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Micromotion in knee arthroplasty

A roentgen stereophotogrammetric analysis of tibial component fixation.

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Definitions

Micromotion: Small movements between the tibial component (called "prosthesis") and bone. Divided into two types – migration and inducible displacement.

Migration: Gradual micromotion over time; the prosthesis is detached from its original position to a new resting position.

Inducible displacement: Instant micromotion occurring in response to external forces; the prosthesis is detachable from its resting position.

Rigid body: A body in which distances remain constant in time.

Rigid-body model (RBM): A configuration of three or more non-collinear markers representing the bodies of interest (prostheses or bone segments), and used for calculation of movements.

Mean error of rigid-body fitting (ME): The degree of conformity of an object to the rigid-body concept. The magnitude of ME is a function of errors of measurements and true movements of the markers within the object.

Unicompartmental prosthesis: A prosthesis that replaces one femoro-tibial compartment in the knee. In this study only one compartment was replaced in each knee, either the medial or the lateral one.

Total prosthesis: A prosthesis that replaces both femoro-tibial compartments.

Coordinate system: The micromotion is expressed in a laboratory coordinate system defined by the calibration cage approximately parallelling the major planes of the body. The X-axis is transverse, the Y-axis is vertical and the Z-axis is sagittal (Fig.1). For rotations, movements are expressed with a positive (+) sign if the anterior part of the prosthesis moves downwards (x-axis), peripherally or laterally for unicompartmental and total prostheses, respectively (y-axis) and if the central or medial side for unicompartmental and total prostheses, respectively, moves downwards (z-axis).
Translations are expressed with a positive sign if peripheral or lateral (x-axis), upwards (y-axis) or forwards (z-axis).

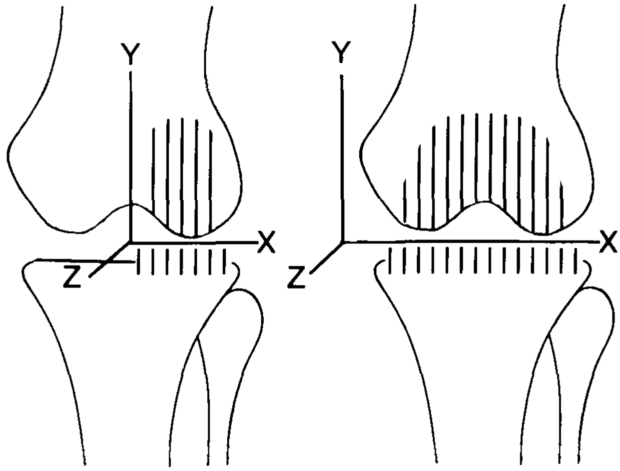


Figure 1. The coordinate system for unicompartmental and total knee prostheses, respectively.

Hip-Knee-Ankle (HKA) angle: The medial angle between the lines from the centre of the hip and the ankle,respectively, meeting at the center of the knee. $<180°$ = varus alignment, $>180°$ = valgus alignment.

Prosthesis-Tibia (PT) angle: The medial angle between the prosthesis and the long axis of the tibia in the frontal projection.

Prosthesis-Tibia-Side (PTS) angle: The posterior angle between the prosthesis and the long axis of the tibia in the lateral projection.

Prosthesis-Hip-Ankle (PHA) angle: The medial angle between the prosthesis and the mechanical axis of the limb, i.e. the line connecting the hip and the ankle, in the frontal projection.

Abbreviations

FEM	Finite element model
HKA	Hip-knee-ankle angle
HSS	Hospital for Special Surgery (score)
ME	Mean error (of rigid-body fitting)
MTPM	Maximum total point motion
PCA	Porous Coated Anatomic knee prosthesis
PHA	Prosthesis-hip-ankle angle
PM	Point motion
PT	Prosthesis-tibia angle
PTS	Prosthesis-tibia-side angle
RBM	Rigid-body model
RSA	Roentgen stereophotogrammetric analysis
SM	Segment motion
TLD	Thermoluminescent dosimeters

Summary

The modern era of endoprosthetic joint replacement started with the introduction of acrylic cement to improve component fixation. Long-term results have, however, indicated that prosthetic fixation remains critical; loosening at the bone-cement interface has become an important problem. Research in recent years has focused on attempts to achieve better fixation by improving cementing techniques, improving prosthetic design by, for example, adding metal support of polyethylene components and by exploring alternative ways to bond prosthetic components to bone without cement.

The mechanical integrity of the bone-cement interface has been studied under laboratory conditions. Because of the *in-vivo* reaction of bone, with the interposition of a fibrous tissue layer at the interface, such studies are not totally valid. Studies on autopsy material, more closely resembling the *in-vivo* situation, are few and there has been only one previous study like the present one. In this study, roentgen stereophotogrammetric analysis (RSA) was evaluated and found to have an accuracy ten times better than conventional radiography. This accuracy was judged adequate for studies of micromotion.

In this work, two types of micromotion of the tibial component were studied; migration, i.e. gradual motion over time, and inducible displacement, i.e. instant motion in response to external forces. Ninety-six knee arthroplasties for gonarthrosis, representing four different types of fixation were studied by roentgen stereophotogrammetric analysis (RSA). Eighty-nine arthroplasties were clinically successful. The follow-up ranged from two to five years. Full post-operative weight-bearing was allowed for all patients, except those operated with a Freeman-Samuelson prosthesis, who were adviced to use crutches for six weeks and partial weight-bearing for another six weeks.

Fifty-one conventionally cemented all-polyethylene prostheses, 27 total and 24 unicompartmental, represented a baseline series. Migration was found for all prostheses, with a mean maximum deflection of 1.2 and 0.9 mm, respectively, after four years. In both groups, the major part of the migration occurred during the first year, after which the majority of the components did not migrate further. Some prostheses, with larger migration during the first year, continued to migrate throughout the investigation. None of the total, but the majority of the unicompartmental prostheses showed signs of cold flow within the polyethylene. All prostheses showed reversible inducible displacement, the maximum deflection ranging from 0.2 to 1.0 mm.

Twenty-six metal-supported prostheses, seven Kinematic, seven PCA (both total) and 12 Prototype (unicompartmental), bonded to bone by cement showed the same amount of both migration and inducible displacement as the all-polyethylene prostheses. No cold flow was registered for the metal-supported prostheses. Since this phenomenon was found for a number of all-polyethylene unicompartmental prostheses, such prostheses may be particularly suitable for metal support under the polyethylene.

Nineteen non-cemented total prostheses, 13 PCA and six Freeman-Samuelson, showed larger migration and inducible displacement than the cemented prostheses. For a number of the PCA and all Freeman-Samuelson prostheses the inducible displacement was not reversible, i.e. was of a plastic nature, which suggests a different type of bond between prosthesis and bone for non-cemented than for cemented prostheses.

None of the clinical or radiographic variables such as clinical score, age, weight, alignment, component position or cortical bone support showed any correlation with the RSA results.

All components developed radiolucent zones in the interface between bone and cement/prosthesis, some of which were judged to be of tensile origin. By a finite element analysis (FEM) and by analyzing marker movements in the upper tibia, it was concluded that both migration and inducible displacement occur within the soft tissue of the bone-cement/prosthesis interface. A progressively *extending* zone was the best indicator of increasing micromotion.

As most of the migration occurred within the first post-operative year, after which the inducible displacement also did not increase, it was concluded that per-operative trauma causing bone death and resorption is the major cause of micromotion. Since the micromotion was larger for non-cemented components, most of the per-operative trauma in knee arthroplasty may be caused by mechanical factors during bone preparation and not by the heat from curing bone cement.

For a minority of the components, continuous migration was found. Such continuous migration must involve bone resoption and it was suggested that this occurred in response to forces in a way resembling orthodontic tooth movements.

All of the prostheses in this work were "technically loose", although the majority of the patients were perfectly satisfied with the procedure and loosening was suspected on clinical grounds in only one case. None of the more modern concepts of prosthetic design or fixation has thus proved superior to conventional techniques, and it was concluded that "functionally" all but one of these prostheses were stable and represented a type of fixation where good results might be expected in long term follow-ups.

Introduction

The introduction of polymethylmethacrylate (bone cement) for bonding endoprosthetic components to bone by Charnley (1960, 1964) triggered the success of total joint replacements. Before that time, knee arthroplasties with interpositioned loose devices such as the MacIntosh (1958) or the McKeever (1960) tibial plateau or hinges (Walldius 1957) had met with limited success. Using Charnley's metal-on-polyethylene "low-friction" concept, Frank Gunston (1971) developed the first "modern" knee prosthesis bonded to bone by cement.

Initial reports on cemented joint replacements in the hip and knee were optimistic (Charnley 1964, Skolnick et al. 1976). After solution of the early problems with infection (Charnley 1972, Carlsson et al. 1978) and, to some extent, prosthetic component wear (Charnley 1979, Clarke & McKellop 1980) non-infectious loosening evolved as the single most frequent cause of failure in long-term follow-ups of both hip (Charnley & Cupic 1973, Beckenbaugh & Ilstrup 1978, Amstutz et al. 1982) and knee replacements (Ahlberg & Lundén 1981, Bryan & Rand 1982, Cameron & Hunter 1982, Insall & Dethmers 1982, Thornhill et al. 1982, Haberman 1984, Bauer et al. 1985). In knee arthroplasty incidences of mechanical loosening of 10–12 percent have been reported (Lacey 1978, Freeman et al. 1978).

Bone cement is not a glue but acts by three-dimensional interlocking with the prosthetic irregularities and bony trabeculae (Charnley 1964). Failure of this mechanical interlock, giving micromotion at the interfaces, is the key event in mechanical loosening (Amstutz et al. 1972, Ducheyene et al. 1978, Lewis et al. 1985). For more stable implant fixation, improved cementing techniques with lavage of the resected bone surfaces (Halawa et al. 1978) and low-viscosity cement (Miller et al. 1979), metal support of the polyethylene tibial components (Walker et al. 1981, Bartel et al. 1982, Reilly et al. 1982) and fixation without cement by bony ingrowth (Galante et al. 1971, Welsh et al. 1971, Hungerford et al. 1982), immediate interlocking (Freeman et al. 1981) or osseointegration (Albrektsson et al. 1981) are being tried.

In 1974, Selvik introduced a system of roentgen stereophotogrammetric analysis (RSA) allowing very accurate measurements in all planes. The system has been extensively used in orthopedic and radiographic research conducted in Lund (Olsson et al. 1976, Lindstrand & Selvik 1976, Baldursson et al. 1979, Bylander et al. 1981, Tjörnstrand et al. 1981, Kärrholm et al. 1982, Mogensen et al. 1982, Ryd et al. 1983, Selvik 1983, Mjöberg et al. 1984, Selvik et al. 1985).

The tibial component is most susceptible to loosening, and might therefore serve as a model in studies of different fixation methods. In 1976 the first knee replacement was prepared for RSA and since 1981 every knee arthroplasty performed for gonarthrosis in our departement has been entered into the project. The prostheses studied are representative of the knee arthroplasties most commonly done in the world today.

This work analyzes:

1. The potential of RSA for studying micromotion in knee arthroplasty.
2. Micromotion of polythethylene tibial components fixed to bone by cement.
3. The effect of metal support of the polyethylene.
4. Micromotion of two different types of non-cemented tibial components.
5. The correlation of the above findings with clinical and radiographic variables.

Patients

Only patients with femoro-tibial gonarthrosis were included in this work. Ahlbäck (1968) Stages I–II (only cartilage loss) were treated with unicompartmental arthroplasty while Stages IV–V (bone attrition >5mm) were treated with total knee arthroplasty. Stage III (bone attrition <5 mm) were treated in either way, depending on the condition of the lateral compartment (Bauer 1982).

The patients with a Freeman-Samuelson prosthesis were operated on at the Orthopedic Department, Östra Sjukhuset, Gothenburg, where a post-operative regimen including mobilization on crutches for six week followed by partial weight-bearing for another six weeks was used. All remaining operations were performed at the Orthopedic Department, Lund, where the post-operative regimen included early mobilization with physiotherapy and immediate full weight-bearing.

Clinical assessment

The clinical results after arthroplasty were assessed by the Lund system (Bauer et al. 1969, Jónson 1981) and by the Hospital for Special Surgery (HSS) score (Ranawat et al. 1976). In the Lund system the patients are analysed in a Venn diagram according to stability, mobility and pain-free walking distance into three groups; satisfactory, acceptable and failure (Fig. 2).

The HSS-score gives a maximum of 30 points for pain, 22 for function, 18 for mobility and 10 each for muscle strength,flexion deformity and instability. Score $\geqslant$ 85 = excellent, 84–70 = good, 69–60 = fair and <60 = poor.

In the total material of 96 cases, there were seven cases assessed as failures and eight as acceptable according to the Lund system, while only four were not excellent or good according to the HSS-score; the HSS score being more permissive than the Lund system (Jónson 1981). No patient in this material has been reoperated thus far.

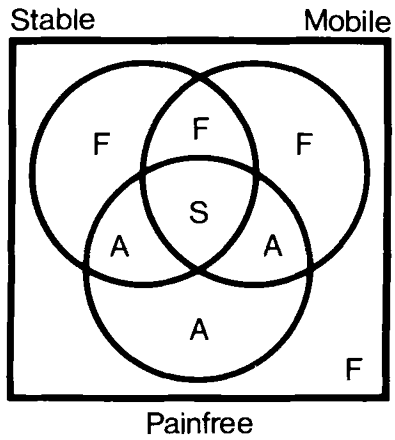


Figure 2. The Venn diagram used in the Lund assessment system. Stable = a varus-valgus instability ≤5°, Mobile = a flexion arc not less than 5-90° and Painfree = a painfree walking distance of >500m. S=satisfactory, A=acceptable and F=failure.

All-polyethylene series

Fifty-one knees had a tibial component made exclusively of polyethylene.

Twenty-seven knees had a **Total Condylar** (Fig. 3) total knee arthroplasty done between January 1980 and September 1981. An additional four knees were prepared for RSA but were excluded; two because of inadequate marker positions, one because of senility and one beacuse the patient moved away from the region. The material was not consecutive, since an additional 25 patients were operated with total knee arthroplasty during this period. There was, however, no predetermined selection of patients for RSA. The operations were performed according to the intentions of the originators (Insall et al. 1976). High viscosity bone cement including barium sulphate was used without lavage or pressurization techniques. The Total Condylar knee is a semi-constrained, non-linked prosthesis. Antero-posterior as well as translatory forces are carried by the prosthesis since the tibial component condyles are deeply cup-shaped, allowing very little translation or rotation under load. The cruciate ligaments are resected.

There were 22 women and five men with a mean age of 71 (56–84) years and with a follow-up ranging from 20 to 51 months. Two cases were classified as failures according to the Lund system at the last follow-up; one (Case 3) because of a pain associated with a medio-lateral instability and the other (Case 21) because of unexplained pain. The remaining 25 cases were all satisfactory. All cases were subjected to stress examinations one to three years after the operation; in eight cases a secondary stress series was performed about a year after the first one.

Twenty-four knees had a unicompartmental **Modular** knee prosthesis (Fig.4) (Marmor 1973) inserted between November 1978 and November 1982. All Modular knees prepared for RSA were included. An additional 21 Modular knee arthroplasties

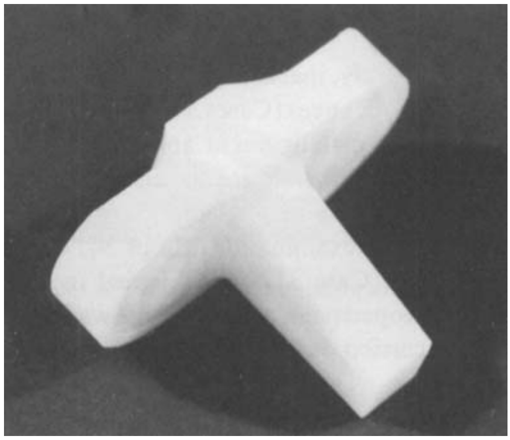


Figure 3. The tibial component of the Total Condylar total knee prosthesis.

were performed during this period but, as with the total arthroplasties, the patients were selected for RSA at random. All prostheses were inserted with conventional high viscosity bone cement including barium sulphate, which penetrated the cancellous bone down to the cortical bone through specially made holes. A guide instrument (Lindstrand et al. 1982) was used to obtain optimum positions of the prosthetic components after an L-shaped complete resection of the subchondral bone of the tibial condyle. The Modular knee is an unconstrained surface replacement prosthesis relying on intact collateral and cruciate ligaments for stability. The femoral component conforms to the natural shape of the femoral condyle, to which it is cemented after a small bone resection. The tibial component is a polyethylene block with a flat upper surface.

There were 21 women and two men with a mean age of 69 (62–75) years. One female patient had bilateral procedures. Three arthroplasties were lateral and 21 medial. The follow-up was between 34 and 66 months.

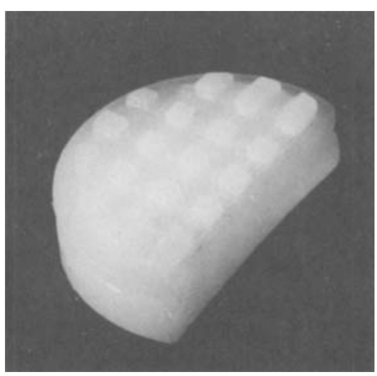


Figure 4. The tibial component of the Modular (Marmor) unicompartmental knee prosthesis seen from underneath.

The clinical results were satisfactory in 20 while four arthroplasties failed according to the Lund system. Case 32 showed large migration and had had pain intermittently during the last two years. Currently, mechanical loosening is suspected and the patient is scheduled for reoperation. Both knees (Cases 38 and 39) in the a woman with bilateral arthroplasties failed because of malalignment and medio-lateral instability giving a walking distance of less than 100 m. Case 41 had pain because of a suspected impingement between the femoral component and the tibial eminence. Seven Modular knees were subjected to a stress examination. Cases 34, 43, 45, 46, 47 and 49 were included in a pilot study while Case 32 was subjected to a stress series on clinical grounds; the patient had developed pain, and loosening was suspected. However, when the stress examination was carried out she was again symptom free.

Metal-supported series

Twenty-six knees had metal-supported tibial components.

Six patients were operated on during the Spring 1982 with seven metal-supported total knee prostheses all of which were prepared for RSA. The prosthesis used was the **Kinematic** knee prosthesis (Fig.5) inserted with the original guiding instruments (Ewald et al. 1984) using high viscosity barium sulphate cement including gentamycin (Palacos). The tibial component of the Kinematic knee, like that of the Total Condylar knee, includes a tibial eminence resisting medio-lateral translation, but the prosthesis is considerably less constrained in rotation. Only the anterior cruciate ligament is resected.

There were five women and one man with a mean age of 76 (69–80) years and with a follow-up ranging from 25 to 30 months. One woman had bilateral procedures. The clinical results after two years were satisfactory in all cases according to the Lund system. Case 58 sustained a traumatic supra-condylar fracture of the femur just after

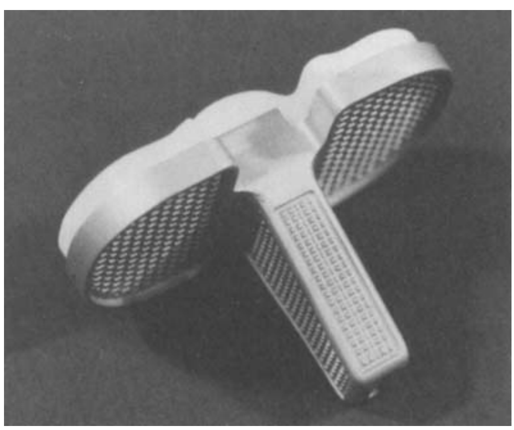


Figure 5. The tibial component of the Kinematic total knee prosthesis.

