Abnormal kinematics of the artificial knee

Roentgen stereophotogrammetric analysis of 10 Miller-Galante and five New Jersey LCS knees

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The in vivo kinematics of two types of unconstrained, posterior cruciate ligament retaining knee prostheses were analyzed 1 year postoperatively using roentgen stereophotogrammetric analysis. Ten knees had the Miller-Galante and five the New Jersey LCS design. The Miller-Galante knees displayed decreased adduction during active flexion when compared with normal knees. In both types of prostheses, there was decreased medial and increased proximal and posterior displacement. The abnormal kinematics probably reflect the design of the articular surfaces, the absence of the anterior cruciate ligament, and the dysfunction of the posterior cruciate ligament.

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The degree of congruity between the articular surfaces of knee prostheses influences joint stability, polyethylene wear, and stresses at the prosthesis-bone interface. Increased congruency stabilizes the joint but may result in altered kinematics (Nilsson et al. 1990), transmittance of higher stresses to the interface, induction of micromotions, and increased risk of clinical loosening (Nilsson et al. 1991, Insall 1988, Rittman et al. 1981, Werner et al. 1978). The desiderata of some less constrained implant designs have been to imitate the anatomy and the kinematics of the normal knee, but the in vivo performances of these designs have not been completely evaluated.

We investigated two types of unconstrained knee prostheses during active motion using roentgen stereophotogrammetric analysis (RSA; Kärrholm et al. 1988, Nilsson et al. 1990, Selvik 1974, 1989).

Patients and methods

Fourteen patients with 15 knee prostheses were examined 1 year after surgery. Nine patients (10 knees) had been operated on in Umeå and constituted 9 in a series of 33 patients operated on with the original Miller-Galante prosthesis (Zimmer, Warsaw, IA, U.S.A.). Five patients (five knees) had been operated on in Gällivare with the New Jersey LCS prosthesis (DePuy, Warsaw, IN, U.S.A.; Table 1). In both series the knees had been selected because they had an optimum configuration of the femoral and tibial tantalum markers. All the patients had given informed consent for the investigation.

The Miller-Galante (MG) tibial component is inserted with a 10° posterior tilt (Landon et al. 1986, Rosenberg et al. 1989). It has comparatively planar articular surfaces with a small anterior lip (Figure 1). There is a slight central elevation with rather sharp edges. The joint area of the femoral component is also flat in the coronal plane, making the contact area between the femoral and tibial components close to a straight line. In the sagittal plane the femoral component is comparatively flat distally with increasing curvature anteriorly and posteriorly. The femoral intercondylar notch is wider posteriorly, allowing increased medial-lateral translations with increasing knee flexion.

The New Jersey LCS (LCS) tibial component is inserted with a 10° posterior tilt (Buechel and Pappas 1989) and has dovetailed tracks, guiding two mobile polyethylene bearings (Figure 1). The curvature of the polyethylene joint area corresponds to the shape of the prosthetic femoral component. The femoral condyles have posteriorly decreasing radii of curvature, and are also convex in the coronal plane.

The operation was performed according to the instructions delivered by the manufacturers. The posterior cruciate ligament (PCL) was retained in all the knees. The anterior cruciate ligament, when present, was resected in all the knees. A patellar prosthesis was only used in the MG knees. At the operation, six to nine 0.8-mm tantalum markers were inserted into the metaphyses of the distal femur and the proximal tibia,

Table 1. Clinical and radiographic data for 15 knee arthroplasties for arthrosis

Case	A	B	C	D	E	F	G	H	I	J	K	L	M
1	MG	F	73	L	92	120	175	9	0	86	82	+1.3	50
2		F	78	R	81	105	180	9	4	88	75	–0.8	50
3		F	74	R	83	105	180	9	6	88	79	–0.5	36
4		F	74	L	83	115	182	10	0	90	79	+0.6	39
5		F	72	R	86	100	174	8	0	85	81	+4.0	38
6		F	70	R	73	90	190	6	1	94	83	–2.0	23
7		F	70	R	88	120	175	8	2	87	80	+2.4	50
8		F	70	L	69	110	184	7	3	90	82	+0.8	58
9		F	75	R	87	115	186	10	0	90	85	+3.5	48
10		M	69	R	67	90	180	9	5	85	75	+1.5	36
11	LCS	F	78	R	85	125	180	11	9	87	80		39
12		M	68	L	88	115	180	10	9	86	81		47
13		F	66	R	84	110	180	4	11	91	77		28
14		M	67	L	96	115	174	9	4	83	82		45
15		F	74	R	88	120	184	10	11	88	86		46

