Cold flow reduced by metal backing

An in vivo roentgen stereophotogrammetric analysis of unicompartmental tibial components

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Thirty-six cases of unicompartmental arthroplasty for gonarthrosis were followed for 6 years. The arthroplasties were prepared for roentgen stereophotogrammetric analysis (RSA) by marking the tibial components and the tibial methaphysis with tantalum balls. Nineteen out of 24 cases with a polyethylene tibial component had an increase in intermarker distances signifying progressive enlargement of the prosthesis due to cold flow. The enlargement was up to 2.8 percent of the circumference and was greater for 9 mm than for 12 mm thick components. Twelve cases with a metal-backed tibial component showed no cold flow. No correlation between cold flow and demographic, clinical, or radiographic data was found.

Deformation of the polyethylene of tibial components has been reported in unicompartmental prostheses (Knutson et al. 1981, Jònson 1981, Marmor 1985), as well as in total knee prostheses (Ducheyene et al. 1978, Wright et al. 1982). Therefore, metal backing to support the polyethylene has been advocated to stiffen the prosthesis and decrease the risk of mechanical loosening (Bargren et al. 1978, Walker et al. 1981).

The problem of polyethylene cold flow or creep, although less often observed in clinical practice, is well recognized (Weightman 1979, Goodfellow and O'Connor 1978, Wright et al. 1982, Walker et al. 1976). Hypothetically, slow expansion of the prosthesis will, with time, create stresses in the interface bond, which may eventually break.

Roentgen stereophotogrammetric analysis (RSA; Selvik 1974, 1989) has been used to measure tibial component fixation (Ryd 1986, Ryd et al. 1986, 1987, 1988). In this study of cold flow in polyethylene, Marmor tibial components were compared with similar, but metal-backed tibial components of the Lund prostheses.

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Material and methods

Thirty-six knees were operated on for femorotibial gonarthrosis, Ahlbäck (1968) Stages I–III. In 24 knees, operated on between November 1978 and August 1981, the Marmor prosthesis with a UHMW polyethylene tibial component was used (Marmor 1973). In 12 knees operated on between November 1981 and November 1982, the Lund prosthesis was used. The Lund prosthesis was a prototype representing a step in the evolution of the PCA unicompartmental knee. The Lund prosthesis had an articulating geometry similar to that of the Marmor prosthesis, but with a metal tray (≈ 3-mm thick) and a keel underneath the polyethylene (Figure 1). These cases were the first arthroplasties in the knee to which RSA was applied. The study was prospective,

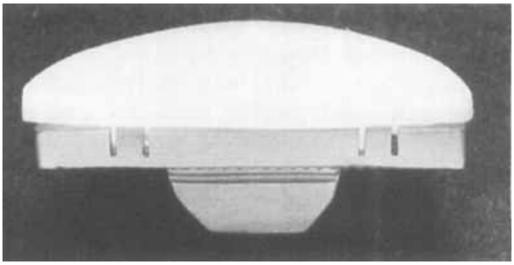

Figure 1. The metal-backed tibial component of the Lund prosthesis seen from the side facing the tibial eminence.

Table 1. Demographic, clinical, polyethylene cold flow, and radiographic data for 24 Marmor knees and 12 Lund knees

Case[a]	Age	Weight	Lund score	Tibial[b] component	Cold[c] flow	PT	PTS	HKA
1	65	72	84	9	1.5	83	91	177
2	64	99	91	12	0.5	86	86	182
3L[a]	72	65	88	9	0	86	83	181
4	66	88	93	12	1.4	87	81	186
5	64	70	76	12	1.2	78	86	173
6	70	70	93	15	0	89	90	173
7	69	97	81	9	2.8	85	95	169
8	68	63	96	12	0	89	89	178
9	65	84	93	12	0	83	89	186
10	70	77	90	9	1.1	83	83	174
11	75	65	58	12	0.8	88	87	196
12L	75	65	59	9	0	94	96	156
13	62	74	80	9	0	85	81	178
14	65	93	65	12	1.1	92	86	180
15	70	70	94	9	0.9	83	85	177
16	68	94	95	12	0.8	92	76	178
17	64	68	98	9	1.9	81	91	172
18	75	70	78	12	0.4	88	82	181
19	73	83	93	9	0	92	85	183
20	65	83	91	12	0.7	75	86	173
21	74	62	95	9	1.3	85	88	181
22	66	82	77	9	2.0	87	93	178
23L	72	81	87	9	0	84	89	178
24	70	69	87	9	0.4	90	94	180
Mean	69	77				86	87	178
25	70	72	92	9	0	81	77	172
26L	75	73	92	9	0	90	92	182
27	78	78	90	12	0	86	90	170
28	66	80	87	9	0	86	90	176
29	72	104	89	9	0	88	92	177
30	65	82	95	9	0	85	89	179
31	77	88	93	9	0	84	83	168
32	78	68	91	9	0	89	88	181
33L	75	72	83	12	0	88	77	180
34	67	85	93	9	0	84	88	176
35	65	82	90	9	0	86	86	177
36	65	80	91	12	0	84	79	175
Mean	71	80				86	86	176