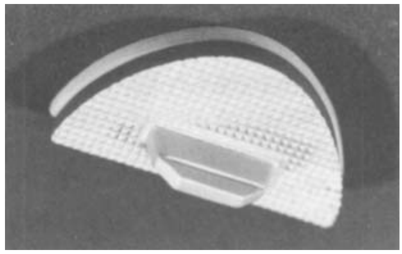


Figure 6. The tibial component of the Prototype unicompartmental knee prosthesis seen from underneath.

the 1-year follow-up. After traction, percutaneous pinning and cast bracing, the fracture healed with a considerable varus alignment. At the 2-year follow-up the patient was ambulatory using a cane but had slight pain on weight-bearing, the score, however, being satisfactory. All cases were subjected to a stress examination after two years.

In 12 knees a unicompartmental **Prototype** arthroplasty with a metal-supported tibial component was performed from January 1981 to November 1982. Three additional arthroplasties of this type were excluded because of improper marking in one case and intercurrent disease in two cases. The prosthesis was produced in the development of what finally became the PCA unicompartmental knee prosthesis. It closely resembles the Modular knee, the only major difference being a vitallium support including a keel of the tibial component polyethylene (Fig. 6). The prostheses were inserted with barium-containing cement in the same way as the Modular knees.

There were 10 women and two men with a mean age of 71 (65–78) years. Ten arthroplasties were medial and two lateral, and the clinical result was satisfactory in all cases. These cases were not included in the stress investigation.

Seven cases had a **Porous Coated Anatomic (PCA)** knee prosthesis inserted with cement (Palacos) using the Universal knee instrument during the Autumn 1982 and Spring 1983. The PCA knee has a porous surface which bonds well with cement (Fig.7). The prosthesis closely resembles the natural knee in motion pattern, being virtually

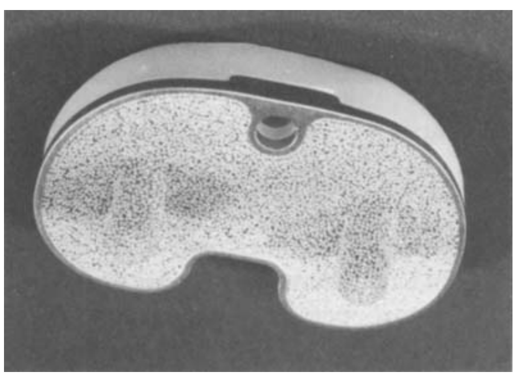


Figure 7. The tibial component of the Porous Coated Anatomic (PCA) total knee prosthesis.

unconstrained in rotation but with a central eminence which prevents translation. Only the anterior cruciate ligament is sacrificed.

The prostheses were inserted with Palacos high viscosity cement in six patients, five women and one man, who had bilateral procedures. The mean age was 71 (62–82) years. Cement was used when cementless insertion (see below) was unstable or judged to be contra-indicated because of poor bone quality or because the arthroplasty was a revision (Cases 62,63 and 64). All cases were successful except one (Case 63), who had unexplained pain and was classified as a failure.

Immediate interlocking series

During the Autumn 1983, seven knees were operated for arthrosis on six patients using the **Freeman-Samuelson** prosthesis. All knees were prepared for RSA. One of the patients was excluded due to an intercurrent mental disorder. The Freeman-Samuelson prosthesis has an all-polyethylene tibial component which is fixed to the bone without cement (Fig.8). Fixation is achieved by two flanged polyethylene pegs, which are inserted into drilled holes in such a way that immediate interlocking of the flanges with the cancellous bone is achieved (Freeman 1981, Blaha et al. 1982). The geometry is basically of the roller-in-a-trough type, being constrained in rotation under load. Both cruciate ligaments are resected.

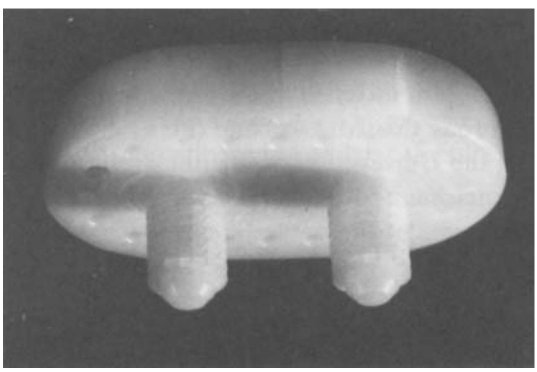


Figure 8. The tibial component of the Freeman-Samuelson total knee prosthesis.

The patients were five women with a mean age of 72 (65–72) years. Due to faulty marker positioning in the tibia, tantalum markers were inserted percutaneously in three cases a couple of months post-operatively. This left only three patients for the longitudinal study. The follow-up period was two years, at which time all cases were clinically successful. All patients were subjected to a stress examination one year post-operatively.

Biological ingrowth series

In May 1982 the first non-cemented **PCA** knee prosthesis was inserted in Lund, and during the first year 13 such arthroplasties were performed. The PCA knee was the first commercially available prosthesis provided with a porous surface. These pores have a mean diameter of 425 μm which is close to the optimal size for bony ingrowth (Bobyn et al. 1980). After meticulous preparation of the bone immediate fixation is achieved by press-fit, together with a bone screw and molly bolt. Long-term fixation by ingrowth, preferably of bone, is aimed at (Hungerford et al. 1982).

Thirteen prostheses were inserted without cement in 13 patients; 12 women and one man with a mean age of 72 (63–79) years. All patients were followed for approximately two years. The clinical results were satisfactory in all cases according to the Lund system. Stress examinations were performed at one and two years.

Apart from from four patients having bilateral operations with the same prosthesis; Modular Cases 38 and 39, Kinematic Cases 54 and 55, PCA Cases 60 and 61 and Freeman-Samuelson Cases 94 and 95, an additional four patients had bilateral operations with different prostheses. Thus Cases 6 and 78 were performed on the same patient, as were Cases 20 and 81, Cases 23 and 82 as well as Cases 29 and 75. Thus, altogether 88 patients participated in the project reported here.

Methods

Conventional radiography

The **position** of the prostheses was determined in a reproducible fashion. The true lateral projection in the weight-bearing position, on which the PTS-angle was measured, was defined under fluoroscopic control by superimposing the posterior aspect of the two femoral condyles. At 90° to this, a true frontal exposure was obtained (Lindstrand et al. 1982) on which the mechanical axis (Maquet 1967) , i.e. alignment (HKA-angle), and the position of the tibial component with reference to the tibia (PT- and PHA-angles) were defined (Fig. 9).

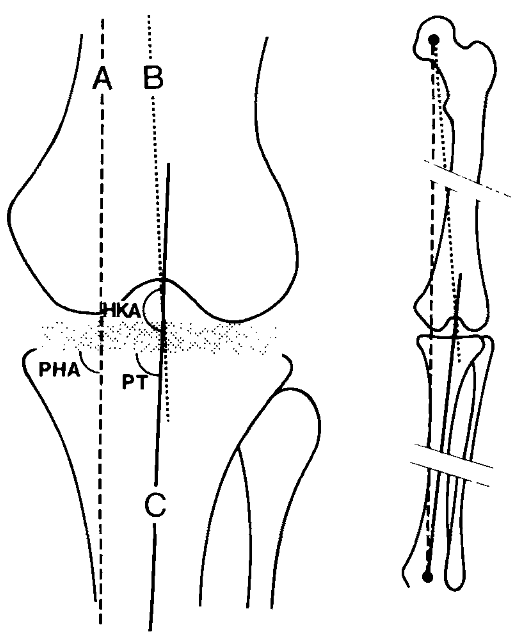


Figure 9. Determination of the HKA-, PHA- and PT-angles. A=line connecting the centre of the femoral head to the centre of the ankle joint. B=line connecting the centre of the femoral head to the centre of the knee. C=the anatomical axis of the tibia.

Cortical bone support was analyzed by measuring the amount of resected upper tibial surface left uncovered on both sides of the prosthesis or cement in both projections on conventional radiographs. The tibial tuberosity projects ahead of the anterior part of the tibial condyles on lateral radiographs, and the amount of anterior cortical bone support of the prosthesis thus cannot be determined. For unicompartmental prostheses, only posterior and peripheral support was therefore analyzed.

A **radiolucent zone** often developes between bone and cement/prostheses. In Total Condylar and Kinematic knees this zone was assessed in three ways: 1. The maximum *zone width* on either projection was determined by direct measurements using a magnifying glass graded to a tenth of a millimeter. 2. The *zone extension* was obtained by measuring the length of the zone in both projections and expressing it as a percentage of the entire bone-cement interface. 3. The *zone value* for Total Condylar and Kinematic knees was obtained by dividing the bone-cement interface into four parts in both projections (Fig.10). The mean width of the zone in the eight sub-parts was measured in steps of 0.5 mm (<0.5 mm = 1, 0.5 – <1 mm = 2, 1 – <1.5 mm = 3, etc.), the sum of which represents the zone value. A complete zone 0.5–<1 mm in width thus has a zone value of 16.

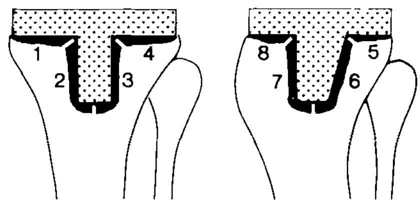


Figure 10. The bone-cement interface of Total Condylar and Kinematic prostheses subdivided into eight parts.

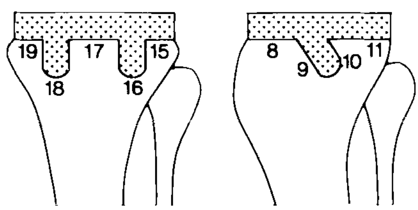


Figure 11. The nine interface regions of the tibial component of the PCA prosthesis numbered according to the originators (Hungerford & Kenna 1983).

The zone in unicompartmental components was assessed by the maximum width, and the extension as a percentage of the entire interface.
For PCA knees the size of nine zone regions (Hungerford & Kenna 1983) (Fig.11) was assessed in steps of 0.3 mm. (<0.3 mm=1, 0.3 – <0.6 mm=2, etc.).

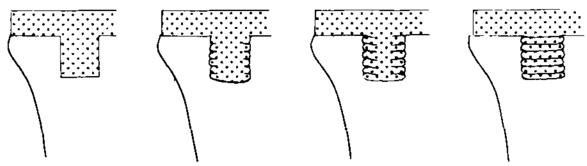


Figure 12. Assessment of the sclerotic reation around the flanged pegs of the Freeman-Samuelson prosthesis.

The interface of the Freeman-Samuelson knees was graded in four stages according to the sclerotic reaction of the bone; a modification of the staging previously proposed (Blaha et al. 1982) (Fig.12).

Roentgen stereophotogrammetric analysis (RSA)

The process of measuring the motion of one object relative to another (kinematics) can be divided into: 1. the determination of the three-dimensional positions of the objects of interest, and 2. the calculation of motion.

Determination of the three-dimensional position of an object in the human body by stereo technique (convergent rays) was achieved soon after the discovery of roentgen rays (Davidson 1898). The obligatory calibration of the stereo set-up can be obtained conveniently by utilizing a three-dimensional test object prior to, or even simultaneously with, the "real" radiographic examination, making the process "simultaneously calibrated", as shown by Torlegård (1967).

The human skeleton has few points that are clearly identifiable from all directions. When viewed from two directions in the stereo situation, the points "seen" might thus not be identical and the measurements invalidated. This problem can be solved by the insertion of radiopaque markers, as first used in orthopedic stereophotogrammetric research by Lysell (1969).

The volume of data and the amount of mathematical processing necessary to determine the 3-D coordinates and obtain a kinematic analysis is enormous and represent a major reason why such work did not progress far until the arrival of modern computer techniques.

Selvik RSA system

In 1974, Selvik presented a comprehensive system of roentgen stereophotogrammetric analysis (RSA) including automatic calibration, bone markers and modern measuring and data processing techniques.

Marking of objects

The objects, i.e the prostheses and the proximal tibia, were marked with spherical markers (ø=0.8 mm) of the metal tantalum. Tantalum has two advantages; a high atomic number (73) giving a high degree of radiopacity and thus distinct images on radiographic films, and a high degree of biocompatibility resulting in virtual osseointegration (Alberius 1983). In the prostheses the markers were inserted into holes in the polyethylene. These holes were drilled with a dentist's drill with a diameter slightly less than 0.8 mm. For the all-polyethylene prostheses, the markers were inserted from underneath whereas the metal-supported prostheses were marked from their periphery. The markers were usually inserted maximally spread out; either anteriorly, posteriorly, laterally and medially or in each quarter of the prosthesis. The proximal tibia was marked with a special instrument with a spring-loaded piston pushing the markers through a cannula (Aronsson et al. 1974). The markers were introduced through the upper tibial surface after resection of the subchondral bone, and were thus lodged in the cancellous bone or the inner side of the cortical bone (Fig.13).

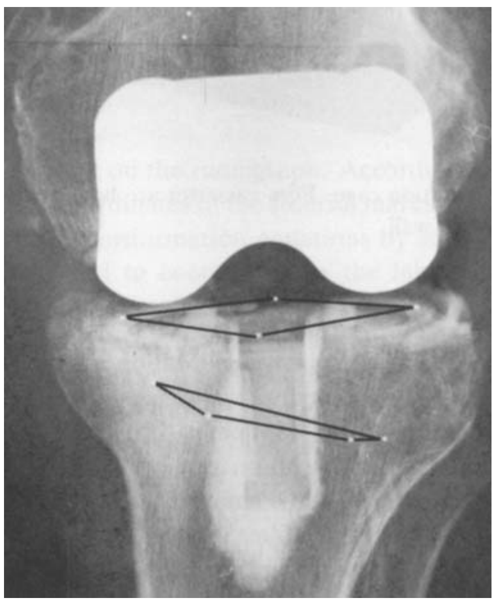


Figure 13. Radiograph of a Total Condylar prosthesis prepared for RSA. The groups of markers in the tibia and the prosthesis, respectively, are modelled as two rigid bodies.

Stereophotogrammetric exposures

The RSA system includes both a situation with convergent rays (40° alternative) and a biplanar situation (90° alternative). In this work the biplanar technique was used.

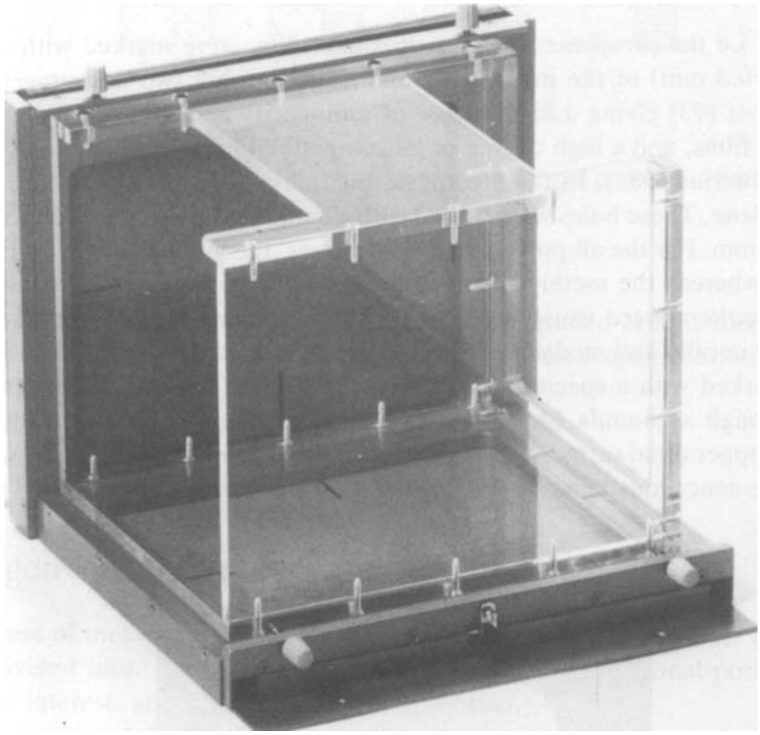


Figure 14. The stereo calibration cage. Film cassettes are held behind and beneath the cage, parallel to the adjacent cage wall.

The calibration of the stereo set-up is carried out simultaneously with the patient examination by putting the knee inside a calibration cage with tantalum markers in exactly known positions in each of the four walls (Fig.14). The position of the roentgen foci relative to the coordinate system defined by the calibration cage is determined for every stereo exposure, and their position can be changed between exposures. A film-focus distance of approximately 100 cm with the central rays perpendicular to the films was aimed for. The radiographic films were positioned in holders at 90° from one another, immediately beneath and behind the calibration cage. Measurements of film coordinates were made with an instrument for cartography (Wild A8) with an accuracy of about 20 µm. These primary data were processed by a computer (Sperry-Univac 1100) using the program X-RAY to obtain 3-D coordinates.

3-D Reconstruction

In radiography, images are obtained by an ideal bundle of rays projecting from the roentgen focus; rays that are not deflected or distorted to any appreciable degree. In RSA the images of the markers in the calibration cage plane adjacent to the film

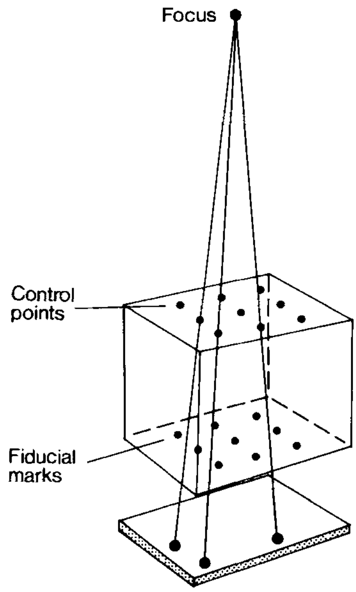


Figure 15. Determination of the position of the roentgen focus. The image and the known position of a control point define a line which can extrapolated towards the focus. A number of such lines intersect at the focus.

(fiducial marks) are measured on the radiograph. According to the law of projective transformation, the image coordinates of the fiducial marks together with their known coordinates determine two transformation equations by which image coordinates of an object can be transformed to coordinates in the laboratory coordinate system defined by the plane of fiducial marks (Selvik 1974). Images of the markers in the calibration cage plane distant from the film (control points) are transformed to the plane of fiducial marks, and thus with two points on a line (the known position of the control point and its image transformed to the plane of fiducial marks), the line can be extrapolated to the focus. A number of such lines will approximately intersect at the focus, the position of which is thus determined (Fig. 15) using a least-squares method. To determine the object point coordinates, its film image is transformed to the laboratory coordinate system. The real position of the object point is somewhere on the line connecting its image with the focus. The position of the object point in space is determined by the intersection of corresponding lines between the two foci and the two transformed images of the object (Fig.16), a computation performed by the program X-RAY.

