Orthop* 1994;306:256-263.
7. Blyme PJ, Kristensen G, Jensen CH, et al. Subtrochanteric fractures following osteosynthesis of femoral neck fractures with the VLC-femoral system. *Orthopedics* 1992;15:1373-1374.
8. Bridle SH, Patel AD, Bircher M, et al. Fixation of intertrochanteric fractures of the femur: a randomised prospective comparison of the

- gamma nail and the dynamic hip screw. *J Bone Joint Surg Br* 1991;73:330–334.
9. Chang WS, Zuckerman JD, Kummer FJ, et al. Biomechanical evaluation of anatomic reduction versus medial displacement osteotomy in unstable intertrochanteric fractures. *Clin Orthop* 1987; 225:141–146.
 10. Choueka J, Koval KJ, Kummer FJ, et al. Biomechanical comparison of the sliding hip screw and the dome plunger: effects of material and fixation design. *J Bone Joint Surg Br* 1995;77:277–283.
 11. Claes L, Becker C, Simnacher M, et al. Die Verbesserung der Primarstabilität von DHS-Osteosynthesen bei instabilen, pertrochantaren Femurfrakturen osteoporotischer Knochen durch einen neuen Glas-Ionomer-Zement. *Unfallchirurg* 1995;98:118–123.
 12. Clark DI, Crofts CE, Saleh M. Femoral neck fracture fixation: comparison of a sliding screw with lag screws. *J Bone Joint Surg Br* 1990;72:797–800.
 13. Desjardins AL, Roy A, Paiement G, et al. Unstable intertrochanteric fracture of the femur: a prospective randomised study comparing anatomic reduction and medial displacement osteotomy. *J Bone Joint Surg Br* 1993;75:445–447.
 14. Dimon JH, Hughston JC. Unstable intertrochanteric fractures of the hip. *J Bone Joint Surg Am* 1967;49:440–450.
 15. Flores LA, Harrington II, Heller M. The stability of intertrochanteric fractures treated with a sliding screw-plate. *J Bone Joint Surg Br* 1990;72:37–40.
 16. Gallagher JC, Melton LJ, Riggs BL, et al. Epidemiology of fractures of the proximal femur in Rochester, Minnesota. *Clin Orthop* 1980;150:163–171.
 17. Geissler N, Meffert O, Stapel A, et al. Ergebnisse der operativen Versorgung instabiler pertrochantarer Femurfrakturen mittels DHS und T-Platte. *Unfallchirurgie* 1994;20:72–75.
 18. Gerard D, Degreif J, Runkel M. Experimentelle Untersuchungen zur Schwächung der Stabilität des Trochanter major durch laterale Fensterung. *Unfallchirurgie* 1993;19:208–212.
 19. Gotfried Y. Percutaneous compression plating of intertrochanteric hip fractures. *J Orthop Trauma* 2000;14:490–495.
 20. Gotfried Y. Percutaneous compression plating of unstable intertrochanteric fractures. *61st Annual Conference of the German Society for Emergency Surgery*, Berlin, 1997.
 21. Hip fractures in the elderly: the coming epidemic. *AAOS Symposium*, September 1993.
 22. Jacobs RR, McClain O, Armstrong HJ. Internal fixation of intertrochanteric hip fractures: a clinical and biomechanical study. *Clin Orthop* 1980;146:62–70.
 23. Jensen JS. Incidence of hip fractures. *Acta Orthop Scand* 1980;51: 511–513.
 24. Larsson S, Friberg S, Hansson LI. Trochanteric fractures: mobility, complications, and mortality in 607 cases treated with the sliding-screw technique. *Clin Orthop* 1990;260:232–241.
 25. Leung KS, So WS, Shen WY, et al. Gamma nails and dynamic hip screws for peritrochanteric fractures: a randomised prospective study in elderly patients. *J Bone Joint Surg Br* 1992;74:345–351.
 26. Lunsjo K, Ceder L, Stigsson L, et al. Two-way compression along the shaft and the neck of the femur with the Medoff sliding plate: one-year follow-up of 108 intertrochanteric fractures. *J Bone Joint Surg Br* 1996;78:387–390.
 27. Martin AD, Silverthorn KG, Houston CS, et al. The incidence of fracture of the proximal femur in two million Canadians from 1972 to 1984: projections for Canada in the year 2006. *Clin Orthop* 1991;266:111–118.
 28. Miller K, Atzenhofer K, Gerber G, et al. Risk prediction in operatively treated fractures of the hip. *Clin Orthop* 1993;293:148–152.
 29. Pauwels F. *Gesammelte Abwandlungen zur Funktionellen Anatomie des Bewegungsapparates*. Springer: Berlin; 1965.
 30. Pitsaer E, Samuel AW. Functional outcome after intertrochanteric fractures of the femur: does the implant matter? A prospective study of 100 consecutive cases. *Injury* 1993;24:35–36.
 31. Radford PJ, Needoff M, Webb JK. A prospective randomised comparison of the dynamic hip screw and the gamma locking nail. *J Bone Joint Surg Br* 1993;75:789–793.
 32. Ramser JR, Mihalko WM, Carr JB, et al. A comparison of femoral neck fixation with the reconstruction nail versus cancellous screws in anatomic specimens. *Clin Orthop* 1993;290:189–196.
 33. Sarmiento A, Williams EM. The unstable intertrochanteric fracture: treatment with a valgus osteotomy and I-beam nail-plate. A preliminary report of one hundred cases. *J Bone Joint Surg Am* 1970;52:1309–1318.
 34. Scheuba G. Die Spontanfraktur an der Nageleinschlagstelle nach Schenkelhalsnagelung. *Arch Orthop Unfallchir* 1969;66:265–272.
 35. Singh M, Nagrath AR, Maini PS. Changes in trabecular pattern of the upper end of the femur as an index of osteoporosis. *J Bone Joint Surg Am* 1970;52:457–467.
 36. Stappaerts KH, Deldycke J, Broos PL, et al. Treatment of unstable peritrochanteric fractures in elderly patients with a compression hip screw or with the Vandeputte (VDP) endoprosthesis: a prospective randomized study. *J Orthop Trauma* 1995;9:292–297.
 37. Swiontkowski MF, Harrington RM, Keller TS, et al. Torsion and bending analysis of internal fixation techniques for femoral neck fractures: the role of implant design and bone density. *J Orthop Res* 1987;5:433–444.
 38. Wallace WA. The increasing incidence of fractures of the proximal femur: an orthopaedic epidemic. *Lancet* 1983;8339:1413–1414.