Bone mineral content and fixation strength of femoral neck fractures

A cadaver study

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We measured the bone mineral content (BMC) of 21 human femoral autopsy specimens using Radiographic Attenuation Technique (RAT). The specimens were then tested for stiffness in a material-testing machine. Osteosynthesis was performed after medial femoral neck osteotomy, using 1 of 3 different pinning methods: 2 hook pins (LIH), 2 cannulated screws (Uppsala), and 3 cannulated screws (CHP). The specimens were allocated to each method by grouping them in triplets with similar BMC. Cyclic loading at 500–1000 N and 1 Hz was performed during 50 minutes. The amount of axial compression during loading was measured. Finally, loading to failure was done and post-fixation stiffness and ultimate compression strength recorded.

With one exception, all cases failed where the BMC was less than 0.4 g/cm^2. No differences between the methods of osteosynthesis were found. The fixation strength was mostly dependent on the strength of the bone, i.e., the degree of osteoporosis.

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We have studied in cadavers the effect of bone quality on the strength of 3 methods for fixation of cervical hip fractures.

Material and methods

Fresh human autopsy specimens of the proximal right femur were obtained from 21 cadavers, 18 women, 3 men, aged 83 (63–91) years. The bones were kept frozen at −18 °C; prior to testing, the specimens were thawed overnight at room temperature.

The bone mineral content (BMC) was measured at the center of the femoral neck, using Radiographic Attenuation Technique (RAT; Jonson et al. 1986, Jonson 1988). This method was chosen because it is not as sensitive to fat as photon methods. The accuracy of the RAT method has been shown to be within 4 ±1 percent in predicting the amount of bone mineral in vertebral bodies (Jonson et al. 1987).

Further evaluation of the strength of the intact specimens was done by submaximal (500N) load-deformation analysis. For this purpose, the specimens were mounted in an aluminum fixture and placed in a computerized material-testing machine (MFL E 20-100, Eiserstedt, Germany; Figure 1). The tests were performed at 0, 10, and 20 degrees of valgus. Similar pretesting is described by MacKechnie-Jarvis (1983) and

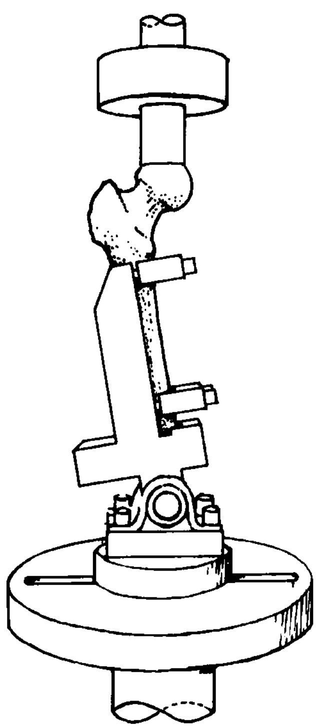

Figure 1. A femur specimen mounted in the material-testing machine.

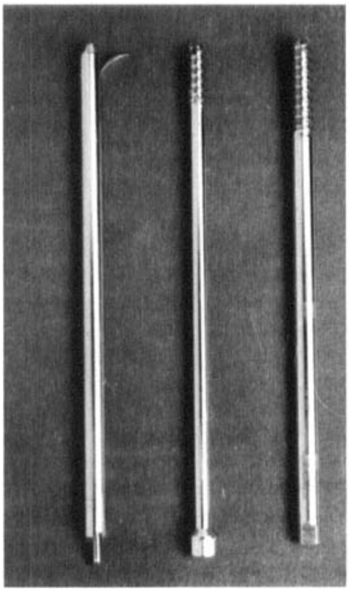

Figure 2. The 3 tested implants. From left, the Hansson hook-pin, the CHP screw, and the Uppsala screw.

Swiontkowski et al. (1987).

The osteotomy was made with a saw medially on the femoral neck and perpendicularly to its axis. For practical reasons, the osteotomy was not totally completed until the osteosynthesis was finished.

The methods chosen for comparison were (A) 2 Hansson hook-pins (LIH, Hansson 1982), (B) 2 cannulated screws (Uppsala, Rehnberg and Olerud 1989a), and (C) 3 cannulated screws (CHP, Lehman et al. 1984), (Figure 2).