A Type of prosthesis
MG Miller-Galante
LCS New Jersey LCS
B Sex
C Age at operation
D Side
E HSS knee score
F Knee flexion (all the knees had full extension)
G Hip-Knee-Ankle angle
(< 180° varus, > 180° valgus alignment)
H Alignment of the femoral component in the frontal plane in relation to the longitudinal axis of femur
(< 7° varus, > 7° valgus alignment)
I Alignment of the femoral component in the sagittal plane in relation to the longitudinal axis of the femur
(< 0° anteriorly, > 0° posteriorly tilted)
J Alignment of the tibial component in the frontal plane
(< 90° varus, > 90° valgus alignment)
K Alignment of the tibial component in the sagittal plane
(< 90° posteriorly, > 90° anteriorly tilted)
L Change in position of the joint line postoperatively
(< 0 mm below, > 0 mm above preoperative joint line)
M Anterior/posterior position of the tibia in relation to the femur, ratio > 36 implies a more anterior, and < 36 a more posterior location of the tibia compared with the mean value of the normal knees (see Figure 2)

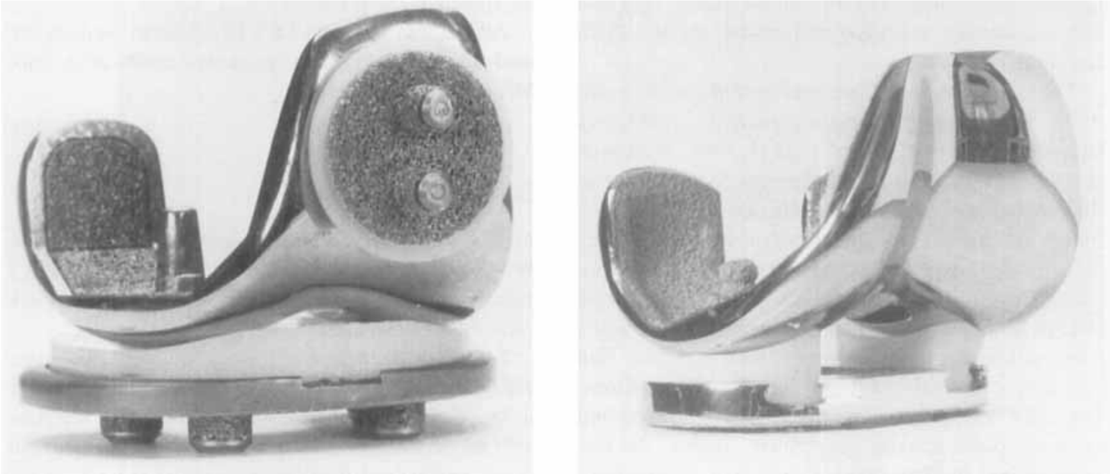

Figure 1. The Miller-Galante prosthesis.

The New Jersey LCS prosthesis.

respectively. At the time for this investigation, all the patients had had a successful outcome and no pain in the operated on knee (Table 1). The Hip-Knee-Ankle angle (HKA angle; Hagstedt et al. 1980) and the alignment of the separate prosthetic components relative to the tibia and femur were recorded (Nilsson et al. 1990).

The change in position of the joint line could only be adequately measured in the MG knees (Figgie et al. 1986) due to inferior quality of the preoperative radiographs of the LCS knees. All the knees were stable at full extension, and did not display any obvious posterior subluxation at 90° of flexion.

The anterior-posterior (AP) position of the tibia in

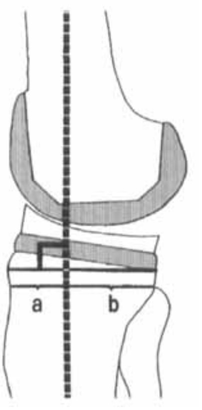

Figure 2. The anterior-posterior position of the tibia in relation to the femur was measured at the reference examination. The reconstructed center line of the femoral diaphysis divides the tibia into two parts. The ratio a/a+b × 100 was determined.

relation to the femur was measured on the lateral radiograph of the reference examination (Figure 2). The magnification factor was determined by dividing the measured AP length of the tibial components by their actual length. Because the radiographic setup was identical for all the knees examined, it was assumed that this magnification was also valid for the normal knees.