Each marker in the patient gives two corresponding images, one on each radiograph. These images should be given the same number in the 2-D measurements, also corresponding to the number given to that particular marker in preceding examinations. The coupling of corresponding images in the two projections is done by comparing the y-parallax, i.e. the vertical component of the directed line segment connecting the two rays at their shortest distance. This is done by an optimization part

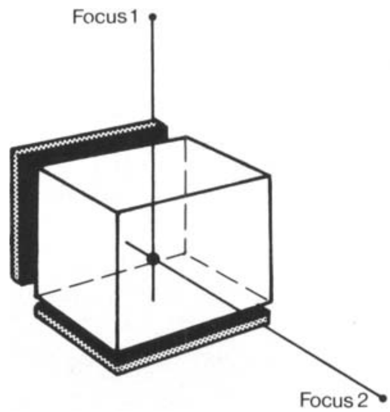


Figure 16. 3-D coordinate determination. The object (tantalum marker) is somewhere on the line which connects its image to the roentgen focus. Two foci determine the 3-D position.

of the program X-RAY where possible combinations (y-parallax <1 mm) are presented in a matrix. The finding of the proper coupling (= "solving of the rigid body") is done partly manually, since the computer's choice, the combination with the smallest y-parallax, might not always be the true one because of technical errors such as film bending, because of patient movement between the two exposures, or because a number of markers are positioned at approximately the same vertical level.

Kinematic analysis

The kinematic analysis is performed using classic rigid-body kinematics according to Euler (1776), who stated that any such motion can be characterized by three rotation and three translation parameters. When comparing the position of the object of interest in the laboratory coordinate system on two occasions, the difference is equal to the motion that has occurred. In this particular work, two objects of interest were used (the proximal tibia and the prosthesis) and motion of the prosthesis was expressed in relation to the tibia.

The motion of the prosthesis was expressed as rotation about and translation along each of the three cardinal axes in relation to the tibia, i.e. *segment motion* (SM). Alternatively, the translation of each of the prosthetic markers could be presented, i.e. *point motion* (PM). The length of the translation vector (having the x-, y- and z-translations as composants) for the marker with the largest total motion was denoted *maximum total point motion* (MTPM). This MTPM is an approximate synthesis of rotation and translation and represents a simple way to express the magnitude of the motion, as it is a scalar in contrast to the rotation matrix and translation vector.

For the kinematic analysis, the objects of interest, characterized by their tantalum markers, are modelled as two *rigid bodies*. The minimum number of markers defining a rigid body is three, not on a line. Mostly, four to six markers were used to overdetermine the rigid body allowing for optimization procedures with redundant observations. The accuracy of the kinematic analysis depends upon lack of deformation of these rigid bodies.

The program for kinematic analysis, KINEMA, initially tests the rigid-body models by calculating inter-marker distances and angles at the reference time, which values subsequently are compared to successive examinations and registered changes listed. Thereafter, computation of rotation matrices and translation vectors for absolute motion of each segment is conducted by minimizing the mean error of rigid body fitting (ME). Finally, the relative displacement between the segments is calculated. The ME can be interpreted as the root mean square distance (error) between segment markers in a rigid-body displacement compared to their actual movement.

$$ME = \sqrt{\frac{d_1^2+d_2^2+d_3^2+\ldots.d_n^2}{n}},$$

where d = the length of the residual motion vector for each marker after optimizing the rotation matrix and subtracting the rigid body displacement using a least-squares method and n = the number of markers. Thus, the calculated position of the rigid-body model (RBM) will optimally represent the true position. For perfect rigid-body behavior of the tantalum marker model, assuming no measurement errors, the ME is zero. Should a marker prove to move in relation to the others in the segment studied, this marker is excluded from the calculations.

Point motion analysis can also be done for parts of the prostheses not marked using a *point transfer* (PT) routine included in the RSA system. Thus, artificial marks can be made on the radiographs at points corresponding to arbitrarily chosen parts of the prosthesis in both projections. These points are measured and the 3-D coordinates of the point most closely corresponding to the two (both projections) marks made are determined as the mid-point of the shortest line segment joining the two rays. This artificial point is transformed to the prosthesis on other occasions and its motion relative to the bone can be determined. The procedure was used to determine the MTPM for 15 Total Condylar prostheses.

Accuracy of RSA

The accuracy of the RSA system was determined on clinical material by repeated examinations presuming stable conditions between the objects studied, i.e. between prosthesis and bone. The standard deviation of the method was calculated from these movements by the formula

$$SD = \sqrt{\frac{\Sigma\ r^2}{n}},$$

where r = x-,y- or z-axis rotation or translation, respectively (Ryd et al. 1983). The accuracy thus calculated depends on a number of factors, such as the size, geometry and number of markers of the RBMs and the geometry of the stereo arrangement (40° or 90° alternative) and can therefore differ depending on the application. In studies using the 90° alternative of facial bone displacements in adults an accuracy of 0.1° for rotation and 0.05 mm for translation (1 SD) has been reported, while the corresponding values in infants were 0.4° and 0.1 mm, respectively (Rune et al. 1979, Rune 1980). Studies on hip arthroplasty have yielded an accuracy of 0.1 mm (1 SD) (Baldurson et al. 1979).

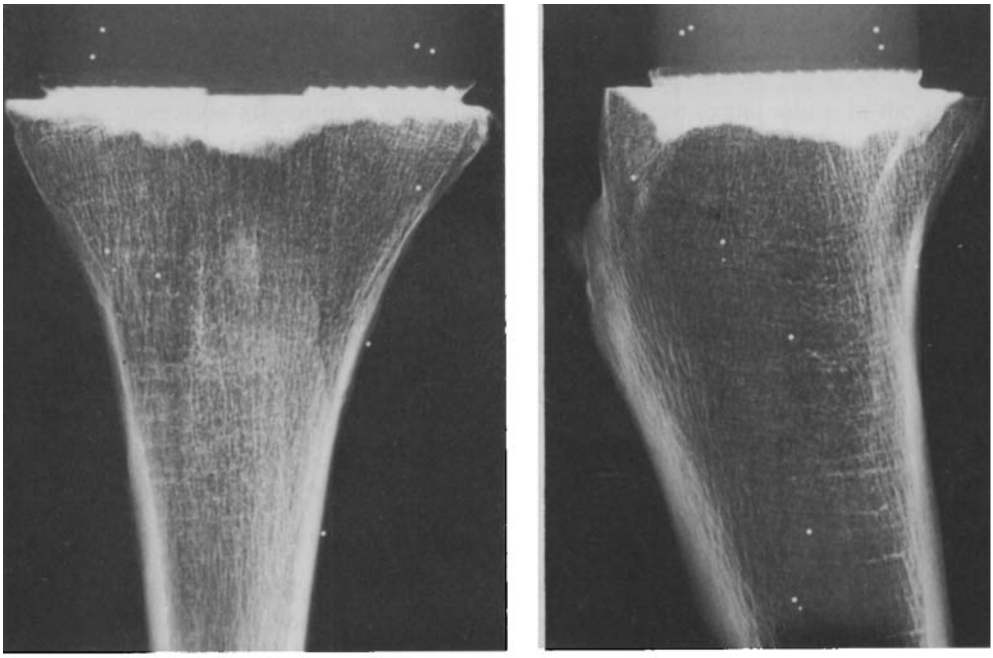
Applied to knee arthroplasty repeated examinations were made on **clinical material** (13 patients) and the motion thus calculated (1 SD) ranged from 0.07°–0.10° for rotation and 26 μm – 62 μm for translation (Table 1). With the stress investigation (see page 33) it became evident that the presumption of stable conditions between the bone and the prosthesis in the repeated examinations was fallacious and that true motion between the prosthesis and bone could interfere with the calculations. Therefore, a series of 16 repeated examinations was done using a **bone specimen** prepared to resemble the real situation (Fig.17). In this way 8 totally independent "motions" were determined and the SD was calculated (Table 2), to represent the optimal technical accuracy of the RSA system applied to knee arthroplasty.

Table 1. Calculation of the SD (accuracy) from repeated examinations in 13 patients ("in-vivo"). The SD of total motion from computer simulation.

	X-axis	Y-axis	Z-axis	Total	
Segment motion					
Rotation	0.10°	0.07°	0.09°	0.10°	(df=13)
Translation	32μm	26μm	48μm	46μm	(df=13)
Point motion	47μm	54μm	62μm	66μm	(df=52)

Table 2. Calculation of the SD (accuracy) from 16 stereo examinations of a bone specimen ("in-vitro"). The SD of the total motion calculated as for single axis motion. Number of markers pertain to the bone rigid bodies.

	Number of markers	X-axis	Y-axis	Z-axis	Total	
Segment motion						
Rotation	4	0.029°	0.038°	0.026°	0.052°	(df=8)
	6	0.035°	0.032°	0.025°	0.056°	(df=8)
Translation	4	21μm	6μm	11μm	24μm	(df=8)
	6	18μm	10μm	11μm	23μm	(df=8)
Point motion						
	4	27μm	16μm	26μm	40μm	(df=32)
	6	24μm	18μm	25μm	39μm	(df=32)



A B

Figure 17. The bone specimen used for accuracy determination. The four tantalum markers in the prosthesis and the six in the tibia are arranged as to represent the clinical situation.

For mathematical reasons, the SD of the *total* vector motion cannot be calculated accurately as above and indeed cannot be determined analytically. By a computer simulation (Monte Carlo) using the values in Table 1, the total accuracy (3 SD) was found to be 0.3°, 0.14 mm and 0.20 mm for segment rotation, segment translation and point motion, respectively. In the stress examinations the largest value out of four is presented. The deliberate choice of a largest value introduces a statistical bias, but this additional error was calculated to be negligible (Sarham & Greenberg 1962).

The accuracy for 4 and 6 markers was almost identical (Table 2). However, this finding pertains only to rigid bodies where the ME is small, representing the error of measurement of film images only. If the true motion of the markers within the object of interest is also introduced, the ME increases and the accuracy of the method deteriorates. In this situation a larger number of markers should give a better optimization and thus more accurate results.

Influence of mean error

To investigate the influence of an increasing ME and an increasing number of markers, the primary data of the bone specimen measurements were manipulated. Thus the

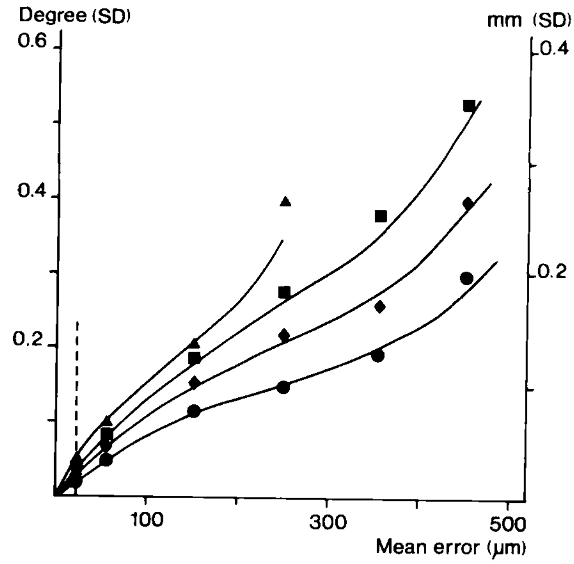


Figure 18. The influence on the accuracy of RSA of an increasing ME and an increasing number of markers. The dashed line indicate the ME obtained under optimal conditions and represents technical errors exclusively. (▲ = 3 markers, ■ = 4 markers, ⧫ = 5 markers, ● = 6 markers.) Left ordinate = single axis rotation, right ordinate = translation (MTPM).

position in space of each bone marker was changed 0.05–0.5 mm in a positive or negative direction along one of the coordinate axes in a random fashion, giving an ME ranging from 50–500 μm for 160 different RBMs. By running the KINEMA program using these manipulated "rigid bodies", including from 3 to 6 bone markers, the influences on the accuracy of an increasing ME and of an increasing number of bone markers were determined (Fig. 18).

From these investigations on the accuracy of the method, the following limits for significant motion were chosen for this work:

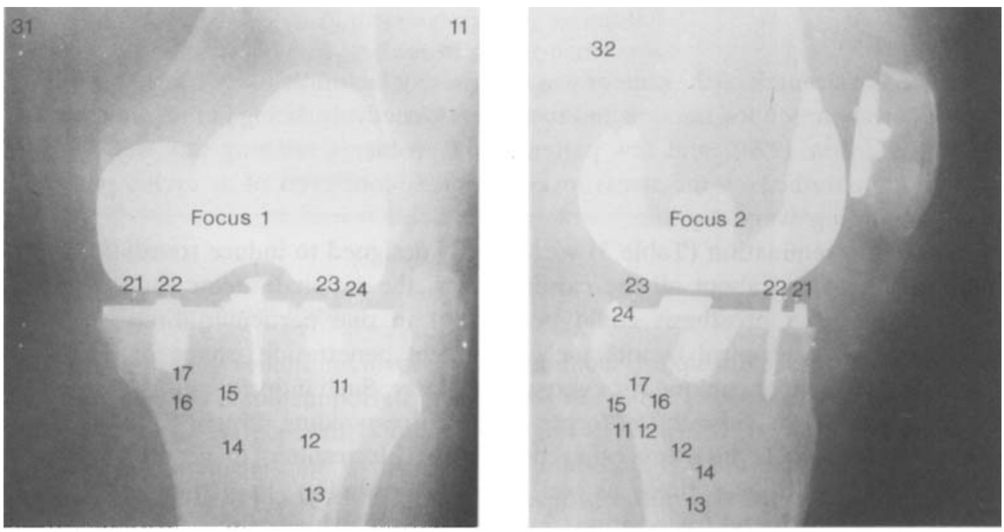
Single axis rotation 0.3°;
Total rotation 0.4°;
Single axis segment translation 0.2 mm;
Total segment translation 0.2 mm;
Single axis and total point motion 0.2 mm.

In this investigation the ME was usually less than 100 μm and an ME above 200 μm was only occasionally accepted. The values reported in this work were thus within the 95–99% confidence limit.

Patient investigations

The migration study

After the operation all patients were subjected to a stereo examination in the supine position on the first or second post-operative day, before they were allowed to put weight on the operated leg. The leg was positioned as straight as possible in the calibration cage to have the principal planes of the body as close to the laboratory coordinate axes as possible (Fig.19). In the KINEMA program the bone RBM is transformed back to the original position, which serves as a reference position. Stereo examinations were done after 1 week (excluding the Modular and the Total Condylar series), 6 weeks, 6 months and yearly, and the difference in prosthetic position relative to the reference examination represents the migration.



A B

Figure 19. Stereo radiographs of a PCA prosthesis. Note cage and patient markers. For metal-supported components, the projections of the tibial component had to be approximately tangential with the metal tray in order to visualize the markers in the polyethylene. The greyish hue is due to the exposure caracteristics chosen in order to minimize the absorbed radiation dose; only the tantalum markers had to be clearly visualized.

The stress study

To obtain stereo radiographs with external forces acting on the knee a special platform was constructed (Fig.20). Forces were standardized by weights acting on the knee; either directly over pulleys or indirectly applied to a rotating plate, and the patients were told to resist the action of the force. In this way the forces became physiological,

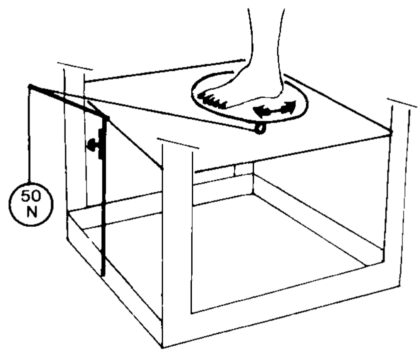


Figure 20. The "stress platform". The patient stands on a rotating plate to which 50 N is applied in either direction to a 20 cm lever arm.

as the muscle strength of the patient was the limiting factor. A 100 N force or a torque of 10 Nm was chosen for traction and rotation, respectively, being in the physiological range (Morrison 1970), and few patients had problems resisting this weight. The micromotion studied by the stress investigation is concieved of as cyclic, probably occurring during every day life.

The stress examination (Table 3) was initially designed to induce translations and rotations along and about all the cardinal axes, the rationale being that inducible displacement of a prosthesis could be present in one particular direction only, depending on mechanical factors such as cement penetration, shape of the bone resections etc., factors unique for every arthroplasty. Such inducible displacement is, hypothetically, revealed only by forces in the corresponding direction, the fixation possibly being stable in every other direction. This resulted in a very elaborate examination, which was time- as well as film-consuming. In a pilot study little displacement was found for positions 2, 4, 6 and 9, which were therefore excluded. In this work the inducible displacement found for positions 5,7,8 and 9 was, in most cases, less than for either weight bearing or rotation. Results are therefore presented only for weight bearing and rotation in both directions. That rotational stress is an important test of inducible displacement has been shown also for hip prostheses (Mjöberg et al. 1984a).

Only inducible displacement corresponding to the external force was given in the results. The weight-bearing position gives a compressive force on the tibial prosthesis, but due the alignment, muscle and ligament balance, cortical bone support etc.; motion in any direction can be seen. Thus vertical translation and total rotation were recorded. Rotatory forces create a counter force in the opposite direction acting on the tibial component, and the rotation occurring between outward and inward rotation was presented.

This work represents more than a hundred stress examinations. On a few occasions slight discomfort was evoked but in no case has there been any adverse effects on the prosthetic function.

Table 3. The initial stress examination. In positions 3–11 the patient is standing on the operated leg only. A stereo exposure is taken in each position.

Position	
1	Supine position.
2	Vertical position with hanging leg (no weight-bearing).
3	Vertical position weight-bearing.
4	10N applied immediately below the joint line pulling laterally (varus).
5	As 4, the weight pulling medially (valgus).
6	Shearing. 10 N applied below the joint line while a counter buttress acts on the femur in the opposite direction. Weight pulling laterally.
7	As 6, the weight pulling medially.
8	As 6, the weight pulling forwards.
9	As 6, the weight pulling backwards.
10	Rotation. 10 Nm acting on a rotatory plate in outward rotation, which the patient is told to resist.
11	As 10, inward rotation.
12	Supine position.

Radiographic exposure burden

RSA examinations sometimes involve a large number of exposures; especially the stress examinations on metal-supported prostheses. For these prostheses the metal bases have to be perfectly parallel to the roentgen rays, which are not controlled by fluoroscopy, in order not to obscure the markers. On four patients, therefore, measurements were made of the radiation exposure during a stress series.

The absorbed dose was measured with thermoluminescent dosimeters (TLD) of lithium fluoride with a measuring range from 20 µGy to 5 Gy. Four TLDs were used for each knee, placed in front of and behind the knee in both projections. The intensifying screens used were the Cronex Quanta Detail from DuPont with an sensitivity of 1200 R^{-1} (ASA 1956). The exposures were made at 90 kV and 1.6 mAs and an average of 7.5 (4–10) double exposures were taken per stress examination. Under these conditions the mean absorbed dose per stress series was found to be 1.86 (0.83–2.90) mGy.

For comparison, the radiation dose during a routine knee examination under weight-bearing conditions was measured on four patients. These examinations included three exposures: a frontal exposure using an AGFA MR 200 screen (2000 R^{-1}), a lateral exposure of the entire lower limb using a KODAK X-omatic Regular screen (2000 R^{-1}) and an axial exposure of the femoro-patellar joint using a DuPont Cronex Detail screen (200 R^{-1}). 70 kV and 100 mAs were used and the absorbed dose was found to be 2.29 (0.27–4.88) mGy. The absorbed dose in a stress series was thus approximately 80% of that of a routine knee examination.

