Calcar unloading after hip replacement

A cadaver study of femoral stem designs

The load transfer from the pelvis to the femur in extension and 45° of flexion was examined in normal hips and hips replaced with 11 different prostheses. The cadaver specimens were mounted in a jig and the load patterns recorded by strain gauges attached to critical points determined in a pilot study, with the aid of a photoelastic coating. All the prostheses except an "isoelastic" model showed marked reduction of strain in the calcar area compared with the normal femur. A collar increased the strain to about 40 per cent of normal. No significant difference was noted between curved and straight stems. There were no major differences between the prostheses on loading in flexion.

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Stress shielding due to load discrepancies between the components of an artificial joint and bone may cause bone loss and loosening (Bobyn & Engh 1983, Malchau et al. 1986). Loosening may also be caused by increased stress, e.g., by rotational forces acting on a hip prosthesis in the flexed position (Carlsson et al. 1977, Hampton et al. 1980, Wroblewski 1979); rotation may interfere with bony ingrowth and anchoring of uncemented prostheses (Albrektsson et al. 1981, Mjöberg et al. 1984). We have compared loading characteristics of different hip prostheses in cadavers.

Materials and methods

The prostheses were cemented or uncemented, with straight or curved stems and with or without a collar. In addition, one so-called isoelastic prosthesis was tested. Two stems were made of a titanium alloy, two of stainless steel, and seven of cobalt-chromium alloy (Figure 1).

The specimens were removed as soon as possible after death from accident-killed men aged 35–45 years. Each specimen consisted of part of the pelvis including the acetabulum, and about 25 cm of the femoral shaft. The hip joint capsule was left intact, but all the other soft

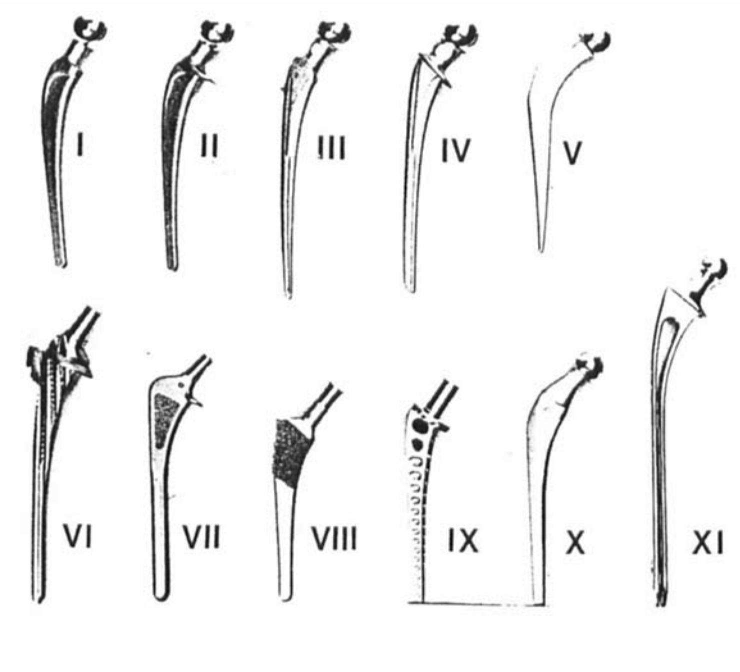
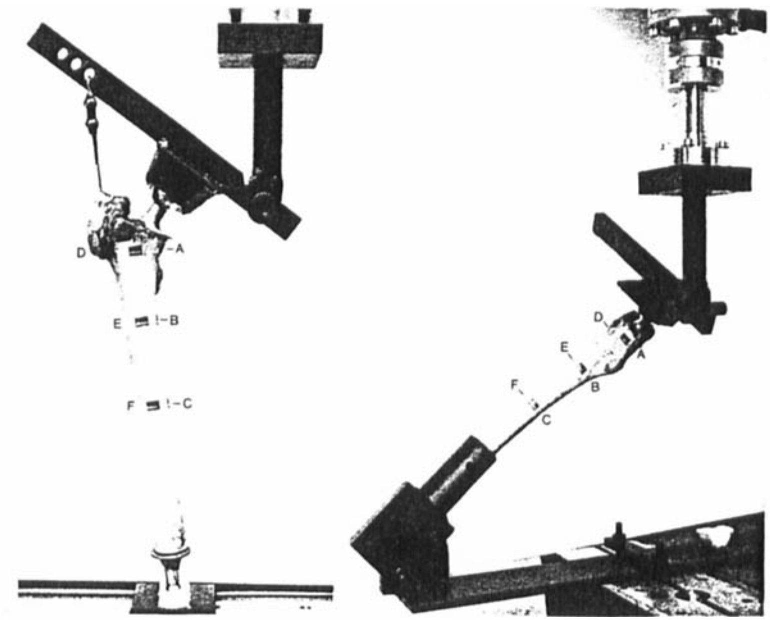