The *healthy knees* of 23 persons (20 men, 3 women; median age 24 years, range 18–40 years) with old ruptures of the anterior cruciate ligament of the opposite knee constituted the control group. They had been examined previously (Jonsson et al. 1989, Kärrholm et al. 1988).

The technique of the stereoradiography used in this study has been presented thoroughly previously (Jonsson et al. 1989, Kärrholm et al. 1988, Nilsson et al. 1990). The examinations were conducted with two film exchangers placed perpendicular to each other. Before the patient was examined, a calibration cage defining the laboratory coordinate system was radiographed with two reference plates firmly fixed in front of each film exchanger. The cage was removed, and a fixed position of the roentgen foci and the film exchangers was assured throughout the examinations. The knee was first radiographed in the relaxed and extended position with the patient supine. At this examination, the knee was placed in a standardized position in relation to the laboratory coordinate system (Jonsson et al. 1989, Kärrholm et al. 1988). All the subsequent knee positions were related to this relaxed and extended position.

Active flexion was studied in the prone position, and extension in the sitting position. All the prosthetic knees were examined with a load of 15 N attached to the ankle. The patients performed a number of trial flexions and extensions before the serial exposures were made. Flexion was studied from a position of maximum extension. The radiographic setup did not permit studies of knee flexion exceeding 55°. When examining knee extension, the knee was flexed as much as the radiographic setup allowed (mean value 35°). The tibia was pushed backwards by the examiner, and a pair of radiographs were exposed. Thereafter, active extension and the serial exposures were started. An average of 10 pairs of radiographs were obtained for each series. At the evaluation, tibial movements (flexion/extension, internal/external rotation, adduction/abduction, and medial/lateral, proximal/ distal and anterior/posterior translations) were calculated in relation to the relaxed and extended position. A central point on the tibial plateau was plotted on the radiographs of the reference position. Its three-dimensional position was calculated and transformed to all subsequent positions of the tibial prosthesis by using the rigid body of the proximal tibia defined by its tantalum markers (Kärrholm 1989). At this calculation, it was presumed that no or negligible movements occurred between the center of the tibial component and the bone. Displacements of this point represented the tibial translations. Because the commencement of the knee movements and the start of the film exchangers could not be exactly synchronized, data were not collected at the same degree of flexion/extension. To standardize the measurements, the recorded rotations and translations for every patient were plotted and interpolated at intervals of 5°.

All the MG (10) and LCS (five) knees were examined during flexion. Extension was analyzed in nine MG and four LCS knees.

Statistics

Analysis of variance with repeated measurements, Wilcoxon's rank sum and signed rank tests, Student's paired t-test, and Spearman's nonparametric rank correlations were used.

To reduce the effect of missing observations at the beginning and the end of the flexion and extension series, the statistical analysis included the intervals with all the observations available in LCS and normal knees, and at least nine observations in the MG knees corresponding to 5°–50° of flexion. For the extension series, the MG knees were analyzed between 30° and 20° of extension, and the LCS knees between 20° and 15°.

The reproducibility of the method was determined earlier (Nilsson et al. 1990). The standard deviations of rotations were 1.5° and 0.3° (internal/external rotation, adduction/abduction), and the translations 0.2 mm (medial/lateral) and 0.5 mm (proximal/distal, anterior/posterior) when measured at 10° intervals of knee flexion.

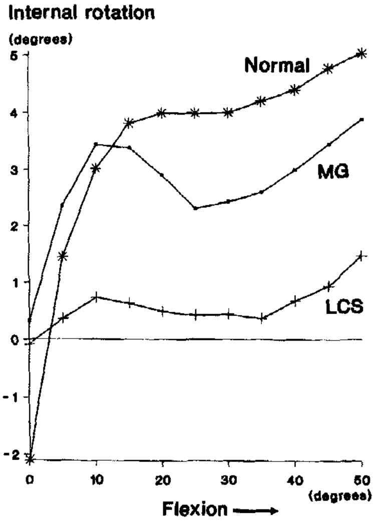

Figure 3. Tibial rotation about the longitudinal axis (internal/external rotation) occurring during knee flexion. Abbreviations Figures 3–7: MG = Miller-Galante knees, LCS = New Jersey LCS knees, Normal = normal knees.

