Kinematics and Fixation of Total Knee Arthroplasties

A clinical, radiographic, scintimetric, and roentgen stereophotogrammetric evaluation

av

Kjell G. Nilsson

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Docent Johan Kärrholm

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Docent Leif Ryd

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Kjell G. Nilsson

From the Department of Orthopaedics, University Hospital of Umeå, S–901 85, Umeå, Sweden.

Abstract

Aseptic loosening of the tibial component is an important cause of failure after total knee arthroplasty. Bone destruction often claimed to be caused by the cement makes the revision difficult. In order to treat younger patients, uncemented fixation has been introduced, but the etiology to loosening is multifactorial and only partly known. Early detection of implant migration facilitates research in this field but is difficult using conventional techniques. In this study modified versions of roentgen stereophotogrammetric analysis (RSA) were developed to obtain accurate and standardized evaluations facilitating comparison between prosthetic designs. The method was used to record the efficacy of cemented and uncemented fixation of different designs of the tibial component, to determine the accuracy of scintimetry in the detection of early aseptic loosening, and to analyse the *in vivo* kinematics of knee arthroplasties with different design and stability between the joint surfaces.

Forty–three arthroplasties with comparatively high inherent stability of the joint surfaces were randomized to cemented or uncemented fixation of the tibial component. In all groups micromovements were rather large, but with no differences between the cemented and uncemented components. The preoperative diagnosis (arthrosis OA, n=25; rheumatoid arthritis RA, n=18) did not influence the magnitude of micromotion.

20 arthroplasties with the same design as above but equipped with an intramedullary stem, were randomized to cemented or uncemented fixation in patients with RA. Cement improved the fixation. Uncemented stemmed components displayed micromovements seemingly larger than unstemmed ones.

34 arthroplasties with an unconstrained design of the joint area and fixed to the tibia with four pegs were randomized to cemented or uncemented fixation in patients with OA. When used uncemented 4 screws were added. Compared with previously investigated designs small micromotions were recorded, and especially in the cemented cases. Uncemented components with thin polyethylene inserts displayed larger initial micromotions. The preoperative deformity influenced the direction of the micromotion.

33 knees were followed prospectively with RSA and scintimetry to evaluate any correlation between these methods. Low activity under the tibial component at 2 years implied prosthetic stability, whereas high activity indicated instability *or* high bone remodelling caused by the preoperative malalignment.

The *in vivo* kinematics in three different designs of knee arthroplasties were analyzed during active flexion and extension without weight–bearing. Each type of prosthesis displayed design–specific abnormalities when compared with a normal material. Pronounced posterior tibial translations were recorded during flexion regardless whether the posterior cruciate ligament had been sacrificed or not. Data from the kinematic and the fixation studies suggest that movements restricted by the design of the joint area are transmitted to the bony interface with design–specific micromotions as the result.

Analysis of knee joint kinematics during extension and weight–bearing revealed small alterations compared with non–weight–bearing. Evaluation of the three–dimensional movements in terms of helical axis rotations and translations confirmed the constrained or unconstrained *in vivo* behaviour of the designs under study. This analysis also facilitated the interpretation of the kinematic behaviour of the prosthetic knees and may be of value in the evaluation of new designs.

Key words: Total knee arthroplasty, tibial component, kinematics, micromotion, loosening, scintimetry, roentgen stereophotogrammetric analysis.

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Contents

This thesis is based upon the following papers, which are referred to in the text by their Roman numerals:

I. Nilsson KG, Kärrholm J, Ekelund L, Magnusson P. **Evaluation of micromotion in cemented *vs* uncemented knee arthroplasty in osteoarthrosis and rheumatoid arthritis.** Randomized study using roentgen stereophotogrammetric analysis. *Journal of Arthroplasty* **6:**265–278, 1991.

II. Nilsson KG, Kärrholm J. **Increased subsidence of uncemented stemmed Tricon-M knee prostheses.** Roentgen stereophotogrammetric analysis of cemented *vs* uncemented tibial components. Submitted.

III. Nilsson KG, Kärrholm J. **Increased varus–valgus tilting of screw fixated knee prostheses.** Stereoradiographic study of uncemented *vs* cemented tibial components. *J Arthroplasty, in press.*

IV. Nilsson KG, Björnebrink J, Hietala S–O, Kärrholm J. **Scintimetry after total knee arthroplasty.** Prospective 2–year study of 18 cases of arthrosis and 15 cases of rheumatoid arthritis. *Acta Orthopaedica Scandinavica* **63:**159–165. 1992.

V. Nilsson KG, Kärrholm J, Ekelund L. **Knee motion in total knee arthroplasty.** A roentgen stereophotogrammetric analysis of the kinematics of the Tricon–M knee prosthesis. *Clinical Orthopaedics and Related Research* **256:**147–161, 1990.

VI. Nilsson KG, Kärrholm J, Gadegaard P. **Abnormal kinematics of the artificial knee.** Roentgen stereophotogrammetric analysis of 10 Miller–Galante and five New Jersey LCS knees. *Acta Orthopaedica Scandinavica* **62:**440–446, 1991.

VII. Kärrholm J, Jonsson H, Nilsson KG, Söderkvist I. **Kinematics of successful knee prostheses during weight–bearing.** Three–dimensional movements and positions of screw axes in the Tricon–M and Miller–Galante designs. Submitted.

Some complementary data (5–year follow–up – Study I), not earlier published, are included in the Results and Discussion sections.

Definitions and Abbreviations

ACL: Anterior cruciate ligament.

ANOVA: Analysis of variance

Condition number of a rigid body: The result of a mathematical calculation determining or describing distances and angles between the tantalum markers in a rigid body. The size of the condition number decreases with increasing scattering of the tantalum markers.

Constrained prosthesis: High congruity between the articulating surfaces of the femoral and tibial components resulting in increased stiffness of the joint in certain directions. Semi– or unconstrained prostheses have less congruity between the articular surfaces and less inherent stability.

F–S: The Freeman–Samuelson knee prosthesis.

Helical axis: A mathematically computed axis with three–dimensional orientation used to describe rigid body motion. Also called screw axis. See also "Rigid body motion".

HKA angle: The Hip Knee Ankle angle is used to define the alignment of the leg. It is the medial angle between two lines, one from the center of the hip and the other from the center of the ankle joint, meeting at the center of the knee; < 180° varus alignment, > 180° valgus alignment.

HSS: Hospital for Special Surgery.

Instant center of rotation: The center of rotation for movements of a rigid body based on reconstructions in two dimensions.

LCS: The New Jersey Low Contact Stress knee prosthesis.

Mean error of rigid body fitting (ME): The result of a mathematical calculation determining the deformation of a rigid body between two roentgen stereophotogrammetric examinations. The smaller the value the better the stability of the tantalum markers defining the rigid body.

M–G: The original design of the Miller–Galante knee prosthesis.

MTPM: Maximum total point motion. The length of the three–dimensional translation vector for the fictive prosthetic marker that moved the most.

Micromotion: Prosthetic movements not detectable with conventional radiography.

Migration: Micromotion between 2 stereoradiographic examination with an interval of at least 4 weeks..

NS: Not significant. The significance level was set at $P < 0.05$ in this study.

OA: Osteoarthrosis.

PCA: The Porous Coated Anatomic knee prosthesis.

PCL: Posterior cruciate ligament

RA: Rheumatoid arthritis

Rigid body: A volume or area defined by constant distances. In this investigation bone and prosthetic components were regarded as separate rigid bodies represented by 4 to 9 tantalum markers each.

Rigid body motion:

1) Movement of a rigid body in a three–dimensional coordinate system with body–fixed axes. This movement has six degrees of freedom: rotations about–, and translations along the three cardinal axes.
2) Movement of a rigid body in relation to one axis (helical axis or screw axis). The screw axis has a variable three–dimensional orientation described by a unit vector or its intersection with the three planes. Movement is described as rotation about–, and translation along this axis.

ROI: Region of interest.

RSA: Roentgen Stereophotogrammetric Analysis. The technique of obtaining accurate three–dimensional measurements from radiographs.

Screw axis: See "Helical axis" and "Rigid body motion".

SD: Standard deviation.

SE(M): Standard error (of the mean).

TKA: Total knee arthroplasty.

Total point motion: The length of the three–dimensional translation vector for each point of measurement.

Tricon–M: The Tricon–M knee prosthesis.

Tricon–M stem: The Tricon–M knee prosthesis with a intramedullary stem added to the tibial component.

UMRSA: Program system for RSA. Main programs are: **MEASURE, Xray90, KINLAB** (SEGMENT MOTION, POINT MOTION, POINT TRANSFER, ROTATE, GROWTH), and **KINERR**.

Introduction

History

The evolution of artificial replacement of the knee joint started about forty years ago with the introduction of hinged prostheses (Shiers 1954, Walldius 1957, 1960) and plates acting as interpositional devices fixed with screws (Townley 1964) or without firm fixation to the bone (MacIntosh 1958, McKeever 1960). An increasing incidence of infection, loosening and even dislocation was reported, partly due to their use in severely destroyed knees, but also because of inadequate primary fixation and kinematics (Ranawat et al 1973, Insall et al 1976a, Jones et al 1979).

Gunston (1971) designed a condylar prosthesis based on anatomical measurements, and contrary to the previous authors/inventors he used polymethylmethacrylate for the fixation to bone. In 1974, the semiconstrained Total Condylar prosthesis (Insall et al 1976b) was introduced based on studies of knee kinematics and implant biomechanics (Walker et al 1972). This prosthesis is still regarded "the golden standard". During the same period the unconstrained Townley prosthesis (Townley 1985) and the more constrained uncemented Freeman–Swanson prosthesis (Freeman et al 1973) were introduced. Due to continuous progress of the clinical results, knee arthroplasty gained an increasing popularity and today about 3000 knee prostheses are implanted each year in Sweden (Lewold 1992, personal communication).

The problem of fixation and wear

The most common cause of failure in total knee arthroplasty during the last 10 years has been aseptic loosening of the tibial and patellar components. Wear of the prosthetic components and especially the polyethylene has also become an increasing cause of implant failure during the recent years (Goodfellow 1992, Goodman and Lidgren 1992). Metal–backed tibial and patellar components with thin polyethylene layer have proven especially prone to wear (Bartel et al 1986, Engh et al 1992, Wright et al 1992). This Thesis deals primarily with the problem of tibial component loosening in bi/tri compartmental total knee arthroplasty.

Clinically, tibial component loosening is usually associated with pain, increasing deformity and reduced mobility. Revision surgery is often technically demanding due to bone destruction, and the results are often inferior to those obtained after primary surgery (Rand and Ilstrup 1991). Aseptic loosening has a multifactorial and only partly known background. Increased knowledge about the etiology of aseptic loosening is therefore desired.

Incidence and definition of loosening

The survival rates using the first generation cemented total knee arthroplasties were relatively low (Tew and Waugh 1982, Lewallen et al 1984, Riley and Woodyard 1985, Tew et al 1985, Knutson et al 1986, Rand and Coventry 1988). The use of cement was blamed for these inferior results (Jones and Hungerford 1987) and a new interest in uncemented alternatives was initiated. However, factors such as poor prosthetic design (Lew and Lewis 1982, Insall 1984), the use of too small components (Laskin 1988), improper alignment of the components and the leg (Lotke and Ecker 1977), and inferior cementing technique (Miller 1984) were also attributed to failure. Later, altered surgical and cementing techniques improved the survival rates of cemented knee arthroplasties, in some instances with unchanged design, to more than 90 or 95 percent survival after 5 to 10 years (Ritter et al 1989, Scuderi et al 1989, Vince et al 1989, Lewold et al 1992).

Analysis of the results after knee arthroplasty is difficult to perform due to the lack of a clear definition of aseptic loosening as reflected in several clinical and radiographic definitions of loosening. Recurrence of pain on weight–bearing, and/or progressive radiolucent zone under the entire component and/or a shift in position has been considered definite signs of loosening (Kaufer and Matthews 1981, Blaha et al 1982, Schneider et al 1982, Ranawat and Boachie–Adjei 1988). However, pain on weight–bearing may have other causes than loosening, and evaluation of radiographic loosening is unpredictable because the skeleton lacks distinct bony landmarks and visualization of radiolucent zones is often difficult (Insall 1984, Tibrewall et al 1984, Ecker et al 1987). Moreover, prosthetic micromovements may occur in the absence of radiolucent zones (Ryd 1986). Therefore, to evaluate the efficacy of different modes of prosthetic fixation, improved technique with higher resolution is needed.

Scintigraphy and Scintimetry

Technetium scintigraphy and scintimetry have been used to detect aseptic loosening of hip prostheses (Mjöberg et al 1985, Snorrason et al 1989). The isotope adheres to the mineral phase of the bone and the increased activity found in such situations is thought to be the result of bone remodelling (Christensen 1985) caused by prosthetic instability. However, its use in knee arthroplasties is controversial (Rosenthall et al 1987, Duus et al 1987, 1990, Hofmann et al 1990). It has been suggested that bone scan can detect loosening of knee prostheses (Gelman et al 1978, Hunter et al 1980), whereas others have found persistent high activity under the tibial component several years after the operation in asymptomatic knees (Rozing et al 1982, Schneider and Soudry 1986, Rosenthall et al 1987, Hofmann et al 1990, Kantor et al 1990). These divergent findings may reflect that the changes of the isotope uptake during the postoperative period are not completely known and that simultaneous recordings of prosthetic micromovements have not been done.

Kinematics of the knee

The knee joint shall provide stability (mainly in extension) and mobility (mainly in flexion) (Kapandji 1970). The joint motion contains 6 degrees of freedom: rotations and translations in the sagittal, frontal and transverse planes. The kinematics of the knee has been investigated *in vivo* using goniometers (Kettelkamp et al 1970, Townsend et al 1977, Isacson 1987), photographic or video techniques using skin–markers (Huntington et al 1979) or skeletal pins (Levens et al 1948, Lafortune and Cavanagh 1985), radiography (Frankel et al 1971), or roentgen stereophotogrammetry (RSA) (Kärrholm et al 1988b, 1989, Jonsson et al 1989). During flexion without weight–bearing the tibia has been found to rotate internally and into adduction, and the reverse movements occur during extension (Kärrholm et al 1988b, Jonsson et al 1989). External tibial rotation close to full extension has been called the "screw–home" mechanism (Smillie 1978, Andriacchi et al 1986), however, some have questioned its existence (Hallén and Lindahl 1966). The anatomy of the joint area and the ligaments, muscular and ground reaction forces may all contribute to the kinematics of the normal knee (Hsieh and Walker 1976, Goodfellow and O'Connor 1978, Markolf et al 1981). Abnormal kinematics could be presumed in artificial knees as indicated from gait analysis (Rittman et al 1981, Andriacchi et al 1982), but detailed *in vivo* studies have, according to my knowledge, not been done.

The early designs of knee prostheses substituted the ligamentous functions by a hinge (Shiers 1954, Walldius 1957, Young 1963). Due to the high frequency of loosening (Bain 1973, Freeman et al 1973, Tew and Waugh 1982, Knutson et al 1986), it became apparent that knee prostheses had to be designed allowing more degrees of freedom. Based on two-dimensional calculations of a changing axis of rotation (Frankel et al 1971, Kettelkamp and Nasca 1973, Harding et al 1977) condylar replacement (Gunston 1971, Walker et al 1972, Waugh et al 1973) aiming at a more physiological reproduction of knee joint motion were developed.

The design of total knee arthroplasty is a compromise between two more or less incompatible requirements; inherent stability by increased congruence between the articular surfaces, and normal joint kinematics requiring decreased congruence between the articular surfaces. Due to concern of instability, many of the first generation condylar replacements were designed with congruent joint areas (= semi constrained prostheses). *In vitro* tests of the inherent stability of these designs displayed increased resistance to rotational forces (Walker et al 1975, Werner et al 1978, Shoemaker et al 1982), implying that the rotational stresses resisted by the prosthesis might be transmitted to the prosthesis–cement or cement–bone interfaces (Werner et al 1978). Preservation of both the cruciate ligaments in prostheses with congruent articular surfaces was found to result in high joint forces also increasing the risk of loosening (Goodfellow and O'Connor 1978, Soudan et al 1980, Lew and Lewis 1982).

In order to minimize the forces transmitted to the interfaces, less constrained knee arthroplasties were introduced (Hungerford and Krackow 1985, Landon et al 1986). However, reducing the congruence between the articular surfaces decreases the contact area and thus increases the contact stresses with a potential of increased polyethylene wear (Bloebaum et al 1991, Engh et al 1992, Jones et al 1992, Wright et al 1992) and especially if thin polyethylene components are inserted (Bartel et al 1986, Engh et al 1992, Wright et al 1992).

The instant center technique first described by Reuleaux (1876) has been used to reconstruct surface joint motion. When obtaining sequential radiographs of the joint in different positions throughout the range of motion in one plane, the Reuleaux method can locate the instant center for each interval of motion, enabling the determination of instant center pathways. In the knee joint a variable axis of rotation has been demonstrated (Frankel et al 1971, Walker et al 1972, Harding et al 1977, Goodfellow and O'Connor 1978). Using this method many of the current designs of knee prostheses have been constructed (Walker et al 1972, Walker and Shoji 1973). However, since the Reuleaux method only allows analysis of motion in one plane condensing a helical axis with a three-dimensional extension into a point, the accuracy of the method to represent knee movements has been seriously questioned (Soudan et al 1979, 1980, Woltring et al 1985). Using RSA, Blankevoort et al (1986, 1988) examined passive knee motion in a few human knee specimens, and even reconstructed helical or screw axis positions in one of them. The method has been used to reconstruct different types of skeletal (Lundberg et al 1989, Ragnarsson and Kärrholm 1992) and implant movements *in vivo* (Kärrholm and Snorrason 1992a, 1992b), but not so far to evaluate the knee kinematics in patients with prosthetic replacements.

The design of the prosthetic tibiofemoral joint is a balance between low and high congruence. The choice between flat or dished design depends on which factor is believed to become the most significant threat to the durability; polyethylene wear or poor fixation (Insall 1988). Detailed and objective analysis of the kinematics and fixation of the knee prostheses representing different degrees of stability or conformity between the joint areas has, however, not been performed.

Knee joint forces

The compressive knee forces varies between the stance- and swing phases during gait and also between normal and diseased knees. They have been calculated at 2–4 times body weight during the stance phase (Morrison 1968, 1969, 1970, Seireg and Arvikar 1975, Harrington 1983). Rotatory torques about the knee has also been analyzed (Morrison 1970). Harrington (1983) calculated the loading patterns in normal knees and in knees with deformities during static and dynamic conditions. The maximum joint loads during walking corresponded to about 3.5 times body weight in normal knees. Lower values were found in knees with varus or valgus deformities (2.1 and 2.4 times body weight, respectively). In the normal knees there were three peaks in joint force: two shortly after heel strike, and one before toe off. The smallest joint force approximated body weight and occurred during mid-stance. In the knees with deformities there were no force peaks, and the greatest joint forces occurred during mid–stance.

In normal knees, the center of joint pressure varies during the stance phase (Harrington 1983, Weidenhielm 1992). Immediately after heel strike it is positioned in the lateral compartment, but thereafter moves to the medial one. In varus knees, the maximum pressure is medially positioned throughout the stance phase. In valgus knees an individual distribution was found; both lateral and medial locations were recorded despite of the deformity.

The forces acting on the knee joint during the swing phase have not been studied, but Nisell et al (1986) calculated forces of about twice the body weight in the non weight–bearing position with a weight at the ankle.

Thus, regardless whether the knee arthroplasty restitutes normal gait or not (Andriacchi et al 1982) the loads that the prosthesis/cement–bone interface has to resist are considerable. The location of the center of joint pressure also varies during the stance phase both in the frontal (Harrington 1983) and the sagittal planes (Grady–Benson et al 1992) implying that the forces can induce rocking movements of the tibial component in patients with total knee arthroplasty (Ryd et al 1991).

Posterior cruciate ligament

Retention of the posterior cruciate ligament (PCL) in knee arthroplasty is thought to contribute to more normal kinematics by controlling the femoral "roll–back" during flexion (Goodfellow and O'Connor 1978, Andriacchi et al 1982, Andriacchi and Galante 1988), leading to increased knee flexion and a more normal pattern of the gait, especially during stairclimbing (Andriacchi et al 1982, Andriacchi et al 1986, Dorr et al 1988). On the other hand, resection of the PCL, as advocated by several authors (Freeman et al 1977, Freeman and Railton 1988, Insall 1988), has not caused any inferior results in clinical practise (Insall 1988). If the PCL is sacrificed the articular surfaces of the knee arthroplasty are usually designed to compensate for this, corresponding to a more constrained prosthesis. On the other hand, retaining the PCL necessitates more unconstrained joint surfaces, resulting in larger areas exposed to translatory movements and loads.

Factors related to prosthetic fixation

Several factors have been claimed to influence the result of knee arthroplasties such as patient selection, surgical technique, loading conditions (Albrektsson et al 1981), and type of implant.

Patient factors. Rand and Ilstrup (1991) reported decreased implant survival rates in patients younger than 60 years, whereas Scuderi et al (1989) and Knutson et al (1986) did not find any age related influences on prosthetic loosening in cemented total condylar prostheses. It is unclear, however, whether some of these results reflect a true variability in fixation between age groups, or merely reflects dissimilar indications for revision surgery in different age groups.

Rand and Ilstrup (1991) noted higher knee prosthesis survival rates in women than in men after 5 and 10 years, but this was not confirmed by Scuderi et al (1989). Increased frequency of patellar complications has been reported in males (Rosenberg et al 1988).

Over–weight is more common in cases of revision surgery (Bryan and Rand 1982, Thornhill et al 1982, Barett and Scott 1987, Moreland 1988, Windsor et al 1989), and Ranawat and Boachie–Adjei (1988) found a correlation between high body weight and radiolucency under the tibial component, suggesting excessive loading of the implant. Hvid (1988b) found a tendency to positive correlation between bone strength and body weight, but after normalization of bone strength with respect to body weight it turned out that this increase in bone strength was not sufficient. However, several authors have not been able to detect any negative effects on implant survival rates in over–weight patients (Scuderi et al 1989, Stern and Insall 1990, Smith et al 1992). Rosenberg et al (1988) found an increased frequency of patellar problems in over–weight payients. In sum, there is no general consensus as regards the relation between patient weight and prosthetic fixation.

Most studies have reported similar revision rates of total knee arthroplasties in patients with rheumatoid arthritis (RA) and arthrosis (OA) (Sledge and Walker 1984, Knutson et al 1986, Hvid et al 1987, Laskin 1990). Rand and Ilstrup (1991) found somewhat smaller revision rates in patients with RA. Hvid (1988a) found almost the same bone strength of the proximal tibial metaphysis in OA and RA patients at the time for knee arthroplasty, and concluded that about equal prosthesis survival rates could be expected.

However, decreased activity in patients with a multiple joint disease could also be expected, which has implications in terms of fixation.

The bone strength at the knee varies with the degree of knee deformity. In normal knees and in arthrotic knees with varus deformity, the metaphyseal bone in the proximal tibia is stronger medially than laterally, whereas in arthrotic knees with valgus deformity the location of maximum bone strength is variable (Hvid 1988a). Further, the bone strength of the unloaded condyle may be so reduced that there will be risk of fatigue failure (Hvid 1988a). In both normal and diseased knees, bone strength decreases with increased distance from the joint surface (Cornell et al 1986, Hvid 1988a), which has resulted in the recommendation to cut as little bone as possible at surgery.

Surgical technique. Restoration of knee alignment and reconstruction of a horizontal joint line has been found important for favourable long-term results of TKA (Lotke and Ecker 1977, Freeman et al 1978, Bargren et al 1983, Insall et al 1983, Landon et al 1985, Andriacchi et al 1986, Uematsu et al 1987, Ryd et al 1990).

Bone cutting may produce temperatures in the bone above the critical temperature for bone necrosis (Eriksson 1984) which might influence the quality of the fixation (Toksvig-Larsen 1992).