Figure 1. The prostheses tested.
Cemented stems:
I. Scanhip® without collar;
II. Scanhip® with collar;
III. Lubinus SP® without collar;
IV. Lubinus SP® with collar;
V. Exeter®.
Uncemented stems:
VI. Rippenprothese Waldemar Link®;
VII. Harris/Galante®;
VIII. PCA®;
IX. Authophor®;
X. Osteonics®;
XI. Butel isoelastique®.
I, II, III, IV, VIII, IX, and X are made of cobalt-chromium, V and XI of stainless steel, and VI and VII of titanium alloy.



A B

Figure 2. The jig simulating single-limb stance in A extension and B 45° of flexion with the strain gauges A–F.

tissues were removed. The specimens were stored at −23° and thawed immediately before the experiment to minimize drying (Sedlin 1965, Evans 1973).

The specimen was mounted in a special jig that allowed loading in an Alvetron TC10® material testing machine (Figure 2). A continuous static load of 500 N was applied in extension and 45° of flexion both on the intact femur and with a hip prosthesis inserted. The sampling of data was finished within 1 minute after loading. Modern cementing technique was used including pressurization by distal plugging and a cement gun. The cement was of a high viscosity type (CHW®) and the procedure was checked by radiography. Calcar-collar contact was secured by reaming, and no cement was allowed to interpose between bone and the collar.

In the extended position the abductor force was simulated by a steel wire attached to the greater trochanter, the distal end of the femur being free to rotate in a ball and socket joint. In the flexed position the distal end of the femur was fixed in a tube with Wood's metal. Measurements during the cadaver dissection gave the length of the lever arms. The load was applied medially to act as partial body weight in single limb stance, transmitted through the moveable lever arm, resulting in compression and abductor forces. In flexion no muscle forces were simulated, and a bending moment was produced where the femur entered the tube, decreasing proximally so as not to interfere with the results. The variation in lever arm and offset was kept as small as possible when the various prostheses were tested. To check the primary stability of the implantation, we used a mechanical dial gauge (Teclock®, precision 0.01 mm), which disclosed no movement between femur and prosthesis in any case. Each prosthesis was tested three times using a new femur and new strain gauges.

Strain measurements. Six strain gauges of rectangular rosette type (CEA-XX-062UR-350, Measurement Group, Inc. North Carolina) were applied to the bone with methyl-2-cyanoacrylate glue after thorough cleaning. The proper positioning of the gauges was determined in an initial experiment on one specimen covered with a photoelastic coating (Photostress®; Measurement Group Inc. North Carolina). The strain pattern could be observed in polarized light when the specimen was loaded. Patterns were determined for the intact specimen and for an uncemented (VIII) and a cemented prosthesis (V). In the intact femur in extension, the area of the maximum stress was located on a level just below the lesser trochanter on both the medial and the lateral aspect. In the prosthesis specimens the high stress area on the medial aspect was situated more distally. On the lateral aspect, stress was even more concentrated to the area around the stem tip. The cemented stem showed less marked stress concentrations than the uncemented one. With reference to this stress analysis, three gauges were applied to the medial aspect and three to the anterior aspect. On the medial aspect the first gauge was placed 5 mm below the osteotomy (A), the second within the area of maximum stress shown in the photoelastic experiment (B), and the third 1 cm below the tip of the prosthesis (C). The remaining three strain gauges were placed on the anterior aspect of the femur at the same levels, corresponding to the areas loaded in the flexed position (C, D, and F; Figure 2). The reduction of the strain gauge data was done by standard equations and the maximum and minimum principle strains (tension and compression), directed at right angles to each other, were calculated (Perry 1979).

The Wilcoxon test was used for statistical analysis for differences between the intact femur and the prosthesis. One-way ANOVA was used for differences between the prostheses, with a further breakdown using the Student's *t*-test (Armitage 1974).