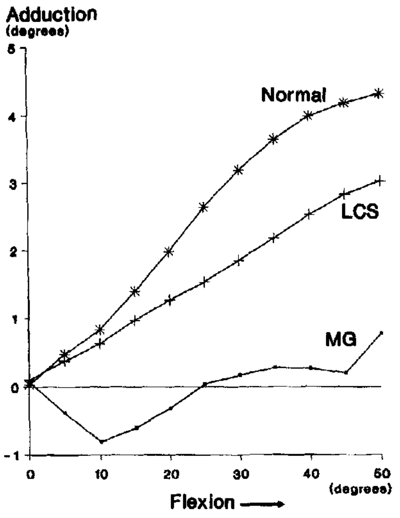

Figure 4. Tibial rotation about the sagittal axis (adduction/abduction) during flexion. For abbreviations, see Figure 3.

Results

Knee flexion

Internal/external rotation. The MG knees started from a neutral position and reached 3.5° of internal rotation at 15° of knee flexion (Figure 3, Table 2). Thereafter, a slight increase to 4° at 50° of flexion was recorded. The pattern was similar to that of the normal knees (NS). The LCS knees displayed minimum internal rotation during knee flexion (mean 0.5°, NS).

Adduction/abduction. The MG knees displayed initial abduction (mean 0.5° at 10° of flexion) and thereafter minimum adduction, which was smaller than in the normal knees (MG vs normal knees $P < 0.05$; Figure 4). The LCS knees showed a pattern closer to the normal knees, with increasing tibial adduction, reaching a mean of 3° at 50° (NS).

Medial/lateral translation. In both prosthetic designs, there was an increasing medial translation of the center of the tibial plateau, reaching 1–2 mm at 50° of flexion (Figure 5). However, this medial translation was not as pronounced as in the normal knees (MG vs normal $P < 0.01$; LCS vs normal $P < 0.05$).

Proximal/distal translation. During the first 20° of flexion, the center of the MG knees displayed distal displacement of slightly more than 1.2 mm, and thereafter proximal translation, reaching a mean of 6 mm at 50° of flexion (Figure 6). The LCS knees displayed an increasing proximal translation during flexion, reaching 12.5 mm at 50° of flexion (MG and LCS vs normal; $P < 0.01$).

Table 2. Pooled standard deviations for each type of movement

Movement	MG	LCS	Normal
Flexion			
Internal/external rotation	3.5°	3.6°	6.5°
Adduction/abduction	2.2°	2.7°	2.3°
Medial/lateral translation, mm	1.8	1.8	0.9
Proximal/distal translation, mm	1.8	1.2	2.3
Anterior/post. translation, mm	2.3	2.6	2.2
Extension			
Internal/external rotation	4.2°	3.0°	4.4°
Adduction/abduction	1.9°	2.7°	1.5°
Medial/lateral translation, mm	1.6	1.4	0.6
Proximal/distal translation, mm	1.3	0.9	1.0
Anterior/post. translation, mm	2.1	3.2	1.1

Posterior translation. Both the MG and LCS knees displayed an increasing posterior translation with proceeding flexion, reaching an average of 30 mm at 50° of flexion (Figure 7). The prosthetic knees had displaced about twice as far posteriorly as the normal ones (MG vs normal; LCS vs normal; $P < 0.0001$).

Knee extension

Paired data for the flexion and extension series for the prosthetic knees were available between 20° and 30°

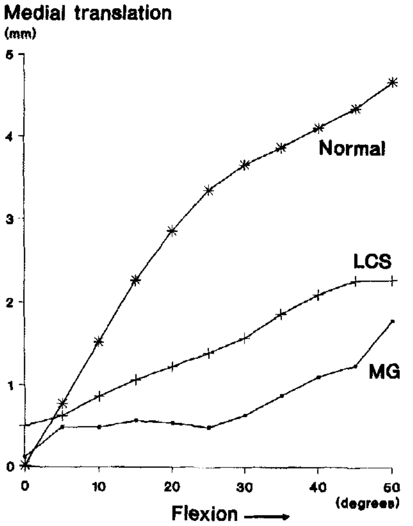

Figure 5. Translation of the center of the tibial plateau in the prosthetic knees, and the tibial intercondylar eminence in the normal knees along the transverse axis (medially/laterally) during flexion. For abbreviations, see Figure 3.