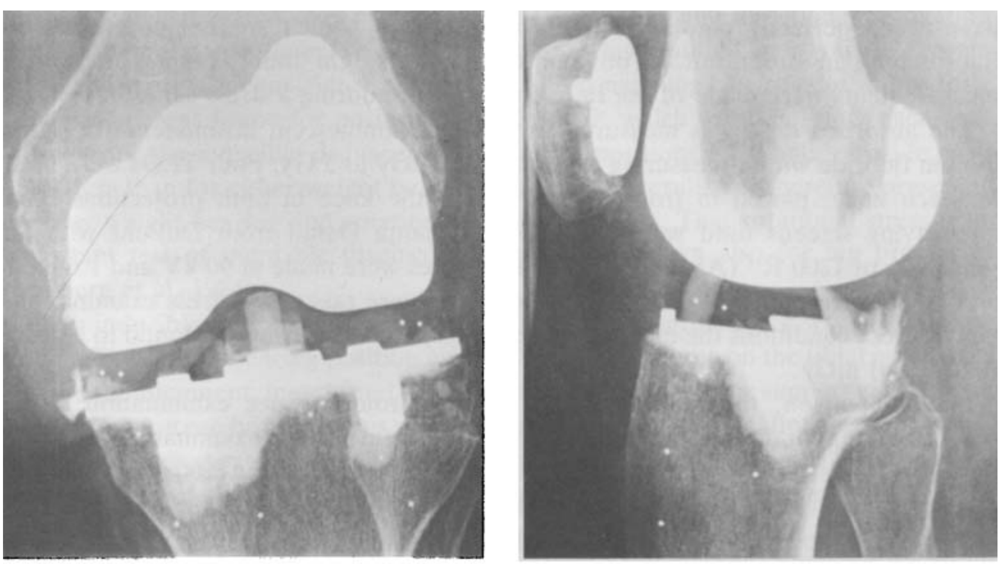
At interfaces between materials of high and low atomic numbers (Z), secondary radiation from the high-Z material, in this case the prosthetic metal or tantalum markers, is absorbed by the soft tissue. Such secondary radiation is of the electron type and has an extremely short range (<20μm). This interface secondary radiation increases dramatically with an increasing Z-value. For steel (Z=26) or vitallium (Z $\approx$ 30) prostheses the absorbed dose increases by a factor of 3–4, while the absorbed dose close to lead (Z=82) increases by a factor of $\approx$ 30 (Alm-Carlsson 1973). The effect of such secondary radiation is presently unknown. In cases with osseointegrated implants of high Z numbers adverse effects on the fixation may possibly be caused by radiation (Alm-Carlsson 1985).

It was concluded that radiation exposure was not a limiting factor for RSA investigations of the knee replacements in this investigation.

Mechanical integrity of metal-supported components

With the introduction of metal-supported tibial components, movements were possible between the plastic and the metal within the prosthetic component. Such movements would invalidate the RSA findings.

To investigate possible motion between plastic and metal, markers were also introduced into the cement in five PCA arthroplasties (Fig.21). Presuming perfect stability between the cement and the porous-coated metal tray, these cases permitted



A B

Figure 21. PCA prosthesis (Case 59) with tantalum markers also in the cement.

Table 4. Motion between polyethylene and metal in 5 cemented PCA prostheses supplied with markers also in the cement. A, B and C were not further included in this work.

Case	Weight bearing	Valgus	Shearing medially	Shearing anteriorly	Rotational	Max.total rotation
62	0.5	−0.4	–	–	0.2	0.5
62	0	–	–	–	0	0
63	0	0	–	–	0	0
63	0	–	–	–	0	0
A	0	0	0	0	0	0
A	0	–	–	–	0.6	0.7
B	0	0	0	0	0.5	0.6
C	0	–	–	–	0	0

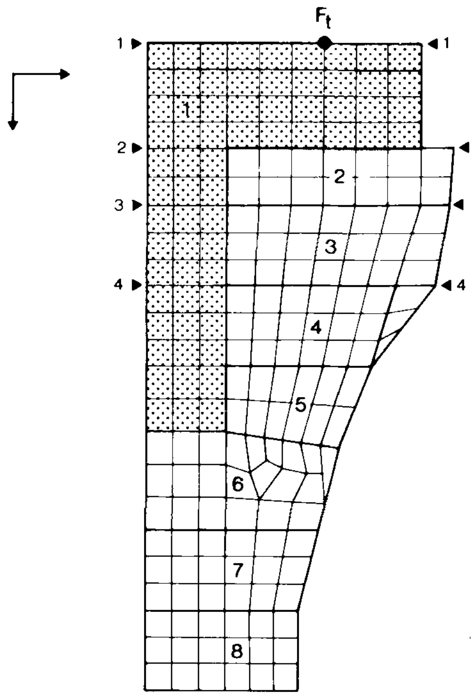
analysis of motion between the plastic and metal of the tibial component. These patients were subjected to a total of eight stress examinations and motion up to 0.7° was found in three prostheses (Table 4).

The investigation indicates that instability within the PCA knees can influence the results. For the Kinematic and Prototype prostheses, where the plastic is more securely bonded to the metal by a circumferential rim, motion within the tibial component probably is negligible.

A finite element analysis of the elastic properties of the upper tibia

While the motion of interest for prosthetic fixation is that between bone on one hand and cement or prosthesis on the other, the motion studied in this work was that occurring between markers in the polyethylene of the tibial component and in the tibial metaphysis. The distance between these two groups of markers (rigid bodies) covers a certain bulk of material, the elastic properties of which may present some relative motion during the stress investigation. To study the magnitude of such elastic motion, presuming rigid interface(s), an analytic investigation using a finite element model (FEM) (Huiskes & Chao1983) was carried out in collaboration with the Biomechanical Laboratory, Department of Orthopedics, Nijmegen, the Netherlands (Rik Huiskes).

A 3-D axi-symmetric model of the proximal tibia including a Total Condylar prosthesis was designed and a finite element mesh was generated by the computer program MENTAT (192 elements, 224 nodes; 4 nodes per element; bilinear displacement field within an element; isoparametric formulation; Fig.22). The dimensions were obtained from direct measurements of the tibia and the prosthesis and the material properties were those reported by Hayes et al. (1978). A model including composite



igure 22. Finite element model of the proximal tibia including a Total Condylar prosthesis. The elative displacement of levels 1–4 was studied. Level 4 represents the average distance of the tantalum markers to the prosthetic bed. 10 Nm rotatory torque was applied at F_t. Young's modulus of elasticity (E) of

region 1 (prosthesis)=$1x10^3$ MPa,
region 2 (cancellous bone)=$0.4x10^3$ MPa,
region 3 (cancellous bone)=$0.3x10^3$ MPa,
region 4 (canc./cortical bone)=$1x10^3$ MPa,
region 5 (canc./cortical bone)=$2x10^3$ MPa,
region 6 (canc./cortical bone)=$4x10^3$ MPa,
region 7 (canc./cortical bone)=$6x10^3$ MPa,
region 8 (canc./cortical bone)=$8.5x10^3$MPa.

elastic moduli for cancellous and cortical bone for regions 4 through 8 was used. This simplification was permissible since local displacements within these regions were not studied. By applying a tangential force of 450 N at point F_t, a torque of 10 Nm (= the rotatory torque during the stress series) was obtained and the displacements in regions 1–4 were analyzed by the FEM computer program MARC.

The maximum displacement occurred at the point where the force was applied and was equal to 8×10^{-4} radians = approximately 0.05° (Fig. 23). The angular displacement between lines 1 and 4 was 3×10^{-4} radians = approximately 0.017°. The external and internal rotatory stress positions thus give a relative elastic deformation of 0.034°.

The axi-symmetric model represents a certain simplification of the true three-

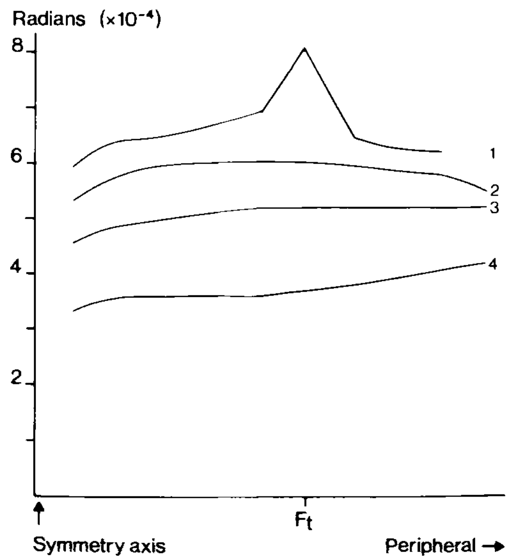


Figure 23. Results of the FEM analysis (1×10^{-4} radians ≈ 0.0057°). Lines 1–4 represent the displacements of levels 1–4 in Figure 22.

dimensional geometry of the upper tibia. However, since the elastic displacements were found to be more than one order of magnitude smaller than the displacements found in the stress examinations on patients, this error was regarded as negligible. If cement was included, the model would be stiffer, since cement is stiffer than both cancellous bone and polyethylene.

It was concluded that inducible displacement, of a magnitude like that found in the stress examinations, could only occur in the interfaces between materials.

Statistical methods

(Multiple) linear regression analysis was used to correlate RSA results to clinical and radiographic parameters. Student's t-test (two-tailed) was used to compare RSA results between groups. The statistical limits chosen were:

$p < 0.001$ = (***)
$0.01 > p > 0.001$ = (**)
$0.05 > p > 0.01$ = (*)
$p > 0.05$ = (–)

Results

Micromotion in cemented all-polyethylene prostheses

Micromotion, both migration and inducible displacement, was found for every single all-polyethylene prosthesis.

For the **Total Condylar** prostheses the mean migration was 0.8 (0.2–2.5) mm (MTPM) at the last follow-up, while the corresponding value for the **Modular** prostheses was 1.2 (0.3–5.4) mm. For both types of prostheses the major part of the migration occurred during the first 6–12 months (Fig. 24), after which time most of the prostheses remained stable.

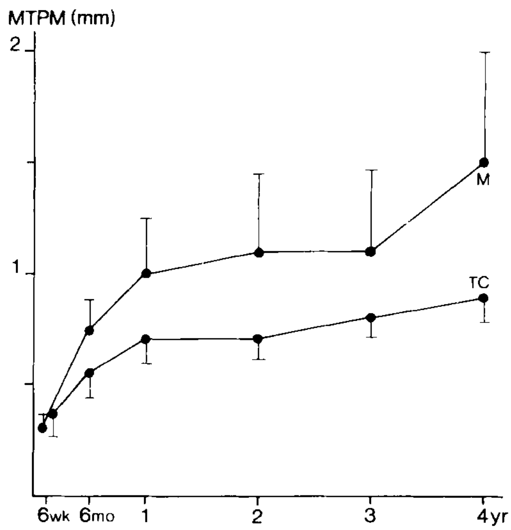


Figure 24. Migration (MTPM±SEM) of all-polyethylene prostheses. (M = 24 Modular prostheses, TC = 27 Total Condylar prostheses.)

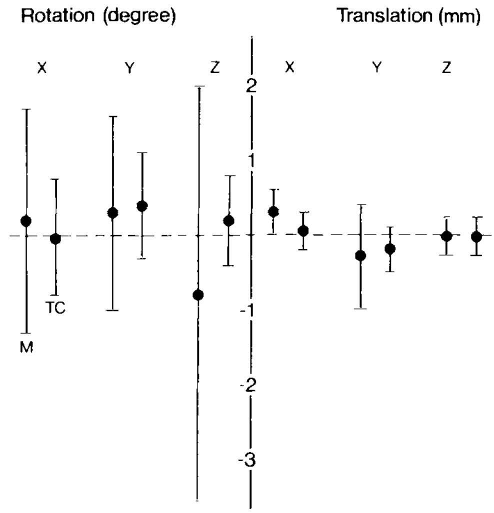


Figure 25. Direction (mean±SD) of the migration after 2 years for 24 Modular and 27 Total Condylar prostheses.

The migration occurred in both directions about or along all three axes (Fig.25). There was a greater number of significant rotations than translations. Most consistently the Total Condylar prostheses rotated outwards about the vertical axis while the unicompartmental prostheses rotated with the anterior part away from the centre of the knee about the same axis. Also, all Modular prostheses that translated along the transverse axis, did so away from the centre of the knee.

Inducible displacement, exclusively as segment rotation, was found for every all-polyethylene prosthesis subjected to a stress examination. Thirty-three stress examinations on Total Condylar prostheses showed a mean inducible rotational displacement of 0.7 (0.3–1.8) degrees, corresponding to a MTPM of 0.4 (0.2–1.0) mm. The rotatory stress positions proved to be the most efficient in revealing inducible displacements. Thus in 18 stress examinations the rotatory stress positions yielded larger displacement than the weight-bearing position while the opposite was true in 10 examinations.

Seven stress examinations on Modular prostheses showed consistent inducible rotational displacement of 0.7 degrees, corresponding to a MTPM of 0.3 (0.2–0.4) mm. No stress position was superior in revealing inducible displacement in unicompartmental prostheses.

In all cases but one the inducible displacement was reversible and the prostheses regained their original position at the last supine exposure in the stress series. Modular Case 32 showed progressive migration, and became symptomatic after approximately 4 years. The stress examination when the the knee again was asymptomatic did not reveal any larger inducible displacement. This case was the only cemented prosthesis that showed a change of position between the first and the last exposures in the stress examination, both in the supine position.

Micromotion in cemented metal-supported prostheses

All metal-supported components that were cemented migrated, predominantly during the first year (Fig.26). The **Kinematic** prostheses had migrated 1.1 (0.3–1.9) mm (MTPM) after 2 years. The corresponding values for **cemented PCA** prostheses was

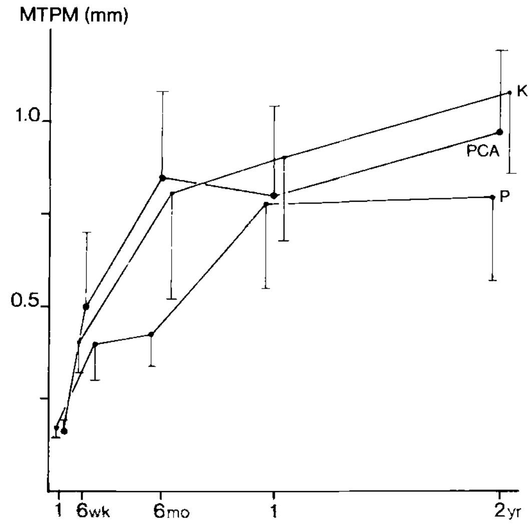


Figure 26. Migration (MTPM±SEM) of 12 Prototype (P), seven cemented PCA and seven Kinematic (K) metal-supported prostheses.

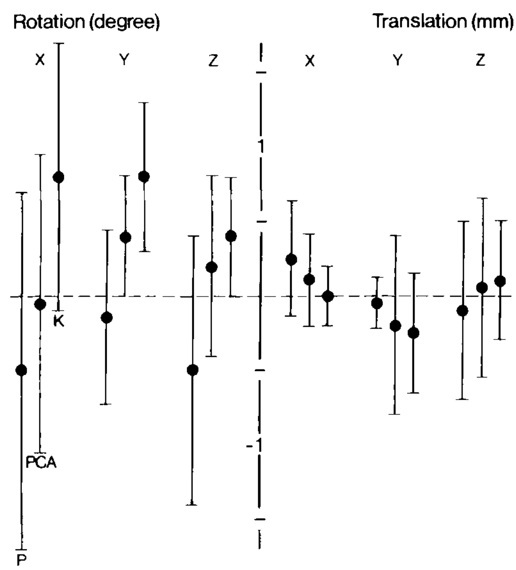


Figure 27. Direction (mean±SD) of the migration at 2 years of 12 Prototype (P), seven cemented PCA and seven Kinematic (K) metal-supported prostheses.

0.9 (0.2–1.6) mm and for the **Prototype** prostheses 0.9 (0.3–3.2) mm. For the total prostheses the most consistent migration was an outward rotation about the vertical axis while the unicompartmental prototype prosthesis consistently translated away from the center of the knee along the transverse axis (Fig.27). Five out of seven Kinematic knees and all cemented PCA knees showed inducible rotational displacements ranging from 0.4–1.2 and 0.3–0.7 degrees, respectively, corresponding to a mean MTPM of 0.6 (0.2–1.0) and 0.4 (0.2–0.8) mm. For both these prostheses the weight-bearing stress position was more efficient than the rotatory positions in revealing inducible displacement.

Micromotion in non-cemented prostheses

Only three of the **Freeman-Samuelson** cases were followed longitudinally and for these a migration of 1.3, 1.5 and 1.2 mm (MTPM) was found, respectively, after 2 years (Fig.28). The migration occurred about and along all three axes (Fig 29).

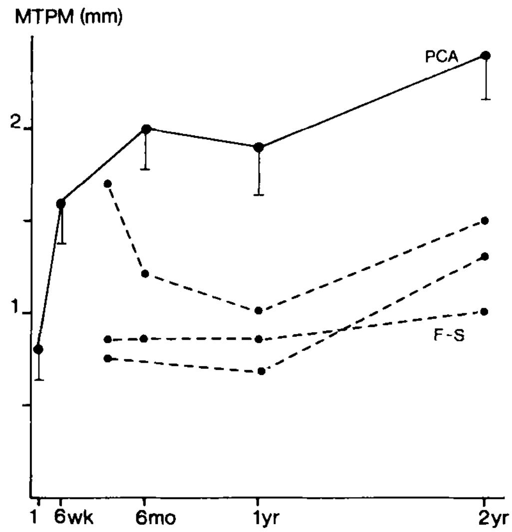


Figure 28. Migration (MTPM±SEM) of 13 non-cemented PCA and three Freeman-Samuelson prostheses.

Inducible displacement was found for all six cases with a mean MTPM of 1.9 (0.5–5.0) mm. Both rotational displacements, ranging from 0.8 to 5.4 degrees, and translational displacements, ranging from 0.2 to 1.4 mm, were found. The inducible displacement was found to be of a plastic nature in these cases; there was a difference in position of the prostheses between the two supine exposures starting and ending the stress examinations, amounting to 0.2 to 2.1 mm (MTPM). In one case (Case 95) the

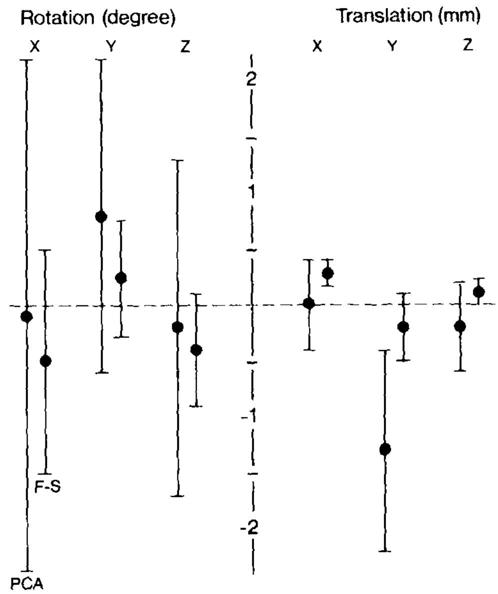


Figure 29. Direction (mean±SD) of the migration at 2 years of 13 non-cemented PCA and at 1 year of three Freeman-Samuelson prostheses.

prosthesis was distorted (ME = 900μm) under the rotatory stress. This distortion was painless and totally reversible. The rotatory stress positions were the most revealing in all cases.

