Intramedullary versus extramedullary fixation of subtrochanteric fractures

A biomechanical study

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We compared two different subtrochanteric fracture fixation techniques, an intramedullary hip screw system (IMHS) and an extramedullary, dual sliding screw-plate system (MSP), to determine relative fixation stability. 6 matched pairs of osteosynthesized osteopenic cadaver femurs were axially loaded to 1000 N with concurrent, simulated abductor forces of 0%, 50% or 86% of the applied head force. The initial loading sequence was made with uniaxial dynamization—the lag screw of the MSP locked and distal locking of the IMHS nail. Femoral head displacement and medial femoral strain were measured for intact femur controls, after fixation of a 2-part reverse oblique subtrochanteric fracture and finally a 3-part reverse oblique subtrochanteric fracture with a lateral wedge defect. The samples were then loaded at 750 N for 10^4 cycles with both devices uniaxially locked, followed by 10^4 cycles with both devices fully biaxially dynamized (unlocked).

For the 2-part subtrochanteric fracture pattern, both devices exhibited similar inferior displacements of the femoral head (average 2.0 mm) and medial femoral strain (~70% of intact). Increasing abductor forces decreased medial compressive strain but did not significantly affect head displacement. For the 3-part fracture model, the MSP demonstrated significantly less inferior displacement of the head (1.6 mm vs. 2.1 mm) and both devices demonstrated significantly decreased medial strain. After cycling, head displacement increased approximately 50% in both devices and medial strain increased slightly. After unlocking and cycling, the MSP group showed significant lateral displacement of the proximal fragment.

The IMHS and MSP devices provide similar stability for fixation of 2-part and 3-part reverse oblique subtrochanteric fractures. In a biaxially dynamized, 3-part reverse oblique fracture, displacement of the proximal fragment can occur with the MSP.

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Although subtrochanteric fractures occur in less than 10% of all hip fractures their treatment is problematic. Various intramedullary and extramedullary devices have been developed in an attempt to address potential complications of device failure, mal- or non-union and postfixation deformity. The classic mechanical study (Cochran et al. 1980) compared several first generation intramedullary devices and concluded that intramedullary fixation (Zickel nail) was superior to extramedullary fixation. They also showed that simulation of abductor forces was important for mechanical testing of these fractures. A later laboratory study (Mahomed et al. 1994) comparing a locked Gamma nail to a standard sliding hip screw for the fixation of stable and unstable subtrochanteric fractures showed that the intramedullary nail was more rigid and permitted less fracture displacement. Newer intramedullary devices such as the intramedullary hip screw (IMHS) can provide controlled collapse of the proximal fragment, if not axially locked, and an internal buttress to medial translation of the femoral shaft. A new extramedullary alternative is the Medoff sliding plate (Medoff and Maes 1991), which allows axial fracture compression, and offers an external buttress to medial translation of the femoral shaft, if the lag screw is locked (Figure 1). However, no laboratory studies comparing these newer devices have been done.

Material and methods

The study was performed at the Department of Bioengineering, Hospital for Joint Diseases Orthopaedic Institute, New York, USA.

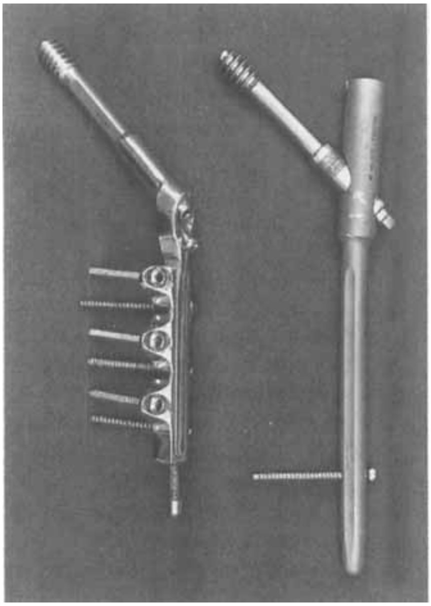

Figure 1. Devices tested were (left) the Medoff Sliding Plate (MSP) and (right) the Intramedullary Hip Screw (IMHS).

6 matched pairs of osteopenic femurs from North-American Caucasian donors aged above 65 years (all female, 66–82 years) were selected on the basis of a bone density of 0.3–0.5 g/cm^2 (QDR-2000 Supine Lateral X-Ray Bone Densitometer, Hologic, Waltham, MA). Radiographs in 2 planes were taken to exclude samples with morphologic abnormalities. The femurs were stripped of all soft tissue and the distal femoral condyles were removed and the femoral shafts potted with methylmethacrylate bone cement in 6 cm diameter steel tubes. The specimens were tightly wrapped in airtight double bags throughout the experiment to avoid desiccation. The constructs were mounted on a MTS servohydraulic mechanical testing machine (MTS Systems, Minneapolis, MN) at 25 degrees adduction to simulate the anatomical loading during one-legged stance (Chang et al. 1987). A superior steel bar in contact with the upper aspect of a femoral head holder was attached at one end to the greater trochanter with several cerclage wires (Figure 2). 3 different fixed reference points on the bar were located medially from the head holder to produce calculated abductor forces of 0%, 50% and 86% of femoral head loads. Vertical compressive loads, as calculated from this lever arrangement, were applied on the bar at these three different abductor force points to obtain femoral head loads of 500 N, 750 N and 1000 N (approximately twice body weight). Measurements of strain and displacements were initially obtained for each femoral head load with the three abductor forces.