Results

The highest strain level was noted in gauge A (Table 1). In general, in the extended position, the strains measured by gauges A and B were lower for the prostheses than for the intact femur. In the flexed position differences were recorded in gauges A, B, D, and E as compared with intact femur (Table 2). All the prostheses showed the same pattern except no. XI, which had no stress reduction at the proximal test points in the extended position (Figure 3). For all the prostheses, there was only slight, though significant, stress reduction at the middle level (gauges B and E); and at the lowest level (gauges C and F), there was almost no difference between the intact femur and the femur with a prosthesis. The uncemented prostheses showed higher tensile strains in gauge A in extension than did cemented ones (hoop stress); and the collared prostheses developed higher strain levels proximally, though still about 40 per cent lower than with the intact femur. No major differences were found between straight or curved stems, even in the flexed position. The significances concerning straight and curved stems (Table 2) indicate only small differences.

In flexion the highest strain occurred distally, partly caused by the bending moment caused by the jig. As in the extended position, the differences between different prostheses were small.

In extension the direction of the principal strains were closely related to the long axis of the femur, and in flexion, as expected, the axis became more oblique. No major differences were noted, neither between the normal femur and prostheses nor among the separate prostheses (Table 3).

Discussion

In our experiment a static load was applied in the test jig, which is a simplification of natural conditions where dynamic loads play an important role. Furthermore, muscle forces can probably produce much higher peak loads than those used in our experiments; only the abductors were simulated in the jib, but all the tests were performed in the same way. Similar experiments have been described by Andriacchi et al. (1976), Brockhurst & Svensson (1977), Oh & Harris (1978), McBeath et al. (1979), Crowninshield et al. (1980, 1981), Rohlman et al. (1983), Huiskes (1980), Voilin (1983), Lewis et al. (1984), and Panjabi et al. (1985).

An important feature of our experiment is the fact that the load patterns caused by the different stems were examined under a standard load. Even if higher dynamic loads are involved under natural conditions, the load pattern may be the same. The load required to dislodge the stem was not examined. Other important differences may exist between the stems in the load-bearing capacity of the bone-stem interface. For example, it is claimed that stem no. X (Osteonics®) is better able to carry a load without stem movement owing to the step pattern on the lateral sides (Murphy et al. 1985).

Our test of uncemented prostheses simulates the period before bony ingrowth. This is important because both the stability of the stem in the femur and the loading pattern of the femur will determine whether ingrowth or bone resorption will occur after implantation (Albrektsson et al. 1981). It could be suggested that a stem that transfers load to the distal end of the stem, reducing load in the proximal region, will be more likely to cause problems in clinical use owing to proximal bone resorption (Bobyn & Engh 1983, Malchau et al. 1986). Proximal stress shielding and distal loading increase the risk of loosening and stem fracture, a risk that seems to be inherent in stem designs nos. VII–X. However, also the cemented designs reduced the load in the proximal femur as compared with the intact femur. The reduction found by us is of the same magnitude as previously described by others. We have confirmed that a collar significantly increases calcar load even with the same stem design (see Oh &

Table 1A. Maximum and minimum principal strain (+=tension and −=compression) in gauges A–F in extension for intact femur and prostheses nos. I–XI. Microstrain (m/m), mean values and SD below

	Extension											
	A		B		C		D		E		F	
	max	min	max	min	max	min	max	min	max	min	max	min
femur	67	−1229	292	−1154	253	−910	363	−453	204	−178	194	−81
	46	112	63	74	43	101	65	75	48	69	78	5
I	103	−53	201	−637	199	−725	508	−116	240	−204	60	−27
	75	48	67	164	81	124	252	11	81	6	14	5
II	164	−478	235	−750	233	−802	241	−100	184	−164	163	−35
	90	95	78	88	31	39	25	31	26	22	40	22
III	28	−23	238	−789	238	−755	195	−138	206	−255	88	−47
	38	7	29	117	68	212	24	59	39	106	15	34
IV	116	−176	184	−780	235	−767	220	−225	202	−193	143	−42
	16	67	23	84	55	98	64	99	24	25	46	29
V	201	−21	250	−834	190	−814	365	−161	278	−184	176	−70
	68	10	26	37	20	20	41	25	48	15	33	11
VI	295	−300	173	−763	242	−834	203	−279	142	−205	53	−79
	94	121	7	42	10	22	71	84	12	19	6	57
VII	292	−63	177	−638	227	−781	304	−119	228	−137	51	−27
	163	17	31	107	23	20	66	29	68	10	29	18
VIII	356	−131	243	−815	263	−957	313	−227	310	−282	183	−80
	262	92	43	98	44	59	67	31	69	64	17	6
IX	478	−188	257	−904	241	−889	300	−122	228	−249	149	−62
	67	85	85	109	55	145	66	57	20	51	34	29
X	226	−45	218	−696	271	−901	264	−229	219	−227	175	−54
	68	29	35	83	66	136	44	79	41	70	18	35
XI	154	−1100	257	−935	195	−823	293	−416	179	−210	69	−81
	48	53	38	111	76	30	46	66	54	90	9	12