All **non-cemented PCA** prostheses migrated predominantly during the first year. The mean MTPM at the last follow-up was 2.4 (0.5–5.0) mm (Fig.28). Migration occurred about and along all three axes. There was a considerable rotation in both directions about the transverse axis. Also, most prostheses rotated outwards about the vertical axis and every single prosthesis subsided (Fig. 29).

Inducible displacement with a mean MTPM of 0.7 (0.4–1.3) mm at 1 year and 0.6 (0.4–1.1) mm at 2 years was found in all cases. Rotational displacement ranging from 0.4–1.8 degrees was found in all cases at both 1 and 2 years, while translational displacement ranging from 0.2 to 0.5 mm was found in ten cases at 1 year and in seven cases at 2 years. In five cases the inducible displacement was of a plastic nature.

Micromotion: comparison between groups

There was no difference between the all-polyethylene prostheses regarding the magnitude of the migration at each follow-up. After 6 months, both the total prostheses (*) and the unicompartmental prostheses (***) that migrated continuously had migrated more than the corresponding prostheses showing only initial migration (Fig. 30).

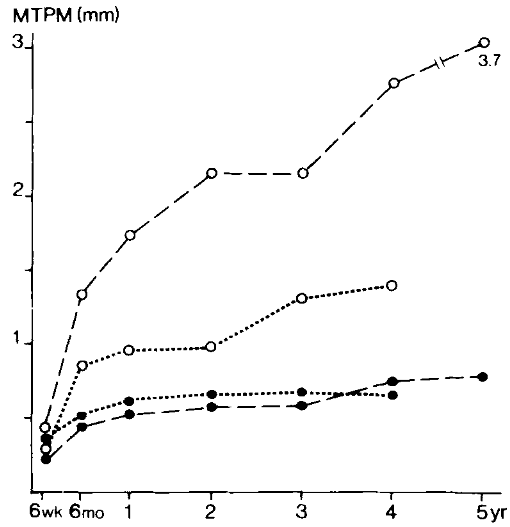


Figure 30. Continuously and non-continuously migrating all-polyethylene prostheses. The difference was significant after 6 months for Total Condylar (*) and for Modular (***) prostheses (●----● = Modular non-continuous n=15, ● · · · ● = Total Condylar non-continuous n=19, o----o= Modular continuous n=9, o · · · o= Total Condylar continuous n=8).

Among the metal-supported prostheses, the Kinematic prostheses showed a larger (*) rotational migration at 2 years than the all-polyethylene total prostheses. There was no difference in magnitude of the migration between cemented PCA prostheses and the total all-polyethylene prostheses, and no difference between the all-polyethylene and metal-supported unicompartmental prostheses.

The migration of non-cemented PCA components was larger than for Total Condylar (***), for Kinematic (**) and for cemented PCA (*) components. The direction of the migration was similar for all total prostheses; a majority of the components rotated outwards about the vertical axis. However, the amplitude of the migration about/along all axes was larger for the non-cemented components than for the cemented ones (**–***). Non-cemented PCA prostheses subsided a mean 1.3 (0.2–3.5) mm into the tibia, which was different (***) from the cemented components of which only six cases subsided. The three Freeman-Samuelson prostheses had migrated 1–1,5 mm (MTPM) after 2 years.

There was no difference regarding inducible displacement between the different types of cemented prostheses. The non-cemented PCA knees displayed an inducible displacement which was larger than that of the Total Condylar (***), the Kinematic (*) and the cemented PCA (**) prostheses, but was smaller (**) than that of the Freeman-Samuelson prostheses.

The results of RSA can be summarized as follows: There was no difference between cemented all-polyethylene and metal-supported prostheses regarding migration and

	A	B	C	D	E
A		NS	NS		NS
B	NS		NS		NS
C	NS	NS			NS
D	***	**	*		NS
E	NS	NS	NS	NS	

A

	A	B	C	D	E
A		NS	NS		
B	NS		NS		
C	NS	NS			
D	***	*	**		
E	***	**	**	**	

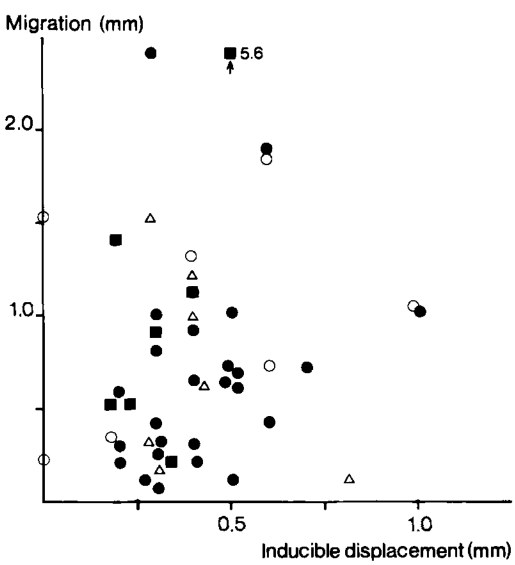
B

Figure 31. Differences in migration (A) and inducible displacement (B) between the total prostheses studied (A=Total Condylar, B=Kinematic, C=PCA cemented, D=PCA non-cemented and E=Freeman-Samuelson).

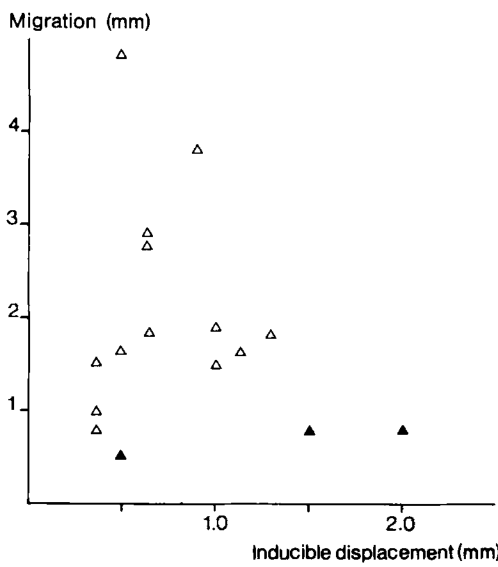
inducible displacement. Unicompartmental prostheses showed as much migration and inducible instability as total prostheses. Non-cemented total prostheses showed significantly more migration and inducible displacement than cemented total prostheses (Fig.31).

No correlation was found between the migration and the inducible displacement calculated either for each series separately or for the entire material (Fig.32).

No correlation was found between the post-operative time and the inducible displacement. In all cases where a secondary stress examination was done, there was a significant agreement between the two examinations; in eight Total Condylar cases (***) as well as for the PCA knees, both cemented (**) and non-cemented (**).



A



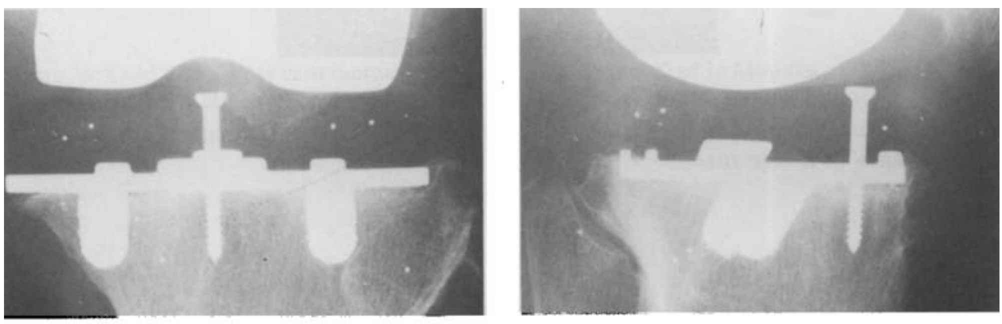
B

Figure 32. Migration versus inducible displacement (MTPM) for cemented (A) and non-cemented (B) prostheses (● = Total Condylar, ■ = Modular, ○ = Kinematic, △ = PCA and ▲ = Freeman-Samuelson).

Micromotion versus the radiolucent zone

Only four of the cemented cases did not develop a radiolucent zone while six developed complete or almost complete (>90% zone extension) zones. For Total Condylar knees the extension of the radiolucent zone correlated (**) with migration at any time, while zone width and zone value correlated at 1 year (***) but not at 3 years. No correlation was found between the size of the zone and inducible displacement. No correlation could be found between the size of zone and micromotion for any of the other prostheses.

The non-cemented PCA prostheses developed a radiolucent zone in all but one case. These zones were most conspicuous in the peripheries, especially regions 8 and 11. Here, the shape often suggested rocking motions in the sagittal plane. All but one of the non-cemented PCA prostheses showed between 1 and 18 loose beads from the porous coating in the interface (Fig 33).



A B

Figure 33. Non-cemented PCA prosthesis (Case 79) with loose beads from the porous coating. Note the tapered form of the zone on the lateral projection.

By point motion analysis, the vertical migration of each prosthetic condyle was assessed with reference to the subjacent radiolucent zone. In ten Total Condylar cases and in one Kinematic case, one of the condyles had migrated upwards, away from the bone surface, while 15 Total Condylar and four Kinematic condyles subsided into the tibia (Fig.34).

The total inducible displacement of the markers in the flanged pegs of the Freeman-Samuelson prostheses ranged from 0.2 to 0.9 mm and the sclerotic reaction ranged from grade 1 to grade 4, having a diameter of 15–18 mm. There was a tendency (–) for larger inducible displacement with less sclerotic reaction, but not even a tendency towards a larger inducible displacement of the peg markers with an increasing diameter of the corresponding sclerotic reaction.

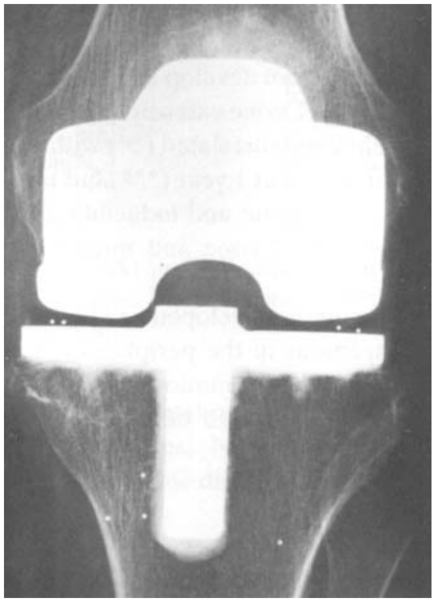


Figure 34. Frontal radiograph of a Kinematic prosthesis. The zone beneath the medial condyle is almost completely caused by a lateral tilt of the prosthesis about the sagittal axis creating a void medially (=tensile zone).

Cold flow of prosthetic components

All prosthetic components except the Modular knee were stable, i.e. the ME of the rigid body did not increase during the investigation and showed no relative movements of the markers during the longitudinal or the stress examinations. Fifteen out of the 24 Modular knees showed an increase of the circumference during the migration study, but not during the stress examinations. The increase (=cold flow) was progressive (Fig.35) and larger (*) for the 9 mm than for the 12 mm components at the last follow-up.

Micromotion versus clinical variables

No correlation could be found between the clinical assessment scores and either migration or inducible displacement.

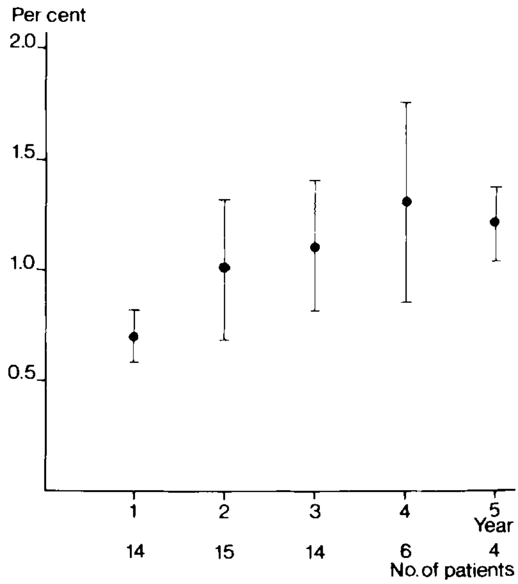


Figure 35. Cold flow (=per cent increase of intermarker distance) of 15 Modular all-polyethylene prostheses.

Micromotion versus radiographic variables

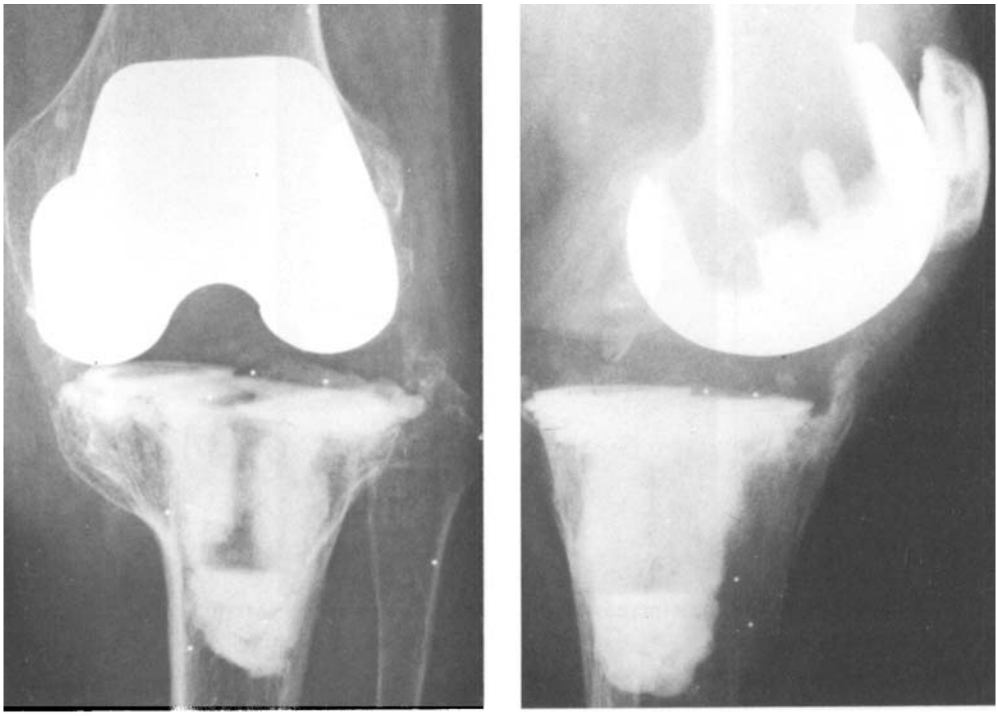
Cortical bone support

The Freeman-Samuelson prostheses showed good cortical bone support in most cases except anteriorly, where some bone was left uncovered in three cases. The Prototype prostheses had good cortical bone support. For the other types of prostheses, a varying amount of bone was left uncovered, ranging up to 15 mm (Total Condylar Case 3). No particular pattern was found, except for the PCA prostheses, where uncovered bone anteriorly was a common finding.

No correlation was found between the lack of cortical bone support and the migration for any of the prostheses.

Alignment

The alignment was generally good in this material. Twenty-three cases were, however, found to be outside ± 6 degrees and five of these were outside ± 10 degrees from the perfect' 180 degree HKA angle post-operatively. No correlation could be found between alignment and migration.



A B

Figure 36. Total Condylar prosthesis (Case 27) with posterior subluxation of the medial tibial condyle giving a varus alignment of the leg.

Component positioning

Nineteen cases were outside ± 5 degrees, and two of these outside ± 10 degrees from the ideal PT angle, while 28 were outside ± 5 degrees, and five of these outside ± 10 degrees from the ideal PTS angle. No correlation could be found between prosthetic position and RSA results. The PHA angle did not influence the RSA results.

Discussion

Forces acting across the knee joint

The everyday forces across the knee joint have been investigated by Morrison (1970), who found compressive forces of up to 4 times the body weight (=approximately 3000 N) during normal walking. He also found a rotatory torque of about 10 Nm acting during the stance phase and a medio-lateral shear force of up to 25% of the body weight. In a theoretical analysis using force vectors Maquet et al. (1967) arrived at a force of 3720 N for a 60 kg person bending the knee 36 degrees.

In the knee with a prosthetic replacement, a bone-cement/prosthesis interface is created across which all forces must pass. The strength of this interface is a crucial factor for prosthetic loosening (Charnley 1965, Amstutz et al. 1972, Ducheyne et al.1978). In the absence of sufficient ligaments, as in rheumatoid arthritis, or when the cruciate ligaments are resected, as in most total knee replacements, this interface is subjected to larger and more unphysiological forces since ligament action must also be carried by the prosthesis. Even a slight 5° medial or lateral tilt of the tibial component gives additional shear forces of about 50% of the body weight (Kagan 1977), and eccentric loading of the prosthesis gives rise to tensile forces (Walker et al. 1976, Bartel et al. 1982, Bargren et al. 1983), which never occur in the normal knee.

The mechanical strength of the interface including the cancellous bone has been subjected to a number of laboratory *in-vitro* investigations (Kölbel et al. 1976, Walker et al. 1976, Bargren et al. 1978, Halawa et al. 1978, Walker et al. 1981, Lewis et al. 1982, Brookes et al. 1984). Bargren et al. (1978) found the compressive strength of a tibial-prosthesis preparation to be 3000–15000 N. They concluded that the strength is critical and advocated the use of large tibial components. The interface withstood a rotatory torque up to 28 Nm without disruption. Lewis et al. (1982) found the stress levels obtained by a load of 2000 N applied asymmetrically to be critical in tension and shear in an analytic study. Walker et al. (1981) applied 2250 N in compression and 23 Nm torque and found no disruption but measurable deflections of up to 0.2 mm. Although these investigations deal with the interface in the optimal, post-operative state, they confirm the clinical experience that the bone-prosthesis bond in the upper tibia does not have a large margin of safety.

Cancellous bone often reacts to replacement surgery using cement by the formation of a fibrous tissue membrane between cement and bone (Charnley 1970, Willert et al.

1974, Vernon-Roberts & Freeman 1976) Thus, the interdigitation of cement with bone trabeculae is lost and the strength of the bond is less than the above investigations indicate (Kaufer et al. 1978, Lewis et al. 1982). In a finite element model, Hori et al. (1982) have studied the influence of the fibrous tissue membrane and concluded that peak stresses were reduced but that displacements were increased by a factor of five. Comprehensive *in-vivo* investigation of the mechanical characteristics of prosthetic fixation on a large clinical material has not hitherto been presented.

Micromotion in cemented all-polyethylene prostheses

All cemented all-polyethylene tibial components in this material showed micromotion, regardless of whether the prosthesis was unicompartmental or total. The migration over time was identical in both groups, the major part occurring during the first 6 post-operative months.

There was no difference between medial and lateral unicompartmental prostheses. Thus, for example, the prostheses in both these types of arthroplasty translated symmetrically along the transverse axis in the peripheral direction.

A smaller number of prostheses in each group showed continuous migration. For both unicompartmental and total prostheses, the group showing continuous migration had migrated significantly more than the others already after six months. An indication can thus be obtained as early as half a year after the operation as to whether the prosthesis will migrate continuously or not.

All of the all-polyethylene prostheses, which were subjected to stress examinations, showed inducible displacement, corresponding to a deformation of the bone-cement interface of up to 1.0 mm. Within the resolution of RSA, these prostheses regained their original position after the stress series. The deformation thus appears to be mainly of an elastic nature.