Implant loading conditions. Johansson (1991) proposed the following theory for the incorporation of a foreign material into bone, based on the theory for fracture healing by Frost (1989). At the implantation there is inevitably a surgical trauma that results in tissue repair. In the first injury phase, starting immediately after implantation, cells are sensitized and there is release of growth factors. Two to 3 weeks after the implantation the granulation phase starts, and connective tissue is formed in the interface. After 4 to 16 weeks the callus phase is initiated when, under optimum circumstances, bone formation is started. This process may take one year or more to be completed. If the stresses applied to this thin fibrous interface are too low, bone formation might be inhibited (Askew and Lewis 1981, Murase et al 1983, Turner et al 1986, Hvid 1988a, Dawson and Bartel 1992) and formation of a wider fibrous layer ensues. If, on the other hand, the stresses are too high, bone will crush and micromovements will ensue, resulting in the formation of fibrous tissue (Aspenberg et al 1992, Söballe et al 1990a, 1990b, 1992a, 1992b), which under certain circumstances might be replaced by fibrocartilage (Pauwels 1980, Ryd and Linder 1989). Continuous micromovements exceeding 150–200 μm have been shown to be incompatible with bone formation (Pilliar et al 1986, Söballe et al 1992). The type and magnitude of the stresses that are optimum for bony ingrowth are not known, but pure compressive stresses have been found to be advantageous (Goldstein et al 1986, Carlsson 1989).

Differences in elasticity at the implant–bone interface during compressive load may also compromise the incorporation. In a finite element analysis, Natarajan and Andriacchi (1988) calculated a tangential displacement between the bone and an implant without pegs of 150 μm at a load of 2 times the body–weight. When the implant had pegs the displacement between the pegs was about ten times smaller.

Fixation with bone cement. When using bone cement, fixation is achieved by interdigitation between cement and bone, resulting in a large contact area and a capacity for load distribution. The quality of the cement fixation is influenced by several factors. Bone cement has in itself several negative side effects that may jeopardize fixation. The monomer is toxic (Feith 1975, Morberg 1991, Morberg and Albrektsson 1991), and curing of cement may lead to thermal injury (Feith 1975, Mjöberg 1986), but this is probably of minor importance in knee arthroplasties (Toksvig–Larsen et al 1991). Bone cement is brittle, weak in tensile and shear strength (Saha and Pal 1984), ages with time (Jasty et al 1991), and degradation particles may elicit bone resorption at the bone cement interface (Jasty et al 1986, Goodman et al 1988).

Several measures have been taken to further improve the properties of the cement. The viscosity has been reduced to improve penetration (Miller et al 1979, Krause et al 1982, Noble and Swarts 1983), but the value of low viscosity cement has been debated (Mjöberg et al 1987, Granhed et al 1991). Centrifugation (Burke et al 1984, Davies et al 1989) or vacuum mixing (Lidgren et al 1987) have been introduced to reduce cement porosity and increase its strength. Partial substitution of polymethylmethacrylate (PMMA) with new forms of methacrylate monomers (DMA and IBMA) (Jensen and Jensen 1991) or reduction of the amount of monomer (Mjöberg 1986, Mjöberg and Selvik 1988, De Bastiani et al 1990) decreases the monomer leakage and bone toxicity (Stürup et al 1992).

Changes of the cementation technique including high–pressure lavage, meticulous cleaning of the cut surfaces and insertion of the cement under pressure have proved to increase the bone–cement interface strength (Harris 1975, Halawa et al 1978, Lange 1979, Krause et al 1982, Miller 1984).

Lewold et al (Lewold et al 1992) found a continuous improvement in cumulative survival rate with time for two types of knee prostheses with unchanged design during the last 15 years, indicating that improvements in surgical technique and the handling of the cement are the most likely causes of increasing survival rates of cemented knee arthroplasties during the last 10–15 years.

Uncemented fixation. Due to the low survival rates of the early designs of cemented knee arthroplasties, extensive research in uncemented fixation was initiated. Although, it has become clear that these inferior early results had a multifactorial background, uncemented fixation has certainly gained popularity during the last decade. Cement toxicity, aging and increasing brittleness of the material (Saha and Pal 1984, Jasty et al 1991, Morberg 1991) may be important especially if total knee arthroplasty is to be considered an alternative for younger patients. If necessary, revision might be easier to perform if the original prosthesis was inserted without cement.

The goal of uncemented fixation is to achieve a close contact between bone and implant without an intervening fibrous tissue membrane (Albrektsson et al 1981). By increasing the area available for contact to bone with sintered beads (Bobyn et al 1980) or fiber metal mesh (Galante et al 1971) bone ingrowth into the pores may occur and enhance the fixation. To allow bone ingrowth a pore size between 50 μm and 400 μm and a porosity volume over 30 percent is required (Galante et al 1971, Bobyn et al 1980, Cameron 1982).

Initial intimate contact between the prosthesis and bone is important to obtain bone ingrowth. However, the cut surface of the proximal tibia in the clinical situation is often uneven to a degree which might be incompatible with bone ingrowth (Toksvig–Larsen 1992). In animal experiments the ability of bone to bridge a gap between implant and bone varies from almost 0 mm to 2 mm (Albrektsson and Albrektsson 1987, Sandborn et al 1988, Carlsson 1989, Söballe et al 1990b). Hydroxyapatite coating may, at least to some extent, reduce the negative effects of a gap between bone and implant (Geesink et al 1987, Söballe et al 1990b 1992a).

The process of bone ingrowth or bony adhesion can be compared with healing of a fracture (Morscher 1987) taking up to a year or more to be completed (Cameron et al 1973, Johansson 1991). During that time stability between implant and bone is required. Both Pilliar et al (1986) and Söballe et al (1992a) found that movements >= 150 μm inhibited bone ingrowth. Even the frequency of the micromotions applied may be of importance (Aspenberg et al 1992).

The early clinical results of uncemented knee prostheses were encouraging and displayed results similar to cemented fixation (Blaha et al 1982, Freeman et al 1985, Hungerford and Krackow 1985, Landon et al 1986, Laskin 1988, 1991, Rorabeck et al 1988, Knahr et al 1990, Rosenberg et al 1990, Kavolus 1991). However, the follow-up periods were short and the patients often selected. No randomization between cemented and uncemented fixation in comparable groups were performed, and the results were assessed only by various knee scores and conventional radiographs, making a proper evaluation of the merits of uncemented *vs* cemented fixation impossible. One exception was Dodd et al (1990), who reported on the results of paired cemented and uncemented PCA knee prostheses in 18 patients followed for a mean of 5 years. They could not detect any differences in clinical and radiographic outcome. However, recently some reports with longer follow-up periods have emerged displaying inferior results using uncemented implants as compared to cemented ones (Albrektsson and Herberts 1988, Moran et al 1991, Rand and Ilstrup 1991, Nafei et al 1992)

The results from studies of implants retrieved for various reasons show that bony ingrowth into porous surfaces vary considerable. Rates of ingrowth ranges from 0 to 40%, and the extent of ingrowth is often so small that the prosthesis could not be considered permanently fixed to bone (Collier et al 1986, Cameron 1986, Ranawat et al 1986, Haddad et al 1987, Cook et al 1988, 1989, Sumner 1989). These observations suggest that the initial stability of many uncemented implants has been insufficient.

Background to this Thesis

Ryd (1986) was the first to measure micromovements of the tibial component using RSA. The migration of certain designs of uncemented arthroplasties were larger than in cemented ones, albeit equal clinical results (Ryd 1986, Ryd et al 1988, 1990). He concluded that micromotions always occur in uncemented knee arthroplasty. *In vitro* tests using bone or plastic have shown that uncemented components with few and small pegs display permanent subsidence and lift-off when loaded asymmetrically; exceeding the limit for bone ingrowth (Volz et al 1988, Branson et al 1989). Adding more or larger pegs, screws and longer stems to the tibial components increased the stability *in vitro* (Strickland et al 1988, Volz et al 1988, Shimagaki et al 1990, Walker et al 1990, Lee et al 1991). However, at the time when this study was planned, no *in vivo* investigations using high resolution technique had been published examining the stability of various types of cemented and uncemented fixation concepts in prostheses with different joint area constraint in prospective randomized trials.

Stable fixation of total knee arthroplasties is essential to avoid long–term complications. Cemented fixation of some prosthetic designs is associated with superior initial stability and high long–term survival rates when used in the elderly (Vince et al 1989), but micromovements seem to be difficult to avoid also when cement is used (Ryd 1986). In uncemented fixation the need for initial stability between implant and bone is imperative. New designs aiming at increased initial stability have been introduced after *in vitro* testing. The *in vivo* performance of such designs has not yet been evaluated. Using clinical and conventional radiographic methods 7–10 years may elapse before the results of such changes may become apparent. The initial stability of the bone–implant interface is also dependent on the load transmission between the components indicating that the inherent kinematics of the prosthesis is of importance. Analysis of the *in vivo* kinematics of knee prostheses, as well as analysis of the relation between kinematics and prosthetic micromotion has not yet been performed due to lack of adequate methodology.

Scintigraphy and scintimetry have been used to evaluate loosening of knee arthroplasties. The results have, however, been difficult to interpret because the natural course of the postoperative scintimetric activity and its relation to the magnitude of prosthetic instability has not been fully investigated.

Roentgen Stereophotogrammetric Analysis (RSA) has a resolution an order of magnitude better than conventional radiography (Selvik 1974, 1989, Kärrholm 1989, Ryd 1992). It allows analysis of small movements between orthopedic implants and bone *in vivo*. Important information concerning the quality of fixation of implants can be obtained within 1 to 2 years postoperatively (Mjöberg 1986, Ryd 1992). RSA also allows an analysis of the three–dimensional kinematics of the knee *in vivo*, thus enabling comparison of different articular designs of knee prostheses and normal knees.

Aims of this study:

A. Tibial component fixation

1. To evaluate if the preoperative diagnosis influenced prosthetic fixation (I).

2. To evaluate the efficacy of cemented *vs* uncemented fixation using porous coating and two flanged polyethylene pegs (I).

3. To evaluate the efficacy of cemented *vs* uncemented fixation using porous coating, two flanged polyethylene pegs and an intramedullary stem (II).

4. To evaluate the efficacy of screws *vs* cement using titanium fibermesh undersurface and 4 pegs for additional fixation (III).

5. To investigate the natural course of the postoperative scintimetric activity under the tibial component, and its relation to prosthetic micromovements (IV).

B. Knee kinematics

6. To analyse the *in vivo* non weight-bearing kinematics of normal knees and three designs of knee prostheses; two using different solutions to obtain a low constrained joint and one design with a more constrained joint area (V, VI).

7. To analyse the three-dimensional movements and positions of the screw axes *in vivo* in normal knees and two designs of knee prostheses during weight-bearing (VII).

Patients and Knee Prostheses

Patients:

A total of 90 patients (103 knees, I - VII), operated on between November 1985 and February 1991, were examined with roentgen stereophotogrammetric analysis. 81 of the patients were operated on in Umeå with either the Tricon–M prosthesis, the Tricon–M Central Tibial Stem prosthesis, or the Miller–Galante prosthesis. In order to achieve a larger material, 4 consecutive patients (4 knees) were operated on in Örnsköldsvik with the Tricon–M prosthesis. 5 patients (5 knees) were operated on in Gällivare with the New Jersey LCS prosthesis (Table 1).

Forty–eight patients had arthrosis (OA stage III–V, Ahlbäck 1968), 35 had rheumatoid arthritis (RA), 6 arthrosis secondary to previous fracture and 1 osteonecrosis. These latter 7 patients were grouped as primary arthrosis in the analyses. Mean age at the time for operation was 70 years (range 59–84) in the patients with OA, and 65 years (36–81) in the RA patients.

The number of patients and knees participating in the different investigations are shown in Table 2 and in the appendix. In the scintimetric study (IV) 33 of the 39 patients operated in Umeå with the Tricon–M prosthesis agreed to participate.

The patients participating in the investigations of joint kinematics (V, VI, VII) were selected based on optimum configuration of the tantalum bone markers in the distal femur and proximal tibia, allowing adequate visualisation also at the flexed positions of the knee at the RSA examination. All these patients had a good or excellent clinical result at the time for the investigation.

Patient losses during the follow up:

In paper I, 4 patients were lost during the investigation. 1 woman (case 958, see appendix) developed radiographic evidence of loosening of the patellar component 15 months postoperatively. In addition she had pain during weight–bearing corresponding to the proximal part of the tibia. At the reoperation 21 months postoperatively, both the patellar and the tibial components were revised. One woman (961) and 1 man (975) died of unrelated causes 15 and 18 months after the operation, respectively. Finally 1 woman (973) refused to attend the 2 year follow–up.

In paper II, one man (368) died of heart failure 12 months after the knee operation.

In paper III, one woman (668) was reoperated 9 months after the index operation due to patellar luxation and instability of the knee joint.

In paper IV, 2 patients refused to attend the 2 year scintimetric examination.

Control group with normal knee joints (V, VI, VII):

The *healthy knees* of 23 patients (20 men, 3 women; median age 24 years, range 18–40 years) with old ruptures of the anterior cruciate ligament in the opposite knee constituted the control group in papers V and VI. Ten patients were examined during knee flexion, and 13 during extension (Kärrholm et al 1988, Jonsson et al 1989).

In paper VII, the control group consisted of the *healthy knees* of 14 subjects (10 men, 4 women; median age 25 years, range 17–39 years) with old ruptures of the anterior cruciate ligament in the opposite knee (Jonsson and Kärrholm 1992).

Table 1. Number of patients operated on with one or two types of knee prostheses in studies I–VII

Tricon–M	37
Tricon–M + Tricon–M Central Tibial Stem	2
Tricon–M + Miller–Galante	3
Tricon–M Central Tibial Stem	17
Miller–Galante	26
New Jersey LCS	5

Table 2. Diagnoses, type of fixation, gender and age of the **patients** (*knees*) in studies I–VII

Study	I				II		III		IV		V, VI			VII	
Diagnoses/ Fixation	OAc	OAuc	RAc	RAuc	RAc	RAuc	OAc	OAuc	OA	RA	Tricon	M–G	LCS	Tricon	M–G
Men	**1** (*1*)	**2** (*2*)	**3** (*3*)	**1** (*1*)	**4** (*4*)	**1** (*1*)	**0**	**3** (*4*)	**3** (*3*)	**3** (*3*)	**1** (*1*)	**1** (*1*)	**2** (*2*)	**0**	**2** (*3*)
Women	**13** (*13*)	**8** (*9*)	**8** (*8*)	**6** (*6*)	**8** (*8*)	**6** (*8*)	**15** (*17*)	**11** (*13*)	**15** (*15*)	**12** (*12*)	**10** (*10*)	**8** (*9*)	**3** (*3*)	**6** (*6*)	**4** (*7*)
Mean age	69	70	68	65	62	68	72	72	70	67	68	72	70	69	73
Range	60–78	63–77	55–78	36–79	41–80	62–81	59–78	63–84	60–78	36–79	62–74	69–78	66–78	67–74	69–76

OAc = arthrosis cemented; OAuc = arthrosis uncemented; RAc = rheumatoid arthritis cemented, RAuc = rheumatoid arthritis uncemented
Tricon = Tricon–M; M–G = Miller–Galante; LCS = New Jersey LCS

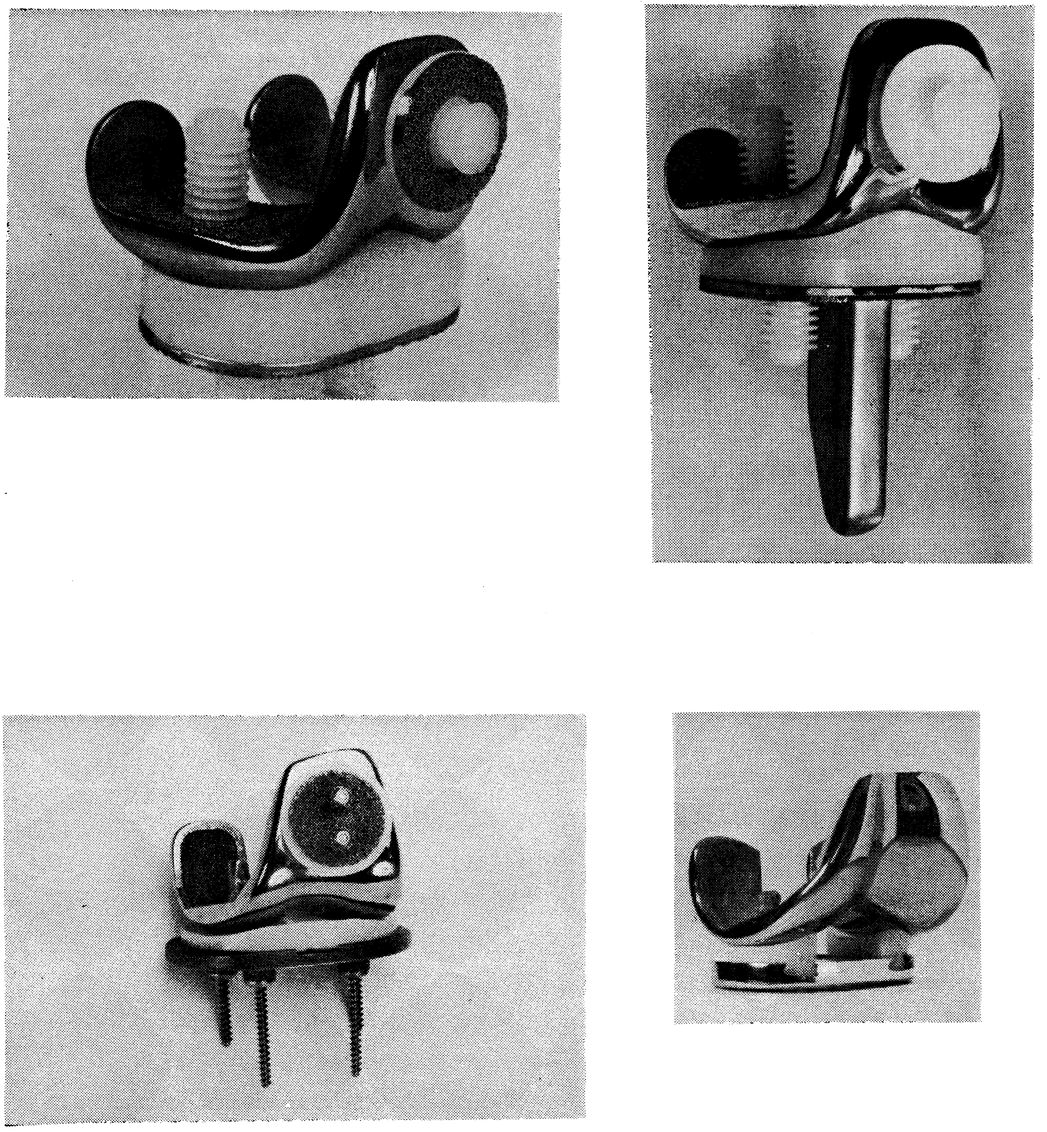


Figure 1. A) *Top left:* Tricon–M prosthesis, **B)** *Top right:* Tricon–M Central Stem Tibial prosthesis, **C)** *Bottom left:* Miller–Galante prosthesis (uncemented version), **D)** *Bottom right:* New Jersey LCS prosthesis.

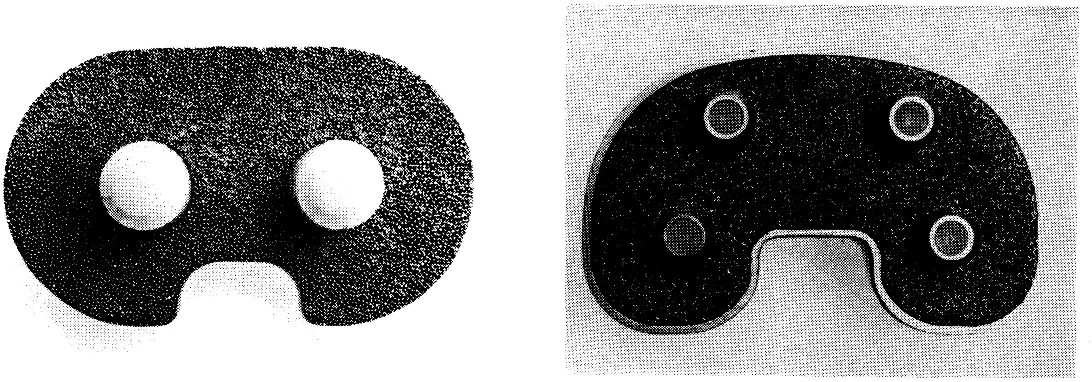


Figure 2. Tibial component undersurface of the Tricon–M prosthesis (*left*), and the Miller–Galante prosthesis (*right*).

Knee prostheses used:

1. Tricon–M: (Fig. 1 A)

The Tricon–M is a tricompartmental knee replacement with cobalt–chrome–molybdenum alloy metallic components and ultra high molecular weight polyethylene components (Laskin 1991) and relatively conforming articular surfaces. The tibial component is equipped with a multilayered surface of sintered cobalt–chrome beads with an average pore size of 250 μm. Two flanged plastic pegs emerge from the undersurface to gain purchase into the metaphysis of the proximal tibia (Fig. 2). The tibial component has a built–in posterior slope of 10° in the sagittal plane. The articular surface has asymmetric medial and lateral facets, both slightly concave in the medial–lateral and anterior–posterior directions. A slight elevation between the two facets may prevent medial or lateral subluxation (Fig. 3 B).

The femoral component has also asymmetrical condylar surfaces, the medial being larger than the lateral one. The articular surfaces are convex both in the medial–lateral, and anterior posterior planes. In the sagittal plane, the radii of curvature is larger distally and smaller anterodistally and posterodistally, giving the prosthesis a "square" appearance (Fig. 3 A).

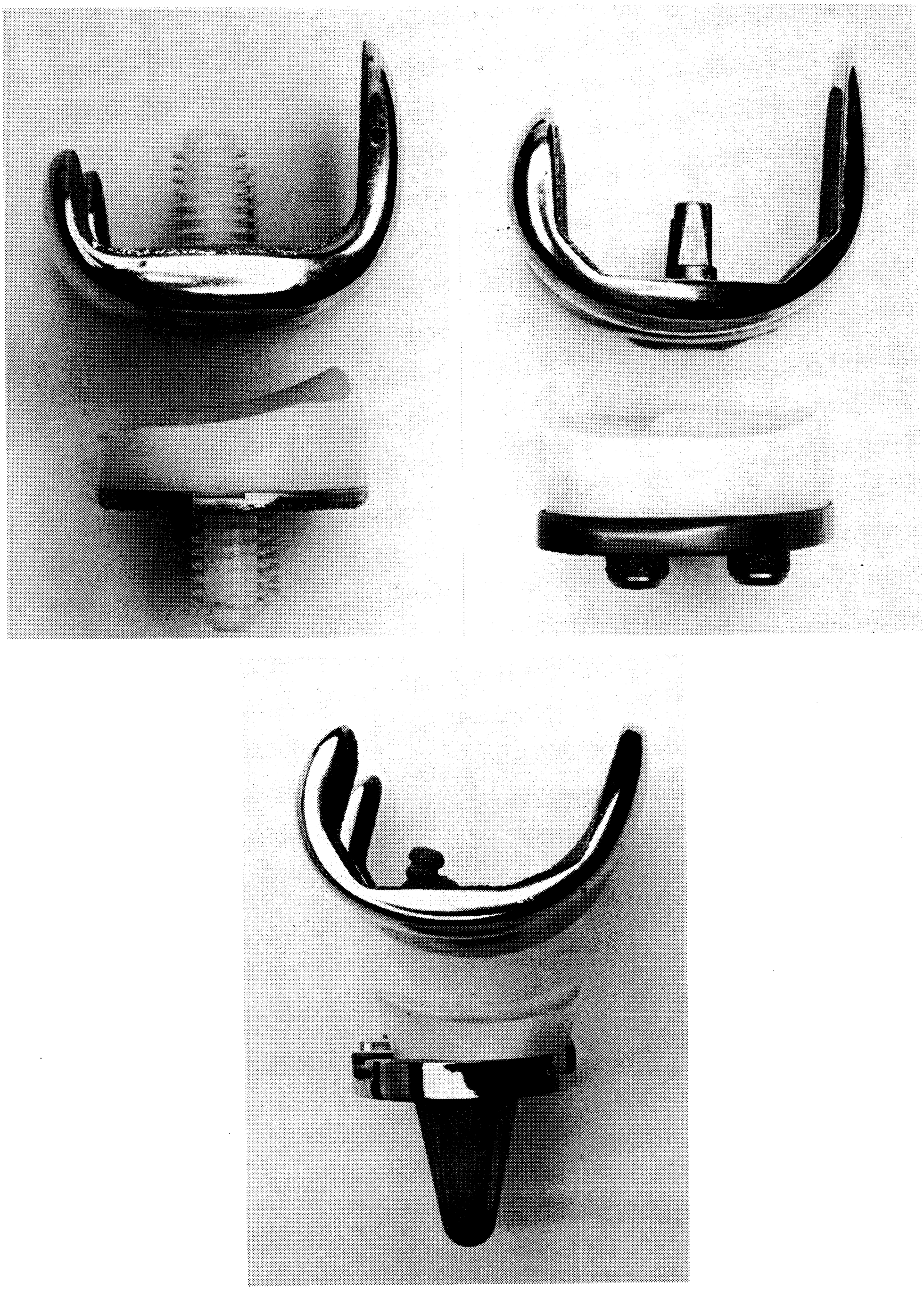


Figure 3 A. The articular surfaces in the sagittal plane of the prosthetic designs studied in the kinematic investigations. *Top left:* Tricon–M, *Top right:* Miller–Galante, *Bottom:* New Jersey LCS.