[a] L Lateral compartment, all other medial.
[b] Height (mm).
[c] Percentage increase of prosthetic circumference.
PT Medial angle between the prosthetic plane and the long axis of tibia.
PTS Posterior angle between the prosthetic plane and the long axis of tibia.
HKA Postoperative mechanical axis of the entire limb (Hip-Knee-Ankle angle).

but not consecutive; the selection of cases, however, was solely related to the availability of RSA during this initial period. There were 4 men and 30 women with a mean age of 69 (62–78) years (Table 1). Thirty-one replacements were in the medial compartment, whereas five—3 Marmor and 2 Lund—were lateral compartment replacements.

All the operations were performed with a positioning jig to optimize the bone resection (Lindstrand et al. 1982), and the components were always fixed to the bone with bone cement containing barium sulfate. Follow-up, including radiographic examination, was at 6 weeks, 6 months, and annually thereafter. All the knees were followed for 6 years with the exception of 6 patients, 4 of whom died after 2, 3, 4 and 5 years, respectively (Cases 26, 29, 10, and 27) and two which failed (Cases 11 and 12). At 6 years, 1 additional case (Case 14) was regarded as a failure and later revised for collision between the femoral component and the tibial eminence. The remaining cases were all satisfactory.

Radiographic examination

The mechanical axes (Hip-Knee-Ankle [HKA] angle) of the lower extremity (Maquet et al. 1967) were determined postoperatively, usually after 6 weeks, using a standardized method (Lindstrand et al. 1982). At this time, the position of the tibial component with reference to the long axis of the tibia in the frontal (PT angle) and lateral (PTS angle) projections was determined.

Roentgen stereophotogrammetric examination

RSA included the peroperative insertion of 0.8-mm tantalum markers in the tibial component and tibial methaphysis and simultaneous biplanar exposures with the knee inside a calibration cage (Selvik 1974). In the Marmor components the markers were inserted from underneath through slightly undersized holes, 2-mm deep, made with a dentist's drill. The markers were instead inserted peripherally in the Lund components using the same technique to avoid the metal backing. A high-resolution digitizing table (Hasselblad Engineering, precision 10-μm) was used, and a computer program "X-RAY" provided the 3-D coordinates of the prosthetic markers. By the program "GROWTH," included in the RSA system, the intermarker distances were obtained and compared with those of the first postoperative examination. The accuracy of RSA in this application is approximately 75 μm at the 99 percent confidence level (Kärrholm et al. 1984). When each intermarker distance increased by more than 100 μm, cold flow was judged to have occurred, and was presented as a percentage of the entire circumference of the prosthesis.

Statistical methods

The Mann-Whitney *U*-test or Wilcoxon rank sum test were used to compare unpaired or paired RSA data, respectively. The Spearman *rho* test was used to correlate RSA findings with clinical and demographic parameters.