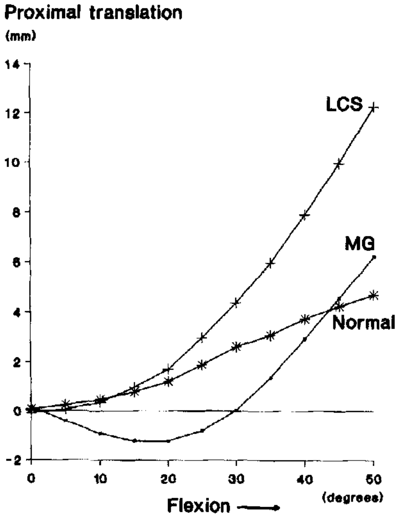

Figure 6. Tibial translations along the longitudinal axis (proximally/distally) during flexion. For abbreviations, see Figure 3.

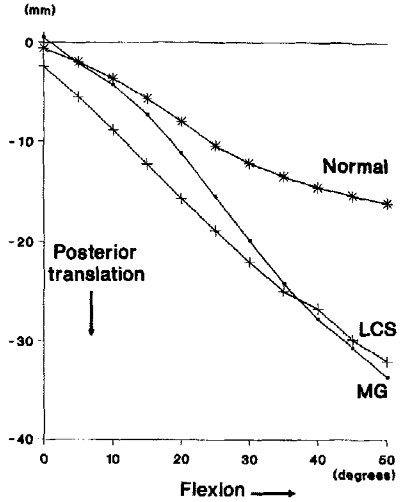

Figure 7. Tibial translations along the sagittal axis (anteriorly/posteriorly) during flexion. For abbreviations, see Figure 3.

of flexion in the MG knees, and between 15° and 20° in the LCS knees. The coupled tibial rotations did not differ between flexion and extension in either of the two designs. During extension from 30° to 20°, the MG knees were more medially translated than during flexion ($P < 0.01$). Proximal/distal and anterior/posterior translations did not differ. The LCS knees displayed somewhat less proximal ($P < 0.05$) and posterior ($P < 0.01$) translation during extension between 20° and 15° than at the corresponding flexion.

Prosthetic positioning

There were no differences in alignment of the tibial component between the MG and LCS knees. The LCS femoral components were, however, significantly ($P < 0.01$) more posteriorly tilted than in the MG knees.

The prosthetic tibiae were more anteriorly positioned in relation to the femur than were the normal tibiae (Figure 2, Table 1). The ratio for the normal knees was 36 ± 8.0 (mean ± SD) and for the prosthetic knees 43 ± 9.1 ($P < 0.05$). This increased anterior location corresponded to approximately 3 mm after correction for the magnification of the radiographs.

In the MG knees, there was a positive correlation between the HKA angle and internal tibial rotation (10° to 40° of flexion; $r = 0.76$, $P < 0.05$), indicating that valgus alignment resulted in increased and varus alignment in decreased internal rotation during flexion. Alignment of the tibial component in the frontal plane also seemed to be of importance (J in Table 1; $r = 0.75$, $P < 0.05$). In the LCS knees, there were no correlations between prosthetic positioning and the recorded movements.

The change in position of the joint line did not influence the recorded movements in the MG knees.

Discussion

The control group in this study was younger than the prosthetic groups, and included more men, which might have influenced the results. On the other hand, it was not appropriate to use the contralateral knees in the prosthetic groups, because they all displayed various degrees of degenerative disease.

In normal knees, there is an initial internal tibial rotation during flexion (Essinger et al. 1989, Kärrholm et al. 1988), and a constantly shifting center of rotation (Kärrholm 1989). Most MG knees also rotated internally during early flexion. The central elevation on the tibial plateau and the posteriorly increasing width of the intercondylar notch might be of importance. The use of movable bearings in the LCS knees did not seem to be superior in the restitution of this "screw-home" mechanism, maybe because the meniscal tracking limits the internal/external rotation to take place about a central axis.

