FEMORAL NECK FRACTURE FIXATION

COMPARISON OF A SLIDING SCREW WITH LAG SCREWS

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The rigidity of a sliding compression screw and three cannulated lag screws in the treatment of subcapital fractures was compared in five pairs of female cadaver femora. There were no significant differences between the compressive strength, bone density, cortical thickness or Singh index of the bones in each pair. A subcapital fracture was standardised using a perpendicular saw cut across the femoral neck. A uniaxial 'load test system' with force and length measurement facilities was used to mimic cyclical stressing applied in vivo at a frequency of 0.5 Hz from 0 to 3 times body-weight. There was no significant difference between the fixation afforded by the sliding compression screw and three lag screws. Bone quality was the single most important factor in the stability of the bone implant unit.

The importance of bone quality in the fixation of femoral neck fractures has been emphasised by in vitro studies (Frankel 1960; Van Audekercke et al 1979; Husby, Høiseth and Fønstelien 1987). However, most in vitro studies have compared fixation devices using femora from different subjects (Van Audekercke et al 1979; MacKechnie-Jarvis 1983; Elmerson et al 1987). The limitation of this technique is that the variation in the mechanical properties of bone from different individuals is not taken into account. The two reports which overcame this problem by using paired femora from the same patient employed a static loading system (Mizrahi et al 1980; Engesaeter et al 1984). This method is unsatisfactory since in life the fixation device must be able to withstand the cyclical loading pattern associated with walking.

We report an in vitro study which investigated the fixation of femoral neck fractures using (1) paired femoral heads from the same patient and (2) a dynamic loading regime to simulate walking.

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MATERIALS AND METHODS

Fourteen pairs of femora were obtained from females over the age of 60. Those with known hip pathology such as a previous arthroplasty, a femoral fracture or a tumour were excluded from the study. The specimens were initially stored at −20°C. Prior to testing, the mechanical properties of the right and left femora were investigated by measuring bone density (Dalén, Hellström and Jacobson 1976), cortical thickness (Griffiths, Swanson and Freeman 1971), the Singh index (Khairi et al 1976) and using a static compression test.

The frozen bones were radiographed in pairs. The Singh index was assessed for each proximal femur by three independent observers and the femoral shaft cortical thickness measured 120 mm below the greater trochanter using vernier calipers. After the bones had been radiographed they were returned to their storage environment. Using a ND 2100 bone density scanner, readings were taken from the proximal two-thirds of each femur, and from the femoral shaft sections. As soon as the measurements had been made all the bones were again returned to storage.

Before mechanical testing, the bones were defrosted for six hours at room temperature, carefully cleaned of muscle and soft tissue, and rehydrated with warm saline. The mechanical tests were performed on a Mayes material testing machine (Fig. 1). Static compression tests were carried out on six pairs of femoral shaft sections. After the bones had defrosted, the length and diameter of each specimen were measured. The length of each sample was adjusted using a band-saw to give a length to diameter ratio of approximately 4 to 1. The specimen was loaded to failure at a rate of 3 mm/min. The maximum load was recorded.

Cyclical compression tests were performed on five pairs of randomly selected femora artificially fractured and fitted with two different fixation devices. After the bones had defrosted, one specimen from each pair was selected at random for fixation with a sliding compression screw, whilst the other was fixed with three self-tapping cannulated lag screws. In order to achieve an anatomical reduction, both devices were initially inserted into the unfractured femur. They were removed and a fracture artificially produced using a saw cut perpendicular to the neck. Purpose-built grips to hold the bone at an anatomical angle of 20° from the vertical were fitted to the test machine (Fig. 2). The specimen was placed in the lower grip. Each bone was covered with damp paper during testing. Each specimen was loaded sinusoidally from 0 to 3 times body-weight (2.1 kN) at a frequency of 0.5 Hz until fracture occurred. The load and displacement were recorded graphically and from a digital display. The deflections after 1 and 30 cycles were recorded. Failure of the system was defined as the point at which a deflection of 125% of the value recorded at 30 cycles was observed (a figure set for the proposed British Standard for femoral head prostheses BSI DD91:1986). The specimens were refrozen after testing and later radiographed to determine the pattern of failure.