The allocation of the specimens to the 3 different osteosynthesis methods was done by stratification according to the BMC value. Thus, the material was grouped into triplets with similar BMC, where each method was represented (Table 1).

The osteosynthesis was performed with the aid of fluoroscopy and according to the manuals of the manufacturers. The implants were placed at a 135-degree angle to the femoral shaft, and were allowed to come within a few millimeters of the contour of the femoral head. In the case of the Uppsala screw, a subchondral position was attempted, as recommended by its designers. With the CHP screw, the osteotomy was gently compressed.

Table 1. Data and results of the 21 bones in the study

A	B	C	D	E	F	G	H	I	J	K	L	M	N
1	91	C	0.681	3.13	2.14	1.56	1.22	0.98	0.14	0.09	2.43	1.39	2030
2	78	H	0.628	1.45	1.63	1.24	2.1	0.77	0.12	0.05	3.0	1.31	2450
3	63	U	0.548	1.73	1.70	1.38	–	0.96	0.08	0.07		1.51	2450
4	72	H	0.515	1.38	1.52	1.24	2.28	0.85	0.15	0.06	3.34	1.23	1750
5	91	C	0.501	1.51	1.57	1.43	1.48	0.97	0.16	0.07	2.68	1.48	1600
6	81	U	0.488	1.54	1.63	1.11	1.6	0.63	0.19	0.12	2.5	1.36	1600
7	74	H	0.467	1.48	1.55	1.15	1.9	0.40	0.13	0.07	2.5	1.18	1875
8	82	C	0.459	1.05	1.17	1.04	2.24	2.03	0.19	0.12	4.57	0.625	1125
9	87	U	0.458	1.38	1.65	1.42	2.5	0.30	0.14	0.10	3.0	1.35	2265
10	78	C	0.456	1.53	1.59	1.59	1.6	0.24	0.18	0.12	2.1	1.46	2235
11	84	H	0.431	1.16	1.28	1.13	1.78	0.56	0.20	0.11	2.65	1.16	1480
12	89	U	0.430	1.14	1.19	0.925	2.3	1.8	0.14	0.12	4.4	1.40	1600
13	85	C	0.411	1.33	1.43	1.20	1.72	0.43	0.11	0.07	2.32	1.19	1750
14	75	U	0.410	0.931	1.05	0.984	3.08	0.55	0.12	0.06	3.81	0.888	1500
15	90	H	0.400	1.75	1.78	1.32	–	–	–	–	–	1.46	2175
16	80	H	0.393	0.915	0.89	0.905	3.5	–	–	–	–	0.694	1025
17	82	C	0.359	1.05	1.19	1.12	3.0	1.2	0.08	0.07	4.35	0.809	1550
18	85	U	0.354	0.625	0.75	0.745	4.8	5.1	1.04	–	–	0.35	1090
19	87	U	0.350	0.945	0.96	0.90	–	–	–	–	–	0.363	560
20	87	H	0.345	0.425	0.55	0.98	–	–	–	–	–	0.413	875
21	91	C	0.319	0.90	0.96	0.89	–	–	–	–	–	0.275	537

A Specimen number
B Age
C Osteosynthesis method
  C CHP screws
  H Hansson hook-pins
  U Uppsala screws
D Bone mineral content, g/cm^2
E Stiffness of intact specimen at 0 degree force/diaphysis angle, MN/m
F Stiffness of intact specimen at 10 degrees force/diaphysis angle, MN/m
G Stiffness of intact specimen at 20 degrees force/diaphysis angle, MN/m
H Compression (deformation) during first cycle, mm
I Compression during cycles 2–1000, mm
J Compression during cycles 1001–2000, mm
K Compression during cycles 2001–3000, mm
L Total compression during cyclic loading
M Stiffness of osteosynthesized specimen after cyclic loading, MN/m
N Ultimate compression strength, N

With the mounted specimen in 10 degrees of valgus, cyclic loading of 500–1000N and 1 Hz was performed for 50 minutes (3000 cycles), roughly corresponding to a walk of about 3000 meters of a patient with crutches (Davy et al. 1988).