Increasing HKA angle implied increased internal rotation of the MG knees. The tibiofemoral compression force increases laterally with increasing valgus alignment, probably displacing the center of internal/external rotation towards the lateral compartment. The result might be a more externally rotated position in extension and less resistance to internal tibial rotation during flexion.

The LCS knees displayed adduction during flexion, as was found in the normal knees. The polyethylene menisci move on a flat and posteriorly sloping surface. Because there was only minimum simultaneous internal/external rotation during flexion, it is less probable that relative posterior and distal displacements of the medial meniscus could explain the adduction. Perhaps these movements were an effect of the femoral condyles displacing out of the meniscal groove. The MG knees showed almost no rotations about the sagittal axis, maybe an effect of the flatness of the articular surfaces in the coronal plane. The muscular forces trying to adduct the knee during flexion are probably further resisted by the central elevation preventing normal medial translations. Instead, the stresses on the tibial plateau increase, resulting in varus/valgus movements of the tibial tray, which seem to be the most pronounced type of micromotion of the MG design when used without cement (Nilsson and Kärrholm 1990).

With a compressive load of the knee, the geometry of the articular surfaces becomes the major determinant for the kinematics (Hsieh and Walker 1976, Markolf et al. 1981, Shoemaker et al. 1982). The tibiofemoral compression forces in the present study were not known, but have been calculated at about twice the body weight in the supine position (Nisell et al. 1986), and five times the body weight during walking (Seireg and Arvikar 1975). Studies of the comparatively constrained Tricon-M knee have indicated that the knee kinematics are also governed by the anatomy of the joint area during active flexion and extension (Nilsson et al. 1990). In unconstrained prostheses the presence or function of the cruciate ligaments could be expected to be more decisive.

Despite that all of the prosthetic knees had their posterior cruciate ligaments retained, posterior translation of the center of the tibia was significantly larger in the prosthetic knees during flexion, and was of the same magnitude as in PCL resected prosthetic knees (Nilsson et al. 1990). Absence of the anterior cruciate ligament and a more anterior tibial position of the prosthetic tibiae in the reference position may only partly explain this finding, because the mean increased anterior position of the prosthetic tibiae when compared with the normal knees was only about 3 mm. The lack of function of the posterior cruciate may be another important cause.

Radiographic measurements of patellar thickness and joint anatomy after prosthetic replacement in the MG group were difficult to make, but theoretically the presence of a thick patella-prosthesis unit could have changed the tibiofemoral relationship due to an altered lever arm. However, both types of prostheses showed about the same amount of posterior displacement. The only significant difference was in adduction-abduction, presumed to mainly be an effect of the configuration of the tibial joint area.

The posterior cruciate ligament is thought to contribute to the femoral "roll-back" during flexion (Andriacchi and Galante 1988, Goodfellow and O'Connor 1978). Visual analysis of the lateral radiographs of the flexion series in the prosthetic knees suggested that the contact point between the femoral condyles and the tibial plateau did not move backwards during flexion, as was observed in the normal knees, but was located centrally or even slightly anteriorly (Soudry et al. 1986), suggesting that no "roll-back" occurred.

An increased posterior slope of the tibial component has been shown to increase the AP laxity in flexion by shortening the distance between the femoral and tibial attachments of the PCL (Whiteside and Amador 1988). Exact tensioning of the PCL during operation is often difficult to achieve (Insall 1988), and elongation of the PCL during the first postoperative year may also occur (Hagena et al. 1989).

The pattern of proximal tibial translation in the prosthetic and normal knees is probably attributable to the different designs of the femoral components in the sagittal plane, the MG prosthesis being more "square" and the LCS having a posteriorly decreasing radii of curvature. Further, the LCS femoral components were significantly more posteriorly tilted, probably increasing the proximal tibial displacement during early flexion.

In conclusion, both of these minimally constrained prostheses displayed abnormal in vivo kinematics, probably reflecting the design of the articular surfaces, the absence of the anterior cruciate ligament, and dysfunction of the posterior cruciate ligament.

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