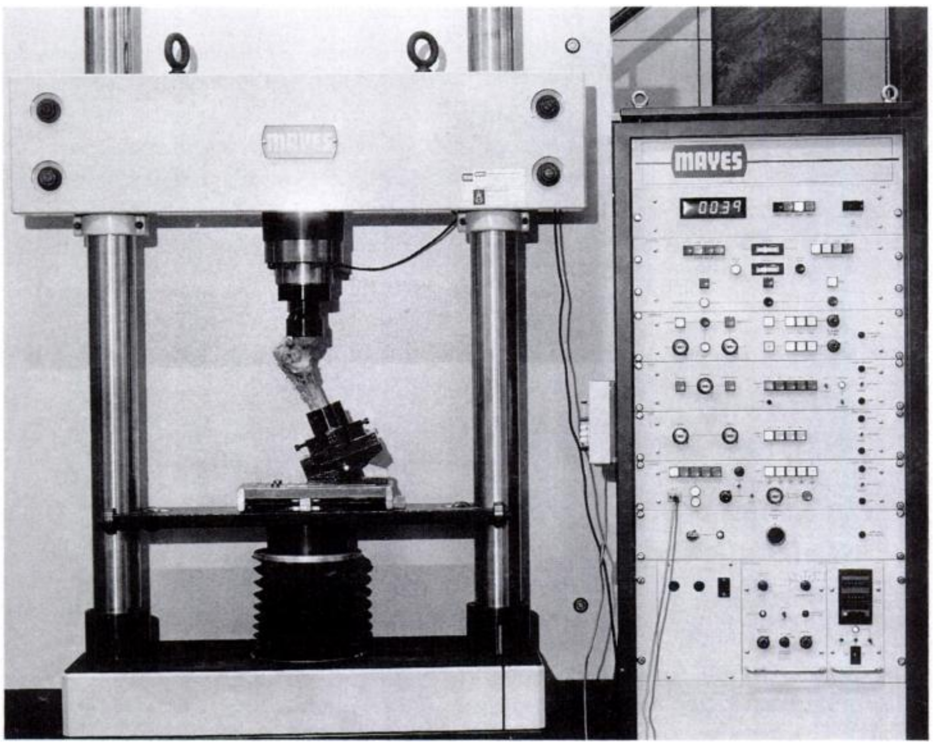

Fig. 1

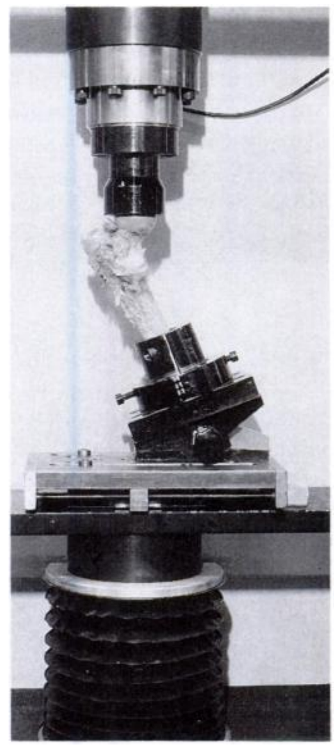

Fig. 2

Figure 1 – The Mayes material testing machine as used in our experiments. Figure 2 – The proximal femur held between the purpose-built grips.

RESULTS

Bone quality. In order to be certain that there was no difference in the bone quality of the paired femora, four pretest investigations were performed: static compression tests, bone density measurements, the Singh index and cortical thickness as measured radiographically. Statistical analysis revealed no significant difference between the right and left femora (Table I).

Rigidity. The deflections recorded during the fatigue testing of the five pairs of proximal femora are given in Table II. There was no significant difference between the deflections recorded after the first or 30th cycles for the bones fixed with compression or lag screws. In addition, there was no significant difference between the number of cycles required to produce failure in the femora fixed with a sliding compression screw and those fixed with three lag screws.

The results in Table III compare the age and bone quality (Singh index and bone density) against deflection of each pair of fixed bones. The pair with the largest deflection (number 4) were from the oldest subject (84 years) and had the weakest bone (Singh index 3) and the lowest bone density of the five pairs. By contrast, the smallest deflection occurred in bones from the youngest subject (65 years) with the best bone quality (Singh index 6) and a high bone density.

Failure mode. The sliding compression screws did not bend in any of the tests. Failure was due to collapse of the trabecular bone in the femoral neck and head. At first the fracture impacted with the screw sliding down the barrel. The head then began to shear across the fracture site due to the screw barrel cutting through the femoral neck trabecular bone (Fig. 3a). During loading the three cannulated lag screws were forced back down the drilling holes. A crush fracture of the inferior surface of the femoral neck then occurred. Finally the head sheared across the fracture line and the screw threads converged and rotated downwards (Fig. 3b). No plastic deformation of any implant was observed.