Cortical strain in the calcar region was determined with a unidirectional strain gauge (EA-06-125-AC-120, Omega Engineering, Stamford, CT), mounted with cyanoacrylate glue on the surface of the bone at the base of the femoral neck, aligned parallel to the femoral shaft. A digital strain indicator (Micromeasurements P-3500, Vishay Technology Inc., Greenboro, NC, USA) was connected to the strain gauge. Inferior displacement of the femoral head and lateral displacement of the femoral shaft were measured using 2 spring-driven, electronic displacement gauges (IDC-25E Mitutoyo Co, Tokyo, Japan), mounted on a collar fixed on the femoral shaft directly below the osteotomy. The inferior head displacement gauge was aligned parallel to the femoral shaft and in contact with the inferior aspect of the femoral head, and the lateral displacement gauge was aligned perpendicular to the femoral shaft in contact with the greater trochanter. This lateral gauge determined lateral or medial displacement of the proximal fragment in relation to the femoral shaft. The femurs were sequentially tested during destabilization in the following steps:

1. Intact femurs tested at 500 N, 750 N and 1000 N with 0%, 50% and 86% concurrent abductor forces, to obtain baseline mechanical properties.

2. With a surgical saw, a reverse oblique osteotomy was made, starting immediately above the lesser trochanter and directed inferolaterally at a 30° angle. This osteotomy was chosen on the basis of previous

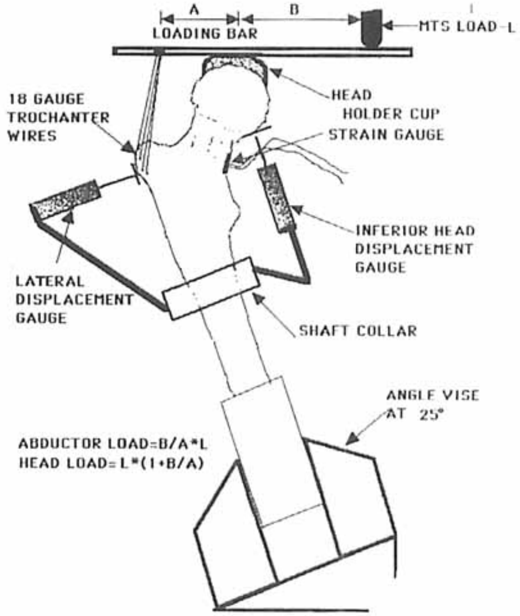

Figure 2. Mechanical testing arrangement showing method and calculation for application of abductor forces and location of femur, strain and displacement gauges.

Table 1. Comparison of medial strains (percent of intact femur) and displacements (mm) for IMHS and MSP for the sequential test conditions (750 N head load; 645N abductor force-standard deviations are approximately 25% for strain and 20% for displacement)

Test condition	Medial strain		Inferior disp.		Lateral disp.	
	IMHS	MSP	IMHS	MSP	IMHS	MSP
Intact femur	100	100	0.61	0.64	–1.02	–0.56
Stable osteotomy	75	69	2.04	1.94	–1.74	–1.06
Unstable osteotomy	13	10	2.14	1.60	–0.86	–0.87
10^4 cycles-locked	17	11	3.27	2.36	–1.11	–1.25
10^4 cycles-unlocked	16	4	3.77	3.14	–0.53	1.94

Table 2. Effect of abductor load on MSP medial strain (percent of intact femur) and inferior head displacement (mm) for a 750 N head load (standard deviations are approximately 25% for strain and 20% for displacement)

	Abductor load as a percent of head load					
Test condition	0%		50%		86%	
	Strain	Disp.	Strain	Disp.	Strain	Disp.
Intact femur	100	0.36	100	0.41	100	0.64
Stable osteotomy	73	1.96	72	2.02	63	1.94
Unstable osteotomy	51	0.70	31	0.98	13	1.60

biomechanical testing (Kummer et al. 1997). Femurs were then instrumented, using standard techniques. Each femur pair was fixed with the 2 following treatment groups (alternating left-to-right assignment): 1) 6-hole Medoff sliding plate (MSP, MedPac, Valencia, CA), with the lag screw fixed by the locking set screw of the MSP to prevent sliding of the lag screw, or 2) Intramedullary Hip Screw (IMHS, Smith & Nephew Orthopedic Division, Memphis, TN) with 1 transcortical distal locking screw to prevent subsidence of the intramedullary nail. In both devices, the same type of lag screw was used. These 2-part fractures were tested at 500 N, 750 N and 1000 N with 0%, 50% and 86% concurrent abductor forces.