Table 2. Statistical analysis. Third row shows one-way ANOVA among all prostheses. Where no differences were found, no further breakdown was made (–). Third row shows intact femur versus the "isoelastic" prosthesis. X=P<0.05, XX=P<0.01, XXX=P<0.001

	N	Extension						45° Flexion					
Gauge		A	B	C	D	E	F	A	B	C	D	E	F
Femur versus cemented	19	XX	XX	NS	NS	NS	XX	XX	XX	NS	XX	XX	NS
Femur versus uncemented	22	XX	XX	NS	XX	NS	NS	XX	XX	NS	XX	XX	NS
Femur versus XI	7	NS	NS	NS	NS	NS	X	NS	NS	XX	XX	NS	NS
ANOVA among prostheses	33	XXX	NS	NS	XXX	NS	XXX	XXX	NS	XXX	XXX	XXX	NS
Cemented versus uncemented	33	XXX	–	–	XX	–	NS	NS	–	X	NS	NS	–
Cemented with versus without collar	15	XXX	–	–	NS	–	NS	X	–	NS	XX	NS	–
Uncemented with versus without collar	18	NS	–	–	NS	–	XXX	XX	–	NS	XXX	NS	–
Cemented straight versus curved stem	15	NS	–	–	NS	–	NS	X	–	NS	NS	NS	–
Uncemented straight versus curved stem	18	NS	–	–	X	–	NS	NS	–	XX	X	NS	–
II versus IV	6	X	–	–	NS	–	NS	X	–	NS	XX	NS	–
IX versus uncemented	18	X	–	–	NS	–	NS	NS	–	NS	NS	NS	–
XI versus uncemented	21	XXX	–	–	XXX	–	NS	XX	–	NS	NS	XXX	–

Table 1B. Maximum and minimum principal strain (+ = tension and − = compression) in gauges A–F in 45° flexion for intact femur and prostheses nos. I–XI. Microstrain (m/m), mean values and SD below

	45° Flexion											
	A		B		C		D		E		F	
	max	min	max	min	max	min	max	min	max	min	max	min
femur	69	−330	252	−275	438	−215	250	−168	443	−173	748	−269
	10	34	53	23	64	17	29	47	24	26	72	29
I	83	−108	211	−146	471	−181	170	−12	294	−116	587	−237
	29	46	56	6	179	37	61	7	34	12	47	56
II	80	−195	170	−160	454	−163	123	−64	307	−139	643	−182
	36	55	13	34	85	13	41	23	10	26	30	27
III	63	−92	215	−138	574	−236	76	−36	410	−155	681	−226
	8	8	68	26	81	26	32	32	20	46	184	86
IV	57	−89	145	−160	313	−196	109	−64	264	−105	597	−192
	14	34	53	48	75	76	22	33	82	28	216	72
V	49	−57	208	−186	267	−178	93	−22	321	−148	529	−332
	4	17	23	17	80	10	18	17	11	14	16	17
VI	69	−62	179	−177	239	−171	121	−60	277	−113	681	−225
	1	30	3	21	35	23	33	27	37	6	31	8
VII	33	−31	155	−110	410	−157	85	−47	322	−145	674	−259
	29	10	10	2	3	30	9	9	9	42	27	35
VIII	77	−93	222	−185	283	−210	231	−87	343	−148	597	−225
	27	129	14	18	84	11	18	15	11	5	20	8
IX	62	−70	253	−164	410	−191	71	−17	343	−152	608	−193
	8	15	58	17	110	±6	15	7	4	28	61	73
X	240	−119	220	−154	337	−182	150	−51	288	−201	675	−247
	22	11	85	58	19	11	37	28	31	61	77	19
XI	39	−190	247	−199	208	−182	129	−55	425	−178	699	−245
	18	36	123	2	20	53	50	23	13	50	21	32