The amount of axial compression was measured with an accuracy of 0.01 mm. Force and displacement were continuously recorded by an x-y recorder and also with a Grass multichannel strip chart recorder. The compressions during the first cycle (Phase 1), and during cycles 2-1000 (Phase 2), 1001-2000 (Phase 3), and 2001-3000 (Phase 4), respectively, were recorded. In one of the bones surviving the cyclic loading, there was a technical complication making the recorded compression higher than the actual one. The compression values of this bone were not included in the results.

After the cyclic loading, a final static loading to failure was done, from which post-fixation stiffness and ultimate compression strength were calculated.

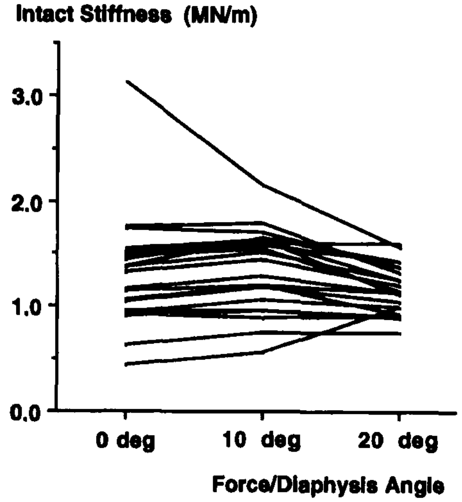

Figure 3. Stiffness at different force/diaphysis angles. Each line connects the measurements of one specimen.

Statistics

The Kruskal-Wallis non-parametric test for comparison of groups was used to compare the 3 osteosynthesis methods. The Spearman rank test was used to evaluate the correlations between the bone quality variables and the fixation strength variables. The Wilcoxon non-parametric rank sum test was used for the analysis of the BMC threshold value. 2-tailed tests were used and *P*-values less than 0.05 were considered significant.

Results

In the submaximal loading test of the intact bones, the stiffness tended to be highest at 10 degrees of valgus (Table 1, Figure 3). The stiffness at 10 degrees of valgus showed a good correlation to the BMC value (r 0.79, $P < 0.001$).

In the 16 specimens which survived all 3000 cycles, the pattern of compression during cyclic loading was consistent. During the first cycle, the osteotomy space of about 1.5 to 2.5 mm was closed. Compression during the first cycle was only marginally lower in the CHP specimens, where intraoperative compression had been performed. During the remaining 3000 cycles, the compression was continuously diminishing, and during the last 1000 cycles it never exceeded 0.12 mm (Table 2).

In 2 specimens, the osteosynthesis failed after about 20 and 1500 cycles, respectively. In these 2 cases, there was a fast compression and no tendency towards stabilization. In 3 other specimens, the osteosynthesis failed during the first cycle, i.e., the ultimate compression strength was less than 1000 N.

The amount of compression during cyclic loading was not correlated to BMC or intact stiffness in the

Table 2. Median (range) of some variables, according to method of osteosynthesis

Method	BMC (g/cm^2)	Intact stiffness at 10 degrees of valgus (MN/m)	Compression cycle 1 (mm)	Compression cycles 2001–3000 (mm)	Post-fixation stiffness (MN/m)	Ultimate compression strength (N)
Hansson hook-pins	0.43 (0.35–0.63) n 7	1.52 (0.55–1.78) n 7	2.1 (1.78–3.5) n 5	0.08 (0.05–0.11) n 4	1.18 (0.41–1.46) n 7	1750 (875–2450) n 7
CHP screws	0.46 (0.32–0.68) n 7	1.43 (0.96–2.14) n 7	1.7 (1.22–3.0) n 6	0.08 (0.07–0.12) n 6	1.19 (0.28–1.48) n 7	1600 (537–2235) n 7
Uppsala screws	0.43 (0.35–0.55) n 7	1.19 (0.75–1.70) n 7	2.5 (1.60–4.8) n 5	0.10 (0.06–0.12) n 5	1.35 (0.35–1.51) n 7	1600 (560–2450) n 7

specimens that completed the 3000 cycles. However, there was a threshold BMC value of 0.4 g/cm^2 below which an osteosynthesis could not be expected to hold during the cycling period ($P < 0.001$).

Both post-fixation stiffness and ultimate compression strength showed a very good correlation with the intact stiffness value (r 0.85, $P < 0.001$ and r 0.89, P 0.0001, respectively). The correlation with BMC was somewhat weaker (r 0.69, $P < 0.01$ and r 0.74, P 0.001, respectively).