These prostheses were all, except one, asymptomatic and represent conventional, widely used types of arthroplasties with excellent results reported in long-term follow-ups (Insall et al. 1983, Bauer et al. 1985, Marmor 1985). It is therefore concluded that micromotion, both migration and inducible displacement, is the rule in such arthroplasties and is quite compatible with successful clinical results.

The effect of metal support

Polyethylene is a rather soft material with a Young's modulus of elasticity in the same order of magnitude as that of cement and cancellous bone (Swanson 1977). A number of reports have been published on the deformation of plastic components (Ducheyne et al. 1978, Marmor 1979, Jónson 1981, Knutsson et al. 1981, Bartel et al. 1982, Scott 1982, Wright et al. 1982). Such deformation allows high peak forces to be transmitted to the underlying bone, possibly increasing the risk of loosening.

Both experimental (Bargren et al. 1978, Walker et al. 1981) and analytical studies using FEM methods (Bartel et al. 1982, Reilly et al. 1982) indicate that a metal tray supporting the polyethylene should place smaller compressive peak stresses on the bone. Moreover, the formation of a fibrous tissue layer in the bone-cement interface may be due to bone necrosis caused by the heat from the curing cement (Huiskes 1980, Mjöberg et al. 1984b). A metal component acts as a "heat sink" and lowers the interface temperature (Reckling & Dillon 1977, Huiskes 1980). For two reasons, therefore, metal-support might give a more secure implant fixation.

The metal-supported prostheses in this material did not migrate less, nor did they show less inducible displacement in response to external forces than the corresponding all-polyethylene prostheses. Indeed, the Kinematic components migrated more than the Total Condylar components (segment rotation). The hypothetical advantage of metal backing could not be verified in this clinical setting during a 2–3-year follow-up.

A number of both metal-supported and all-polyethylene components actually migrated upwards, that is, tilting of the prostheses made parts of them deflect upwards, which must have created tensile forces in the interfaces. Such tensile forces are augmented by metal-support (Walker et al. 1981, Bargren et al. 1983), a fact not emphasized. Therefore, it may be that the trade-off concerning metal support may be negative in some cases.

Among the cemented prostheses, the sensitivity to rotatory forces decreased progressively from the Total Condylar and Freeman-Samuelson prostheses through the Kinematic to the PCA prostheses. This finding can be explained by a corresponding decrease in constraint; the two former knees being of the roller-in-a-trough type while the PCA knee being virtually unconstrained in rotation (Hungerford et al. 1984). A tibial component with two condylar pegs should also, for mechanical reasons, be more resistant to rotation, as has been suggested by laboratory studies (Walker et al. 1981). In this respect the flanged pegs of the Freeman-Samuelson prosthesis do not appear to be as effective.

Plastic deformation, i.e. creep, or cold flow, is a physical characteristic of all polymers (Weightman 1977) and can give rise to potentially hazardous tensions in the interfaces. In the migration study, a progressive increase in the circumference of the Modular components was often observed and this finding was interpreted as representing such cold flow. No corresponding cold flow at the base of the polyethylene was observed in the Total Condylar series, which may be partly explained by the fact that these components were thicker. Another reason could be that in the Modular knee there is considerably less conformity between the femoral and tibial component. This gives rise to higher contact stresses in the polyethylene (Bartel et al.1984), which may explain the differences between unicompartmental and total prostheses.

It has previously been found (Marmor 1979, Knutsson et al. 1981) that the 6 mm prostheses, originally on the market, were too thin and easily deformable. The cold flow in this study was larger for 9 mm than for 12 mm prostheses, which may indicate that 9 mm prostheses may also be susceptible to plastic deformation and loosening in the very long term. These findings suggest that metal support of the polyethylene may be particularly advantageous in unicompartmental knee prostheses.

Cementless fixation

As mechanical loosening is the most common cause of failure in knee arthroplasty, it was natural to look for new ways of bonding prosthetic components to bone. Bone cement is mechanically brittle and weak in tension (Charnley 1970, Haas et al. 1975). It has also been suggested that it causes fibrous tissue formation because of heat from the curing process (Willert et al. 1974, Feith 1975, DiPisa et al. 1976) and monomer toxicity (Feith 1975). The macrophage response in the interface has also been attributed to the cement (Freeman et al. 1978, Freeman et al. 1982). Attempts have therefore been made to bond prostheses without cement, a development which has continued along two different lines.

Immediate interlocking fixation

The Freeman-Samuelson prosthesis seeks to achieve fixation to bone by immediate interlocking. When the prosthesis is inserted into the bone, the polyethylene flanges are bent and act as barbs preventing distraction of the component. The laboratory pull-out strength has been reported to equal that of an AO cancellous bone screw, and the design aimed to give a perfectly secure fixation in order to prevent polyethylene debris formation (Freeman et al. 1981).

The inducible displacement of the Freeman-Samuelson prosthesis was the largest found in this study, with motion between prosthesis and bone up to 5 mm, in Case 95 combined with a considerable distortion of the prosthesis during the stress series. Even though Case 95 may be a special case not representative of this type of fixation, considerable micromotion generally occurred in the stress examinations, simulating every day life, in immediate interlocking prostheses. This type of fixation does not give bony ingrowth, and a fibrous tissue interface is always present (Freeman et al. 1982). Additional inducible displacement could be anticipated because of bending of the polyethylene flanges. This magnitude of inducible displacement is more than was intended by the originators and is also not consistent with what they have reported in two revision cases; at reoperation after 10 and 36 months no motion could be detected and the strength of the fixation was such that the entire leg could be lifted by pulling one of the pegs (Blaha et al. 1982). The reason for this discrepancy is unclear. The forces produced across the knee in this investigation were, however, much greater and applied in other directions than those that could have been applied by the surgeon during the revision operations.

In the post-operative period, the flanged pegs cannot be seen on radiographs. With the passage of time, a sclerotic reaction, which can be staged in four degrees, occurs. It has been suggested that such a reaction represents a more stable and mechanically secure fixation in cemented prostheses (Tibrewall et al. 1984) and could also give a better resistance to micromotion for these prostheses. This is especially so if the diameter is small giving only a little fibrous tissue interposed between the bone and the prosthesis. This hypothesis was in part confirmed by the findings in this work. The sclerotic reaction is sometimes not complete until after two years and, with longer follow-up and a larger material, further support might be found.

This type of cementless fixation does not aim for bony ingrowth, and therefore rigid fixation is not a requirement post-operatively. However, the considerable inducible displacement found, ranging up to 5.0 mm, raises the possibility of polyethylene debris formation and fatigue failure.

Biological ingrowth

Bone has been shown to grow into a variety of porous materials such as bone cement (Ypma 1981, DeVijn 1982), polyethylene (Spector et al. 1975, Klawittel et al. 1976), titanium fibre mesh (Galante et al. 1971), stainless steel (Nilles et al. 1973, Ducheyne et al. 1977) and cobalt-chrome alloys (Cameron et al. 1972, Bobyn et al. 1980, Hedley et al. 1982). The PCA prosthesis in this study represents the first commercially available knee prosthesis with porous coating (Hungerford et al. 1982) and preliminary clinical results have been excellent (Hungerford & Kenna 1983). To achieve bony ingrowth, perfect stability in the post-operative period has been shown to be a prerequisite (Cameron et al. 1973, Ducheyne et al. 1977, Hungerford & Kenna 1983). In a finite element study, using multiple pyramidal projections from the prosthesis into cancellous bone, Lewis et al. (1985) found it possible to achieve such stability in the immediate post-operative period. In the PCA knee immediate fixation is obtained by the press-fit of two posteriorly inclined posts and a screw/molly-bolt.

The migratory pattern for the non-cemented PCA prostheses in this study was similar to that of cemented prostheses, the major part occurring during the initial six months. The magnitude of the migration was, however, larger than for cemented prostheses. The direction of the migration was also similar to the cemented prostheses, most commonly occurring as outward rotation about the vertical axis. However, all non-cemented prostheses subsided into the tibia to an extent not encountered in cemented prostheses.

Inducible displacement was significantly larger for non-cemented than for cemented PCA prostheses. The inducible displacement was also larger than can be explained by the intra-prosthetic instability, in which respect the cemented cases also represented a control group. With the possible exception of the non-cemented Cases 79, 80 and 90, where the inducible displacements could be explained by prosthetic instability, it has to be concluded that bony ingrowth did not occur.

In most of these cases there was a conspicuous radiolucent zone between the prostheses and the bone, a zone that made bony ingrowth improbable on plain radiographs. This finding is in contrast to what was reported by the originators, who found a radiolucent zone in only one out of 56 non-cemented cases. One reason for this discrepancy could be the different post-operative regimen with immediate weight bearing. In the two Cases 9 and 13, however, the interface did not show any trace of radiolucency, despite inducible displacement, indicating that plain radiography might not be an adequate way to assess bony ingrowth. Again, for these two cases, intra-prosthetic instability could possibly be the source of the inducible displacement.

The magnitude of the subsidence was such that it can only be explained by bone resorption, which probably results in micromotion. The molly-bolt loses its grip and as a result, a rocking motion occurs in the sagittal plane in response to extension-flexion exercises, as indicated by the shape of the zone in the lateral projection (Fig.33). The motion thus permitted is shown in stress studies to be about 0.7 mm, which is greater than the pore size and thus prevents ingrowth. When the bone necrosis is caused by heat, such bone has been shown to have reduced healing capability (Eriksson 1984). The preconditions for bony ingrowth may thus not have been met in these cases. Nevertheless, bony ingrowth *can* occur and during reoperation on two rheumatoid patients in our department, the femoral component was found to be soundly attached by ingrowth (Knutsson 1984). It is important to emphasize that 6 to 12 weeks of partial weight bearing, as proposed by the originators (Hungerford & Krackow 1985), was not a part of the post-operative regimen for these patients. Such immediate weight bearing has been suggested as a general cause of fibrous tissue interfaces (Albrektsson et al. 1981), and may be a major reason why micromotion occurred in these cases. However, post-operative physiotherapy, a necessity in joint arthroplasty, also gives considerable loads across the joint even in a non-weight-bearing situation (Morrison 1968, Reilly et al. 1972) as does manipulation under anaestesia. Whether no or only partial weight-bearing in the post operative period gives better fixation remains to be determined. In this respect the potential of the PCA concept has not been fully evaluated by this study.

In all cases but one plain radiography revealed loose beads from the porous coating within the radiolucent zone or in the uppermost cancellous bone region. Some of these beads were loosened by the per-operative handling of the prostheses. Few were, however, seen on the reference stereo radiographs (Fig.19) which in most cases were taken in a plane almost perfectly tangential with the metal base. Some beads must therefore have become loose post-operatively, which is not surprising considering the grating effects of the movements between porous coating and bone. With bony ingrowth, resulting in a bone-metal composite in the porous coating region, the individual beads should be fixed in their position and loosening and migration of individual beads should not occur. It is therefore concluded that loose beads probably indirectly indicate that bone ingrowth has not occured.

In summary all tibial components in this investigation of mostly asymptomatic patients showed micromotion, both as migration and inducible displacement. The newer concepts of metal-support and non-cement fixation do not appear to offer a more rigid fixation.

The site for micromotion

The movements studied in this work could theoretically have occurred anywhere within the sandwich structure, the top and bottom of which were defined by the markers in the prosthesis and the tibia, respectively, i.e. within the polyethylene, cement or bone, or the interfaces between these materials.

Regarding inducible displacement, the FEM analysis indicated that the total material elasticity in the upper tibia sandwich could not account for any appreciable amount of motion in response to the forces used in this work. This statement is further supported by laboratory investigations. Thus Walker et al. (1981) found micromotion up to 0.2 mm using axial loads of three times body weight and a rotatory torque of 23 Nm. Brooks et al. (1984) found maximum deflections of 0.09 mm measured over 10 mm bone substance, using a force of 1780 N in a study on the significance of different designs in the presence of deficient bone stock. Thus, the inducible displacement occurred in the interfaces. The integrity of the prosthesis-cement interface was not particularly controlled in this study. However, there were no signs of cement fracture, which is probably a prerequisite for motion of this magnitude to occur. Moreover, if there were fractures, the displacement would probably be of a more plastic nature. Finally, the exclusion of the prosthesis-cement interface did not decrease, but rather increased, the inducible displacement. It is therefore concluded that the inducible instability found in this work occurred within the soft tissue of the *bone-cement/prosthesis* interface.

The mechanical properties of interface tissue from dogs have been examined. The maximum compressibility was found to be approximately 50% (Hori & Lewis 1982). The findings in this work corresponded well to that value; the mean maximum deformation during the stress examinations being about 1/3 of the mean maximum zone width.

The migration cannot have occurred within the polyethylene, cement or the interface between these materials because the migration was at least as large and had a similar time-pattern for metal-supported prostheses and for prostheses without cement. The migration could possibly have occurred as gross remodelling of the proximal tibia, but the fact that bone markers placed close to the interface remained stable relative to the other bone markers does not support this theory. The migration correlated with the size of the bone-cement interface measured as a radiolucent zone for the Total Condylar prostheses. It is therefore suggested that the migration occurs in the interface between bone and prosthesis/cement; either by bone resorption, by shearing or by distraction caused by tensile forces.

Radiographic assessment of loosening

Apart from clinical examination, radiography offers a major way to assess joint replacements, and various radiographic findings are considered as signs of loosening.

Migration, i.e. a change of the position of the prosthesis on repeated radiographs, is considered by some to be a definite sign of loosening (Kaufer et al. 1981, Schneider et al. 1982, Boegård et al. 1984, Harris 1984). Tilting of components in both frontal and lateral projections, and subsidence are detectable with an accuracy of ± 3 degrees and ± 2 mm, respectively (Blaha et al. 1982, Ryd et al. 1983) It is, however, difficult to assess subsidence because of the lack of adequate reference marks (Schneider et al. 1982, Cameron & Hunter 1982). Cameron & McNeice (1981) assessed the radiographs from 39 revision cases and found five different failure modes: tilt and sink, compression, torsion, toggle and combination.

This investigation shows that migration is a common occurrence in knee arthroplasty and, since there was no correlation between migration and inducible displacement, a large migration, also detectable by radiography, does not seem to indicate an inferior fixation. However, when migration is detectable by radiography, it is probably continuous, since only continuously migrating prostheses reached the necessary magnitude of migration.

The development of a radiolucent zone in the bone-cement interface is a common finding (Ahlberg & Lindén 1977, Ducheyne et al. 1978, Goldberg & Henderson 1980, Kaufer & Mattews 1981, Insall et al. 1983, Eftekhar 1983, Ewald et al. 1984, Tiberwall et al. 1984). It has been attributed to bone necrosis and subsequent resorption because of the per-operative trauma (Charnley 1970, Willert et al. 1974), to the inclusion of blood and bone debris in the interface (Miller et al. 1976, Ducheyne et al. 1978), to mechanical factors resulting in micromotion and fatigue fractures of the subchondral bone (Vernon-Roberts & Freeman 1977, Ducheyne et al. 1978) or to enzymatic activity in response to micromotion (Goldring et al. 1983). Ducheyne et al. (1978) reported one case where a tibial component varus tilt was not caused by medial subsidence but by the upward migration of the lateral plateau. That finding is confirmed in this work; a number of zones were voids created by the prosthesis withdrawing from the bone. Thus, altogether 11 Total Condylar and Kinematic condyles migrated upwards and all Modular prostheses migrated peripherally, leaving a void behind which was recognized as a radiolucent zone on radiographs. To arrive at an accurate estimate of the amount of bone resorption that has occurred, the subsidence should be added to the width of the radiolucent zone while the amount of upward migration should be subtracted. Most of the cases with cemented prostheses developed a zone to some extent, but the size of this zone did not correlate with the degree of inducible displacement present. Radiography relies on very precise angles of exposure to obtain an accurate and reproducible view of the bone-cement interface. The shape of the bone-cement interface is unique for every arthroplasty, because of the uncontrollable penetration of cement into the cancellous bone, and even the slightest obliquity of the radiographic rays relative to the interface can obscure even a conspicuous zone (Insall 1984, Tibrewall et al. 1984). Finally it must be emphasized that radiography can give no information regarding the mechanical properties of the interface, where liquid is as radiolucent as fibrocartilage. For these reasons the absolute size of the radiolucent zone, even if it is wide or covers the whole interface, may have little correlation with the amount of fibrous tissue between bone and cement and none at all with the mechanical strength of the interface.

A progressively enlarging radiolucent zone conceptually indicates a progressive biolgical process. Such zones also correlated with the migration for the Total Condylar prostheses in this material and thus appear to better reflect the mechanical situation in the interface.

Importance of the position of the prosthesis

Malalignment and component tilt (Lotke & Ecker 1977, Freeman et al. 1978, Coventry 1979, Rand & Bryan 1982, Bargren et al. 1983, Insall et al. 1983) and lack of total tibial component bone coverage, including the cortical rim (Bargren et al. 1978, Freeman et al. 1978, Cameron & Hunter 1982) are regarded as factors contributing to loosening.

A malaligned knee carries a higher load on the part of the prosthesis on the concave side, which could be expected to subside more than the contralateral side. However, in this study, there was no correlation between the alignment and the migration. One reason for this could be that none of these knees was grossly malaligned and few were outside the acceptable range defined as ± 5° (Insall et al. 1983) or ± 6° (Bargren et al. 1983). These knees thus represent a range of alignment which appears to have little influence on migration.

Lack of cortical bone support did not give larger migration in this study. It has recently been shown that coverage of the peripheral rim of the uppermost tibia does not give an increase of the total strength (Figgie et al. 1984, Hvid et al. 1985). This suggests that the cortical shell is unimportant regarding load transfer from a prosthetic component, which is confirmed by the findings in this study.

Rotatory malignment of the tibial component has been given little attention. Bargren (1980) has reported a case of luxation due to a 30° inwards rotation of the tibial component, and Cameron & Hunter (1982) reported three such cases in a series of 94 failed knee arthroplasties. An outward rotation about the vertical axis was the most consistent migratory movement of total prostheses. A tibial component placed in inward rotation would create a force acting in the opposite direction. The per-operative eversion of the patella causes an outward rotation of the tibia during the insertion of the tibial component. This fact may cause a consistent error in the rotatory position of the tibial component, which could explain the migratory pattern (Fig. 36).

The cancellous bone of the upper tibia has been reported as being weaker distally (Bargren et al. 1978, Sneppen et al. 1981). This was not confirmed, however, by Hvid et al. (1983). A thicker component is equivalent to a larger bone resection but there was no correlation between the thickness of the prostheses and micromotion. It has been suggested that the cancellous bone is weaker on the convex, mostly the lateral, side of the knee due to disuse atrophy (Hungerford & Krackow 1985). The degree of varus or valgus deformity pre-operatively, however, did not influence the micromotion. Differences in cancellous bone strength are therefore probably not important for the development of micromotion.

Implications of micromotion for prosthetic fixation

Apart from reports on micromotion in knee arthroplasty from the Orthopedic Department in Lund (Ryd et al. 1983) there has been only one previous report on micromotion using a comparable technique (Green et al. 1983). In that work 19 knee prostheses, 14 of which were of the Total Condylar type, were followed for 2 years. Migration (MTPM) occurred in 14 cases ranging up to 5.8 mm and "reversible displacement" (inducible displacement) up to 6.7 mm was found in all cases. The frequency of micromotion in that study was consistent with the results in this work although the magnitude was greater. The reason for this might be that 12 of their cases were symptomatic and three were revised for loosening. A number of their prostheses were therefore grossly loose and larger micromotion can be anticipated for such prostheses.