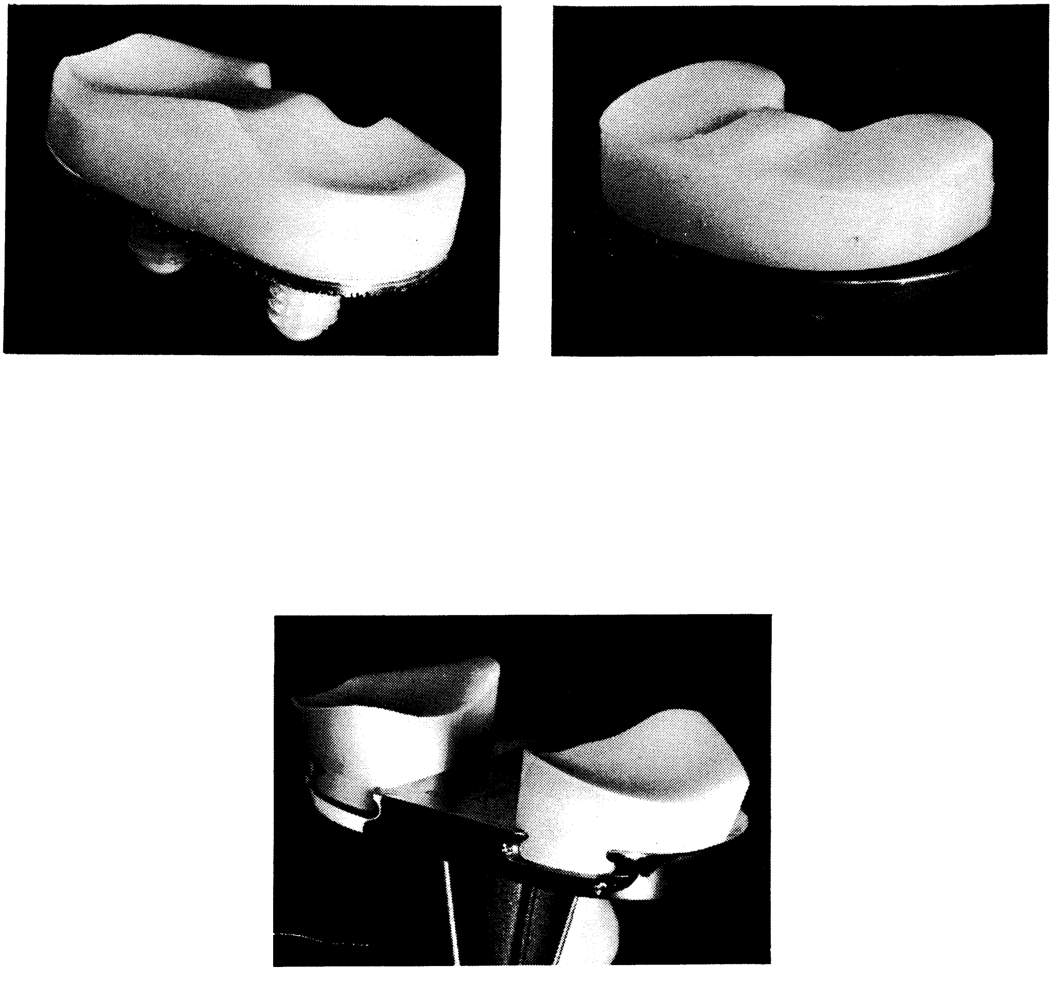


Figure 3 B. The articular surfaces of the tibial components. *Top left:* Tricon–M, *Top right:* Miller–Galante, *Bottom:* New Jersey LCS.

The Tricon–M prosthesis is relatively constrained. The combination of the flanged pegs (used many years in uncemented knee arthroplasties, Blaha et al 1982, Freeman et al 1985) and a porous coated undersurface was regarded an interesting design for uncemented fixation. This prosthesis was therefore chosen for the studies on migration (I) and kinematics (V, VII).

The patella was resurfaced in all but 4 knees using a metal–backed patellar prosthesis.

2. Tricon–M Central Stem Tibial Prosthesis: (Fig. 1 B)
The tibial component is identical to the Tricon–M design as regards articular surface, porous coated metal–backed undersurface, and flanged polyethylene pegs. In addition, it is also equipped with an 8 cm central I–beam metallic stem, bevelled posteriorly to facilitate approximation to the inner posterior wall of the proximal tibia (Laskin 1991). It is intended to be used with the standard Tricon–M femoral component.

To investigate if the addition of a stem to the tibial component improved the fixation in patients with assumed inferior bone quality, this prosthesis was used in patients with RA (II).

All knees operated on with this prosthesis were resurfaced with a patellar prosthesis, 16 of which were metal–backed, and 5 were all plastic.

3. Miller–Galante: (Fig. 1 C)
The Miller–Galante prosthesis used in this Thesis is an unconstrained tricompartmental knee replacement. The tibial base plate is made of titanium–vanadium–aluminium alloy with four circular pegs on the undersurface. The pegs and the undersurface is covered with a fiber mesh made of commercially pure titanium (Galante et al 1971, Landon et al 1986) (Fig. 2). When inserted uncemented, 4 cortical screws, 3 or 4 cm long, were introduced through the pegs and into the tibial metaphysis.

The tibial component is inserted with a 10° posterior tilt. It has comparatively planar articular surfaces with a small anterior lip (Fig. 3 B). There is a slight central elevation with rather sharp edges. The joint area of the femoral component is almost flat in the frontal plane, making the contact area between the femoral and tibial components close to a straight line. In the sagittal plane the femoral component is comparatively flat distally with increasing curvature anteriorly and posteriorly, giving it a "square" appearance (Fig. 3 A). The femoral condyles diverge posteriorly, allowing increased medial–lateral translations with increasing knee flexion.

The M–G knee represents a relatively unconstrained design. The titan fiber–mesh coated undersurface in combination with 4 pegs and screws was thought of as an interesting improvement in uncemented tibial component fixation, at least in bone of presumed good quality. Therefore, patients with OA were operated on and the migration (III) and the kinematics (VI, VII) were studied.

All knees were resurfaced with a patellar prosthesis, 32 were metal–backed, and 2 all plastic.

4. New Jersey LCS: (Fig. 1 D)
The New Jersey LCS is an unconstrained total knee arthroplasty. The tibial component is inserted with a 10° posterior slope, and has two dovetailed tracks, guiding two mobile polyethylene bearings, which are congruent with the joint area of the femoral component (Buechel and Pappas 1989) (Fig. 3 B). In the sagittal plane, the femoral condyles have posteriorly decreasing radii of curvature (Fig. 3 A), and are also convex in the frontal plane.

The LCS knee represents an alternative solution of obtaining a relatively unconstrained prosthesis without reducing the contact area between the femoral and tibial components. Its kinematics was studied in paper VI.

Patellar prostheses were not inserted.

Methods

Clinical evaluation

All the studies on prosthetic fixation (I – IV) were prospective with regular follow-ups. The patients were assessed preoperatively, and at 6, 12, and 24 months postoperatively using the Hospital for Special Surgery (HSS) knee score (Ranawat and Shine 1973). In paper I and III the patients were followed during 2 years, and in paper II during the first postoperative year. In the studies of knee kinematics (V – VII), the patients were clinically evaluated at the time for the kinematic investigation.

In the HSS system pain (at weight–bearing and at rest) is assigned a total maximum of 30 points, function (walking, stairclimbing, transferring ability) 22 points, range of motion 18 points, and 10 points each for muscle strength, flexion deformity, and instability. Subtractions are made for use of external support, extension lag and leg deformity. Maximum score is 100 points, 85–100 is regarded excellent, 70–84 good, 60–69 fair, and < 60 poor (Insall 1984).

Randomization

In the studies on prosthetic fixation, the mode of fixation was randomized. Patients born on an uneven date received uncemented fixation, and those born on an even date received cemented fixation. It was not necessary to change the mode of fixation in any of the patients.

In order to evaluate the influence of diagnosis on the fixation, the patients in paper I were stratified depending on diagnosis (OA and RA), and randomization was then performed within each stratum.

Surgical Technique

In Umeå all of the knees were operated on by, or under the supervision of, two surgeons performing about half of the operations each. A tourniquet was used routinely. The knees were approached through a straight midline or slightly medial skin incision, and medial parapatellar or midline capsular incision. The posterior cruciate ligament was sacrificed in the knees receiving the Tricon–M prosthesis (I), whereas it was retained in the knees receiving the Tricon–M Central Stem and Miller–Galante prostheses (II, III). The tibial components were chosen to cover the cut proximal tibia as completely as possible. In 4 knees (960, 971, 986, 367) a bone transplantation was performed to reconstruct defects of the medial or lateral tibial plateau (Windsor et al 1986)

The uncemented prostheses were fixed with press fit, and in the Miller–Galante knees 4 cortical screws were used in addition to obtain purchase to the tibial metaphysis.

When cement was used, the cut surfaces were brushed, irrigated, thoroughly cleaned, and dried. The cement was mixed manually without vacuum for 2 minutes and was then finger–packed on to the bone and the tibial prosthesis. The tibial component was hammered against the bone with the impactor and thereafter compressed against the femoral trial component with the straight leg raised 50° while the cement was setting. In paper II and III Palacos with Gentamicin was used (Schering Corp), whereas in paper I Palacos with or without Gentamicin was used, depending on type of operation theatre (with or without laminar airflow).

The day after the operation, straight–leg–raising exercises were begun. Continuous passive motion exercises were started the first or second postoperative day. At the first or second postoperative day, the patients were allowed to ambulate with two crutches, touching the floor with the operated leg as much as they needed.

Conventional radiography

1. Alignment of the leg and the prosthetic components

The Hip–Knee–Ankle (HKA) angle (Rydberg 1983) was determined pre– and postoperatively. The alignment of the separate prosthetic components relative to the long axis of the bone was measured on the AP and lateral view of the first postoperative examination (Fig. 4).

2. Joint line

The joint line position before and after surgery was determined according to Figgie et al (1986).

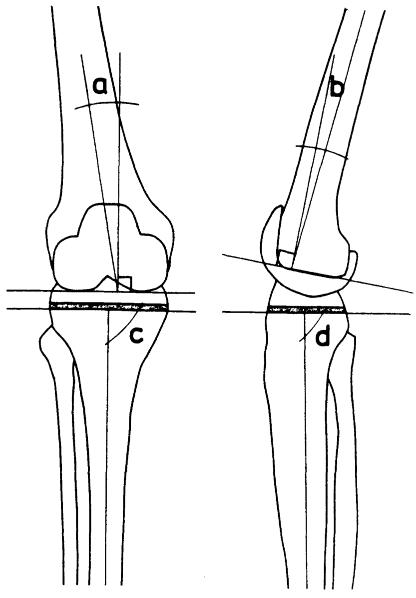


Figure 4. Reconstructed lines used to measure the alignment of the prosthetic components. Two midpoints between the mediolateral and anteroposterior cortex of the distal tibia and the proximal femur were used to reconstruct the center lines.

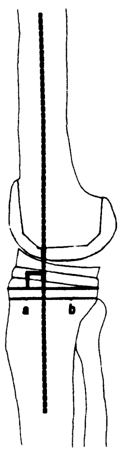


Figure 5.

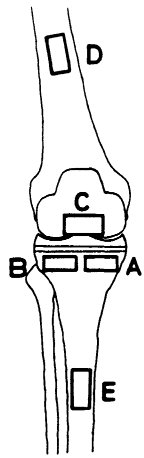


Figure 6.

Figure 5. The anteroposterior position of the tibia in relation to the femur was measured at the reference examination. The reconstructed center line of the femoral diaphysis divides the tibia into two parts. The ratio a/(a+b) x 100 was determined.

Figure 6. Five regions of interest were outlined on the scintigraphs. Areas A and B were used to evaluate the activity under the tibial component. The activity in areas C, D, and E was determined to locate the most suitable reference region.

3. AP–position of tibia in relation to femur (V – VII)
In the kinematic investigations, the AP–position of the tibia in relation to femur was measured on the lateral radiograph of the reference examination (Fig. 5). In study V and VI the magnification factor was determined by dividing the measured AP length of the tibial component by their actual length. In study VII the procedure was changed and simplified by using the known magnification of the cage markers. The subsequent calculations were based on almost constant distances between the film, the knee under study and the lateral x–ray focus.

Scintimetry

Scintimetric examinations were done 3, 6, 12 and 24 months postoperatively. The bone scans were obtained 3 to 4 hours after intravenous administration of 550 M Bq of ^{99}Tcm–MDP with a gamma camera (General Electric 535) using a low–energy general–purpose collimator connected to a computer system (Digital Equipment PDP 11/34 gamma–11). Knee images were recorded in the anterior projection, as this was the routine at the Department of Nuclear Medicine at the time for the investigation.

At the scintimetric analysis, 5 regions of interest (ROI) were outlined on the scans (Fig. 6). To determine the optimal location of the ROI planned to be used as reference, 2 ROI:s were outlined corresponding to the ipsilateral distal femoral and proximal tibial diaphyses, respectively, and one ROI in front of the metallic part of the femoral component. The activity under the tibial component was evaluated using one medial and one lateral ROI in the proximal tibial metaphysis. Counts per pixel for each ROI were recorded. The activity under the tibial component was expressed as a ratio of the chosen reference region.

Roentgen Stereophotogrammetric Analysis (RSA)

RSA according to Selvik (1974, 1989), including the modifications made by Söderkvist (1990) and Nyström (1990) was used to record micromovements of the tibial component and the kinematics of the knee. During the time period for

Table 3. Number of knees investigated with RSA examinations in studies I–III.

Time after surgery		<= 1 week	6 weeks	3 months	6 months	12 months	24 months
Tricon–M (I)	OAc	14	7	10	13	14	12
	OAuc	11	5	8	10	9	11
	RAc	11	5	8	10	10	9
	RAuc	7	6	7	7	7	7
	Total	43	23	33	40	40	39
Tricon–M Stem (II)	RAc	12	9	10	12	10	
	RAuc	8[a]	6[a]	8[a]	8[a]	8[a]	
	Total	20	15	18	20	18	
Miller–Galante (III)	OAc	17	10	17	16	17	16
	OAuc	17	10	17	17	15	16
	Total	34	20	34	33	32	32

OAc = arthrosis, cemented. OAuc = arthrosis, uncemented. RAc = rheumatoid arthritis, cemented. RAuc = rheumatoid arthritis, uncemented
[a]Further one uncemented knee was radiographed, but could not be stereoradiographically evaluated due to poor positioning of the tibial markers.

this investigation, the soft–ware was continuously modified and improved to facilitate the evaluations.

In the studies on prosthetic micromovements (I – IV), RSA was performed within one week after the operation and thereafter at 6 weeks and 3, 6, 12, and 24 months postoperatively. The time interval between the RSA examinations, and the number of knees examined at each time are shown in Table 3. Except for the losses previously accounted for, a few further investigations could not be analyzed due to poor radiographic visualization of the tantalum markers at those occasions. Due to lack of resources, it was not possible to examine more than about half of the patients at the 6 weeks follow–up.

The kinematic studies (V, VI, VII) were done 7 months to 5 years after the operation.

RSA involves the following steps:

1. Implantation of the tantalum markers

Before the insertion of the prosthesis, five to nine 0.8 mm tantalum markers were inserted into the proximal tibial metaphysis, using a hypodermic needle and a trocar. The markers were spread apart as much as possible in the bone. The polyethylene part of the tibial component was supplied with four to six 0.5 and 0.8 mm tantalum markers. The Tricon–M components were also supplied with 2 markers in each plastic peg. The markers were inserted after predrilling into the polyethylene.

Tantalum markers were also inserted into the distal femoral metaphysis in a number of the Tricon–M and Miller–Galante knees to prepare for the subsequent kinematic analyses.

In the normal knees used as control population in the kinematic analyses, tantalum markers were inserted percutaneously at the time for arthroscopy of the contralateral and injured knee (Kärrholm et al 1988b, Jonsson et al 1989).

2. Radiographic examination

a. Prosthetic micromovements (I–III)

The knees were examined with the patient supine and with the knee inside a Plexiglas calibration cage, supplied with tantalum markers with known 3–D coordinates in all four walls. The x–ray tubes were placed perpendicular to each other, with a film–focus distance of 100 cm. Anteroposterior and lateral radiographs were exposed simultaneously. By examining the knee inside the calibration cage, the calibration of the stereo set–up was performed simultaneously with the patient examination. At the first postoperative RSA examination, the knee was examined i maximum extension with the tibia aligned with the longitudinal axis of the laboratory coordinate system, the posterior edges of the tibial prosthesis parallel to the transverse axis, and perpendicular to the longitudinal axis. The sagittal axis of the knee was perpendicular to both the transverse and longitudinal axes. The position of the knee at this examination was called tibial reference position and was identified in all knees to assure a standardized orientation of the knee, necessary for a proper comparison of three–dimensional prosthetic motions between the different knees. Two or three pairs of radiographs were obtained at each occasion to determine the accuracy of the measurements.

b. Studies of the non weight–bearing kinematics (V – VI)

The examinations were accomplished with two film exchangers placed perpendicular to each other. Before examination of a patient, a calibration cage defining the laboratory coordinate system was radiographed with two reference plates firmly fixed in front of each film exchanger. The cage was removed, and a fixed position of the roentgen foci and the film exchangers was assured throughout the examinations. The knee was first radiographed in the relaxed and extended position with the patient supine (reference examination). At this examination the distal femur was positioned with the longitudinal axis of the bone parallel to the longitudinal axis of the cage and the posterior vertex of the femoral condyles parallel to the transverse axis. All subsequent knee positions during flexion/extension were related to this relaxed and extended position (femoral reference position, Kärrholm et al. 1988b).

Active flexion was studied in the prone position, and extension in the sitting position (Fig. 7). A weight of 15 N or 30 was attached to the ankle of the control persons in the flexion and extension series, respectively. In the prosthetic patients the weight was standardized to 15 N, because some of the patients could not preform an even moment using the heavier weight. The patients performed a number of trial flexions and extensions before the serial exposures were made. Flexion was studied from a position of maximum extension. The radiographic set–up did not allow studies of knee flexion past 55°. When examining knee extension, the knee was flexed as much as the

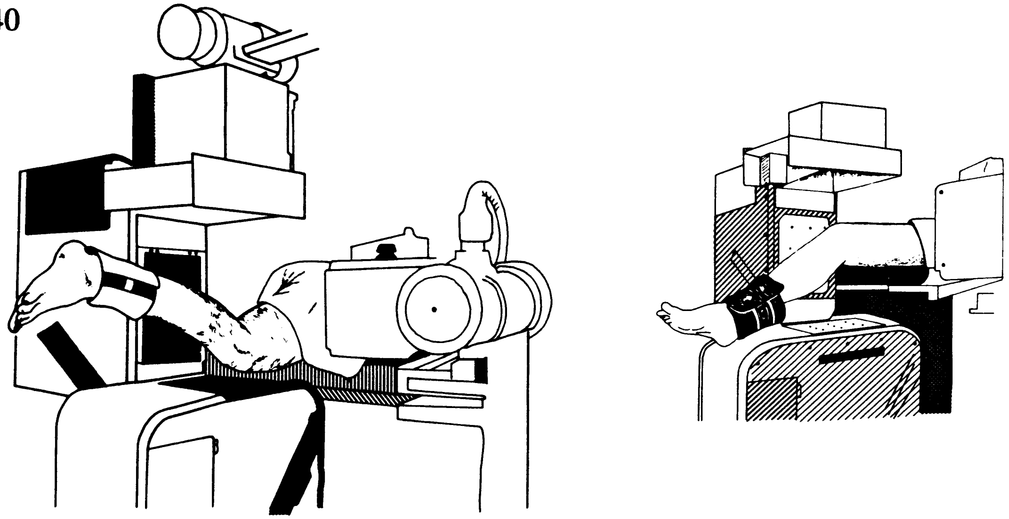


Figure 7. The radiographic set–up at the examination of the non weight–bearing kinematics. Two x–ray tubes were used for simultaneous exposure. One film exchanger was placed below, and the other on the medial or lateral side of the knee. One reference plate supplied with tantalum markers was attached in front of the grid of each film exchanger. Knee flexion was studied in the prone position (*left*), and extension in the supine or sitting position (*right*). The weight was placed at the level of the ankle of the patient.

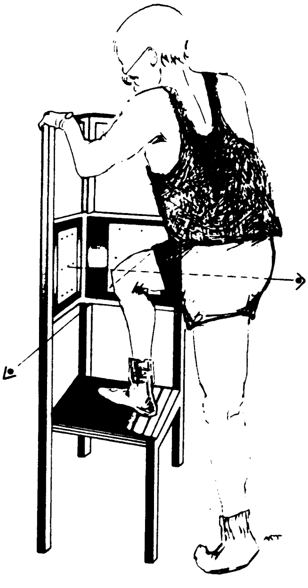


Figure 8. The radiographic set–up at the examination of the weight–bearing kinematics. At the examination the patient was ascending a platform, the starting position is illustrated. Before this examination, the calibration cage and the two perpendicular reference plates were radiographed.

radiographic set–up allowed (mean value 35°). The tibia was pushed backwards by the examiner and a pair of radiographs was exposed. Thereafter, active extension and the serial exposures were started. An average of 10 pairs of radiographs were obtained for each series.

Because the initiation of the knee movements and the start of the film exchangers could not be exactly synchronized, data were not collected at the same degree of flexion/extension in each patient. To standardize the measurements, the recorded rotations and translations for every patient were plotted and interpolated at intervals of 5°.

To reduce the effect of missing observations at the beginning and end of the flexion/extension series, the statistical analysis included the intervals with all observations available. In the flexion series these intervals were: 15° to 40° of flexion (Tricon–M) and 5° to 50° (M–G and LCS). The corresponding intervals in the extension series were 20° to 10° of extension (Tricon–M), 30° to 20° (M–G), and 20° to 15° (LCS)

c. Studies of the kinematics during weight–bearing (VII)

Stereoradiographs were exposed when the patients ascended a platform about 40 cm above the floor using the knee under study (Fig. 8). The platform was attached to a metal frame supplied with one adjustable cassette–holder designed for two film cassettes placed perpendicular to each other. One Plexiglas reference plate supplied with tantalum markers was attached in front of each film cassette. After adjustment for the leg length of the individual patient, the cassette–holder and the x–ray tubes were fixed. A calibration cage was exposed together with the reference plates before the patient examination. The patient placed the foot in a neutral position on the platform (about 90–100° of knee flexion), pressing it against the floor of the platform corresponding to the first stereoradiographic exposure. Thereafter the examination proceeded with the patient weight–bearing on one leg and at decreasing amount of flexion of the knee. Four to five exposures were obtained in each knee.

Before the weight–bearing examination took place, the knee was radiographed without weight–bearing corresponding to the femoral reference position (Kärrholm et al 1988b).

The tibial rotations and the translations of the knee center were plotted in relation to the degree of knee flexion in each knee. The recorded values were interpolated at intervals of 20°.

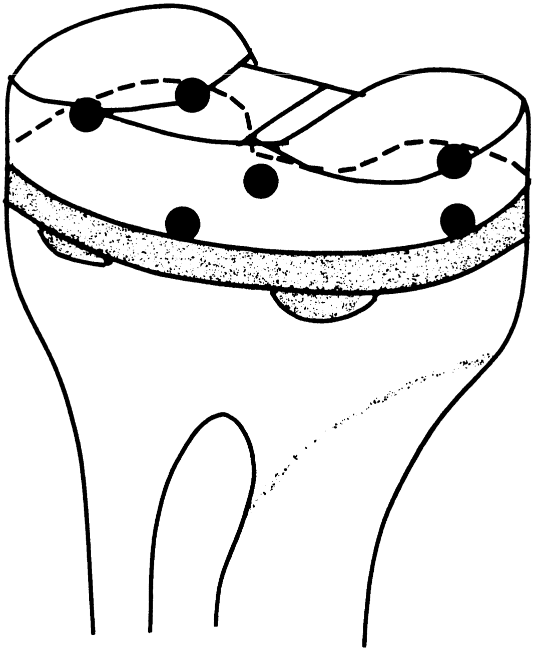


Figure 9. Six fictive points were plotted on the anteroposterior and lateral radiographs of the reference examination in all knees to obtain standardized recordings of translations.