Results

The intermarker measurements showed an increase in the circumference by cold flow of 19 Marmor prostheses after 6 years. This cold flow ranged from 0.4 to 2.8 percent. The cold flow was progressive

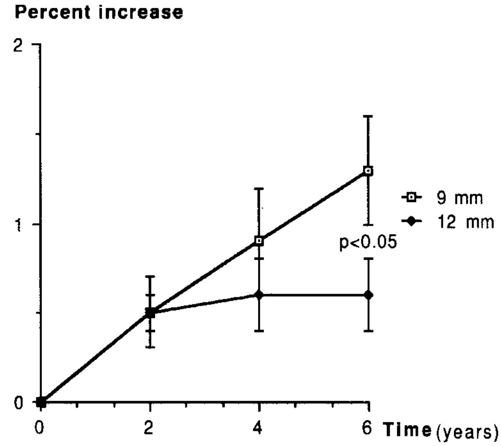

Figure 2. Polyethylene cold flow as percentage increase of prosthetic circumference (mean ± *SEM*) of Marmor 9-mm- (n 13) and 12-mm- (n 10) thick tibial components. (One 15-mm component did not show any cold flow.)

for the 9-mm components throughout the study period and for the 12-mm components only up to 4 years (Figure 2). At 6 years, the mean cold flow was 1.3 percent for the 9-mm components and 0.6 percent for the thicker components ($P< 0.05$). There was no cold flow of the Lund prostheses. The cold flow of the Marmor prostheses did not correlate with the clinical results, nor did it correlate with the HKA angle or other radiographic data.

Discussion

The use of RSA to study cold flow within a prosthesis is novel. RSA has previously been used to study micromotion between prostheses and bone (Ryd 1986, Mjöberg 1986). In studies of micromotion the accuracy of the kinematic analysis of RSA depends on the of the rigid-body models, i.e., the markers within the objects of interest should not move with reference to one another. However, these rigid bodies are never perfect due to measurement errors and the fact that markers sometimes move, especially in bone. To determine the degree of nonrigidity, a mathematical expression, mean error of rigid body fitting (ME), was included in the computations as a way of checking the validity. The effect of an increasing ME has been studied previously (Ryd 1986), and an ME < 200 μm can be considered "rigid." In this study "nonrigidity" of the prosthetic rigid bodies was used to study cold flow. This apparent paradox is possible because the ME is calculated by an elaborate formula (the root square sum of residual vectors after optimization of the rigid body fit), whereas the cold flow was obtained by simple linear measurements. Measurable cold flow is therefore possible also with an acceptably small ME. The markers in the Marmor prostheses were inserted from underneath, and the motion of the markers indicating cold flow was in a plane perpendicular to that of the insertion holes. Intermarker motion could not, therefore, be accounted for by motion of the markers within these holes.

Distortion and cold flow of the tibial component polyethylene is considered to be a major cause of loosening (Ducheyne et al. 1978) especially in the Marmor prosthesis (Jónson 1981, Knutsson et al. 1981, Marmor 1985). The Marmor prosthesis was originally marketed with a 6-mm component, which was often found to be distorted in clinical materials (Jònson 1981, Marmor 1985). In this study an almost linear progression of cold flow was found for the 9-mm components, but not for the 12-mm components. This finding may indicate that components thicker than 6 mm are susceptible to clinically significant cold flow in the long term. An increased load would increase the rate of cold flow. This may be a relative contraindication to the use of even this "conservative" prosthesis in young people. Bartel et al. (1986) found large polyethylene stresses in prostheses with less conforming articular surfaces. Their conclusion was that a polyethylene layer of at least 8 mm was necessary to avoid serious surface damage of the material. In terms of geometric constraint, these unicompartmental prostheses represent an extreme, the tibial component being virtually flat. Our study indicates that even thicker polyethylene may be required to avoid cold flow in nonmetal-backed components.

Cold flow has not been found in the bicondylar polyethylene components of the Total Condylar (Ryd et al. 1986) or the Freeman-Samuelson (Ryd et al. 1988) knees studied by RSA. This may be because these total prostheses are thicker and have more conforming articular surfaces that diminish polyethylene stress, as suggested by Bartel et al. (1986).

No cold flow was found in any Lund prosthesis. This may be due to the metal backing alone or because the metal tray has a peripheral rim gripping the outer margin of the polyethylene. Indeed, a metal ring around the polyethylene has been proposed precisely for this purpose (Goodfellow and O'Connor 1978). Cold flow has not been found in the Kinematic (Ryd et al. 1986) or the PCA (Ryd et al. 1990) types of total knee prostheses, which have metal backing.

The results of this study indicate that also 9–12-mm polyethylene tibial components show cold flow that seems to be preventable by metal backing. Metal backing may therefore be particularly advantageous in the prevention of mechanical loosening in such implants. The study does not address the problem of polyethylene wear or whether metal-backing has any bearing also on this phenomenon.

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