It is generally believed that absolutely rigid fixation is to be preferred. Thus, a number of authors suggest that micromovements in the interface give stress concentrations and fatigue fractures of the cancellous bone leading to loosening (Amstutz et al. 1972, Reckling et al. 1977, Ducheyne et al. 1978, Lintner et al. 1982, Radin et al. 1982, Lewis et al. 1985). Absolute rigidity may be difficult to achieve in a mechanical system like knee arthroplasty, involving materials with different physical characteristics. Discrepanicies in Young's moduli and Poisson's ratios always give rise to stresses and may cause movements in the interface ranging up to 10 μm (Charnley 1965, Kaufer et al. 1978, O'Connor et al. 1982, Radin et al. 1982). Shrinkage of the cement by 4–5% occurs after the curing process (Ohnsorge & Grötz 1974, Miller et al. 1978) and can give movements in the range of 0.1 mm. As stated above, histological studies have shown that a fibrous tissue membrane often forms between cement and bone. Willert et al. (1974) found minute but "not significant" movements in this fibrous tissue, and Kaufer et al. (1978), in a post-mortem laboratory investigation, have reported movements up to 0.1 mm for a tibial component from a previously asymptomatic patient. Other authors have suggested the possibility of motion occurring (Reckling et al. 1977, Freeman et al. 1982, Goldring et al. 1983, Haberman 1984). Therefore, from a technical point of view, prosthetic components, both cemented and uncemented, may seldom be absolutely rigidly bonded to the bone.

The radiographically detected change of position of the prosthesis and radiolucent zones represent other aspects of such technical loosening, and patients free from symptoms are said to suffer from "asymptomatic loosening" (Rand & Bryan 1982, Schneider et al. 1982), "stable loosening" (Convery et al. 1980) or "biomechanical loosening" (Larsson & Ahlgren 1979). Kaufer et al. (1978) have staged such "loosening" as simple, symptomatic, progressive and functional. This rather confusing situation can be clarified by stating that all cases in this study suffered from asymptomatic loosening since all prostheses migrated and all but three prostheses, subjected to a stress examination, showed inducible displacement. The important point is that most of these arthroplasties were successful and that excellent long-term results have been reported with the systems used (Marmor 1985, Insall et al. 1983, Ewald et al. 1984, Bauer et al. 1985). This degree of looseness represents the normal state of tibial components, which are functionally stable. Indeed the fibrous tissue membrane may have a beneficial effect by dissipating stresses caused by different material properties (Charnley 1965, Connor et al. 1982, Hori & Lewis 1983, Tibrewall et al. 1984) and such non-rigidity is a rather universal rule in nature, for example, trees bending with the wind instead of breaking. Whether newer cementing techniques with lavage (Halawa et al. 1978), low viscosity cement and cement pressurization (Miller et al. 1980) will give a better fixation from a technical point of view remains to be determined.

The migration over time was identical for all types of prostheses in this study, the major part occurring during the first 6 post-operative months. The migration did not start until after the 1-week stereo examination in most cases. This is in perfect agreement with the healing process of the bone of the implant bed; a healing process, which starts about three weeks post-operatively and continues for 1–2 years. (Willert et al. 1974). The subjacent bone necrosis has been attributed to mechanical trauma (Rhinelander et al. 1979), cement monomer toxicity (Willert et al. 1974) or heat from the curing cement (Charnley & Crawford 1968, Willert et al. 1974, Feith 1975, Huiskes 1980, Mjöberg et al. 1984). It has been shown that high temperatures occur while drilling cortical bone in man (Eriksson 1984). Knee arthroplasty does not include cutting of cortical bone, but the subchondral bone is often extremely hard in arthrotic cases and significant temperatures are probably reached during cutting. Since, the initial migration was the same or larger for non-cemented than cemented prostheses, it is suggested that the bone necrosis may be caused primarily by the per-operative mechanical trauma and not by curing cement in resurfacing arthroplasty of the upper tibia.

After the initial six months, a minority of the prostheses showed continuous migration, which must indicate continuous bone resorption occurring after the initial bone necrosis has healed. Such resorption may be caused by micromotion causing stress fractures of the bone as previously suggested (Amstutz et al. 1972, Ducheyne et al. 1978). However, the cancellous bone at this stage is not necrotic and should be able to accumulate load cycles without fatigue failure. The majority of the Total Condylar tibial components rotated outwards. Due to the rotatory constraint of the Total Condylar knee, a per-operatively caused rotatory mis-match between the prosthetic components and the soft tissue around the knee, perhaps augmented by the action of the popliteus tendon, may give rise to a continuous rotatory force acting on the tibial component. In orthodontology such small, but continuous, forces can move teeth over long distances. Histo-chemically such forces have been shown to cause an intense macrophage response with enzyme activity in the periodontal membrane (Lilja 1983) closely resembling findings reported for the tissue in the bone-cement interface (Goldring et al. 1983). Macrophages, abundant in the interface (Charnley 1970, Freeman et al. 1982), are the precursors of osteoclasts (Göthlin & Ericsson 1973, Chambers 1980). It is therefore suggested that incompatibilities between the prosthetic components and the soft tissue create continuous forces sufficient to initiate a macrophage-osteoclast response with bone resorption, progressing zone formation and migration of the prosthesis, in much the same way as a tooth is moved within the jaw. During this process, the prosthesis, like the tooth, is "technically" loose as indicated by the stress investigation. However, fuctionally, both the tooth and the prosthesis are stable. A continuously migrating prosthesis may, because of this biological process, be called "biologically loose".

If this migration causes an increasing varus alignment, for example, it is deleterious and might lead to loosening. Indeed, Hvid et al. (1983) have found a correlation between the zone formation and the post-operative alignment for rheumatoid but not for arthrotic cases. Rotation of the tibial component does not, however, cause a deteriorating mechanical situation but alleviates the rotatory tension; such continuous migration might therefore not have any prognostic significance.

A number of the non-cemented PCA components and all Freeman- Samuelson components showed a change of position before and after the stress series suggesting different, more plastic, mechanical properties of the non-cemented fibrous tissue interface. This finding may represent a fundamental difference between cemented and non-cemented interfaces. One cemented prosthesis also showed such a plastic response to the stress investigation. Modular Case 32 was intermittently symptomatic in the period before the stress examination, which did not reveal a larger inducible displacement than other cemented prostheses. She is now scheduled for revison surgery for suspected loosening. Perhaps, the prosthesis was loose during the stress examination and the change of position between unloaded exposures (plastic response) may be a more sensitive indicator of functional loosening than the magnitude of inducible displacement.

Prognostic significance of micromotion

Most arthroplasties in this material were successful and therefore no conclusions can be drawn as to what pattern or magnitude of micromotion may be ominous in longer follow-ups. A number of prostheses, however, showed continuous migration, which probably indicates a less stable interface. Perhaps future failures will be found in this group, although it is emphasized that prostheses in this group were not looser, i.e. did not have larger inducible displacement, than in the group without continuous migration within the time spanned by this work. The only patient with clinical, or "functional", loosening in this material (Modular case 32) belonged to the continuously migrating group.

Fixation by a fibrous tissue membrane was the case for most of the prostheses disregarding the type, and it is concluded that such fixation represents the normal state of the tibial component in knee arthroplasty. This fibrous tissue membrane has only a certain mechanical strength which can be exceeded during a fall or heavy physical exercise, thereby causing rupture. Willert et al. (1976) reported tears and cleft formation within the fibrous tissue in clinically loose femoral components in hip arthroplasty, and such clefts can be visualized by arthrography (Gelman 1978, Hendrix & Anderson 1981). The technically or biologically loose prosthesis then also becomes functionally loose, and symptoms, necessitating revision, may arise.

Hypothetically, fixation without a fibrous tissue membrane may be preferred, but the question is whether such fixation can be achieved without too much inconvenience to the patient. Older patients may not be able to tolerate staged procedures and rheumatoid patients in particular may have difficulties with partial weight-bearing. The optimum solution might instead lie in further refinement of established principles like cementing techniques, precise surgery and patient-prosthesis matching.

Conclusions

1. Roentgen stereophotogrammetric analysis (RSA) offers a sensible *in-vivo* method of assessing the mechanical integrity of prothestic fixation; it may detect differences between fixation concepts as early as 6–12 months postoperatively.
2. *Technically* all tibial components were loose, showing micromotion, both as migration and/or inducible displacement. However, only a minority of the cases showed continuous migration; they were *biologically* loose. Most of the knees showed satisfactory clinical results; they were *functionally* stable loose. This mechanical situation represents the normal state of tibial components in knee arthroplasty.
3. Metal support of the polyethylene did not decrease the micromotion, but, in unicompartmental prostheses, it prevented cold flow of the polyethylene, which may have a bearing on the fixation in the longer term.
4. The immediate interlocking type of fixation displayed the largest inducible displacement of all types of fixation.
5. After full immediate weight-bearing, the biologic ingrowth type of fixation displayed micromotion, both as migration and inducible displacement. Stable bony ingrowth was ruled out in most cases.
6. The migration occurred predominantly as a result of resorption of the bone subjacent to the prosthesis/cement. The inducible displacement occurred within the soft tissue of the bone-cement/prosthesis interface.
7. A radiolucent zone which is progressive and has a large extension is an indicator of migration. The width of the zone is less relevant in this respect.
8. Radiolucent zones may be of both tensile and compressive origin.
9. Alignment of the operated limb, prosthetic position and cortical bone support did not correlate with the micromotion present.

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Coded data

Prosthesis		Total Condylar													
Case		1	2	3	4	5	6	7	8	9	10	11	12	13	14
Age		83	76	75	84	72	65	75	84	72	72	70	72	63	72
Sex		F	F	M	F	M	F	F	M	F	F	F	F	F	F
Weight		52	93	98	64	79	72	66	75	65	69	70	47	69	71
Follow-up (months)		51	36	23	48	48	48	48	36	35	35	48	38	38	37
Lund score		S	S	F	S	S	S	S	A	S	S	S	S	S	S
HSS score		92	92	59	88	94	94	96	66	73	93	87	95	89	86
PT		82	88	86	81	88	87	90	91	92	85	85	86	87	89
PTS		88	88	93	90	88	86	94	99	99	93	94	95	99	98
postHKA		170	182	171	177	178	184	174	185	181	173	175	173	178	185
preHKA		170	173	161	170	174	187	178	205	205	—	166	150	186	205
Zone width (mm)		4.6	0	—	0.7	2.2	2.8	1.0	0.9	0.4	0.6	1.3	2.3	1.7	1.5
Zone extension (%)		54	0	22	17	65	50	18	13	21	21	63	29	60	45
Zone value		23	0	—	3	18	18	8	4	3	4	14	15	13	14
Component height (mm)		10	20	10	10	10	20	15	12	15	10	10	15	25	20
Cortical support[1]	med.	0	0	16	2	6	4	7	0	1	2	5	2	0	2
	lat.	0	5	5	1	5	6	2	7	2	2	2	0	0	3
	post.	0	5	10	6	2	1	0	10	0	0	2	3	0	0
	ant.	2	8	15	0	3	4	11	5	3	3	0	5	5	1
Migration															
MTPM	1w	—	—	—	—	—	—	—	—	—	—	—	—	—	—
	6w	0.5	0.5	0.5	—	—	0.3	0.4	—	—	—	—	0.2	—	—
	6m	—	—	—	—	—	1.5	0.6	0.2	—	—	—	0.2	1.6	0.3
	1y	—	—	—	0.6	—	1.8	0.6	0.3	0.4	0.5	0.6	0.3	—	0.5
	2y	1.0	0.5	1.0	0.7	1.1	2.1	0.8	0.6	0.5	0.7	0.7	0.3	2.0	0.7
	3y	1.0	0.6	—	1.0	1.1	2.5	0.9	0.8	0.4	0.7	0.9	0.3	2.0	0.6
	4y	1.1	—	—	1.6	1.0	2.5	0.7	—	—	0.7	—	—	—	—
Migration 2 years	X	0	+0.7	−0.4	−0.4	−1.0	+0.5	+0.6	0	+0.7	+0.5	0	+0.5	−0.6	+1.4
Rotation	Y	+0.6	+0.4	−0.3	+0.9	+1.1	+3.6	+0.5	0	+0.5	+1.0	+1.0	+0.3	+0.9	0
	Z	+1.2	−0.3	+0.4	+0.3	−0.4	0	−0.6	−0.8	+0.7	−0.2	0	+0.3	+2.1	0
Translation	X	−0.3	0	+0.2	−0.2	+0.4	+0.2	+0.3	0	−0.2	+0.4	0	−0.2	−0.4	0
	Y	−0.2	0	0	−0.3	+0.3	−0.6	0	0.2	0	0	+0.3	0	−1.1	−0.3
	Z	0	+0.4	−0.6	0	−0.3	−0.5	0	0	+0.2	+0.4	0	0	0	+0.2
Stress															
weight-bearing[2]		1.2	—	—	0.7	0.6	0	0.8	0.5	0	0.4	0	0	0.5	0.4
rotational		1.4	—	—	0.7	0.5	0.4	0.3	0	0.9	0.6	0.8	0.6	1.2	0
max. segm. rotation		1.8	—	—	0.8	0.7	0.5	0.8	0.6	1.0	0.6	0.8	0.6	1.3	0.5
MTPM		1.0	—	—	0.4	0.5	0.3	0.5	0.5	0.6	0.4	0.5	0.3	0.6	0.2
weight-bearing[3]		—	—	—	—	0.8	—	—	—	—	—	0	—	0.8	0
rotational		—	—	—	—	0.4	—	—	—	—	—	0.6	—	0.3	0.3
max. segm. rotation		—	—	—	—	0.8	—	—	—	—	—	0.6	—	0.9	0.6
MTPM		—	—	—	—	0.4	—	—	—	—	—	0.3	—	0.5	0.2

[1] amount of bone left uncovered (mm)
[2] primary stress series
[3] secondary stress series

Prosthesis		Total Condylar												
Case		15	16	17	18	19	20	21	22	23	24	25	26	27
Age		58	72	70	64	83	73	68	70	70	70	56	68	72
Sex		M	F	F	F	F	F	F	F	F	F	M	F	F
Weight		94	90	71	86	67	70	53	63	81	82	89	85	58
Follow-up		35	35	37	34	24	20	25	25	24	26	25	24	24
Lund score		S	S	A	A	S	S	F	S	A	S	S	S	S
HSS score		94	90	82	79	88	91	73	93	79	72	88	76	92
PT		87	92	88	91	87	89	88	90	81	88	82	89	96
PTS		87	98	98	94	94	94	102	104	96	94	93	81	97
postHKA		175	193	190	190	180	176	182	180	174	173	176	186	187
preHKA		—	200	157	170	160	185	170	181	170	—	180	180	165
Zone width		0.8	2.3	2.1	1.2	0.5	1.0	0.9	0.8	0.6	0.6	1.3	2.1	0.9
Zone extension		36	24	90	27	11	13	16	67	23	12	29	64	24
Zone value		5	9	21	11	5	6	7	15	6	4	11	11	7
Component height		20	10	15	20	10	15	15	10	15	15	10	15	15
Cortical support	med.	3	4	2	3	10	6	2	2	4	4	9	0	5
	lat.	10	4	0	4	1	0	2	0	1	5	2	4	1
	post.	0	0	0	0	5	2	0	3	0	0	8	0	0
	ant.	5	8	5	4	8	0	4	7	4	2	9	3	2
Migration														
MTPM	1w	—	—	—	—	—	—	—	—	—	—	—	—	—
	6w	0	—	0.5	0.4	0.2	0.2	0.4	0	0.2	0.2	0.2	0.3	0
	6m	0.2	—	0.8	0.2	0.2	0.7	0.4	0.6	0.3	0.4	0.2	1.0	0
	1y	0.3	0.3	0.9	0.3	0.2	0.8	0.4	0.7	0.4	0.4	0.2	1.2	—
	2y	0.4	0.4	1.1	0.2	0.3	0.8	0.4	0.8	0.2	0.5	0.3	0.9	0.2
	3y	0.5	0.5	1.8	0.2	—	—	—	—	—	—	—	—	—
	4y	—	—	—	—	—	—	—	—	—	—	—	—	—
Migration 2 years	X	−0.3	0	−0.9	−0.4	0	0	+0.3	−1.4	0	+0.6	0	−1.3	0
Rotation	Y	0	+0.5	+1.1	0	+0.4	+0.3	0	−0.3	0	0	+0.6	0	0
	Z	0	+0.3	0	0	0	0	−0.4	0	0	−0.3	0	+1.1	0
Translation	X	0	0	+0.4	0	0	0	+0.2	−0.2	+0.2	0	0	−0.3	0
	Y	−0.2	0	−0.4	0	0	0	0	0	0	+0.2	0	−0.3	0
	Z	0	0	−0.2	0	0	0	+0.2	−0.6	0	+0.3	0	−0.4	0
Stress														
weight-bearing[1]		0	0.5	0.5	0	0.5	0	0	0	0.9	0	0	0	0
rotational		0.5	0.3	0.7	0.3	0	0.4	0.4	0.4	0.4	0.3	0.3	0	0
max. segm. rotation		0.6	0.6	0.9	0.4	0.7	0.4	0.4	0.7	1.0	0.5	0.4	0.6	0.6
MTPM		0.4	0.2	0.3	0.3	0.4	0.5	0.3	0.7	0.5	0.3	0.2	0.3	0.3
weight-bearing[2]		0	—	—	0.4	—	—	—	—	0.3	—	0.2	—	—
rotational		0.3	—	—	0	—	—	—	—	0.3	—	0.3	—	—
max. segm. rotation		0.4	—	—	0.4	—	—	—	—	0.7	—	0.3	—	—
MTPM		0.2	—	—	0.3	—	—	—	—	0.3	—	0.2	—	—

[1] primary stress series
[2] secondary stress series

Prosthesis		Modular											
Case		28	29	30	31	32	33	34	35	36	37	38	39
Age		65	64	72	66	64	70	69	68	65	70	75	75
Sex/compartment		F/m	F/m	F/l	F/m	F/m	F/m	F/m	F/m	F/m	M/m	F/m	F/l
Weight		72	99	65	88	70	70	97	63	84	77	65	65
Follow-up		65	60	62	61	62	66	49	46	50	48	48	48
Lund score		S	S	S	S	F	S	S	S	S	S	F	F
HSS score		84	91	88	93	76	93	81	96	93	90	58	59
PT		83	86	86	87	78	89	85	89	83	83	94	88
PTS		91	86	83	81	86	90	95	89	89	83	96	87
postHKA		177	182	181	186	173	173	169	178	186	174	156	196
PHA		85	88	86	87	82	92	90	90	79	87	107	80
Zone width		1.3	1.1	0.4	1.7	1.0	1.0	0.4	0	1.0	0.7	1.4	0
Zone extension		90	80	12	50	100	45	25	0	50	33	13	0
Component height		9	12	9	12	12	15	9	12	12	9	12	9
Cold flow (%)		1.5	0.5	0	1.4	1.2	0	2.8	0	0	1.1	0	0
Cortical support													
periphery		0	0	0	0	0	1	0	0	2	0	0	0
posterior		0	2	0	2	8	7	0	3	0	4	0	0
Migration													
MTPM	1w	—	—	—	—	—	—	—	—	—	—	—	—
	6w	0.4	—	—	0.9	—	—	—	—	—	—	—	—
	6m	—	0.8	0.6	1.6	1.8	—	0.5		—	—	—	—
	1y	0.3	1.2	0.6	2.4	4.6	—	—	—	0.5	0.5	0.5	—
	2y	0.5	1.1	0.7	3.0	5.0	—	1.2	0.3	0.8	0.6	0.7	0
	3y	0.5	1.0	0.6	—	5.6	0.7	1.2	0.4	0.6	0.7	0.8	0.2
	4y	0.6	0.9	0.6	3.8	5.6	0.8	1.3	—	0.7	0.7	0.8	0.3
	5y	0.6	1.0	0.6	5.1	5.4	0.8	—	—	—	—	—	—
Migration 2 years	X	−0.5	−0.5	−1.5	−0.6	+1.3	−0.5*	−1.5	0	−0.9	−1.4	+1.2	0
Rotation	Y	−0.4	+1.3	0	+4.1	+2.3	+0.4	+1.2	+0.4	+0.8	+0.5	−0.4	+0.3
	Z	−0.6	−0.8	0	+2.4	−12.6	−1.4	−1.2	0	−2.0	0	+1.0	−0.3
Translation	X	0	0	+0.2	+1.4	+1.0	+0.5	+0.3	+0.2	+0.5	0	+0.2	0
	Y	0	+0.7	0	−0.9	−2.6	+0.2	+0.3	0	−0.2	−0.2	0	0
	Z	−0.3	−0.3	−0.3	−0.5	0	0	0	0	0	0	+0.3	0
Stress													
weight-bearing		—	—	—	—	0.6	—	0.4	—	—	—	—	—
rotational		—	—	—	—	0.3	—	0.7	—	—	—	—	—
max. segm. rotation		—	—	—	—	0.7	—	0.7	—	—	—	—	—
MTPM		—	—	—	—	—	—	0.4	—	—	—	—	—

* Modular Case 33, values of 3 years.