The mode of failure was in all cases a longitudinal fracturing of the calcar femorale, as the pins moved medially and distally in the collum.

No differences were found between the 3 different osteosynthesis methods when comparing compression, post-fixation stiffness, or ultimate compression strength (Table 2).

Discussion

Frankel (1959) pointed out that differences in the performance of implants during loading could be masked by differences in the strength of the bone. Using the contralateral femur as control (Engesæter et al. 1984, Husby et al. 1987, Swiontkowski et al. 1987, Clark et al. 1990, Goodman et al. 1992), pre-loading (MacKechnie-Jarvis 1983, Swiontkowski et al. 1987), or bone mineral measurements (Swiontkowski et al. 1987) can standardize for this. The specimens in this study were stratified according to the bone mineral content measured by a radiological method, RAT, and were also tested for stiffness prior to osteotomy, to reduce a possible bias due to bone strength.

Deep-freezing of bone does not cause any major alteration in compressive stiffness or hysteresis (Goh et al. 1989). Complete drying of bone can cause a 5 percent reduction in strength (Currey 1988). However, the specimens in this study were kept humid, and this reduced the error (Nachemson 1960).

In general, we chose older female femora to simulate the clinical situation as much as possible. This may explain differences in the ultimate strength values, as compared to those of van Audekercke et al. (1979), for example.

The compressive loading was accomplished at 10 degrees of valgus, where the specimens were stiffest. The force angle recorded in vivo from telemeterized hip prostheses is between 15 and 20 degrees (Rydell 1966, Davy et al. 1988). Further, there was only one direction of force. These factors, like the osteotomy itself, represent differences from the clinical situation, but, on the other hand, make the experimental situation reproducible. Such methodological biases are not systematically in favor of any of the tested devices.

Our results are in accordance with those of Swiontkowski et al. (1987) and Springer et al. (1991) in that they were not able to show any significant differences in bending strength between different multiple pinning fixations.

A power analysis of our results showed that 78 more specimens would be needed in order to show significance regarding any one of the result parameters recorded, more exactly, the rate of compression during Phases 2 , 3, and 4 (39 CHP and 39 Hansson hook-pins). To obtain significance between the same 2 methods regarding ulitimate compression strength, another 788 specimens would be needed (80 percent power). Such an extension of the material would not help when applying the results to the clinical situation. MacKechnie-Jarvis (1983), using an experimental setup similar to ours, found that a highly rigid screw-plate assembly resisted a compression force better than multiple pins or a trifin nail. Clark et al. (1990), also using cyclic loading, did not find any difference between a sliding screw-plate and three cannulated lag screws. However, screw-plate devices offer poor rotational stability (Husby et al. 1989) and also introduce a rotational momentum during insertion of the screw (Elmerson et al. 1987).

Husby et al. (1987), using a vertical osteotomy of the femoral neck, were able to show that von Bahr, AO, and Haukebø screws gave somewhat better results than the other tested methods, although each group consisted of only 6 specimens. In our study, with 7 specimens in each group, no difference was found. The transverse type of osteotomy in our study, which better resembles the situation in vivo, causes a greater part of the loading force to be taken up by bone. This might explain why we found no significant differences between the three osteosynthesis devices, but a significant relation between BMC and the strength of the fixed oseotomies.

In clinical trials, the flanged nail has been shown to be inferior to several other methods (Barnes et al. 1976, Strömqvist et al. 1984, Nilsson 1989). This results partly from damage to the circulation to the femoral head during the operative procedure (Strömqvist et al. 1983). The sliding screw-plate was better than the flanged nail, but inferior to 4 cancellous screws in prospective randomized studies from Denmark (Linde et al. 1985, Nordkild et al. 1985, Madsen et al. 1987). Considering these data, we decided to choose only modern multiple pins/screws for our experiment.

Our results focus clearly on the importance of bone quality for the fixation strength of a fragility fracture like the femoral neck fracture. The finding of a threshold BMC value of 0.4 g/cm^2, below which all but one specimen failed during cyclic loading, suggests that a primary arthroplasty should be considered in the most osteoporotic cases of femoral neck fracture.

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