Statistical comparison between the normal and prosthetic knee movements when measured in relation to the body fixed axes was performed only between 60 and 20 degrees of flexion when observations from all knees were available.

3. Reconstruction of anatomical landmarks

In order to obtain standardized points for measurements of the translations (I – III), 6 fictive points were plotted on the AP and lateral views of the reference radiograph in each patient; 5 points at the prosthetic edge and one at the center of the tibial component (Fig. 9). The two–dimensional positions of these points were measured, enabling calculation of their three–dimensional positions in the RSA program **Xray90**. By using the three–dimensional location of the tantalum markers in the tibial component, the positions of the plotted standardized points could be transformed to their corresponding positions at the subsequent examinations using the RSA program POINT TRANSFER. These calculations were based on an acceptable rigidity and configuration of the polyethylene markers as verified in the initial kinematic evaluation (see 5a, below).

In the kinematic analyses (V – VII) a corresponding fictive point was plotted at the center of the tibial joint area. Because the polyethylene markers could not always be visualized, the tibial bone markers were used in the transformation, presuming that comparatively small movements occurred between the tibial center and the bone during knee motion. In the normal knees a corresponding plotting and transformation of the two tips of the tibial intercondylar eminence was performed (Kärrholm et al 1988a, Jonsson et al 1989).

When determining the positions of the screw axes (VII), eight positions corresponding to radiographically visible femoral landmarks were plotted on the stereoradiographs of the extended reference position. Further one point, used as origo in the subsequent evaluations, was reconstructed based on observations of ten cadaver knees. In these knees the anterior cruciate ligament (ACL) was removed and the midpoint and periphery of its femoral insertion was marked with tantalum balls. Each cadaveric knee was radiographed inside the calibration cage with the same orientation as at the reference position in the patient examinations. From these radiographs the mean relative position of the femoral insertion of the (ACL) in the frontal and sagittal planes was calculated. These data were used to plot the corresponding position of the femoral insertion of the ACL on the two radiographs of the unloaded extended knee (= femoral reference position) in each patient (Jonsson and Kärrholm 1992). To facilitate the reconstruction of the screw axes geometry, the origo of the laboratory coordinate system was positioned at the actual or previous femoral insertion of the ACL.

4. Identification of markers and measurements on the radiographs

The images of the cage and patient markers were numbered according to a standardized pattern. The two–dimensional coordinates of the images were digitized and thereafter stored using a measuring table (Wild Autograph A7Z, Heerbrugg, Switzerland) connected to a computer.

5. Mathematical calculations

The following software for the mathematical calculations (Nyström 1990, Söderkvist 1990) were used:

> **Xray90** calculates the three–dimensional coordinates of the markers from the measured two–dimensional film coordinates.

KINERR detects and corrects erroneous identification and coupling of the markers between x–ray focus 1 and 2.

SEGMENT MOTION calculates the three–dimensional motions of rigid bodies. The calculations consists of two steps: 1) Determination of the rigidity of the tantalum markers representing each bone segment (see below), and 2) Analysis of absolute and relative movements.

POINT MOTION calculates the three–dimensional translations of one or several points. In the first step the rigidity and absolute motions of the reference or fixed segment is calculated. In the second step the relative translations of the markers in the moving segment are computed.

a. Determination of the mean error of rigid body fitting

The markers in each segment (i.e. tibial prosthesis and proximal tibia [I – IV], or proximal tibia and distal femur [V – VII]) were modelled as rigid bodies. The stability of the markers implanted is determined by calculation of the mean error (ME) of rigid body fitting (Selvik 1974, 1989, Ryd 1986) The smaller the value the better the stability of the markers between two evaluations. If one or more tantalum marker moves or is poorly defined, the distorsion of the segment will be detected by an increasing ME. The soft–ware enables setting of a maximum value of ME and excludes loose markers that may distort the rigidity. Provided that at least three markers have sufficient stability to fulfil the requirements of the chosen mean error the calculations are completed.

In addition, the configuration of each segment was determined by calculating its condition number (Nyström and Söderkvist 1990). The condition number decreases with increasing scattering of the markers implanted.

A mean error of rigid body fitting of 0.235 or less (mean 0.066, range 0.011–0.221) and a condition number of 100 or less (mean 32, range 18–100) was used in the present studies (I–VII).

b. Analysis of movements

Movements were calculated as rotations about, and translations along the three coordinate axes, or a single axis called the screw axis. The absolute and relative tibial component (I–III) or tibial (V–VII) movements between two examinations were computed using the proximal tibia (I–III) or the distal femur (V–VII) as the fixed reference segment. The relative movements of a rigid body (segment motion) was only accounted for in terms of rotations (I–VII) or translations along the helical axis (VII). Translations of fictive points (point motion) represented the translatory migration (I–IV). In RSA, rotations are calculated in a sequential order in relation to the laboratory coordinate system (Euler angles). In this study this is equivalent to the following order: rotations about the transverse (x), longitudinal (y), and sagittal (z) axes.

6. Presentation of the RSA calculations

a. Analysis of tibial component micromotion

Rotations of the entire tibial plate and translations of the standardized points at the prosthetic edge and center were recorded.

Rotations were expressed as rotations about the transverse (x–axis), longitudinal (y–axis), and sagittal (z–axis) axes. Since rotations can have both positive and negative values depending on the direction of rotation, the magnitudes of the recorded rotations were expressed as the absolute values.

Translations were measured as: 1) Proximal or distal translation of the standardized points at the prosthetic edge, corresponding to subsidence and lift–off. The largest values recorded, irrespective of location, was denoted maximum subsidence and lift–off; 2) Distal translation of the center of the prosthesis, corresponding to midpoint subsidence; 3) *Total point motion*, which is the length of the three–dimensional translation vector for each point of measurement. Maximum Total Point Motion (MTPM, Ryd 1986) denotes the total point motion of the standardized point that moved the most, and minimum total point motion denotes the translation of the point that moved the least.

b. Analysis of knee kinematics

In the kinematic analyses, the three–dimensional movements of the tibia in relation the femur during flexion and extension of the knee was determined using femur as the fixed reference segment. Rotations of tibia in relation to the laboratory coordinate system (Euler angles, Selvik 1974, 1989) were computed in the following order flexion/extension, internal/external rotation, and adduction/abduction. Tibial rotations during flexion–extension were presented as internal/external rotation and adduction/abduction. Translations of the center of the knee/tibial component were presented as medial/lateral, proximal/distal, and anterior/posterior translations.

With the aid of the plotted point coordinates, the projection of the distal femur and the screw axes in the frontal and horizontal planes were reconstructed. Further, the point of intersection of the screw axes with a sagittal plane through the femoral insertion of the ACL was determined. The recorded movements were separated into six 20° intervals of knee flexion, and the representative axis or point for each interval was plotted on a drawing of a standardized knee. These plottings were condensed by calculating the mean position of the helical axes in relation to anatomical landmarks or their intersections with the plane chosen at each interval.

7. Reproducibility

To determine the reproducibility of the method, repeated examinations were performed the same day. In the studies on prosthetic micromotion, 76 and 102 double examinations were done, respectively (I, III). The standard deviations of the differences from zero were calculated, and the 99 % confidence limit was used to represent significant micromotion. This calculation included the technical accuracy and any inducible displacement unintentionally caused by the examiner or the patient between the examinations (Table 4).

The accuracy of the knee kinematics were determined by repeating the examination i 3 normal knees (Kärrholm et al 1988b) and in 3 prosthetic knees (V). Significant motions were expressed at the 95% confidence limit (Table 4).

To estimate the accuracy of screw axis positioning, a computer simulation was performed based on the marker positioning (condition number) and the stability (ME of rigid body fitting) observed in the patients. Landmark positions were sampled from a uniform distribution in the range 0 – 45 mm, corresponding to a mean condition number of 59.5 (patient examinations 61.3). The perturbations of the landmarks were sampled using an unbiased normal distribution (SD 0.05 mm) causing a mean value for the mean error of rigid body fitting of 0.09 mm (patient examinations 0.088 mm)

Statistics

The statistical methods used in this Thesis are indicated in *italics* for each calculation performed.

Table 4. The accuracy of the RSA measurements. Significant translations and rotations at $P < 0.01$ (prosthetic micromotion) and $P < 0.05$ (knee kinematics)

		Translation (mm)			Rotation (degrees)		
	Number of double examinations	Med/Lat (x–axis)	Prox/Dist (y–axis)	Ant/Post (z–axis)	Transverse (x–axis)	Longitudinal (y–axis)	Sagittal (z–axis)
Micromotion							
Tricon–M	76	0.3	0.2	0.3	0.6	0.4	0.6
Miller–G	102	0.3	0.2	0.4	0.35	0.3	0.3
Knee kinematics							
Tricon–M	3	0.2	0.5	0.5		1.5	0.3
Normal knees[1]	3	0.3	0.8	1.2		1.1	0.3

[1]Kärrholm et al 1988

Summary of papers I–VII and Results

I. Evaluation of micromotion in cemented *vs* uncemented knee arthroplasty in osteoarthrosis and rheumatoid arthritis

Randomized study using roentgen stereophotogrammetric analysis

Patients and preoperative data

Forty–two patients (43 knees) operated on with the Tricon–M prosthesis were studied during a period of 2 years. The distribution after randomization is shown in Table 2.

The preoperative clinical and radiographic data did not differ between cemented and uncemented fixation in neither the RA nor OA patients (*Kruskal–Wallis H–test*). The patients with RA had significantly lower weight and HSS score than the patients with OA ($P < 0.01$ and $P < 0.05$, respectively, *Wilcoxon rank sum test*).

Clinical and radiographic results

In all the groups the HSS knee score increased during the follow–up period (6 months: 72–78; 24 months: 84–88; $P < 0.05$ *signed rank test*). At 6, 12, and 24 months the clinical results did not differ in terms of total HSS score, distribution of excellent and good results, and pain score during walking and at rest (*Kruskall–Wallis H–test*). Neither did the mean postoperative joint line position, the postoperative HKA–angle, and the alignment of the separate prosthetic components differ significantly between the groups (*Kruskal–Wallis H–test*).

One patient (958) developed radiographic evidence of loosening of the patellar component 15 months after the operation. In addition she had pain in the proximal tibia when walking. RSA at 18 months, 3 months prior to the reoperation, revealed that the tibial component had rotated 3° outwards and tilted 1.5° anteriorly, corresponding to a MTPM of 2.5 mm (mean value cemented group at 2 years: 1.3 mm) and midpoint subsidence of 0.8 mm. Histologic investigation of the tibial cement–bone interface revealed highly differentiated fibrocartilage.

RSA results

Most of the micromotion (rotations and translations) occurred during the first 3 to 6 months regardless preoperative diagnosis and type of fixation. The uncemented RA knees seemed to stabilize after 6 months, whereas the other groups displayed a tendency to continuous micromotion.

At 2 years the mean rotations about the transverse axis were 0.7° to 1.2°. Mean internal–external rotations of 0.6° to 0.7° were recorded after 6 months. In three of the groups these values did not change during the subsequent 18 months. In the fourth (uncemented OA), the rotations continued during the whole observation period and reached 1.2° at 2 years. The majority of the tibial components (29 of 36) displayed an external rotation in relation to the proximal tibia. Rotations about the sagittal axis reached 0.5° to 0.7° at 2 years.

The tilting in the sagittal- and frontal planes corresponded to either subsidence of one side and lift–off of the opposite one, **or** subsidence of the entire component (Figs. 10, 11). The mean maximum subsidence in the frontal plane (medial or lateral edge) was –0.7 to –0.9 mm, and in the sagittal plane (anterior or posterior edge) –0.6 to –1.0 mm. The mean maximum subsidence (irrespective of location) was –0.8 to –1.0 mm, and the mean maximum lift–off was 0.2 to 0.6 mm. The MTPM reached 1.0 to 1.5 mm after 2 years (Fig. 12).

None of the parameters evaluated (translations or rotations) differed significantly between the cemented and uncemented tibial components in the two groups of diagnoses (*repeated measurements ANOVA*). Neither did the prosthetic micromovements differ significantly between the RA and OA patients (*Wilcoxon rank sum test*).

There were no significant correlations between micromotion and pre– and postoperative knee alignment (HKA–angle) (*nonparametric correlation*). In three of the groups there were no significant correlations between prosthetic component alignment and micromotion. However, in the uncemented OA group there was a positive correlation between alignment of the tibial

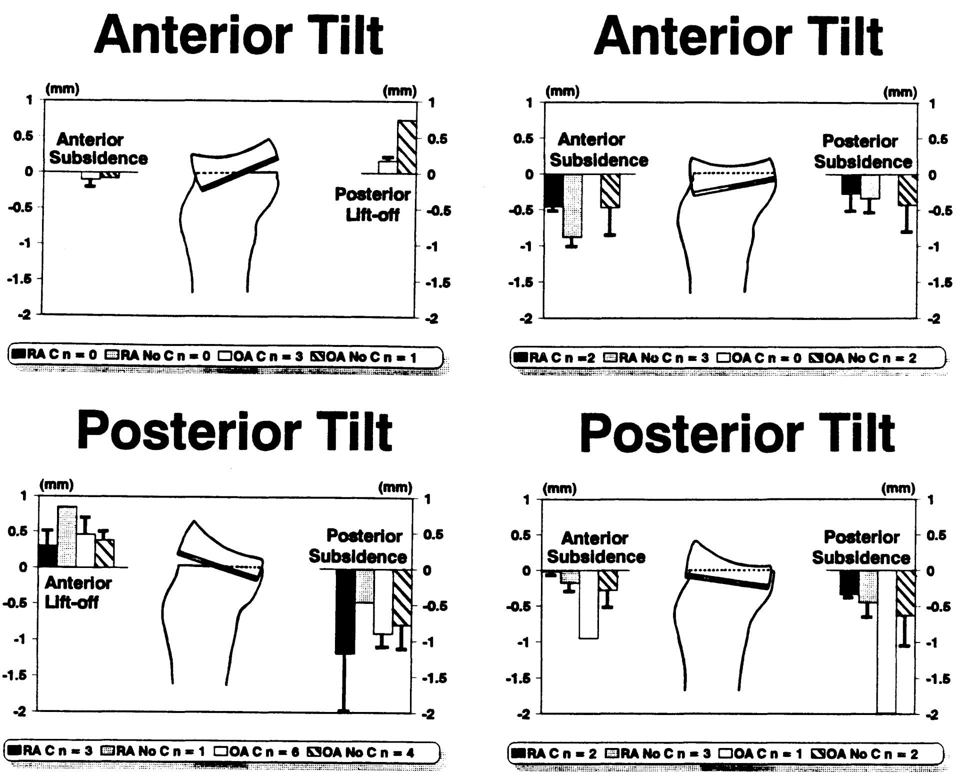


Figure 10. Tricon–M knees (I): Subsidence and lift–off of the tibial components in the sagittal plane at 2 years postoperatively (mean, SEM). Bottom line of each drawing: Diagnosis (RA or OA); Type of fixation (C = cement, No C = no cement); Number of observations (n).

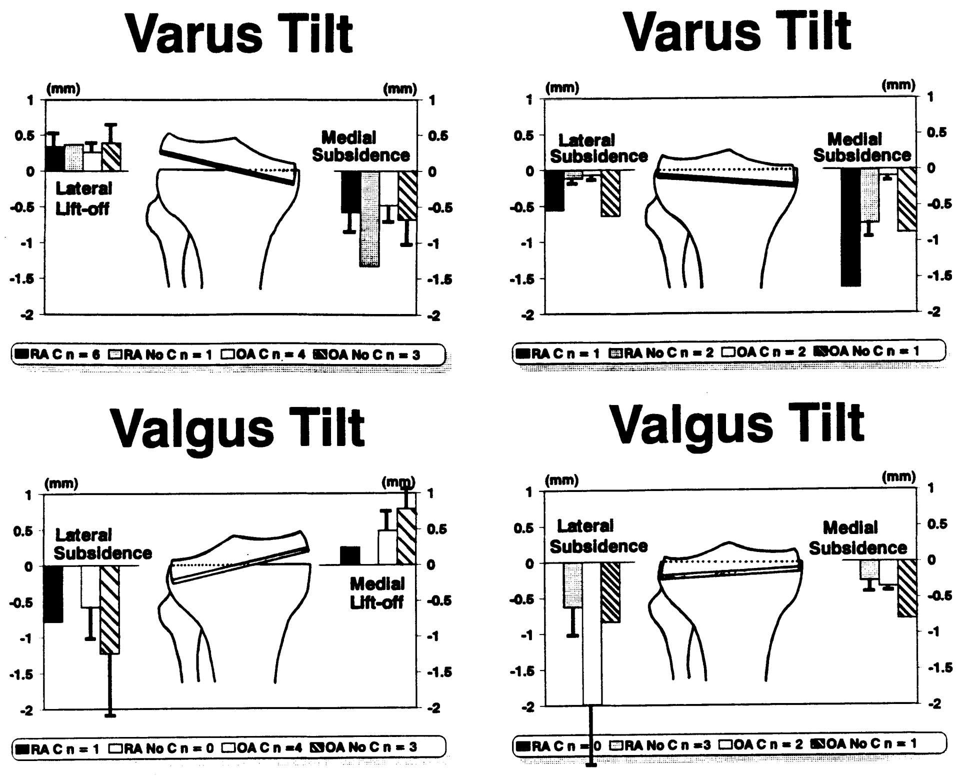


Figure 11. Tricon–M knees (I): Subsidence and lift–off of the tibial components in the frontal plane at 2 years postoperatively (mean, SEM). See also legend to Fig. 10.

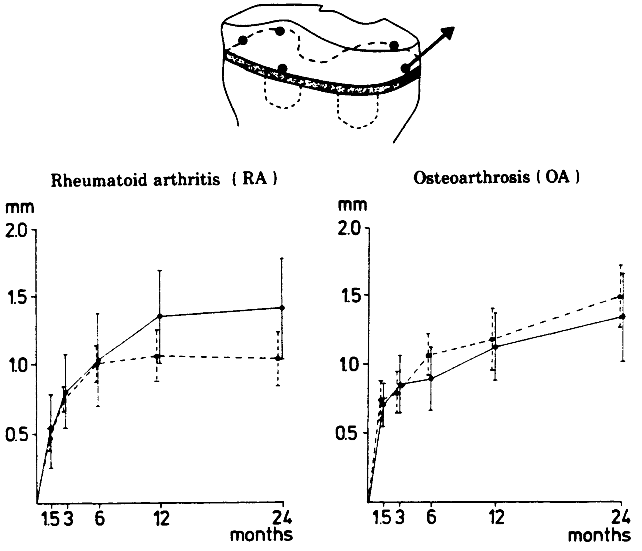


Figure 12. Maximum migration (MTPM) of the Tricon–M prosthesis in the four groups during the 2 postoperative years (mean, SEM). Solid lines, cemented prostheses; dashed lines, uncemented prostheses. Cemented *vs* uncemented: NS (*repeated measurements ANOVA*).

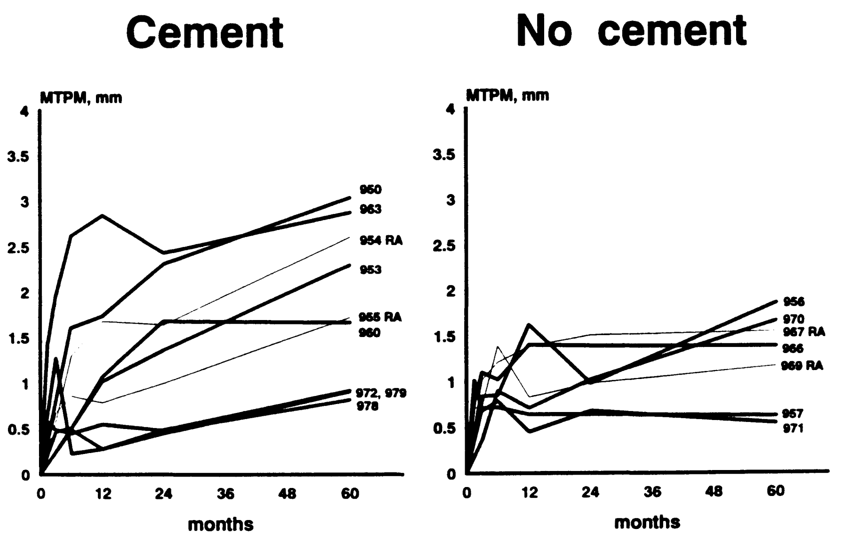


Figure 13. Maximum migration (MTPM) for each of the Tricon–M knees followed for 5 years. Cemented (*left*), and uncemented knees (*right*). Thin lines, RA knees. Between 2 and 5 years, two knees (1 cemented, 1 uncemented) were reoperated, one due to aseptic loosening and the other due to tibial component failure.

component in the sagittal plane and the rotations about the transverse axis (r = 0.83, $P < 0.01$, *nonparametric correlation*).

Conclusion

Cement did not improve the fixation of the Tricon–M tibial component in terms of migration. The preoperative diagnosis had no influence on the early micromotion of the implants.

5 year results

Out of the patients operated in Umeå, 18 knees have so far been examined with RSA at 5 years. Two of these could not be fully evaluated due to inaccurate radiographs.

Between the 2– and 5–year follow–ups, two patients (959, 968) had died of unrelated causes. One patient (964) had been revised 59 months postoperatively due to fracture of the tibial component in combination with subsidence and loosening. RSA could not be performed prior to revision because the polyethylene had worn down posteriorly and the remaining tantalum markers in the tibial component could not be visualized. This knee had displayed a MTPM of 2.9 mm at the 2–year follow–up. Another knee (954) had had a patellar revision 43 months postoperatively because of loosening. Finally one patient (965) refused to attend the 5–year follow–up. On inquiry by telephone she reported no pain from her operated knee.

The mean HSS knee scores were 83 and 85 in the cemented and uncemented groups, respectively (NS, *Wilcoxon rank sum test*). All uncemented knees were painfree when walking, whereas 4 cemented knees (950, 954, 955, 960) complained of pain during walking.

All but 1 of the cemented knees displayed increasing migration between 2 and 5 years (Fig. 13). The mean migration (MTPM) increased from 1.3 (± 0.8) mm (mean, SD) at 2 years to 1.9 (± 0.9) mm at 5 years ($P < 0.01$, *signed rank test*). Continuously increasing anterior/posterior tilt and internal/external rotation were the dominant types of micromotion recorded. In contrast, most of the uncemented knees did not migrate further between 2 and 5 years (mean values: 1.0 ± 0.3 and 1.3 ± 0.5 mm; NS, *signed rank test*).

The migration (MTPM) in the cemented knees with pain (n = 4) was 2.3 (± 0.7) mm, and in those without pain (n = 5) 1.6 (± 1.0) mm (NS, *Wilcoxon rank sum test*).

II. Increased subsidence of uncemented stemmed Tricon–M knee prostheses

Roentgen stereophotogrammetric analysis of cemented *vs* uncemented tibial components

Patients and preoperative data

Nineteen patients (21 knees) with RA operated on with the Tricon–M Central Tibial Stem prosthesis were studied during a period of 1 year. The distribution after randomization is shown in Table 2. The preoperative clinical and radiographic data did not differ between the groups (*Wilcoxon rank sum test*).

Clinical and radiographic results

The mean HSS score increased significantly between 6 and 12 months in the cemented group (6 months: 80, 12 months: 89; $P < 0.01$ *signed rank test*), whereas it increased from 70 to 72 in the uncemented group (NS) At 1 year the mean HSS knee score was higher in the cemented group ($P < 0.01$ *Wilcoxon rank sum test*). One patient with bilateral uncemented prostheses suffered from transient hemiparesis due to cerebrovascular ischemia immediately after her second knee operation. She had no pain, but the delayed rehabilitation resulted in inferior knee mobility and thus low knee score. After one year, 10/11 cemented and 7/9 uncemented knees were painfree during walking. The postoperative HKA angle and the alignment of the separate prosthetic components did not differ between the groups (*Wilcoxon rank sum test*).