3. A second osteotomy more distally located was then made in the proximal fragment, removing a triangular wedge starting one centimeter above the lateral cut end and ending at the medial cut. Thus, a lateral fracture gap was created to simulate loss of lateral support. A complete gap would allow collapse when both devices were biaxially dynamized. These 3-part fractures were tested at 750 N with 0%, 50% and 86% concurrent abductor forces.

4. The specimens were loaded at 1 Hz for 10^4 cycles with a 750 N load and a concurrent abductor force of 86%.

5. The devices were biaxially dynamized by removal of the locking set screw of the MSP and the distal locking screw of the IMHS and the specimens again loaded at 1 Hz for 10^4 cycles with a 750 N load and a concurrent abductor force of 86%.

Analysis of data

The strain and displacement data for the IMHS and MSP matched pairs were compared with the Wilcoxon signed-rank test to assess differences between treatment techniques and effects of testing within a technique; a significance level of $\alpha = 0.05$ was used.

Results

For the 2-part subtrochanteric fracture, both devices exhibited similar inferior head displacement (mean 2.0 mm, range 1.4–3.2 mm) and medial strain (approximately 70% of intact; Table 1). Increasing abductor forces reduced the medial compressive strain and, in general, increased inferior head displacement, as illustrated in Table 2 for the MSP constructs; the IMHS constructs demonstrated similar behavior. For the 3-part fracture model, the MSP demonstrated significantly less inferior displacement of the proximal fragment (1.6 mm vs. 2.1 mm) and both device constructs showed significantly reduced medial strain (approximately 15% of intact). After cycling, head displacement increased approximately 50% for both devices and medial strain increased slightly. After unlocking, both groups exhibited the maximum inferior displacements of the femoral head; the MSP group also showed a 2 mm (range 1.5–2.9 mm) lateral displacement of the proximal fragment.

Discussion

This test model attempted to simulate abductor forces that would be applied during single leg stance and ambulation. However, due to the mechanical linkage, abductor forces were always less than head forces, whereas physiological forces are at least twice as large. To duplicate in vivo loading correctly, a separate load applicator is required. In contrast to the findings of a prior study (Cochran et al. 1980), which used a similar loading mechanism, medial strain for the intact femur decreased as abductor force increased. This result may be because the location of the strain gauge in our test was much more proximal than in the previous study and because they tested the femurs at much lower abduction angles (0° and 10°) than the 25° used in this experiment. With a proximal strain gauge, the upward tensile forces created by the abductor force on the trochanter through the proximal femur partially counteract the medial compression of the head loading. After simulated fracture and fixation, both studies showed a reduction in medial strain due to some of the loads being borne by the devices. For the 2-part subtrochanteric fractures, there were no significant differences between the devices with respect to inferior head or lateral shaft displacements. The MSP provided slightly less medial cortical strain, possibly due to greater load bearing. Increased inferior femoral head deflections for both devices after cycling were due to crushing of the medial point of bone contact or to a medialization of the femoral shaft along the angled osteotomy. The MSP showed less inferior displacement of the proximal fragment, which may be attributed to an increased capacity to maintain the apposition of the cortices of the medial part of the fracture, due to the buttress effect of the device, while the IMHS may tilt within the medullary canal because of the single distal locking. When biaxially dynamized, the IMHS could rotate in the femoral canal, but this was not observed because of the loading orientation without femoral version. Similarly, the MSP allows translation if biaxially dynamized as shown by the lateral translation seen in this test model and in previous Sawbones studies (Kummer et al. 1997). Locking of the lag screw to prevent sliding can prevent this translation.

The major limitation of this experiment is the multiple testing of the same femur and the small sample size. Although the devices were compared in matched pairs following similar protocols, differences in consolidation of supporting bone, particularly during cycling, would affect strains and displacement measurements. The osteotomies are smooth cuts and have no interdigitation that could occur clinically to increase stability. Previous studies comparing intramedullary to extramedullary fixation have used an unstable fracture model with a gap (Tencer et al. 1984, Curtis et al. 1994, Mahomed et al. 1994) or medial wedge (Shaw and Wilson 1993, Kraemer et al. 1996, Wheeler at al 1997), with devices that were not axially dynamized and did not simulate abductor forces. The various intramedullary devices tested were said to be better for subtrochanteric fracture fixation, because they were more rigid and stronger. In the present study, there was a medial bone buttress and both devices, in general, provided similar stability.

Detailed radiographs after completion of the test sequence were not obtained, since the samples were not tested to failure. Data on inferior head displacement, however, suggest that migration of the IMHS lag screw in the femoral head could have occurred, since this displacement is not explained by medialization of the femoral shaft.

In conclusion, the IMHS and MSP devices can provide sufficient stability for fixation of reverse oblique 2-part and 3-part subtrochanteric fractures. If biaxially dynamized in a 3-part reverse oblique fracture, lateral displacement of the proximal fragment can occur with the MSP.

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