Table I. Comparison of the bone quality in the right and left femora from the same patients

	Pairs tested	Right	Left
Compressive strength (kN)	6	14.24±2.79	12.53±2.4
Bone density (g/cm²)	12	0.97±0.06	0.96±0.05
Singh index	14	5.2±0.26	5.2±0.30
Cortical thickness (mm)	14	7.21±0.27	6.97±0.36

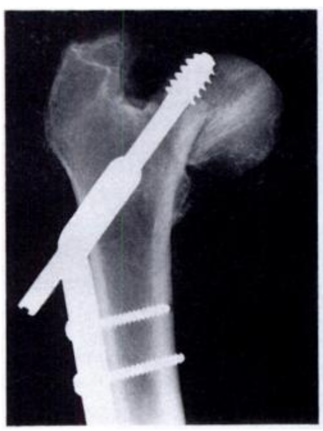

Fig. 3a

Fig. 3b

Mode of failure. Figure 3a – of the sliding screw. Figure 3b – of the lag screws.

DISCUSSION

The rigidity of the Richards compression screw and three lag screws used currently in the treatment of femoral neck fractures have been compared. In order to reflect the population at risk, femora were obtained from females over the age of 60 (Lewinnek et al 1980). To exclude bone quality as a source of error, paired femora from the same subject were tested. The bones from each pair were shown to be matched mechanically by four pretesting investigations. In addition an attempt was made to reduce experimental variables to a minimum by using a standardised fracture (Klenerman and Marcuson 1970) and ensuring an anatomical reduction. A saw cut perpendicular to the neck was chosen in accordance with Klenerman's observations that subcapital fractures have a similar angle. During walking, the angle of the resultant load on the femoral head has been shown to be 10° to 15° from the midline (Inman 1947; Elson and Charnley 1968). The offset angle of the femoral shaft is 7° to 10° (Williams et al 1989). The optimum loading angle for the femoral head in vitro is therefore 17° to 25° from the vertical; 20° was chosen in this study. A cyclical rather than a static loading regime was applied in order to simulate walking. A loading regime of 0 to 3 times body-weight (2.1 kN acting sinusoidally) was chosen for two reasons. First, three times body-weight is within the peak theoretical load acting on the hip joint (Rydell 1966; Paul 1967; Crowninshield, Brand and Johnston 1978). Secondly, the British Standards Institute recommends a

Table II. Deflections recorded during fatigue testing of the proximal femora

Pair	Age (yr)	Screw used	Deflection after 1 cycle (mm)	Deflection after 30 cycles (mm)	Number of cycles to failure
1	71	Compression	7.20	12.93	66
		Lag	2.88	11.18	67
2	65	Compression	1.65	4.71	55
		Lag	1.74	3.45	65
3	65	Compression	3.44	6.45	65
		Lag	1.94	5.71	38
4	84	Compression	18.50	24.97	5
		Lag	11.40	22.21	3
5	81	Compression	2.75	4.30	75
		Lag	2.94	5.46	150

Table III. Age, bony quality and deflection under testing of each pair of fixed femora

Pair	Age (yr)	Deflection after 30 cycles (mm)	Singh index	Bone density (g/cm²)
1	71	12.93	5	0.95
		11.18	6	0.98
2	65	4.71	6	0.96
		3.45	6	1.00
3	65	6.45	6	0.75
		5.71	6	0.69
4	84	24.97	3	0.53
		22.21	3	0.56
5	81	4.30	4	0.89
		5.46	4	0.79

2.1 kN sinusoidal loading pattern when fatigue testing femoral heads.

The choice of the three lag screws was based on biomechanical investigations (Mizrahi et al 1980; Rubin et al 1981). Mizrahi found a triangular three-screw orientation of two superior and one inferior to be the optimum configuration, whilst Rubin showed the importance of parallel screw placement. Both authors, however, found no improvement in fixation if more than three screws were used. The compression hip screw has a sliding barrel for controlled impaction at the fracture site (Clawson 1964). Its blunt nose prevents joint penetration. Biomechanical testing has shown that a large threaded screw provides better internal fixation than a triflanged nail (Brodetti 1961). The advantage of the side plate to provide improved fixation to the femoral shaft was demonstrated by Haboush (1952). In both devices the initial event in failure was impaction of the femoral neck. This underlines the importance of parallel placement of the lag screws allowing backing out, and the controlled sliding action permitted by the compression hip screw.

Frankel (1960) showed that in saw cut femoral neck fractures fixed in vitro, the bone absorbed 75% of the load applied to the femoral head and the device the remaining 25%. The results in Table III confirm Frankel's findings and suggest bone quality to be the most important factor in the rigidity of the bone implant unit.

Using an unphysiological static load and femoral heads from different subjects, other workers have shown that both the sliding compression screw and three lag screws have a superior holding power to the Smith Peterson nail (Brodetti 1961; Van Audekercke et al 1979). However, when comparing parallel lag screws to the sliding compression screw there is disagreement. Van Audekercke et al (1979) and MacKechnie-Jarvis (1983) found both devices equally effective. Husby et al (1987) showed that the sliding compression screw was superior to three lag screws, whilst Elmerson et al (1987) found the reverse, three lag screws providing better fixation than the Richards sliding compression screw. All these studies used unmatched femoral heads from a range of male and female subjects. This factor could have significantly influenced the results since it does not allow for possible variations in bone quality of the different femora under test.