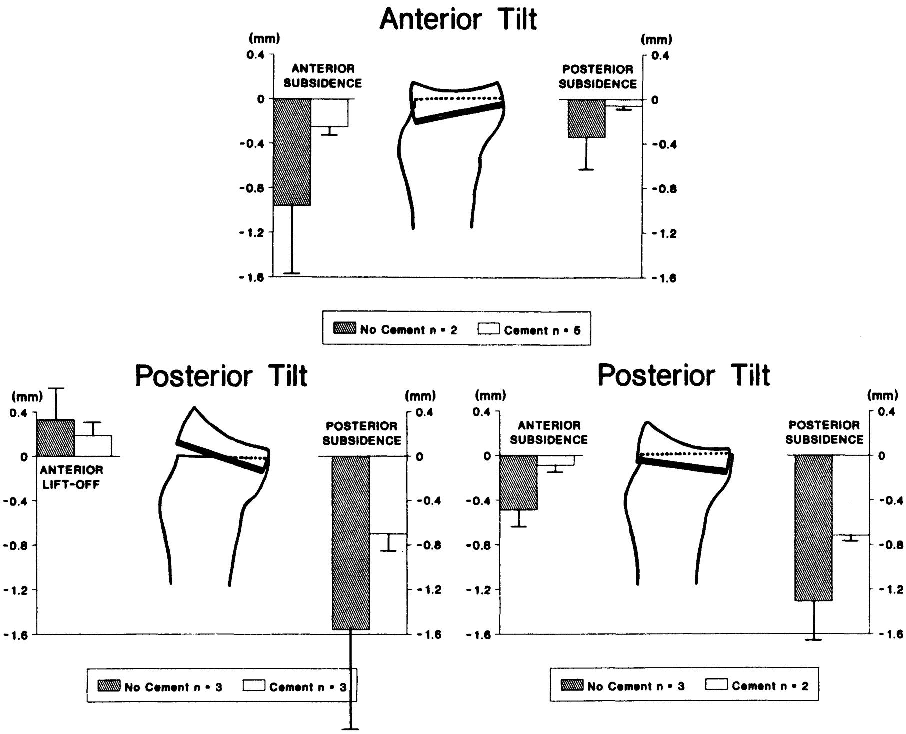


Figure 14. Tricon-M Tibial Stem knees (II): Subsidence and lift-off of the tibial components in the sagittal plane in the cemented and uncemented groups at 1 year postoperatively (mean, SEM). In the cemented knees the largest subsidence was evenly distributed between the anterior and posterior edges. In the uncemented knees 6 of 8 tilted posteriorly.

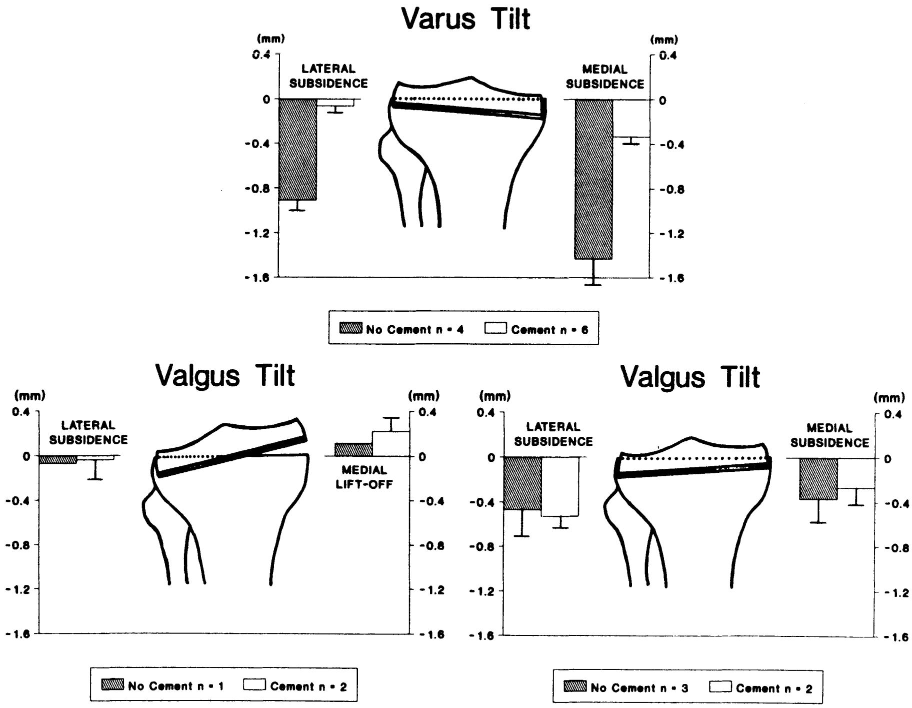


Figure 15. Tricon–M Tibial Stem knees (II): Subsidence and lift–off of the tibial components in the frontal plane in the 2 groups at 1 year postoperatively (mean, SEM). In the cemented knees, varus or valgus tilting implied mainly subsidence of one of the edges and no or minimum movements of the opposite side. In the uncemented knees, tilting was associated with varying amount of subsidence of both edges, which was pronounced when varus tilting occurred.

RSA results

Increasing rotations and translations were recorded in both groups. During the postoperative year the mean rotations about the transverse axis reached 0.5° (± 0.4) in the cemented and 1.4° (± 1.7) in the uncemented group (NS, *repeated measurements ANOVA*). The corresponding rotations about the longitudinal and sagittal axes were 0.3° (± 0.2) and 0.3° (± 0.2) in the cemented group and 0.7° (± 0.7) and 0.3° (± 0.3) in the uncemented group, respectively (NS, *repeated measurements ANOVA*).

Tilting in the sagittal plane corresponded either to subsidence of one side and lift–off of the opposite side, or to pure subsidence of the entire component (Fig. 14). In the cemented knees the largest subsidence was localized anteriorly or posteriorly in 5 prostheses each; in the uncemented 6 of 8 knees tilted posteriorly.

External rotation was found in 5 uncemented and 4 cemented components, whereas internal rotation was found in 3 and 6 knees, respectively.

Tilting in the frontal plane was associated with variable amount of distal movements of the medial and lateral edges of the prostheses (Fig. 15). The pattern of edge subsidence was influenced by the type fixation. In the cemented knees varus or valgus tilting implied mainly subsidence of one of the edges and no or minimum movements of the opposite side. In the uncemented knees, varus or valgus tilting was associated with varying amount of subsidence of both edges, which was pronounced when varus tilting was recorded.

The mean maximum subsidence (irrespective of location) was larger in the uncemented knees ($P < 0.05$ *repeated measurements ANOVA*). At 1 year the cemented knees had sunk −0.5 mm (± 0.3) and the uncemented ones −1.4 mm (± 1.1). The mean maximum lift–off in the cemented knees was 0.2 mm (± 0.2 mm) and in the uncemented knees 0.3 mm (± 0.4) (NS, *repeated measurements ANOVA*).

The MTPM was larger in the uncemented knees (1 year, uncemented: 1.6 [± 1.2] mm; cemented: 0.9 [± 0.5] mm, $P < 0.05$; *repeated measurements ANOVA*) (Fig. 16).

The recorded micromovements were not influenced by pre– and postoperative HKA angle, alignment of the prosthetic components, weight, or pre– and postoperative range of motion (*nonparametric correlation*).

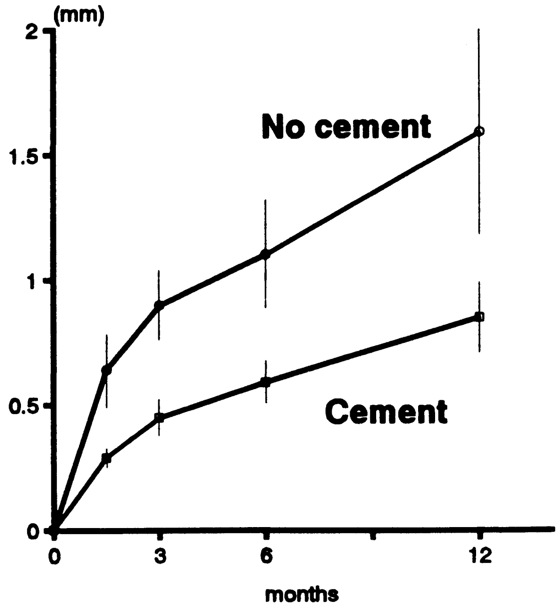


Figure 16. Maximum migration (MTPM) of the Tricon–M Tibial Stem prosthesis during the first postoperative year (mean, SEM). Cemented *vs* uncemented: $P < 0.05$ (*repeated measurements ANOVA*).

Conclusion

Cement improved the fixation of the stemmed Tricon–M tibial component by reducing the amount of subsidence. When used with cement the stemmed Tricon–M tibial component displayed smaller micromovements than was observed in the unstemmed Tricon–M knee (I). The uncemented stemmed component displayed almost twice as large subsidence as was found when no stem was used.

III. Increased varus–valgus tilting of screw fixated knee prostheses

Stereoradiographic study of uncemented *vs* cemented tibial components

Patients and preoperative data

Twenty–nine patients (34 knees) with OA operated on with the Miller–Galante prosthesis were studied during a period of 2 years. The distribution after randomization is shown in Table 2. The preoperative clinical and radiographic data did not differ between the groups (*Wilcoxon rank sum test*).

Clinical and radiographic results

The cemented and uncemented knees had about the same HSS score throughout the observation period. The percentage of good and excellent results in the two groups was about the same at 6, 12, and 24 months (*Fisher's test*). However, the pain while walking seemed to decrease more slowly in the uncemented knees (6 *vs* 24 months; cemented NS, uncemented $P < 0.01$, *signed rank test*).

There were two reoperations during the investigation period. One woman (OA, cement) was reoperated 7 months after the index operation due to patellar luxation and instability of the knee. One woman (OA, no cement) developed increasing medial instability due to malalignment of the femoral component. A Miller–Galante revision femoral component was inserted 12 months after the first operation.

RSA results

In both groups, the rotations and translations were largest during the first 6 weeks to 3 months. In the uncemented knees, the prostheses seemed to stabilize after 3 months, whereas the cemented knees displayed a tendency to continuously increasing micromovements during the investigation period.

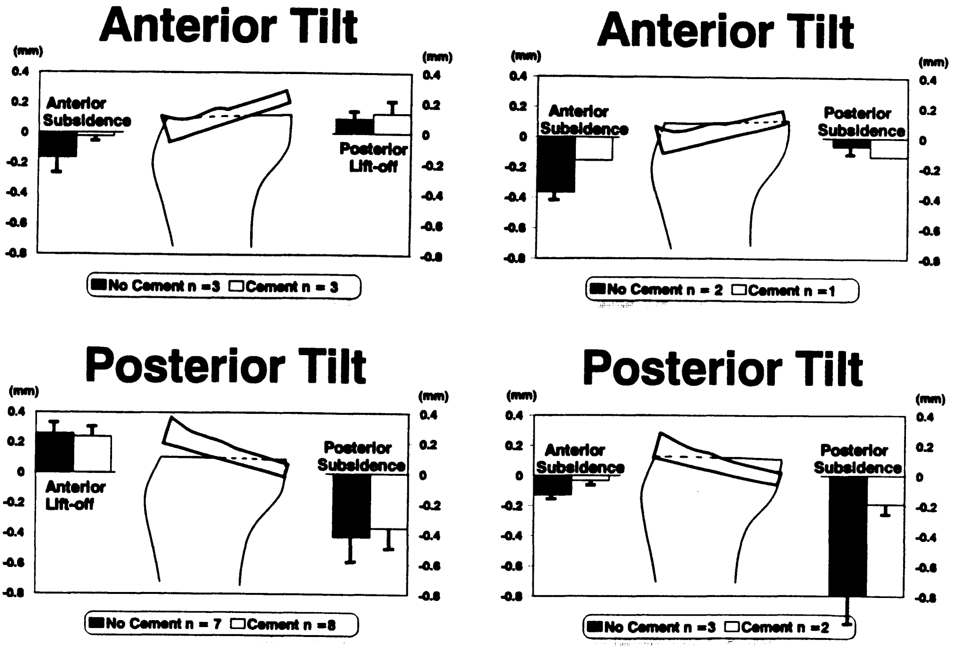


Figure 17. Miller–Galante knees (III): Subsidence and lift–off of the tibial components in the sagittal plane in the 2 groups of fixation at 2 years postoperatively (mean, SEM). The mean maximum subsidence did not differ between the groups (*Wilcoxon rank sum test*).

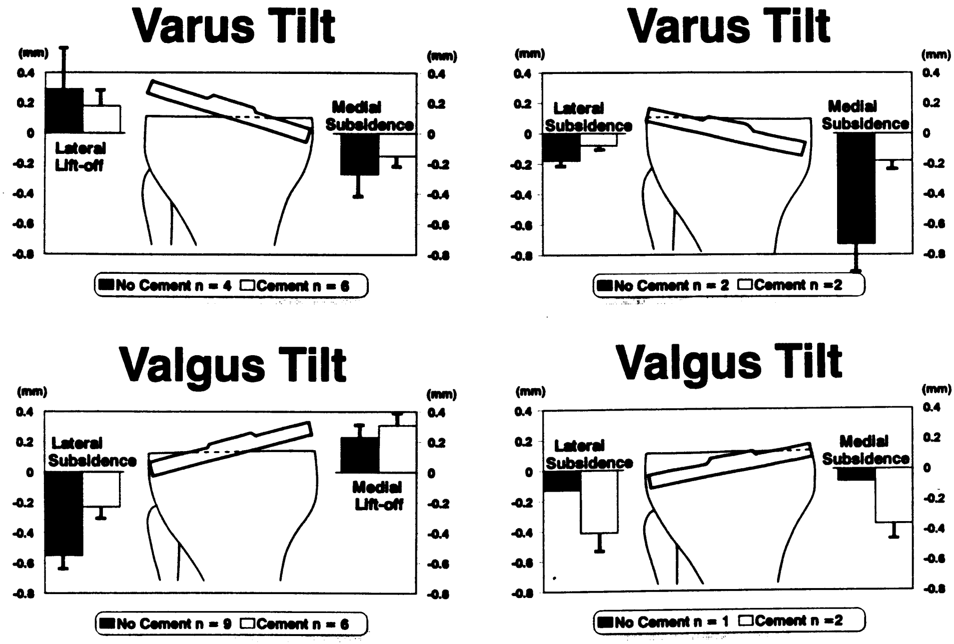


Figure 18. Miller–Galante knees (III): Subsidence and lift–off of the tibial components in the frontal plane at 2 years postoperatively (mean, SEM). The mean maximum subsidence was larger in the uncemented knees ($P < 0.05$, *Wilcoxon rank sum test*).

The mean rotations about the transverse and the longitudinal axes did not differ between the two groups and reached about 0.5°–0.6° and 0.3°, respectively, at 24 months (NS, *repeated measurements ANOVA*).

Larger mean rotations about the sagittal axis were recorded in the uncemented knees as compared to the cemented knees ($P < 0.01$ *repeated measurements ANOVA*). At 24 months the uncemented knees had rotated about twice as much as the cemented ones (cemented 0.3° ± 0.2, uncemented 0.6° ± 0.4).

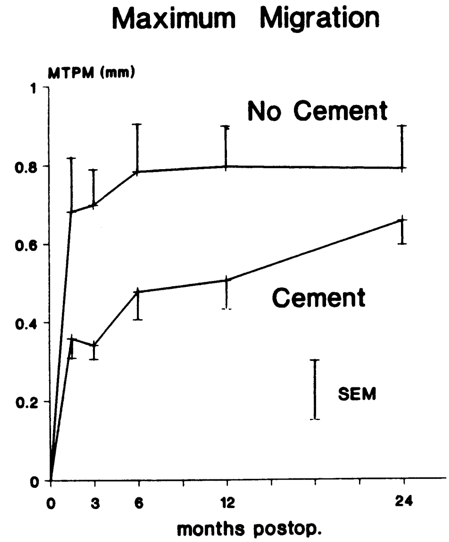
The tilting in the sagittal– and frontal planes corresponded either to subsidence of one side and lift–off of the opposite side, or (more seldom) subsidence of the entire component (Figs. 17, 18). The mean maximum subsidence at 24 months in the frontal plane (medial or lateral edge) was larger in the uncemented (0.5 ± 0.3 mm) than in the cemented knees (0.2 ± 0.2 mm; $P < 0.05$ *Wilcoxon rank sum test*). The mean maximum subsidence in the sagittal plane (anterior or posterior edge) did not differ between the two groups (uncemented 0.4 ± 0.4 mm, cemented 0.3 ± 0.3 mm; NS, *Wilcoxon rank sum test*).

Throughout the follow–up, the mean maximum subsidence of the prosthetic edge, irrespective of location, was about 2 to 3 times larger in the uncemented than in the cemented knees ($P < 0.01$ *repeated measurements ANOVA*), reaching −0.7 (± 0.4) mm and −0.3 (± 0.3) mm, respectively, at 24 months. The mean maximum lift–off, on the other hand, did not differ (uncemented 0.3 ± 0.2 mm, cemented 0.3 ± 0.2 mm; NS *repeated measurements ANOVA*).

In the uncemented knees, the center of the tibial component was the most stable part in 7 knees, and in further 6 knees, maximum stability was found anteriorly or posterolaterally. In the cemented knees, the position of the most stable part was more variable.

The MTPM was larger in the uncemented knees ($P < 0.05$ *repeated measurements ANOVA*) (Fig. 19). In the uncemented knees almost all migration occurred during the first 3 months, and thereafter the movements were minimal (3 – 24 months: NS, *signed rank test*). The cemented knees, on the other hand, displayed initially small but continuously increasing MTPM during the investigation period (3 – 24 months: $P < 0.05$ *signed rank test*).

Three months postoperatively, uncemented knees with preoperative flexion less than 110° displayed larger subsidence posteriorly than those uncemented knees with more than 110° of flexion ($P < 0.05$, *multiple*



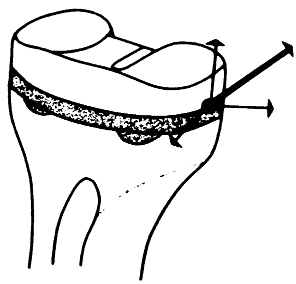


Figure 19. Maximum migration (MTPM) of the Miller–Galante prosthesis during the 2 postoperative years (mean, SEM). Cemented *vs* uncemented: $P < 0.05$ (*repeated measurements ANOVA*).

regression analysis). This difference disappeared at the subsequent examinations.

Uncemented knees with thin (= 8.5 mm) components displayed significantly larger varus–valgus tilting than uncemented knees with thicker components ($P < 0.001$, *multiple regression analysis*) at the 3 months follow-up. This difference disappeared at the subsequent examinations. In the cemented knees, the magnitude of preoperative knee flexion or tibial component thickness did not influence the prosthetic micromotions.

There were no correlations between the micromovements recorded and the postoperative HKA angle, alignment of the prosthetic components, or weight (*nonparametric correlation*). At 2 years, 8/10 uncemented knees with preoperative varus deformity subsided laterally, and the same ratio was found at 3, 6, and 12 months. Uncemented knees with valgus deformity and all cemented knees subsided with an equal distribution between the medial and lateral sides.

Conclusion

The micromotions were in general small. Cement improved the early fixation of the Miller–Galante prosthesis, seemingly reducing the influence of tibial component thickness and bone quality. When used without cement, this prosthetic design displayed increased varus–valgus tilting.

IV. Scintimetry after total knee arthroplasty

Prospective 2–year study of 18 cases of arthrosis and 15 cases of rheumatoid arthritis

Patients

Thirty–three of the patients (33 knees) operated on in Umeå with the Tricon–M prosthesis (paper I) agreed to participate in the scintimetric study. Eighteen knees were cemented (10 OA, 8 RA) and 15 uncemented (8 OA, 7 RA) according to the randomization protocol in paper I.

Clinical results

All but one patient was painfree during the investigation period. In this patient, previously accounted for, the tibial and patellar components were revised 21 months after the index operation.

Scintimetric results

The activity in front of the femoral component was lower than that of the ipsilateral femoral or tibial diaphyses ($0.0005 < P < 0.05$, *ANOVA*) and did not change between 3 and 24 months. This ROI was therefore used as the reference in the subsequent calculations.

In both OA and RA knees the mean activity of the femoral diaphysis decreased between 3 and 24 months, whereas the activity of the tibial diaphysis increased. At 3 months, patients with preoperative varus deformity had higher activity in both the femoral and tibial diaphyses than those with valgus deformity ($P < 0.05$, *Wilcoxon rank sum test*). This difference disappeared at the subsequent examinations. In the total material, the activity under the tibial component decreased from 7.8 ± 2.5 at 3 months to 6.5 ± 1.5 at 24 months ($P < 0.05$, *paired t–test*). Statistical analysis (*ANOVA taking into account effects of*

Table 5. Changes of activity in the tibial metaphysis (A+B/C) 3 to 24 months postoperatively (**mean**, *SD*)

Category	Months postop. 3		6		12		24	
Preop. varus deformity (*n 16*)	**9.13**	*2.13*	**8.54**	*2.36*	**7.74**	*2.61*	**6.94**	*1.39*
	$P < 0.05$[a]		NS[a]		NS[a]		NS[a]	
Preop. valgus deformity (*n 13*)	**6.33**	*2.0*	**7.44**	*2.49*	**6.73**	*1.75*	**6.23**	*1.73*
OA (*n 18*)	**8.50**	*2.29*	**8.43**	*1.83*	**7.45**	*2.43*	**6.92**	*1.41*
	NS[a]		NS[a].		NS[a]		NS[a]	
RA (*n 15*)	**7.12**	*2.62*	**6.98**	*2.86*	**7.18**	*2.06*	**6.10**	*1.67*
Cemented knees (*n 18*)	**8.89**	*2.63*	**8.21**	*2.33*	**7.85**	*2.43*	**6.95**	*1.55*
	NS[a]		NS[a]		NS[a]		NS[a]	
Cementless knees (*n 15*)	**6.89**	*2.09*	**7.28**	*2.49*	**6.70**	*1.88*	**6.19**	*1.54*
"Unstable" (significant rotations)	**8,66**	*2.26*	**8,12**	*2.56*	**7.64**	*2.17*	**6.84**	*1.45*
	$P < 0.05$[b]		NS[b]		$P < 0.05$[b]		$P < 0.05$[b]	
"Stable" (insignificant rotations)	**5,07**	*1.15*	**6,94**	*1.81*	**5,92**	*2.14*	**4.78**	*1.07*
Case 5 (revised due to loosening)			**11.7**		**10.5**		**8.9**[c]	

[a] Analysis of variance taking into account effects of synergy
[b] Logistic regression analysis
[c] At 21 months, prior to revision

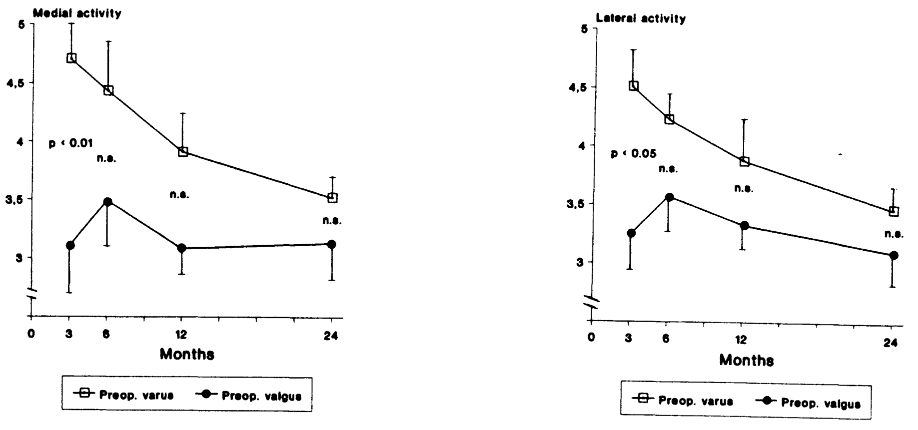


Figure 20. Scintimetric activity (mean, SEM) under the tibial components in preoperative varus knees (*ruta*) and valgus knees (*fylld ring*), (*Wilcoxon's rank sum test*). *Left:* Activity in the medial condyle, *Right:* Activity in the lateral condyle.

synergy) revealed that the increased activity at 3 months could only be explained by the preoperative knee alignment, but not by diagnosis, mode of fixation, or weight (Table 5). At 3 months, patients with a varus deformity had 1.4 times higher activity under the tibial component than those with a valgus deformity. The activity was increased both medially and laterally (Fig. 20).

Scintimetry and micromotion

There were no numeric correlations at any time between the scintimetric activity and the translatory micromotions or single axis rotations of the tibial components. Tibial components displaying significant rotations (exceeding the 95 percent confidence level of the RSA) about any of the three coordinate axes were classified as **unstable**, and those displaying smaller rotations were classified as **fixed**. At 3 months 19 out of 24 tibial components were unstable and at 24 months 24 out of 28. The activity under the tibial component was higher in the knees with unstable components than in those that were stable (Table 5). However, at 3 months patients with preoperative varus deformity had numerically the highest activity. Increasing magnitude of tibial component rotation between 3 and 24 months was not associated with a corresponding increase in the scintimetric activity.