Table 3. The direction (°) of the minimum principle strain (compression). 0 is parallel to the middle segment of the gauge (the long axis of femur). − indicates clockwise rotation

	Extension						45° Flexion					
Gauge	A	B	C	D	E	F	A	B	C	D	E	F
femur	−11	−1	0	50	55	71	27	29	22	59	75	82
I	14	−3	−1	62	43	73	49	42	68	75	80	84
II	6	−1	−1	62	48	86	46	41	63	50	80	83
III	12	0	1	46	47	82	62	44	75	84	82	85
IV	7	2	1	46	46	87	38	33	66	68	81	83
V	4	0	−1	53	58	89	55	39	54	69	79	81
VI	9	−1	0	55	27	89	29	32	56	47	74	84
VII	5	1	0	59	42	90	54	41	67	72	83	84
VIII	24	5	0	42	49	71	44	38	46	71	76	82
IX	1	1	−2	52	54	79	67	44	63	83	85	84
X	8	1	0	43	55	79	40	41	60	62	81	84
XI	−6	1	0	44	47	80	52	32	62	62	80	85

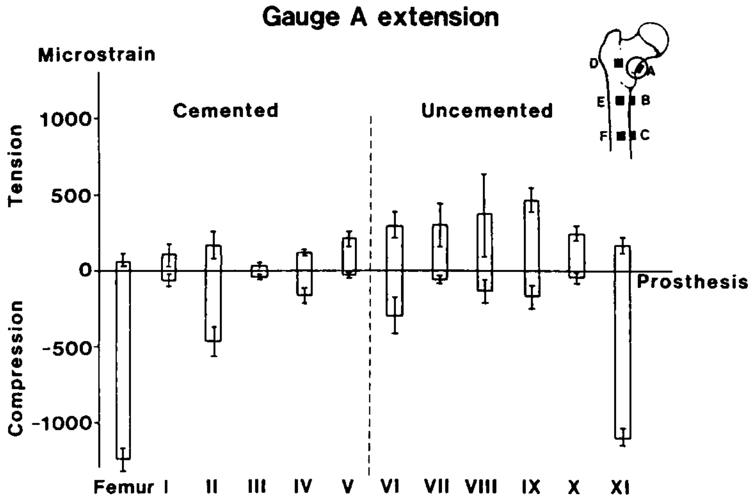


Figure 3. Maximum (tension) and minimum (compression) principal strains for intact femur and all the tested prostheses in gauge A extension. Load on the jig is 500 N. Mean values, microstrain (*m/m*).

Harris 1978, Lewis et al. 1984, Crowninshield et al. 1980, 1981, Ytteram et al. 1980). The load reduction in the proximal femur seems to be large enough to explain the resorption of the calcar sometimes seen clinically (Hierton et al. 1983), and there is probably no need to explain it by a disturbance in blood supply, as was suggested by Blacker & Charnley (1977). The load increase caused by the collar was almost matched by the hoop stress created by the conical uncemented stems, even though the normal load was much higher in gauges A and D in extension. In flexion, differences between the prostheses and the normal femur were as a rule small. There was no major difference between straight and curved stems, which shows that good load transfer can be obtained with both types of stems even in flexion.

The best prosthesis of our experiment was the "isoelastic" no. XI, in which an attempt has been made to reduce the stiffness of the prosthesis to that of the femur (Voilin 1983); the stiffness is considerably reduced compared with the other prostheses in the test. Through the rod construction of the stem, volume is maintained; this allows good contact, and the surface area is increased by increasing the stem length. The prosthesis is equipped with a collar to improve calcar load. Our experiment showed that the designer has succeeded to a considerable extent in preserving the normal load pattern of the femur after implantation. However, there may still be deficiencies in the ability of the prosthesis to carry high peak loads under activity without displacement or fracture. This will have to be tested by further experiments and clinical experience. Yet another "isoelastic" prosthesis, not available to us, has been described by Morscher et al. (1981) and Bombelli & Mathys (1982); it consists of a central stem surrounded by polyacetal.

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