The only report to use paired femoral heads from the same subject compared two Von Bahr lag screws with a sliding compression screw (Engesaeter et al 1984). A static loading regime was used and no significant difference was found between the two devices. The results of our study, using a dynamic loading regime confirm Engesaeter's static test results and show that three parallel screws provide as rigid a fixation as the larger sliding compression screw. In addition, by quantifying the mechanical strength of the femora, we have shown that bone quality was the most important factor in influencing the outcome of fixation.

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REFERENCES

Brodetti A. An experimental study on the use of nails and bolt screws in the fixation of fractures of the femoral neck. *Acta Orthop Scand* 1961; 31:247-71.

BSI. Draft for development. Method for determination of endurance properties of stemmed femoral components of hip joint prostheses 1986. *BSI* (DD91:1986).

Clawson DK. Trochanteric fractures treated by the sliding screw plate fixation method. *J Trauma* 1964; 4:753-6.

Crowninshield RD, Brand RA, Johnston RC. The effects of walking velocity and age on hip kinematics and kinetics. *Clin Orthop* 1978; 132:140-4.

Dalén N, Hellström L-G, Jacobson B. Bone mineral content and mechanical strength of the femoral neck. *Acta Orthop Scand* 1976; 47:503-8.

Elmerson S, Andersson GB, Pope MH, Zetterberg C. Stability of fixation in femoral neck fractures: comparison of four fixation devices in vivo and in cadavers. *Acta Orthop Scand* 1987:58:109-12.

Elson RA, Charnley J. The direction of the resultant force in total prosthetic replacement of the hip joint. *Med and Biol Eng* 1968; 6:19-27.

Engesaeter LB, Asserson O, Molster A, et al. Stability of femoral neck osteotomies fixed by Von Bahr screws or by compression hip screw. *Eur Surg Res* 1984; 16 suppl 2:37-40.

Frankel VH. Mechanical factors for internal fixation of the femoral neck. *Acta Orthop Scand* 1960; 29:21-42.

Griffiths WEG, Swanson SAV, Freeman MAR. Experimental fatigue fracture of the human cadaveric femoral neck. *J Bone Joint Surg [Br]* 1971; 53-B:136-43.

Haboush EJ. Photoelastic stress and strain analysis in cervical fractures of the femur. *Bull Hosp Joint Dis* 1952; 13:252-8.

Husby T, Høiseth A, Fønstelien E. Strength of femoral neck fracture fixation: comparison of six techniques in cadavers. *Acta Orthop Scand* 1987; 58:634-7.

Inman VT. Functional aspects of the abductor muscles of the hip. *J Bone Joint Surg* 1947; 29:607-19.

Khairi MRA, Cronin JH, Robb JA, et al. Femoral trabecular-pattern index and bone mineral content measurement by photon absorption in senile osteoporosis. *J Bone Joint Surg [Am]* 1976; 58-A:221-6.

Klenerman L, Marcuson RW. Intrascapular fractures of the neck of the femur. *J Bone Joint Surg [Br]* 1970; 52-B:514-7.

Lewinnek GE, Kelsey J, White AA, Kreiger NJ. The significance and a comparative analysis of the epidemiology of hip fractures. *Clin Orthop* 1980; 152:35-43.

MacKechnie-Jarvis AC. Femoral neck fracture fixation: rigidity of five techniques compared. *J R Soc Med* 1983; 76:643-8.

Mizrahi J, Hurlin RS, Taylor JK, Solomon L. Investigation of load transfer and optimum pin configuration in the internal fixation, by Muller screws, of fractured femoral necks. *Med Biol Eng Comput* 1980; 18:319-25.

Paul JP. Forces transmitted by joints in the human body. *Proc Inst Mech Engrs* 1967; 181(3J):8-13.

Rubin R, Trent P, Arnold W, Burstein A. Knowles pinning of experimental femoral neck fractures: a biomechanical study. *J Trauma* 1981; 21:1036-9.

Rydell NW. Forces acting on the femoral head-prosthesis: a study on strain gauge supplied prostheses in living persons. *Acta Orthop Scand* 1966; 37 (Suppl 88):1-132.

Van Audekercke R, Martens M, Mulier JC, Stuyck J. Experimental study on internal fixation of femoral neck fractures. *Clin Orthop* 1979; 141:203-12.

Williams PL, Warwick R, Dyson M, Bannister LH. *Gray's Anatomy.* 37th ed. Edinburgh, Churchill Livingstone, 1989:435.