Conclusion

The choice of location of the ROI to be used as the reference is critical, because the activity in the ipsilateral femoral and tibial diaphyses changes with time after operation, and the activity in the corresponding contralateral diaphyses may be influenced by any progression of the usually present arthrosis or subsequent knee replacement surgery.

During the 2 postoperative years, the scintimetric activity under the tibial component is significantly influenced by the type of preoperative deformity, but not by diagnosis, mode of fixation or weight.

At 2 years postoperatively, low scintimetric activity indicates small tibial component rotations, whereas high activity indicates prosthetic instability **or** a high bone remodelling for some other reasons.

V - VI. Knee motion in total knee arthroplasty

A roentgen stereophotogrammetric analysis of the kinematics of the Tricon–M knee prosthesis

Abnormal kinematics of the artificial knee

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

Patients

Twenty–five patients (26 knees) with three designs of knee prostheses were studied 7 to 36 months after the operation (Table 2). At surgery, the posterior cruciate ligament (PCL) had been resected in the Tricon–M knees, and retained in the M–G and LCS knees. The preoperative diagnoses were OA in 22 patients and RA in 3 patients (all with Tricon–M prosthesis).

Twenty–three *healthy knees* (20 men, 3 women; mean and median age 24 years, range 18–40 years) with old ruptures of the anterior cruciate ligament on the *opposite side* were used as controls.

Results

Kinematics of knee flexion

Internal/external rotation (Fig. 21): The Tricon–M and LCS knees displayed no or minimum internal rotation during flexion, whereas the M–G knees displayed a pattern more similar to that of the normal knees. Statistical analysis (all calculations in V–VI: *repeated measurements ANOVA*) revealed no significant differences between the prosthetic and normal knees.

Adduction/abduction (Fig. 22): The Tricon–M knees displayed an initial tendency to abduction which increased after 25° of flexion (Tricon–M *vs* normal: $P < 0.05$). A similar tendency was noted in the MG knees during the first 20° of flexion, but thereafter minimum adduction was recorded (M–G *vs* normal: $P < 0.05$). The LCS knees showed a pattern closer to the normal knees (LCS *vs* normal: NS).

Medial/lateral translation (Fig. 23): The center of the Tricon–M tibial components displayed no medial–lateral translation during the first 30° of flexion; thereafter an increasing lateral translation was observed (Tricon–M *vs* normal: $P < 0.05$). In both the M–G and LCS knees, there was an increasing medial translation of the tibial plateau center with increasing flexion. However, this medial translation was not as pronounced as in the normal knees (M–G *vs* normal: $P < 0.01$; LCS *vs* normal: $P < 0.05$).

Proximal/distal translation (Fig. 24): The translations of the Tricon–M knees did not significantly differ from the normal ones (Tricon–M *vs* normal: NS). During the first 20° of flexion, the center of the M–G knees displayed distal displacement of about 1 mm, and thereafter proximal translation, reaching a mean of 6 mm at 50°. The LCS knees displayed an increasing proximal translation during flexion, reaching 12.5 mm at 50° (M–G and LCS *vs* normal: $P < 0.01$).

Anterior/posterior translation (Fig. 25): In all three designs of prostheses, increasing posterior translation of the tibial plateau center was observed with increasing angle of knee flexion. The differences between the prosthetic and normal knees were 16–18 mm at 50° (MG and LCS *vs* normal: $P < 0.0001$; Tricon–M *vs* normal: $P < 0.0005$).

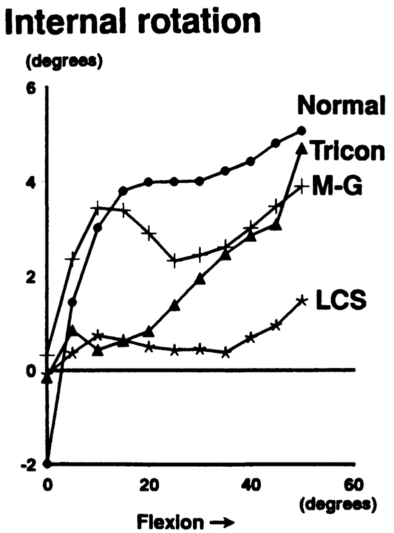


Figure 21.

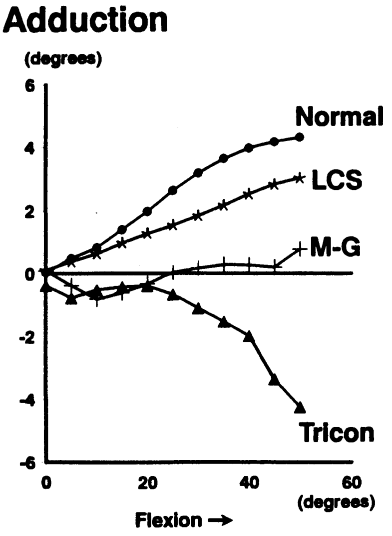


Figure 22.

Figure 21. Tibial rotations (mean values) about the longitudinal axis (internal/external rotation) occurring during knee flexion without weight-bearing. Abbreviations Figs. 21–25: M–G = Miller–Galante knees, LCS = New Jersey LCS knees, Tricon = Tricon–M knees, Normal = normal knees. For numbers of observations, see text.

Figure 22. Tibial rotations (mean values) about the sagittal axis (adduction/abduction) during knee flexion. (See also legend to Fig. 21).

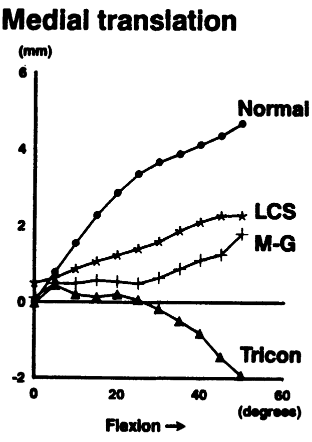


Figure 23.

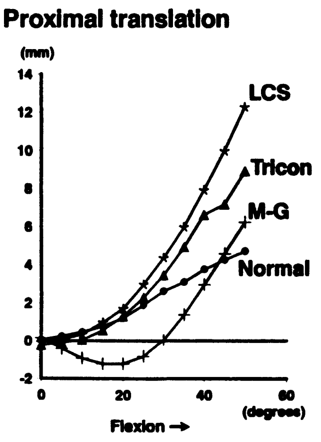


Figure 24.

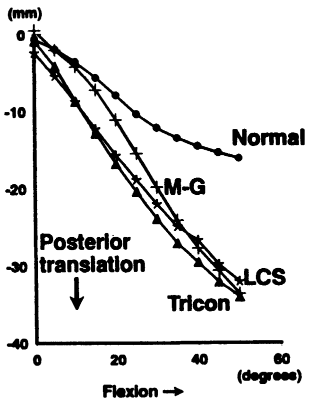


Figure 25

Figure 23. Translation of the center of the tibial plateau (mean values) in the prosthetic knees, and the tibial intercondylar eminence in the normal knees along the transverse axis (medially/laterally) during flexion. (See also legend to Fig. 21).

Figure 24. Tibial translations (mean values) along the longitudinal axis (proximally/distally) during flexion. (See also legend to Fig. 21).

Figure 25. Tibial translations (mean values) along the sagittal axis (anteriorly/posteriorly) during flexion. (See also legend to Fig. 21).

Kinematics of knee extension

There were, in general, small differences in the motions recorded between knee flexion and knee extension in each of the prosthetic designs. The M–G knees were somewhat more medially translated during extension than during flexion, and the LCS knees displayed somewhat less proximal and posterior translations of tibia during extension as compared to flexion ($P < 0.05$, *paired t–test*).

Conclusion

All three designs of knee prostheses examined displayed abnormal kinematics *in vivo*. The less constrained designs tended to display smaller deviations from the normal. Increased posterior translations occurred in all three designs questioning the function of the posterior cruciate ligament in two of the designs, where it had been preserved.

VII. Kinematics of successful knee prostheses during weight–bearing

Three–dimensional movements and positions of screw axes in the Tricon–M and Miller–Galante designs

Patients

Twelve patients (16 knees) with 2 designs of knee prostheses were studied. Six knees had the Tricon–M and 10 knees the Miller–Galante prostheses (Table 2). Eight of the patients had OA, 2 had RA (Tricon–M), and 2 arthrosis secondary to fracture (Tricon–M). The PCL was absent in the Tricon–M knees, but was retained in the Miller–Galante knees. The examinations were done 7 months to 5 years after the operation, and all patients had a successful outcome at the time for the investigation.

The *healthy knee* of 14 patients (14 knees; 10 men, 4 women; mean and median age 25 years, range 17–39 years) with old ruptures of the anterior cruciate ligament on the *contralateral side* were used as controls.

Results

Cardinal axis rotations

Both the Tricon–M and Miller–Galante knees started at a slightly internally rotated position at 80° of flexion (Fig. 26). The M–G knees displayed a small external rotation during extension, while the Tricon–M knees did not (M–G: $P < 0.01$; Tricon–M: NS, *sign test*). Compared to the normal knees, both designs had a more externally rotated position throughout extension (Tricon–M: $P < 0.05$; M–G: $P < 0.05$, *repeated measurements ANOVA*).

The Tricon–M knees displayed a mean of 11° of adduction at 80° of flexion, which decreased during extension (Tricon–M *vs* normal: $P < 0.05$, *repeated measurements ANOVA*) (Fig. 27). Small adduction–abduction movements were recorded in the normal and M–G knees (M–G *vs* normal: NS, *repeated measurements ANOVA*).

Cardinal axes translations

At 80° of flexion the center of the Tricon–M tibial component was displaced 6 mm laterally, whereas the center of the normal and MG knees were 2.5 to 4 mm medially positioned in relation to the reference position. In the Tricon–M knees a more lateral position was noted between 60 and 20 degrees of extension whereas the MG knees maintained a position more closer normal (Tricon–M *vs* normal: $P < 0.001$; M–G *vs* normal: NS, *repeated measurements ANOVA*) (Fig. 28).

The proximal–distal position of the Tricon–M knees did not differ from the normal knees, whereas the M–G knees tended to have a more distal position (Tricon–M *vs* normal: NS; M–G *vs* normal: $P < 0.05$, *repeated measurements ANOVA*) (Fig. 29).

Increased posterior displacement was recorded at the flexed positions in the prosthetic knees (Tricon–M *and* M–G *vs* normal: $P < 0.001$, *repeated measurements ANOVA*) (Fig. 30). At 80° of flexion the center of the tibial

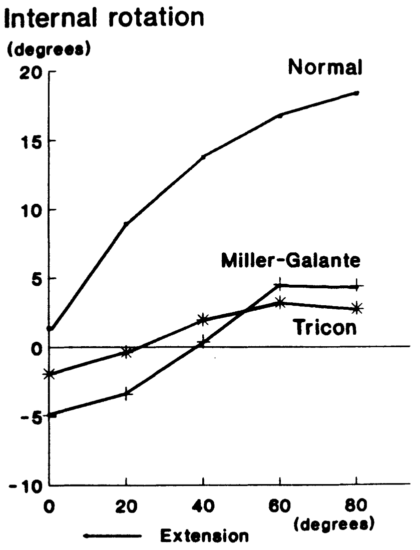


Figure 26.

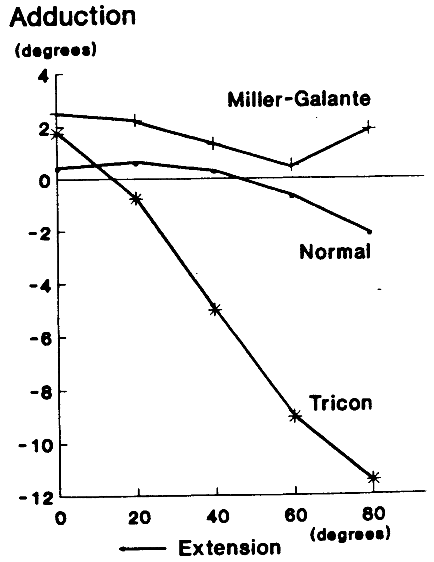


Figure 27.

Figure 26. Tibial rotations (mean values) about the longitudinal axis (internal/external rotation) occurring during knee extension at weight–bearing. For numbers of observations, see text.

Figure 27. Tibial rotations (mean values) about the sagittal axis (adduction/abduction). (See also legend to Fig. 26).

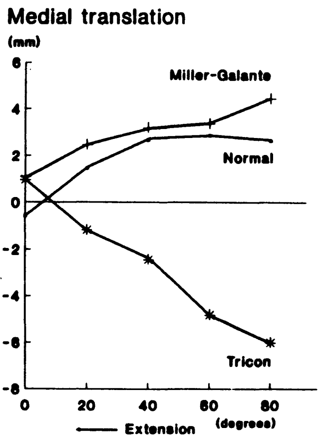


Figure 28.

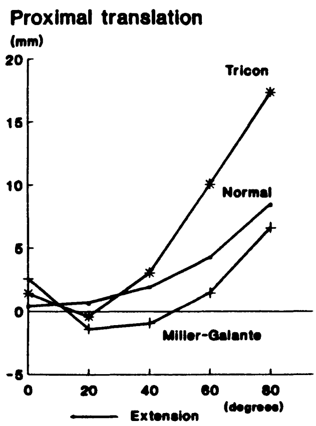


Figure 29.

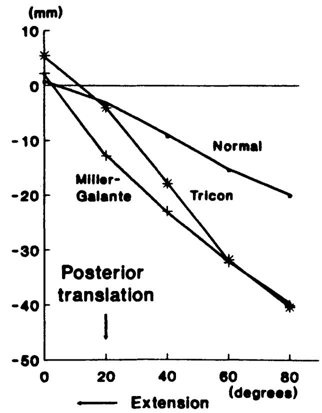


Figure 30.

Figure 28. Translation of the center of the prosthetic knees and the tibial intercondylar eminence (mean values) in the normal knees along the transverse axis (medially/laterally). (See also legend to Fig. 26).

Figure 29. Tibial translations (mean values) along the longitudinal axis (proximally/distally). (See also legend to Fig. 26).

Figure 30. Tibial translations (mean values) along the sagittal axis (anteriorly/posteriorly). (See also legend to Fig. 26).

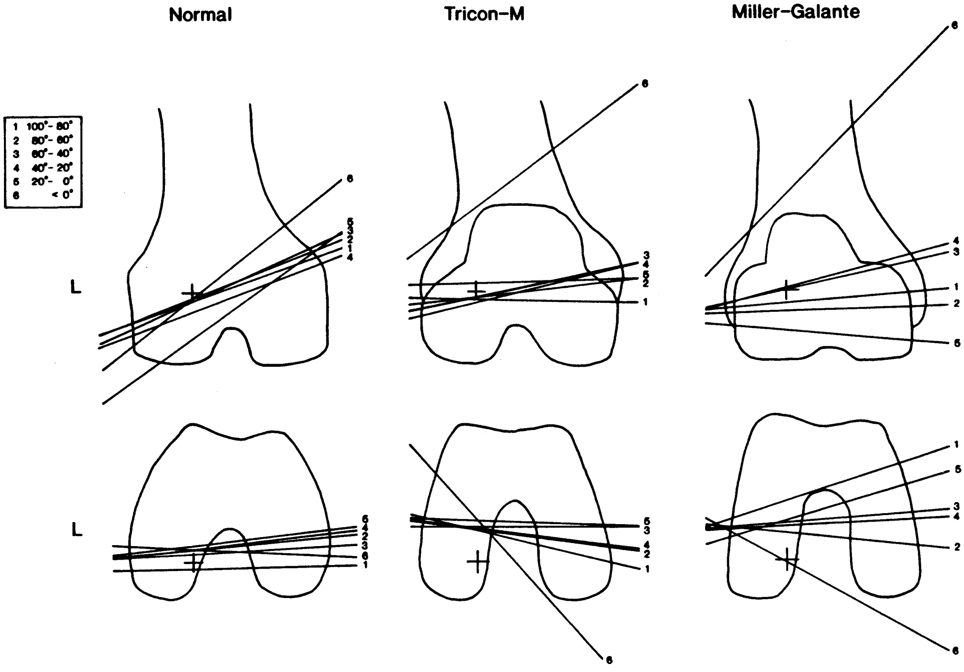


Figure 31. Mean screw axis positions for the 6 chosen intervals of extension. Projection of the screw axes in the frontal (*top*), and horizontal (*bottom*) planes.

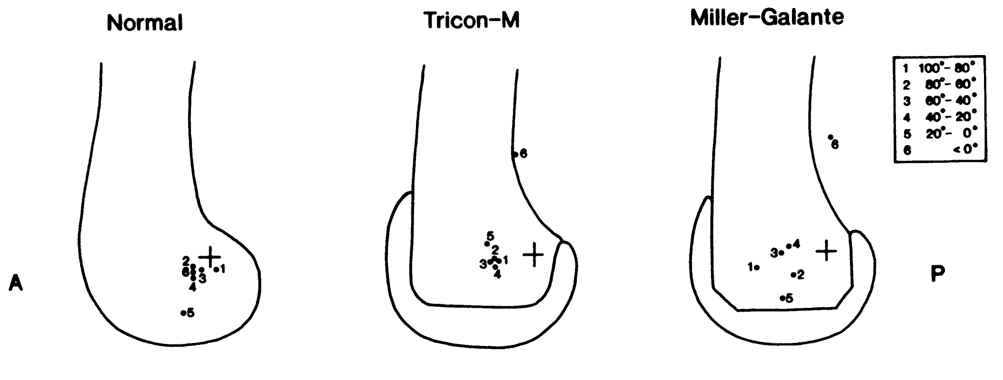


Figure 32. The mean intersection of the screw axes through a sagittal plane at the calculated position of the femoral insertion of the anterior cruciate ligament.

component in both designs was about 20 mm more posteriorly positioned than in the normal knees.

Helical axes

In the frontal plane the helical axes of the Tricon–M knees tended to have a more transverse course than in the normal knees. In the M–G knees the screw axes seemed to converge at a point positioned about 2 cm lateral to the femoral condyle (Fig. 31). In both designs the axes had a more oblique course at 60° to 20° and at hyperextension, indicating that gliding was more pronounced on the medial side of the joint. In the horizontal plane the axes of the M–G knees converged at a point localized laterally corresponding to the point of convergence in the frontal plane (Fig. 31). A similar tendency was found in the Tricon–M knees, but the scatter was less pronounced medially.

In the normal knees the mean helical axes intersections with the chosen sagittal plane ran close to the femoral insertion of the anterior cruciate ligament (Fig. 32). In the M–G knees the mean axes tended to be more scattered and in both designs slightly more anteriorly positioned. Translations along the helical axes were larger than normal in the M–G knees (mainly up to 40° of extension) and smaller than normal in the Tricon–M knees (M–G *vs* normal: $P < 0.01$; Tricon–M *vs* normal $P < 0.05$, *Wilcoxon rank sum test*).

Conclusion

In the Tricon–M knees, the kinematics displayed small differences when compared with active flexion without weight–bearing (V). In the Miller-Galante knees, weight–bearing seemed to change the pattern of rotations about the longitudinal and sagittal axes as compared to the non weight–bearing situation (VI). Decreased screw axis translations in the Tricon–M knee, and increased translations in the Miller–Galante knee reflect differences in joint area constraint between the two designs.

Discussion

The evolution of total knee arthroplasty during the last decade has hitherto only to a certain amount extended the survival rates (Goodfellow 1992), and optimization of the tibial component fixation is still an important issue. Previous RSA studies have indicated that cemented fixation provides a better initial stability. Other factors such as the design of the prosthetic undersurface and the joint area influences prosthetic fixation. In this thesis randomized studies were designed mainly to evaluate the efficacy of cemented vs. uncemented fixation. Unbiased selection of patients were performed and the surgical technique was standardized. Nevertheless, the studies could probably be further optimized by performing the randomization after cutting of the proximal tibia, which was not done in the present study due to practical reasons.

Methodological aspects

In previous RSA studies of tibial component fixation, movements have been presented either as rotations of the entire component and as translation of the prosthetic marker that moved the most (MTPM), or as translation of the center of the plateau markers (Ryd 1986, Ryd et al 1988, 1990, Albrektsson et al 1990, 1992, Carlsson et al 1991, Toksvig–Larsen 1992), which might not be the same as the center of the prosthesis. Sometimes it is difficult to visualize all prosthetic markers at one or several of the radiographic examinations during the follow–up series. This implies that the evaluation of the MTPM depend on the visualisation of the marker representing the most unstable part of the prosthesis throughout a series of radiographs. Further, loosening of single tibial plateau marker could significantly influence the results. Similarly, the results of "segment translation" may vary depending on location and number of stable prosthetic markers visualized at each subsequent examination.

In order to avoid these problems and obtain more standardized and reliable positions of measurements, we used five plotted points at standardized positions at the prosthetic edge and one at the center of the tibial component. This implied that the measurements were not restricted to the fate of a single tantalum marker with a potential risk of loosening or suboptimal location, jeopardizing the evaluation of one or several of the subsequent radiographs.

The accuracy of the initially plotted and subsequent imaginary points was instead influenced by the configuration and rigidity of all available prosthetic markers. This allowed an accurate analysis of micromotions measured at different parts of the prostheses. When rotations of the tibial tray occurs, the MTPM measured in this manner could be expected to be somewhat larger than previously published (Ryd 1986, Ryd et al 1988, 1990, Albrektsson et al 1990, 1992, Carlsson et al 1991, Toksvig-Larsen 1992) as a result of a more peripheral location of the points of measurement than the prosthetic tantalum markers.

The use of cement and tibial component design

Contrary to some other prosthetic designs (Ryd 1986, Ryd et al 1990, Albrektsson et al 1992), the micromovements of the Tricon–M prosthesis did not differ between cemented and uncemented fixation. Medial and posterior subsidence, with or without concomitant lift–off of the opposite side, was the most pronounced motion recorded. Rotations in the sagittal plane were somewhat larger than in the frontal plane. Shimagaki et al (1990) also found larger retro– or anteversion *in vitro* of the tibial component when compared with varus–valgus tilt However, the mean maximum subsidence in both planes in study I was about the same. This implies that the *in vivo* tilting in the frontal plane was often associated with subsidence of both edges of the prosthesis. This difference between *in vitro* and *in vivo* behaviour might be an effect of postoperative bone remodelling, which can not be accounted for in *in vitro* studies.

The migration (MTPM) of the uncemented Tricon–M knees was lower than previously reported for the uncemented PCA (original) and all–polyethylene Freeman–Samuelson (F–S) knee prostheses (Ryd et al 1988, 1990). It might be that the Tricon–M design gives a better press–fit than the PCA or the F–S designs when used uncemented. Another explanation may be that the transformation of forces to the implant–bone interface varies due to differences in joint congruity. When used with cement the Tricon–M knees displayed larger MTPM values than is reported for cemented Total Condylar or F–S knees (Ryd 1986, Albrektsson et al 1992), suggesting that the combination of a stem and cement in these two designs is important.

In vitro tests have shown positive effects on uncemented fixation of the tibial component using intramedullary stems 5 cm or longer, especially in osteoporotic bone (Dempsey et al 1989, Lee et al 1991), and adding a stem has been advocated to reduce subsidence (Laskin 1991). The current study could, however, not verify this statement, because significant subsidence was recorded both peripherally and centrally in the stemmed and uncemented Tricon–M components. In fact, the RA patients in paper II displayed nearly twice as large subsidence as found in the RA patients with uncemented and unstemmed Tricon–M prostheses (paper I), questioning whether the stem has any advantages when used uncemented. The findings in this thesis are in agreement with those of Albrektsson et al (1990) and Nilsson et al (1989), who did not find any effect of a stem in reducing subsidence of uncemented Freeman–Samuelson prostheses. A metal stem carries considerable load due to its stiffness. This will result in "by–pass" of load and stress–shielding of the bone under the tibial plate (Reilly et al 1982), especially if metal backed components with stems longer than 5–6 cm are used (Askew and Lewis 1981, Reilly et al 1982). Bone resorption under the tibial plate and subsidence will ensue since the loads applied cannot be resisted by the tip of the stem. These effects may not be observed *in vitro*, because such studies cannot account for the dynamic bone remodelling occurring after the operation. It seems that the uncemented stemmed design can resist forces trying to tilt the tibial tray into varus or valgus, but instead transforms them into increased compressive loads which become concentrated at the distal part of the stem. These observations was not confirmed *in vitro* by Lee et al (1991), probably due to reasons given above.