Prosthesis		**Modular**											
Case		40	41	42	43	44	45	46	47	48	49	50	51
Age		62	65	70	68	64	75	73	65	74	66	72	70
Sex/compartment		F/m	M/m	F/m	F/m	F/m	F/m	F/m	F/m	F/m	F/m	F/l	F/m
Weight		74	93	70	94	68	70	83	83	62	82	81	69
Follow-up		37	35	38	36	36	36	34	41	41	37	40	40
Lund score		F	S	S	S	S	S	S	S	S	S	S	
HSS score		80	65	94	95	98	78	93	91	95	77	87	87
PT		85	92	83	92	81	88	92	75	85	87	84	90
PTS		81	86	85	76	91	82	85	86	88	93	89	94
post HKA		178	180	177	178	172	181	183	173	181	178	178	180
PHA		—	82	82	94	85	87	90	79	84	88	86	90
zone width		1.0	1.0	0.5	0.9	1.3	0.5	0.3	0.7	1.0	0.9	0.6	0.5
Zone extension		50	75	30	65	78	10	10	40	40	30	40	75
Component height		9	12	9	12	9	12	9	12	9	9	9	9
Cold flow (%)		0	1.1	0.9	0.8	1.9	0.4	0	0.7	1.3	2.0	0	0.4
Cortical support													
periphery		0	0	0	0	0	0	0	0	0	0	0	0
posterior		1	0	3	10	2	4	0	0	0	5	0	10
Migration													
MTPM	1w	—	—	—	—	—	—	—	—	—	—	—	—
	6w	0.4	0	0.3	0.5	0.4	0	0.2	0.2	0	0.3	0.3	0
	6m	0.6	1.8	0.5	0.9	1.0	0.3	0.5	0.4	0.4	0.7	0.3	0.3
	1y	0.6	2.7	0.5	1.2	1.4	0.3	0.5	0.6	0.4	0.7	0.3	0.2
	2y	0.5	3.4	0.4	1.5	1.6	0.3	0.6	0.6	0.5	1.0	0.4	0.3
	3y	0.6	3.5	0.3	1.7	1.6	0.4	0.6	0.7	0.5	1.2	0.6	0.3
	4y	—	—	—	—	—	—	—	—	—	—	—	—
Migration 2 years	X	0	+4.9	0	+0.9	+3.8	0	0	−0.7	0	+1.1	0	+0.3
Rotation	Y	0	−2.9	+0.8	+1.3	−0.6	0	0.7	0	0	+0.3	−0.7	0
	Z	−0.8	+3.2	0	−1.4	−1.3	0	−0.8	−1.2	+0.3	0	0	−0.3
Translation	X	+0.4	0	+0.2	+0.7	+0.2	0	+0.3	+0.2	+0.2	+0.3	0	0
	Y	0	−1.6	0	−0.3	−0.7	0	−0.2	−0.2	−0.2	−0.6	+0.2	0
	Z	+0.2	−0.2	0	+0.8	+0.2	+0.2	+0.2	0	+0.2	+0.3	+0.2	0
Stress													
Weight-bearing		—	—	—	0.3	—	0.3	0	0.3	—	0.4	—	—
rotational		—	—	—	0.5	—	0	0	0.7	—	0.4	—	—
max. segm. rotation		—	—	—	0.6	—	0.7	0.7	0.7	—	0.6	—	—
MTPM		—	—	—	0.2	—	0.3	0.2	0.2	—	0.3	—	—

Prosthesis	Kinematic							PCA cement						
Case	52	53	54	55	56	57	58	59	60	61	62	63	64	65
Age	80	78	76	76	77	69	73	68	62	62	69	75	82	79
Sex	F	F	F	F	F	F	M	F	M	M	F	F	F	F
Weight	50	68	70	70	67	83	78	84	106	106	122	71	78	56
Lund score	S	S	S	A	S	S	S	S	A	S	A	F	S	S
HSS score	86	90	89	83	85	82	89	88	82	90	86	56	94	93
PT	91	88	90	92	88	87	88	86	85	80	86	86	83	88
PTS	91	85	87	91	91	95	94	90	85	88	84	86	92	94
postHKA	179	179	180	175	174	177	177	176	174	171	183	177	176	178
preHKA	156	161	165	172	161	176	157	167	165	157	199	175	198	188
PHA	92	88	90	94	91	89	89	88	89	85	85	87	85	89
Zone width (8)[1]	0.8	0	0.4	2.8	1.6	0.4	1.4	2	3	1	0	1	0	0
Zone extension (9)	68	0	6	88	15	17	100	2	2	0	0	0	0	0
Zone value (10)	11	0	4	16	6	3	16	2	2	0	0	0	0	0
(11)								3	2	1	0	0	0	2
(15)								2	2	3	1	1	1	1
(16)								3	2	2	0	0	2	0
(17)								2	2	3	0	0	0	0
(18)								3	3	2	0	0	1	2
(19)								2	3	3	1	1	1	0
Component height	18	12	9	12	9	9	18	7	9	11	9	9	15	15
Cortical support med.	0	0	0	0	1	4	1	2	4	0	0	0	2	0
lat.	0	0	1	0	3	0	9	3	9	10	0	0	0	3
post.	8	0	0	0	7	0	0	0	0	0	0	2	0	0
ant.	6	6	0	5	1	7	11	10	5	7	4	6	0	6
Migration														
1w	—	—	—	—	—	—	—	0.4	0.2	0.2	0.2	0.2	0.3	0
6w	—	0.3	0.2	0.2	0.9	0.2	0.5	1.5	0.3	0.3	0.2	—	0.5	0.2
6m	0.8	0.7	—	0	1.0	0.4	2.0	1.8	0.9	1.5	0.3	0.5	0.8	0.2
1y	0.9	0.7	0.3	0.4	1.5	0.5	2.0	1.7	0.7	1.4	0.3	0.3	1.0	0.2
2y	1.4	1.1	0.3	0.8	1.6	0.4	1.9	1.6	0.7	1.6	0.4	0.3	1.3	0.2
Migration 2 years X	+1.3	+0.3	+0.3	—1.0	—1.8	+0.3	—1.3	2.7	+1.3	0	0	+0.4	+1.3	0
Rotation Y	+0.6	+1.0	0	0	+0.7	0	+0.5	+1.0	+0.8	+1.4	0	+0.3	+1.4	0
Z	0	0	0	+0.6	—0.6	0	+1.3	+1.0	+0.6	+0.9	0	0	+0.4	0
Translation X	+0.2	0	0	0	+0.4	+0.3	−0.3	−0.3	0	0	0	0	+0.3	0
Y	−0.9	0	0	−0.2	+0.4	0	−1.0	0	0	−1.1	0	0	−0.3	0
Z	+0.4	+0.5	+0.2	−0.4	−1.1	+0.2	+0.4	+1.0	0	−0.4	−0.2	+0.2	−0.2	0
Stress														
weight-bearing[2]	0.4	0.4	0	0.6	0	0	0.6	0.4	0.4	0	0.3	0	0	0.3
rotational	0	0.7	0	0	0	0	0	0	0	0	0	0	0.5	0.3
max. segm. rotation	0.5	1.2	0	0.8	0	0.4	0.8	0.6	0.7	0.4	0.4	0	0.5	0.7
MTPM	0.4	1.0	0	0.6	0	0.2	0.6	0.3	0.4	0.3	0.3	0.3	0.4	0.8
weight-bearing[3]									0.6	—	0.5	0	0	
rotational									0.4	—	0	0.3	0	
max. segm. rotation									0.6	—	0.5	0.4	0	
MTPM									0.4	0.2	0.5	0.2	0.2	

[1] numbers within parenthesis pertain to PCA zones
[2] primary stress series
[3] secondary stress series

Prosthesis		**Prototype**											
Case		66	67	68	69	70	71	72	73	74	75	76	77
Age		70	75	78	66	72	65	77	78	75	67	65	65
Sex/compartment		F/m	F/l	M/m	F/m	M/m	F/m	F/m	F/m	F/l	F/m	F/m	F/m
Weight		72	73	78	80	104	82	88	68	72	85	82	80
Follow-up		36	24	25	37	37	36	36	23	23	24	23	24
Lund score		S	S	S	S	S	S	S	S	S	S	S	S
HSS score		92	92	90	87	89	95	93	91	83	93	90	91
PT		81	90	86	86	88	85	84	89	88	84	86	84
PTS		77	92	90	90	92	89	83	88	77	88	86	79
postHKA		172	182	170	176	177	179	168	181	180	176	177	175
preHKA													
PHA		84	86	92	88	89	86	91	88	88	85	88	87
Zone width		0.8	0.9	1.0	1.0	0.6	0.7	1.1	0.9	3.0	1.0	0.9	1.0
Zone extension		25	17	26	25	8	4	11	25	50	10	62	75
Component height		9	9	12	9	9	9	9	9	12	9	9	12
Cortical support													
periphery		0	0	0	0	1	0	0	0	5	0	0	0
posterior		0	0	0	0	0	0	0	0	0	0	0	0
Migration													
MTPM	1w	0.3	0.2	0	0.3	0	0.2	—	0.2	0	0	0	0.2
	6w	0.6	1.0	—	0.4	0.3	0.3	—	0	0.5	0	0.2	0.6
	6m	0.5	0.9	0.2	0.7	0.2	0.2	0.2	0.2	0.2	0.2	0.7	0.9
	1y	0.6	2.7	0.4	1.3	—	—	0.2	0.2	0.5	0.3	0.7	0.9
	2y	0.7	3.2	0.3	1.5	0.3	0.3	0.2	0.3	0.7	0.6	0.7	1.0
	3y		0.8	—	0.3	1.6	0.5	0.4	0.3	—	—	—	—
Migration 2 years	X	−1.3	−3.6	+0.3	+1.5	0	+0.4	−0.3	−0.3	0	−0.8	−1.1	−0.7
Rotation	Y	0	−2.0	0	+0.4	0	0	0	+0.5	−0.3	−0.3	−0.3	+0.4
	Z	−1.4	−2.6	+0.3	−1.6	−0.3	−0.6	0	0	−0.8	+0.3	+1.0	0
Translation	X	+0.2	+1.4	0	+0.7	0	+0.2	0	0	+0.4	−0.2	−0.3	+0.5
	Y	0	−0.5	0	−0.2	0	0	0	0	0	+0.2	+0.2	−0.5
	Z	−0.3	−1.7	+0.2	+0.9	0	0	0	−0.2	+0.2	−0.2	−0.2	−0.4

Prosthesis		**PCA non-cement**												
Case		78	79	80	81	82	83	84	85	86	87	88	89	90
Age		67	77	78	74	71	79	73	77	65	67	63	73	73
Sex		F	F	F	F	F	F	F	F	M	F	F	F	F
Weight		72	76	69	71	81	88	86	65	79	75	69	77	62
Lund score		S	S	S	S	S	S	S	S	S	A	S	S	S
HSS score		93	86	85	94	91	94	84	86	93	88	91	95	91
PT		83	89	93	88	91	87	85	85	87	91	90	89	87
PTS		96	89	89	94	92	85	81	83	89	80	84	87	87
postHKA		179	184	184	172	183	177	172	185	181	184	182	182	177
preHKA		163	199	175	165	164	—	160	163	167	172	172	164	166
PHA		83	90	90	92	89	89	89	82	87	89	91	89	89
Zone	(8)	2	1	1	1	1	2	3	2	0	2	2	2	1
	(9)	2	1	1	3	1	0	4	1	0	2	1	3	0
	(10)	2	1	1	3	1	0	4	1	0	2	1	3	0
	(11)	2	1	1	2	2	2	3	2	0	3	2	2	2
	(15)	1	1	1	1	1	3	3	3	0	2	4	3	1
	(16)	5	2	1	2	0	3	5	2	0	4	3	5	1
	(17)	3	1	1	1	1	2	2	1	0	2	2	2	1
	(18)	4	1	1	4	4	0	5	1	0	3	2	2	3
	(19)	2	1	1	3	4	2	3	2	0	1	1	2	1
Cortical Support	med.	2	0	0	2	0	0	0	0	2	0	0	0	0
	lat.	1	3	6	0	5	0	3	9	0	6	9	3	6
	post.	0	5	0	0	1	0	0	3	0	0	2	4	3
	ant.	4	4	5	6	3	3	1	2	3	0	3	0	4
Loose beads		18	3	1	4	7	6	15	11	1	9	7	14	0
Migration														
MTPM	1w	0.6	0.3	3.0	—	0.8	0.9	0.5	0.5	0.5	0.2	1.2	0.3	0.3
	6w	1.7	0.6	4.3	0.6	1.1	1.2	1.5	0.9	1.0	1.4	3.1	2.0	0.7
	6m	1.7	—	4.4	1.4	1.9	1.5	2.2	2.1	1.5	1.4	3.4	2.0	0.6
	1y	1.9	1.0	—	1.7	1.9	1.6	2.7	2.0	1.6	1.5	3.6	2.7	0.6
	2y	2.4	1.0	5.0	2.0	2.0	1.6	3.6	3.3	1.7	1.6	3.9	2.9	0.5
Migration 2 years	X	+1.6	0	−5.8	−0.6	+1.7	+1.0	+0.4	+2.6	0	−1.6	−3.3	+2.9	0
Rotation	Y	+2.6	+1.4	+0.6	+1.7	+1.3	+1.8	−2.8	0	+1.3	−0.3	−0.5	+2.7	+0.5
	Z	−2.1	−0.3	0	+2.1	+0.5	+0.3	−1.3	+0.4	+0.8	−0.4	−3.8	+1.2	0
Translation	X	−0.5	0	0	−0.7	+0.4	+0.2	−0.7	−0.6	+0.4	0	+0.6	0	+0.3
	Y	−1.0	−0.4	−3.5	−0.4	−1.1	−1.1	−1.4	−3.1	−1.1	−0.8	+1.5	−1.0	−0.2
	Z	−0.4	0	−1.0	−0.3	+0.7	−0.4	−0.6	−0.6	+0.2	−0.6	−0.3	+0.3	−0.2
Stress														
weight-bearing[1]		1.2	0	0.5	1.4	1.7	0	0.5	0.4	0	0	0.6	0.8	0.4
rotational		+0.9	+0.3	0	+1.3	+1.5	+1.2	+0.8	+0.9	+0.8	+0.8	+1.5	+0.6	+0.5
max. segm. rotation		1.8	0.4	0.6	1.6	1.7	1.5	0.9	1.0	0.8	0.8	1.5	1.0	0.7
max. segm. translation		0.3	0	0.2	0.3	0.4	0.5	0.4	0	0	0.3	0.3	0.2	0.2
MTPM		1.3	0.3	0.5	1.1	1.0	1.0	0.6	0.6	0.5	0.4	0.9	0.6	0.4
TPM 1–12		0.2	0	—	0.3	0.2	0	—	0.2	0	0.3	—	0.3	0.3
weight-bearing[2]		0.8	0	—	1.5	0.5	0	0	0	0	0	0.6	0.7	0
rotational		+0.8	+0.4	—	+1.9	+1.0	+0.7	+1.1	+1.1	+0.9	+0.4	+0.6	+0.4	+0.6
max. segm. rotation		1.0	0.4	—	1.9	1.1	1.1	1.1	1.1	0.9	0.4	1.1	0.9	0.6
max. segm. translation		0.4	0	—	0.5	0.2	0	0.3	0	0	0	0.2	0	0.2
MTPM		0.7	0.3	—	1.1	0.8	0.6	0.8	0.6	0.5	0.3	0.6	0.6	0.4
TPM 1–12		0.4	—	—	0	0	—	—	0	0	0	0.3	—	0

[1] primary stress series
[2] secondary stress series

Prosthesis		**Freeman-Samuelson**					
Case		91	92	93	94	95	96
Age		76	69	71	77	78	65
Sex		F	F	F	F	F	F
Weight		79	75	66	59	59	100
Lund score		S	S	S	S	S	S
HSS score		89	88	90	90	90	86
PT		88	87	83	90	88	88
PTS		91	87	88	90	88	84
postHKA		181	177	170	180	182	177
PHA		90	90	87	90	89	91
Peg sclerosis, medial		4	3	2	2	2	3
	lateral	3	3	3	1	1	3
Sclerosis diameter,							
medial		17.5	17.0	16.0	15.0	16.0	17.5
lateral		17.5	17.0	16.0	16.0	15.5	18.0
Cortical support	med.	—	0	0	0	0	0
	lat.	—	0	0	0	0	0
	ant.	—	0	0	4	5	5
	post.	—	0	0	0	0	3
Migration	3m	—	−1.7	0.9	—	—	—
MTPM	6m	0.8	1.2	0.9	—	—	—
	1y	0.7	1.0	0.9	—	—	—
	2y	1,3	1.5	1.2			
Migration 1 year	X	−0.7	−1.8	+1.1	—	—	—
Rotation	Y	0	−0.4	+0.9	—	—	—
	Z	0	0	−1.1	—	—	—
Translation	X	+0.5	+0,3	+0.2	—	—	—
	Y	+0.2	−0.5	−0.3	—	—	—
	Z	0	+0.3	0	—	—	—
Stress							
weight-bearing		0.4	2.0	0.9	0	0.8	0.7
rotational		+0.8	+1.3	+2.2	+1.1	+5.3	+2.4
max. segm. rotation		0.8	2.7	2.2	1.2	5.4	2.5
max. segm. translation		0.2	1.1	0.4	0.4	1.4	0.3
MTPM		0.5	2.1	1.5	1.2	5.0	1.5
TPM 1–12		0.2	2.7	0.5	—	0.5	1.0