The clinical reports of a higher incidence of radiolucent zones under the tibial tray in uncemented stemmed total knee arthroplasties (Brodie et al 1992), and the high incidence of loosening of the uncemented stemmed Kinemax prosthesis reported by Nafei et al (1992) further questions the advantages using intramedullary stems in uncemented fixation.

The cemented stemmed implants displayed maximum translations (MTPM) comparable to cemented stemmed Total Condylar and Kinematic knee prostheses (Ryd 1986) but slightly larger than cemented and stemmed Freeman–Samuelson knees (Albrektsson et al 1992). The small movements recorded in cemented stemmed tibial components suggests that this design is favourable when used with cement.

The optimum stem configuration and the need and design of any additional fixation devices is not known. Preliminary data from investigations of the Tricon II prosthesis fixed with a 6 cm polished stem and 2 additional cancellous sliding screws, have indicated smaller subsidence than recorded in the Tricon–M stem prosthesis (II), especially when used with cement, but also when hydroxyapatite coating is used (unpublished data).

In vitro studies have shown that tibial components with 4 screws with or without pegs display small micromotions, but not as small as when cement is used (Volz et al 1988, Branson et al 1989, Dempsey et al 1989, Lee et al 1991). The results of the Miller–Galante prosthesis appear to confirm these findings. The fixation achieved seemed to be similar or slightly superior to other designs investigated with RSA (Ryd 1986, Ryd et al 1988, 1990, Albrektsson et al 1990, 1992, Toksvig–Larsen 1992) including the Tricon–M prosthesis with or without a stem.

According to Volz et al (1988) prevention of lift–off should also counteract subsidence, but this finding could not be supported in this study. Although lift–off was equal in the two groups of M–G knees, the uncemented components subsided more and especially at the medial or lateral edges.

In most of the uncemented M–G knees, small translations (total point motion sometimes less than 0.15 mm) was found either close to the center of the tibial component, or anteriorly, or posterolaterally as an effect of subsidence of one of the prosthetic edges and lift–off of the opposite one. This magnitude of micromotion is believed to be the upper limit for bone ingrowth (Pilliar et al 1986) and was recorded at the same locations as bone ingrowth has been observed in uncemented Miller–Galante prostheses retrieved for "reasons unrelated to fixation" (Sumner et al 1989).

The small migration observed for the M–G knees supports that a low constrained prosthesis is beneficial. In this study only arthrotic knees were investigated and it might be that the comparatively small pegs makes this design less efficient when there is poor bone quality. Further, the M–G knees might be more sensible to poor alignment as reflected by the two revisions due to instability in this series.

For comparison, the results of maximum migration (MTPM) for the different prostheses in this Thesis are displayed together with results obtained from the literature (Fig. 33).

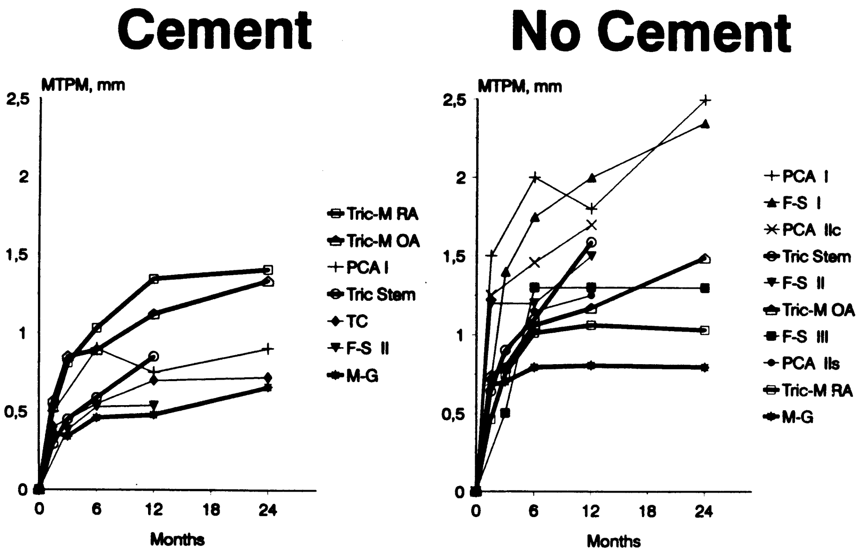


Figure 33. Mean maximum tibial prosthetic migration (MTPM) during two years for various designs of tibial components. *Left:* cemented knees, *Right:* uncemented knees. **Bold** lines: prostheses studied in this investigation, *thin* lines: results from the literature. **Tric–M RA:** Tricon–M, rheumatoid arthritis; **Tric–M OA:** Tricon–M, osteoarthrosis; **PCA I:** Original PCA; **Tric Stem:** Tricon–M Tibial Stem; **TC:** Total Condylar; **M–G:** Miller–Galante; **F–S I:** All–polyethylene Freeman–Samuelson; **F–S II:** Metal–backed Freeman–Samuelson with 80 mm stem; **F–S III:** Metal–backed Freeman–Samuelson with 110 mm stem; **PCA IIs:** PCA version 2, operated with standard saw blade; **PCA IIc:** PCA version 2, operated with cooled saw blade.

Diagnosis and micromotion

In the Tricon–M study (paper I) the patients with RA displayed similar mean micromotions as did the patients with OA regardless of cement being used or not, and in accordance with other studies (Andersen 1979, Laskin 1981, 1990, Sledge and Walker 1984, Knutson et al 1986, Rand and Ilstrup 1991). This might be due to small variations of the strength (Hvid 1988b) and hardness (Lereim et al 1974) of the proximal tibial bone between OA and RA patients. The tendency to generalized bone loss in RA may be counteracted by the bone pathology related to the development of secondary arthrosis (Havdrup et al 1976), and, according to Hvid (1988a), major differences in the results of modern knee replacement between RA and OA would not be expected. An other explanation might be that the RA patients had significantly lower weight, and were probably less physically active than the OA patients.

Leg alignment and prosthetic positioning

Postoperative malalignment of the knee is considered to be of importance in the process of loosening (Lotke and Ecker 1977, Ducheyne et al 1978, Freeman et al 1978, Coventry 1979, Rand and Bryan Orthop 1982, Bargren et al 1983, Insall et al 1983, Tew and Waugh 1985). In this investigation there were no consistent correlations between micromotion of the tibial component and the leg alignment measured (HKA angle). One reason might be that only few knees were aligned outside the ± 5° or ± 6° range of acceptable alignment (Bargren et al 1983, Insall et al 1983).

Even if the knee is normally aligned, the separate prosthetic components may be malaligned. In the uncemented OA group with Tricon–M prostheses there was a positive correlation between alignment of the tibial component in the sagittal plane and subsidence at the posterior edge. The reason for this is not known, but it might be that a posterior slope of the tibial tray displaces the tibio–femoral contact area and the maximum pressure more posteriorly.

Internal–external rotations of the tibial component on the tibia was larger in the Tricon–M than in the Miller–Galante prostheses. Ryd (1986) found that increased conformity of the joint area resulted in larger external rotatory migration of the tibial component which may explain this finding. The majority of the unstemmed Tricon–M prostheses displayed external rotation, whereas in the stemmed Tricon–M and the Miller–Galante components rotations were evenly distributed between external and internal rotation. Rotatory malalignment of the tibial (and femoral) components may be one explanation to this difference suggesting a more accurate prosthetic positioning in the two later series and that measurements of the rotatory position of the tibial components (Ryd 1985) would have been of interest.

Bone quality

In the normal tibia the medial condyle has the strongest bone (Hvid 1988a). Varus malalignment is always associated with stronger bone medially, whereas valgus malalignment has a less predictable influence on the distribution of the bone strength (Hvid 1988a). In the less loaded condyle the bone is rarely so compromised that immediate postoperative failure is to be expected, but there may be a significant risk of fatigue failure with prolonged activity (Hvid 1988a). The majority of the uncemented Miller–Galante tibial components subsided on the side with the weaker bone indicating that fatigue failure occurs. The degree of trabecular bone fatigue failure depends on the loads applied and the individual capacity of the bone to remodel. In the cemented M–G knees, subsidence was smaller and did not depend on the preoperative deformity, suggesting that the cement to some extent compensated for the inferior bone quality.

In the other two designs of prostheses there was no obvious relationship between location of subsidence and preoperative deformity. The reason for this is not known, but may partly be explained by the overall larger migration reflecting a more multi–factorial background.

Thickness of the tibial cut

The bone strength of the proximal tibia decreases significantly with increasing distance from the subchondral bone (Hvid 1988a), and it has been recommended to cut as little bone as possible. On the other hand the polyethylene must be of sufficient thickness (Bartel et al 1986) in order to minimize wear. When metal–backed tibial components are used more bone has to be cut to avoid insertion of thin polyethylene inserts. Cornell et al (1986) found that tibial resection up to a thickness of 13 mm did not influence the outcome. In this investigation, the thinnest uncemented M–G tibial components displayed significantly larger micromotion (*varus–valgus tilt*) initially. The postoperative joint level increased only marginally when thicker components were inserted, indicating that these thicker components were inserted after more bone had been cut from the proximal tibia. Thin polyethylene components inserted uncemented may transmit higher loads to the prosthesis–bone interface causing early and large micromotions.

Due to the elastic properties of the polyethylene, thicker components can probably more successfully distribute the loads to the metal backing and the bone. Thus, the advantages of thicker tibial components outweighs the disadvantages of cutting more bone, at least when uncemented fixation is used. In cemented M–G knees the thickness of the polyethylene did not influence the micromotion. It seems as the mechanical properties of the cement–bone composite to some extent can compensate for the disadvantage of the thin polyethylene, at least in terms of early fixation. The majority of the Tricon–M and Tricon–M stem knees were inserted with the thinnest (6 mm) polyethylene components, thus unabling the evaluation of the influence of polyethylene thickness on micromotion.

Pattern of migration

In all studies, migration was most pronounced during the first 6 months, in accordance with previous observations (Ryd 1986, Ryd et al 1988, 1990, Albrektsson et al 1990, 1992, Carlsson et al 1991, Toksvig-Larsen 1992). In the uncemented M-G knees, almost all migration occurred during the first 6 weeks. In both the cemented and uncemented tibial trays migration was most pronounced in the distal direction, sometimes in combination with lift-off from the underlying bone of one part of the component. In the uncemented knees, this initial migration was always larger than in the cemented knees. Toksvig-Larsen and Ryd (1991) found that a seemingly flat tibial cut surface had a maximum roughness between uppermost and lowermost points of 1.0 to 2.4 mm, and calculated that an uncemented tibial component before impaction of the component would rest on only 1 to 2 percent of the cut tibial surface. During the insertion, postoperative flexion-extension exercises and increasing postoperative weight-bearing, bone trabeculae will crush and the prosthesis will subside until the processes of microdamage and remodelling will come to an equilibrium (Hvid 1988a). This balance will vary depending on individual bone quality, the amount of trauma during surgery, and the capacity for bone remodelling. Of the factors analyzed in this study, the choice of cement and fixation devices such as pegs, screws or a stem influences the fixation. Finally the design of the joint area and the kinematic behaviour of the implant are most certainly important also for the early fixation, because loads and moments induced by extension-flexion exercises start to act within the first days after surgery.

After the initial months of early migration, the uncemented Tricon-M prostheses in RA patients and the uncemented M-G prostheses seemed to settle, whereas the other groups displayed a continuous migration, but with decreasing speed. The implications of an early and short period of migration followed by stability as in the M-G knees is not known, but it might be that this pattern of motion is compatible with at least partial ingrowth of bone (Sumner et al 1989). In the Tricon-M knees followed 5 years, it was evident that the majority of the uncemented knees seemed to have settled within the first 2 years, since no further migration occurred during the subsequent 3 years. This implies that the processes of microdamage and remodelling of the bone under the tibial components might have come to an equilibrium. The cemented

Tricon–M knees displayed, however, a significant increase in migration (MTPM) between 2 and 5 years. It might be that the toxic products released from the cement may induce necrosis of the bone, contributing to the continuous migration during the reparative phase.

The continuous migration encountered might also in part be the result of too heavy loads and rocking movements of the tibial tray during gait. The size of these inducible displacements will depend on the pattern of load transmission during flexion and extension, which probably varies considerably between different prosthetic designs. If the movements exceed certain limits bone resorption will probably become dominant resulting in clinical loosening (Hvid 1988a).

Severe osteolysis in knee arthroplasty has been reported (Peters et al 1991, Kilgus et al 1992, Nolan and Bucknill 1992), but does not seem to be as common as in hip arthroplasty (Anthony et al 1990, Maloney et al 1990, Santavirta et al 1990) maybe because long intramedullary stems are more rarely used and the joint fluid and particulate debris do not so easily become trapped under pressure or gain access to the prosthetic undersurface far from the joint (Anthony et al 1990, Lee et al 1992).

Clinical results and micromotion

The HSS knee scoring system is widely used and allows direct comparison with other studies. This system has, however, certain limits and the total score becomes unreliable when other diseases impair knee function. The total knee score did not differ between cemented and uncemented components in any of the series, despite the fact that the uncemented components in 2 of the 3 studies displayed larger micromotions. Absent correlation between prosthetic micromotion and pain might depend on that pronounced migration only occurred in a few patients and did not cause symptoms in all of them.

Clinical relevance of micromotion

The clinical relevance of early prosthetic micromovements is still not quite clear. However, there are indications that the magnitude of migration as measured by RSA has clinical implications.

Grewal et al (1992) analysed the cumulative survival rates of three different designs of the Freeman–Samuelson knee arthroplasty. Comparison with the stereoradiographic data of these prosthetic designs in the studies of Albrektsson et al (1990, 1992) revealed that the design displaying the lowest survival rates at 5 years showed the largest migration during the first 1 to 2 years and vice versa, suggesting that early instability correlated with late aseptic loosening. Pronounced migration of the uncemented PCA prosthesis was reported by Ryd (Ryd 1986, Ryd et al 1990) and later Moran et al (1991) reported high failure rates of uncemented PCA arthroplasties 5 to 7 years after operation. Ryd et al (1990) also reported a correlation between continuously increasing migration and malalignment of the knee or the tibial component. Because such malalignment increases the risk of future failure (Bargren et al 1983, Boegård et al 1984, Tew and Waugh 1985, Ecker et al 1987), Ryd (1992) postulated that progressive migration may indicate future loosening. As regards hip prostheses stereoradiographic studies of uncemented ribbed femoral components (Kärrholm and Snorrason 1992b) and screw cups (Snorrason and Kärrholm 1990) have indicated an association between high early micromovements and pain during the postoperative years.

There is also a coupling between prosthetic instability/loosening and wear. Polyethylene damage will change the forces acting on the prosthesis/cement/bone interfaces and has been associated with late loosening of cemented acetabular cups (Garcia–Cimbrelo 1992). However, early prosthetic instability might also be important by restricting the area of the articulation between the components, subjecting the polyethylene to increased shear forces and fatigue failure.

Patients prospectively followed with RSA and later subjected to revision of loose tibial components due to aseptic loosening have displayed large micromotions one and 2 years after the primary surgery. Ryd (1992) reported 3 such cases (2 cemented PCA and 1 uncemented F–S) where the loosening was confirmed intraoperatively. These prostheses had displayed micromovements (MTPM) between 1.2 mm and 3.3 mm one year after the primary surgery and between 4.3 mm and 5 mm after 2 years. These values were considerably higher than those mean values recorded for the corresponding cemented and uncemented prostheses not being revised.

The two Tricon–M prostheses revised within 5 years postoperatively in study I displayed migration (MTPM) of 2.2 mm and 1.8 mm at one year and 2.4 mm and 2.9 mm, respectively, after 18 months (before revision) and 2 years. Further one patient with an uncemented hydroxyapatite coated Tricon II tibial component (unpublished data), who had onset of pain about 2 months after the operation, has been revised. Six weeks after surgery the migration was 0.5 mm, but increased to 5.5 mm at 3 months, and reached 9.3 mm at 6 months when subsidence but almost no radiolucent zones were observed on conventional radiographs. At the revision 6 months after primary surgery the prosthesis could easily be removed.

Based on available data, small migration (MTPM < 0.5 – 1 mm) during the postoperative year is most certainly an indication of a durable fixation provided proper prosthetic alignment and control of wear. This probably implies that the processes of trabecular microfractures, remodelling and patchy ingrowth or bonding to hydroxyapatite have reached an early equilibrium. Migration exceeding 2 mm during the first 1 to 2 postoperative years indicates that such an equilibrium has not taken place, and may raise concerns about the long–term fixation and especially in the active individual.

Scintimetry

In scintimetry the identification of the region of interest used as a reference is important. Ideally it should display low and constant activity. We found that ROI:s located in the ipsilateral femoral and tibial diaphyses displayed changing activity during the investigation period, making their use as reference insecure. The ROI located in front of the femoral component, however, fulfilled the criteria for a useful reference.

The activity under the tibial component decreased during the investigation period, but persistent high activity (6.5 times that of the reference ROI) was found after 24 months. The postoperative level and the change of scintimetric activity during the 2 postoperative years was found to be influenced by the type of preoperative knee deformity, but not by diagnosis or mode of fixation. The technetium–isotope ($^{99}Tc^{m}$–MDP) adheres to the mineral phase of the bone, and the increased activity seen in arthrotic joints reflects the increased bone remodelling processes of the subchondral bone (Christensen 1985). The bone strength at the proximal tibia is influenced by the preoperative deformity (Hvid 1988a), and correlates with bone density as measured by quantitative computed tomography (Bentzen et al 1987). Bone density is higher in preoperative varus knees than in valgus knees irrespective of diagnosis (Hvid et al 1988a). The bone remodelling in the knee is generally higher in RA than in OA, but increases in both diagnoses when there is preoperative malalignment (Havdrup et al 1976). This may explain why the RA knees with varus deformity showed as high (or higher) mean activity as the OA knees with varus deformity. After total knee arthroplasty, the bone density of the preoperatively more loaded condyle decreases with time, whereas the less loaded condyle has unchanged density (Hvid et al 1988). In knees with varus deformity, this decrease in bone mass takes longer time than in the lateral condyle of valgus knees (Hvid et al 1988). Initially increased bone mass in preoperative varus knees and postoperative changes of the bone metabolism due to altered loading conditions may explain the natural course of condylar scintimetric activity during the 2 postoperative years found in our investigation.

As a result of the large number of unstable tibial components as assessed by RSA, it was difficult to make a proper evaluation of the influence of micromotion on the scintimetric activity. Stable components seemed to display lower activity throughout the follow–up period when compared to the unstable components. However, this relation was obscured by the effect of the normal postoperative remodelling on the activity. Two years after surgery low scintimetric activity under the tibial component indicates absence of prosthetic tilting, whereas high activity indicates prosthetic instability *or* a high bone remodelling for some other reasons. This finding does not necessarily imply that pure subsidence does not influence the uptake of the isotope, but rather that this issue could not be adequately evaluated, because of too few observations without this migratory pattern.

Knee kinematics

Neither of the two experimental set–ups corresponded to the pattern of knee motion occurring during gait. The method used in papers V and VI was dynamic but performed in a non weight–bearing position and the effect of gravity was different from the situation during gait. Probably the recorded motions most closely represent the knee forces acting during the swing phase of the gait. The tibiofemoral compression forces in this investigation were not known, but have been calculated at about twice the body weight in the supine position (Nisell et al 1986). The method used in paper VII was performed in a weight–bearing position, but the kinematics was not evaluated dynamically. The patients adjusted the knee at an equilibrium at each position of knee flexion. The results thus obtained must therefore be interpreted in the light of these limitations.

The prosthetic knees all displayed the same degrees of freedom regarding rotational and translational movements as the normal knees, although the kinematics were more or less different. The most constrained prosthesis (Tricon–M) displayed the largest deviations from the normal. The micromotions of the Tricon–M knees were larger than those of the Miller-Galante knees, which partly may be an effect of the different kinematics.

Compared with the kinematics during non weight–bearing, the M–G knees displayed no distinct "screw home" mechanism at the last degrees of extension in the weight–bearing situation. Increased compressive forces at weight–bearing might have prevented this pattern of motion (Hsieh and Walker 1976, Bourne et al 1978, Shoemaker et al 1982). The subtle increase of internal–external rotation observed in the M–G knees between 0° and 15° of flexion in the non weight–bearing situation might also have been lost due to the wide interval between the recordings in the weight–bearing situation.

In the M–G knees no or minimum changes in tibial adduction during knee extension and flexion was found both in the weight–bearing and the non weight–bearing situations. However, compared to the unloaded reference, a minor adduction was noted in the weight–bearing position. In the Tricon–M knees the rotations about the sagittal axis during weight–bearing were rather similar to those observed without weight–bearing as were the translations of the knee center in both designs. Because the tibiofemoral compression forces are considerable also at unloaded flexion–extension of the knee (Nisell et al 1986), the joint surface design will become a major determinant for the knee joint kinematics regardless whether weight–bearing is performed or not (Hsieh and Walker 1976, Markolf et al 1981, Shoemaker et al 1982).

In the Tricon–M knees, rotations about the longitudinal axis were small during the first 25° of flexion probably due to the high conformity of the joint surfaces near extension. Beyond 25° the congruity decreases allowing more pronounced rotation. This reduced normal internal tibial rotation may induce torques forcing the tibial component to rotate outwards on tibia as was found in the majority of the tibial components in paper I, and in a previous study of the congruent Total Condylar prosthesis (Ryd 1986). Positioning of the tibial component in internal rotation relative the tibial bone could be another explanation to this migratory pattern (Ryd 1986).

In the normal knee the internal rotation of tibia occurs about a constantly shifting center of rotation (Kärrholm 1989). The meniscal bearings of the New Jersey LCS prostheses did not contribute to any internal rotation, maybe because the fixed meniscal trackings limits the internal–external rotation to take place about a central axis.

The M–G knees showed almost no rotations about the sagittal axis during flexion–extension of the knee. This may be an effect of the flatness of the articular surfaces in the frontal plane. The forces trying to adduct tibia during flexion may be transformed to the prosthetic–bone interface resulting in varus–valgus movements of the tibial tray, which seemed to be the most design specific type of micromotion in the uncemented M–G design.

The prosthetic knees displayed larger posterior translations of the tibial center than did the normal knees. This increased sliding of tibia in relation to femur was reflected by increased translations for a certain amount of flexion and a tendency to a larger distance between the helical axes and the joint area. The oblique course of the axes in the frontal plane also indicates that in the normal knees there is more sliding medially and more rolling laterally. In the M–G knees the inclination of the screw axes in the frontal plane changed more dramatically than in the Tricon–M knees, but at 80°–60° and 40°–20° flexion intervals the axes were rather close to the joint line indicating more rolling at these intervals. Femoral rolling probably occurs when the anterodistal and posterodistal parts of the femoral component, with the smallest radii of curvature in the sagittal plane contact the tibial component, and sliding when the distal and rather flat femoral surface articulates against the tibial component.

The translations along the helical axes were larger than normal in the M–G knees (mainly up to 40° of extension) and smaller than normal in the Tricon–M knees, indicating a less controlled coupling between rotations and translations in the former design. In the M–G knees, the position of the helical axes shifted during extension in the horizontal plane suggesting continuous changes in the location of the loaded area. This unconstrained behaviour of the M–G prosthesis implies that motions are to a greater extent governed by the soft tissues, which would counteract the build–up of high moments in the prosthesis–bone interface. Consequently small micromotions were found not only in this study, but also in a previous investigation of the same design (Carlsson et al 1991). Increased translatory motions might, however, increase the polyethylene wear (Bloebaum et al 1991, Kilgus et al 1991, Blunn et al 1992, Engh et al 1992, Lindstrand et al 1992, Lindstrand and Stenström 1992). However, others factors such as contact area, joint load, quality and thickness of the polyethylene and maybe also the fixation of the prosthetic components have to be considered (Blunn et al 1992 Bartel et al 1986).

Posterior cruciate ligament

Although the posterior cruciate ligament (PCL) was retained in both the M–G and the LCS knees, posterior translation of tibia was as large as in the Tricon–M knees where the PCL had been resected. Resection of the PCL results in anterior luxation of the femur on the tibia (Insall 1984, Kurosawa et al 1985, Townley 1988), and absence of the anterior cruciate ligament (ACL) may have the opposite effect. All 3 prosthetic designs displayed a tendency to anterior tibial displacement at full extension, perhaps an effect of absent ACL and posterior sloping of the tibial joint surface (Townley 1988). This anterior displacement was, however, too small to explain the pronounced posterior translations in the prosthetic knees. Absent PCL in the Tricon–M knees and impaired function of the PCL due to improper tensioning (Insall 1988) or postoperative elongation (Hagena et al 1989) in the other designs might be an explanation. Increased posterior displacement of both the medial and lateral tibial condyles could be presumed, because in all designs the internal tibial rotation was rather small in the flexed positions.

The PCL is thought to contribute to the femoral "roll–back" during flexion (Goodfellow and O'Connor 1978, Andriacchi and Galante 1988). Visual analysis of the lateral radiographs in the prosthetic knees in papers V and VI showed that the contact point between the femur and the tibia did not move backwards on tibia during flexion, as was observed in the normal knees, but was located centrally or even anteriorly, suggesting that no "roll–back" took place. Soudry et al (Soudry et al 1986) found that the contact point displaced anteriorly on tibia after PCL resection, especially if the joint area was flat. The tendency to more distal displacement of the M–G tibial components during flexion suggests that the anterior and highest part of the tibial component contacts the femoral component, further questioning the function of the PCL. In dished tibial components subjected to high loads, the contact point is more centrally located (Soudry et al 1986). If the contact point does not move posteriorly in the flexed positions, the central part of the tibia has to slide more posteriorly on femur for a certain amount of flexion than if a "roll–back" took place.

Implications of this Thesis

Cemented fixation is the golden standard in total knee arthroplasty and provides high percentage of good or excellent clinical results in the elderly. Uncemented fixation displayed larger migrations than cement in 2 of the 3 studies in this Thesis. However, the continuously increasing migration observed in the cemented knees between 2 and 5 years raises concerns about the long-term results. This might be the result of toxic products *and/or* cement particles released from the cement, indicating that there is a place for improvements of the cement and the cementing technique.

The results of uncemented alternatives have so far not always fulfilled the expectations, indicating the need for further research. In order to treat younger and/or more active patients, the development of improved fixation concepts is mandatory. Perhaps bioactive coatings such as hydroxyapatite (Geesink et al 1987) or arthroplasties based on fixation by direct bone apposition based on surgery in two steps (Carlsson 1989, Röstlund 1990) will result in better long-term fixation. Newer designs based on knowledge of the three-dimensional kinematics of the knee may also contribute to lesser micromotion.

According to this Thesis prostheses with unconstrained joint surfaces induce less micromotion than do the more constrained designs. Planar surfaces or meniscal bearings decreases the inherent stability of the joint, and may induce increased risk for polyethylene wear at the joint surface or at the metallic fixation devices for the meniscal bearings. The risk of wear decreases with increasing thickness of the polyethylene (Bartel et al 1986) indicating that thin polyethylene should be avoided. As found in this Thesis, cutting more bone, within certain limits, from the proximal tibia in order to increase the thickness of the tibial component does not increase the risk for micromotion.

Conclusions

A. Tibial component fixation (1 – 2 years postoperatively)

1. The micromotion did not differ significantly between patients with OA and RA (I).

2. There were no differences between cemented and uncemented fixation when using a tibial component with porous coating and two flanged polyethylene pegs (I).

3. Adding a stem to the tibial component enhanced the fixation when used with cement. In uncemented knees the stem seemed to induce increased subsidence (II).

4. Low constrained tibial component with 4 pegs and fiber–mesh undersurface displayed the smallest micromotions. Cement improved the early fixation by reducing the influence of tibial component thickness and bone quality. When used uncemented increased varus–valgus tilting was found (III).

5. During the 2 postoperative years, the scintimetric activity under the tibial component was significantly influenced by the type of preoperative deformity. During that time period low activity indicated absence of tibial component rotations, whereas high activity could imply instability of the prosthesis or high bone remodelling for some other reasons (IV).

B. Knee kinematics

6. All three different designs of knee prostheses displayed abnormal kinematics *in vivo*. The less constrained designs displayed smaller deviations from the normal than the constrained one. Increased posterior translation during knee flexion occurred in all designs. The size of this displacement was unaffected by the presence or absence of the PCL (V – VII).

7. Weight–bearing changed the kinematics only marginally, and only in the unconstrained design. Determination of screw axes positions provided additional information about the kinematics of the knee. Decreased translations along the screw axis in the Tricon–M knee and increased translations in the M–G knee may reflect differences in joint area constraint between the two designs (VII).

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References

Ahlbäck S (1968). Osteoarthrosis of the knee: A radiographic investigation. *Acta Radiol (Suppl 277):1.*

Albrektsson T, Brånemark P-I, Hansson H-A, Lindström J (1981). Osseointegrated titanium implants. Requirements for ensuring a long-lasting, direct bone-to-implant anchorage in man. *Acta Orthop Scand 52:155-170.*

Albrektsson T and Albrektsson B (1987). Osseointegration of bone implants. A review of an alternative mode of fixation. *Acta Orthop Scand 58:567-577.*

Albrektsson BEJ and Herberts P (1988). ICLH knee arthroplasty. A consecutive study of 108 knees with uncemented tibial component fixation. *J Arthroplasty 3:145-156.*

Albrektsson BEJ, Ryd L, Carlsson LV, Freeman MAR, Herberts P, Regnér L, Selvik G (1990). The effect of a stem on the tibial component of knee arthroplasty. A roentgen stereophotogrammetric study of uncemented tibial components in the Freeman-Samuelson knee arthroplasty. *J Bone Joint Surg 72-B:252-258.*

Albrektsson BEJ, Carlsson LV, Freeman MAR, Herberts P, Ryd L (1992). Proximally cemented versus uncemented Freeman-Samuelson knee arthroplasty. A prospective randomized study. *J Bone Joint Surg 74-B:233-238.*

Andersen JL (1979). Knee arthroplasty in rheumatoid arthritis. An analysis of 240 cases of hemi-, hinge and resurfacing arthroplasties. *Acta Orthop Scand (Suppl 180):*1-117.

Andriacchi TP, Galante JO, Fermier RW (1982). The influence of total knee-replacement design on walking and stair-climbing. *J Bone Joint Surg 64-A:1328-1335.*

Andriacchi TP, Stanwyck TS, Galante JO (1986). Knee biomechanics and total knee replacement. *J Arthroplasty 1:211-219.*

Andriacchi TP and Galante JO (1988). Retention of the posterior cruciate in total knee arthroplasty. *J Arthroplasty 3 (Suppl):13-19.*

Anthony PP, Gie GA, Howie CR, Ling RSM (1990). Localised endosteal bone lysis in relation to the femoral components of cemented total hip arthroplasties. *J Bone Joint Surg 72-B:971-979.*

Askew MJ and Lewis JL (1981). Analysis of model variables and fixation post length effects on stresses around a prosthesis in the proximal tibia. *J Biomech Eng 103:239-245.*

Aspenberg P, Goodman S, Toksvig-Larsen S, Ryd L, Albrektsson T (1992). Intermittent micromotion inhibits bone ingrowth. Titanium implants in rabbits. *Acta Orthop Scand 63:141-145.*

Bain AM (1973). Replacement of the knee joint with the Walldius prosthesis using cement fixation. *Clin Orthop 94:65-71.*

Bargren JH, Blaha JD, Freeman MAR (1983). Alignment in total knee arthroplasty. Correlated biomechanical and clinical observations. *Clin Orthop 173:178–183.*

Barrett WP and Scott RD (1987). Revision of failed unicondylar unicompartmental knee arthroplasty. *J Bone Joint Surg 69–A:1328–1335.*

Bartel DL, Bicknell VL, Wright TM (1986). The effect of conformity, thickness, and material on stresses in ultra-high molecular weight components for total joint replacement. *J Bone Joint Surg 68–A:1041–1051.*

Bentzen SM, Hvid I, Jorgensen J (1987). Mechanical strength of tibial trabecular bone evaluated by X-ray computed tomography. *J Biomech 20:743–752.*

Blaha JD, Insler HP, Freeman MAR, Revell PA, Todd RC (1982). The fixation of a proximal tibial polyethylene prosthesis without cement. *J Bone Joint Surg 64–B:326–335.*

Blankevoort L, Huiskes R, De Lange A (1986). Helical axes along the envelope of passive knee joint motion. *32nd Annual ORS 11:410.*

Blankevoort L, Huiskes R, De Lange A (1988). The envelope of passive knee joint motion. *J Biomech 21:705–720.*

Bloebaum RD, Nelson K, Dorr LD, Hofmann AA, Lyman DJ (1991). Investigation of early surface delamination observed in retrieved heat-pressed tibial inserts. *Clin Orthop 269:120–127.*

Blunn GW, Joshi AB, Lilley PA, Engelbrecht E, Ryd L, Lidgren L, Hardinge K, Nieder E, Walker PS (1992). Polyethylene wear in unicondylar knee prostheses. 106 retrieved Marmor, PCA, and St Georg tibial components compared. *Acta Orthop Scand 63:247–255.*

Bobyn JD, Pilliar RM, Cameron HU, Weatherly GC (1980). The optimum pore size for the fixation of porous-surfaced metal implants by the ingrowth of bone. *Clin Orthop 150:263–270.*

Boegård T, Brattström H, Lidgren L (1984). Seventy-four Attenborough knee replacements for rheumatoid arthritis. A clinical and radiographic study. *Acta Orthop Scand 55:166–171.*

Bourne RB, Goodfellow JW, O'Connor JJ (1978). A functional analysis of various knee arthroplasties. *24th Annual ORS 3:156.*

Branson PJ, Steege JW, Wixson RL, Lewis J, Stulberg SD (1989). Rigidity of initial fixation with uncemented tibial knee implants. *J Arthroplasty 4:21–26.*

Brodie MR and Hofmann AH (1992). Stemmed versus resurfacing tibial components in cementless total knee arthroplasty. In: *Transactions of the AAOS 59th Annual Meeting:136.*

Bryan RS and Rand JA (1982). Revision total knee arthroplasty. *Clin Orthop 170:116–122.*

Buechel FF and Pappas MJ (1989). New Jersey low contact stress knee replacement system. Ten year evaluation of meniscal bearings. *Orthop Clin North Am 20:147–177.*

Burke DW, Gates EI, Harris WH (1984). Centrifugation as a method of improving tensile and fatigue properties of acrylic bone cement. *J Bone Joint Surg 66–A:1265–1273.*

Cameron HU, Pilliar RM, MacNab I (1973). The effect of movement on the bonding of porous metal to bone. *J Biomed Mater Res 7:301–311.*

Cameron HU (1982). The results of early clinical trials with a microporous coated metal hip prosthesis. *Clin Orthop 165:188–190.*

Cameron HU (1986). Six–year results with a microporous–coated metal hip prosthesis. *Clin Orthop 208:81–83.*

Carlsson LV (1989). On the development of a new concept for orthopedic implant fixation. *Thesis. University of Gothenburg, Sweden.*

Carlsson LV, Ryd L, Herberts P (1991). An *in vivo* roentgen stereophotogrammetric analysis of the Miller–Galante tibial component. *37th Annual ORS 16:168.*

Christensen SB (1985). Osteoarthrosis. Changes of bone, cartilage and synovial membrane in relation to bone scintigraphy. *Acta Orthop Scand (Suppl 214):1–43.*

Collier JP, Mayor MB, Townley CO, Fenning JB, Buechel FF (1986). Histology of retrieved porous coated knee prosthesis. *Transactions of the AAOS 53th Annual Meeting: 41.*

Cook SD, Thomas KA, Haddad RJ (1988). Histologic analysis of retrieved human porous–coated total joint components. *Clin Orthop 234:90–101.*

Cook SD, Barrack RL, Thomas KA, Haddad RJ (1989). Quantitative histologic analysis of tissue growth into porous total knee components. *J Arthroplasty 4 (Suppl):33–43.*

Cornell CN, Ranawat CS, Burstein AH (1986). A clinical and radiographic analysis of loosening of total knee arthroplasty components using a bilateral model. *J Arthroplasty 1:157–163.*

Coventry MB (1979). Two–part total knee arthroplasty: Evolution and present status. *Clin Orthop 145:29–36.*

Davies JP, O'Connor DO, Burke DW, Greer JA, Harris WH (1989). Comparison and optimization of three centrifugation systems for reducing porosity of Simplex P bone cement. *J Arthroplasty 4:15–20.*

Dawson JM and Bartel DL (1992). Consequences of an interference fit on the fixation of porous–coated tibial components in total knee replacement. *J Bone Joint Surg 74–A:233–238.*

De Bastiani G, Gabbi C, Magnan B, Regis D, Ricci M (1990). Studio sperimentale dell'interfacia cemento–osso: effetto del calore di polimerizzazione. *Biomateriali 3:85–93.*

Dempsey AJ, Finlay JB, Bourne RB, Rorabeck CH, Scott MA (1989). Stability and anchorage considerations for cementless tibial components. *J Arthroplasty 4:223–230.*

Dodd CAF, Hungerford DS, Krackow KA (1990). Total knee arthroplasty fixation. Comparison of the early results of paired cemented versus uncemented porous coated anatomic knee prostheses. *Clin Orthop 260:66–70.*

Dorr LD, Lockwood Ochsner J, Gronley J, Perry J (1988). Functional comparison of posterior cruciate–retained *versus* cruciate–sacrificed total knee arthroplasty. *Clin Orthop 236:36–43.*

Ducheyne P, Kagan A, Lacey JA (1978). Failure of total knee arthroplasty due to loosening and deformation of the tibial component. *J Bone Joint Surg 60–A:384–391.*

Duus BR, Boeckstyns M, Kjaer L, Stadeager C (1987). Radionuclide scanning after total knee replacement: Correlation with pain and radiolucent lines. A prospective study. *Invest Radiol 22:891–894.*

Duus BR, Boeckstyns M, Stadeager C (1990). The natural course of radionuclide bone scanning in the evaluation of total knee replacement. A 2–year prospective study. *Clin Radiol 41:341–343.*

Ecker ML, Lotke PA, Windsor RE, Cella JP (1987). Long–term results after total condylar knee arthroplasty. Significance of radiolucent lines. *Clin Orthop 216:151–158.*

Engh GA, Dwyer KA, Hanes CK (1992). Polyethylene wear of metal–backed tibial components in total and unicompartmental knee prostheses. *J Bone Joint Surg 74–B:9–17.*

Eriksson AR (1984). Heat induced bone tissue injury. An *in vivo* investigation of heat tolerance of bone tissue and temperature rise in the drilling of cortical bone. *Thesis. University of Gothenburg, Sweden.*

Feith R (1975). Side–effects of acrylic cement implanted into bone. A histologic, (micro)angiographic, fluorescence–microscopic and autoradiographic study in the rabbit femur*Acta Orthop Scand (Suppl 161):1–136.*

Figgie HE, Goldberg VM, Heiple KG, Moller HS, Gordon NH (1986). The influence of tibial–patellofemoral location on function of the knee in patients with the posterior stabilized condylar knee prosthesis. *J Bone Joint Surg 68–A:1035–1040.*

Frankel VH, Burstein AH, Brooks DB (1971). Biomechanics of internal derangement of the knee. Pathomechanics as determined by analysis of the instant centers of motion. *J Bone Joint Surg 53–A:945–962.*

Freeman MAR, Swanson SAV, Todd RC (1973). Total replacement of the knee using the Freeman–Swanson knee prosthesis. *Clin Orthop 94:153–170.*

Freeman MAR, Insall JN, Besser W, Walker PS, Hallel T (1977). Excision of the cruciate ligaments in total knee replacement. *Clin Orthop 126:209–212.*

Freeman MAR, Todd RC, Bamert P, Day WH (1978). ICLH arthroplasty of the knee: 1968 - 1977. *J Bone Joint Surg 60–B:339–344.*

Freeman MAR, Samuelson KM, Bertin KC (1985). Freeman–Samuelson total arthroplasty of the knee. *Clin Orthop 192:46–58.*

Freeman MAR and Railton GT (1988). Should the posterior cruciate ligament be retained or resected in condylar nonmeniscal knee arhroplasty? The case for resection. *J Arthroplasty 3 (Suppl):3–12.*

Freeman PA (1973). Walldius arthroplasty. A review of 80 cases. *Clin Orthop 94:85–91.*

Frost HM (1989). The biology of fracture healing. An overview for clinicians. *Clin Orthop 248:283–309.*

Galante JO, Rostocker W, Lueck R, Ray RD (1971). Sintered fiber metal composites as a basis for attachment of implants to bone. *J Bone Joint Surg 53–A:101–114.*

Garcia–Cimbrelo E (1992). Acetabular cup loosening with time in low–friction arthroplasty. *Acta Orthop Scand 63 (Suppl 248):38.*

Geesink RGT, DeGroot K, Klein CPAT (1987). Chemical implant fixation using hydroxyl–apatite coatings. The development of a human total hip prosthesis for chemical fixation to bone using hydroxyl–apatite coatings on titanium substrates. *Clin Orthop 225:147–170.*

Gelman MI, Coleman RE, Stevens PM, Davey BW (1978). Radiography, radionuclide imaging, and arthrography in the evaluation of total hip and knee replacement. *Radiology 128:677–682.*

Goldstein SA, Ku JL, Hollister S, Kayner DC, Matthews LS (1986). The effects of applied stress on experimentally controlled remodeling in trabecular bone. *32nd Annual ORS 11:432.*

Goodfellow J and O'Connor J (1978). The mechanics of the knee and prosthesis design. *J Bone Joint Surg 60–B:358–369.*

Goodfellow J (1992). Knee prostheses – one step forward, two steps back. Editorial. *J Bone Joint Surg 74–B:1–2.*

Goodman SB, Fornasier VL, Kei J (1988). The effects of bulk *versus* particulate polymethylmethacrylate on bone. *Clin Orthop 232:255–262.*

Goodman S and Lidgren L (1992). Polyethylene wear in knee arthroplasty. A review. *Acta Orthop Scand 63:358–364.*

Grady–Benson JC, Kaufman KR, Irby SE, Colwell CW (1992). The effect of posterior tibial slope on tibiofemoral forces after total knee arthroplasty. *38th Annual ORS 17:270.*

Granhed H, Malchau H, Herberts P, Johansson O (1991). A 7–9 – year follow–up of THR operated on with low viscosity cement. *Acta Orthop Scand 62 (Suppl 246):2.*

Grewal R, Rimmer MG, Freeman MAR (1992). Early migration of prostheses related to long–term survivorship. Comparison of tibial components in knee replacement. *J Bone Joint Surg 74–B:239–242.*

Gunston FH (1971). Polycentric knee arthroplasty. Prosthetic simulation of normal knee movement. *J Bone Joint Surg 53–B:272–277.*

Haddad RJ, Cook SD, Thomas KA (1987). Current concepts review: Biological fixation of porous–coated implants. *J Bone Joint Surg 69–A:1459–1466.*

Hagena FW, Hofmann GO, Mittlmeier T, Wasmer G, Bergmann M (1989). The cruciate ligaments in knee replacement. *Int Orthop 13:13–16.*

Halawa M, Lee AJC, Ling RSM, Vangala SS (1978). The shear strength of trabecular bone from the femur, and some factors affecting the shear strength of the cement–bone interface. *Arch Orthop Trauma Surg 92:19–30.*

Hallén LG and Lindahl O (1966). The "screw–home" movement in the knee–joint. *Acta Orthop Scand 37:97–106.*

Harding ML, Harding L, Goodfellow JW (1977). A preliminary report of a simple rig to aid study of the functional anatomy of the cadaver human knee joint. A technical note. *J Biomechanics 10:517–523.*

Harrington IJ (1983). Static and dynamic loading patterns in knee joints with deformities. *J Bone Joint Surg 65–A:247–259.*

Harris WH (1975). A new approach to total hip replacement without osteotomy of the greater trochanter. *Clin Orthop 106:19–26.*

Havdrup C, Hulth A, Telhag H (1976). The subchondral bone in osteoarthritis and rheumatoid arthritis of the knee. *Acta Orthop Scand 47:345–350.*

Hofmann AA, Wyatt RWB, Daniels AU, Armstrong L, Alazraki N, Taylor A (1990). Bone scans after total knee arthroplasty in asymptomatic patients. Cemented versus cementless. *Clin Orthop 251:183–188.*

Hsieh H–H and Walker PS (1976). Stabilizing mechanisms of the loaded and unloaded knee joint. *J Bone Joint Surg 58A:87–93.*

Hungerford DS and Krackow KA (1985). Total joint arthroplasty of the knee. *Clin Orthop 192:23–33.*

Hunter JC, Hattner RS, Murray WR, Genant HK (1980). Loosening of the total knee arthroplasty: Detection by radionuclide bone scanning. *Am J Roentgenol 135:131–136.*

Huntington LJ, Kendall JP, Tietjens BR (1979). A method of measuring from photographic records the movements of the knee joint during walking. *Eng Med 8:143–148.*

Hvid I, Kjaersgaard–Andersen P, Wethelund J–O, Sneppen O (1987). Knee arthroplasty in rheumatoid arthritis. Four– to six–year follow–up study. *J Arthroplasty 2:233–239.*

Hvid I, Bentzen SM, Jörgensen J (1988). Remodeling of the tibial plateau after knee replacement. *Acta Orthop Scand 59:567–573.*

Hvid I (1988a). Mechanical strength of trabecular bone at the knee. *Dan Med Bull 35:345–365.*

Hvid I (1988b). Trabecular bone strength at the knee. *Clin Orthop 227:210–221.*

Insall JN, Ranawat CS, Aglietti P, Shine J (1976a). A comparison of four models of total knee–replacement prostheses. *J Bone Joint Surg 58–A:754–765.*

Insall JN, Ranawal CS, Scott WN et al (1976b). Total Condylar knee replacement. Preliminary report. *Clin Orthop 120:149–154.*

Insall JN, Hood RW, Flawn LB, Sullivan DJ (1983). The Total Condylar knee prosthesis in gonarthrosis. *J Bone Joint Surg 65–A:619–628.*

Insall JN (1984). Total knee replacement. *In:* Insall JN ed: *Surgery of the knee. Churchill Livingstone, New York, Edinburgh, London, Melbourne:587–696.*

Insall JN (1988). Presidential address to the Knee Society: Choices and compromises in total knee arthroplasty. *Clin Orthop 226:43–48.*

Isacson JBI (1987). The knee joint in rheumatoid arthritis. An epidemiologic, radiologic and goniometric study. *Thesis, Karolinska Institutet, Stockholm, Sweden.*

Jasty M, Floyd WE, Schiller AL, Goldring SR, Harris WH (1986). Localized osteolysis in stable, non–septic total hip replacement. *J Bone Joint Surg 68–A:912–919.*

Jasty M, Maloney WJ, Bragdon CR, O'Connor DO, Haire T, Harris WH (1991). The initiation of failure in cemented femoral components of hip arthroplasties. *J Bone Joint Surg 73–B:551–558.*

Jensen LN and Jensen JS (1991). Improved bone response to a new bone cement. *Acta Orthop Scand (Suppl 246):6.*

Johansson CB (1991). On tissue reactions to metal implants. *Thesis. University of Gothenburg, Sweden.*

Jones WN, Aufranc OE, Kermond WL (1967). Mold arthroplasty of the knee. *J Bone Joint Surg 49–A:1022.*

Jones EC, Insall JN, Inglis AE, Ranawat CS (1979). GUEPAR knee arthroplasty results and late complications. *Clin Orthop 140:145–152.*

Jones LC and Hungerford DS (1987). Cement disease. *Clin Orthop 225:192–206.*

Jones SMG, Pinder IM, Moran CG, Malcolm AJ (1992). Polyethylene wear in uncemented knee replacements. *J Bone Joint Surg 74–B:18–22.*

Jonsson H, Kärrholm J, Elmqvist L–G (1989). Kinematics of active knee extension after tear of the anterior cruciate ligament. *Am J Sports Med 17:796–802.*

Jonsson H and Kärrholm J (1992). Three–dimensional knee joint movements and positions of helical axes during weight–bearing. Evaluation after anterior cruciate ligament rupture. *Submitted.*

Kantor SG, Schneider R, Insall JN, Becker MW (1990). Radionuclide imaging of asymptomatic versus symptomatic total knee arthroplasties. *Clin Orthop 260:118–123.*

Kapandji IA (1970). The physiology of the joints. Vol II. *E&S Livingstone (Eds) Edinburg–London.*

Kaufer H and Matthews LS (1981). Spherocentric arthroplasty of the knee. Clinical experience with an average four–year follow–up. *J Bone Joint Surg 63–A:545–559.*

Kavolus CH, Ritter MA, Keating EM, Faris PM (1991). Survivorship of cementless total knee arthroplasty without tibial plateau screw fixation. *Clin Orthop 273:170–176.*

Kettelkamp DB, Johnson RJ, Smidt GL, Chao EYS, Walker M (1970). An electrogoniometric study of knee motion in normal gait. *J Bone Joint Surg 52A:775–790.*

Kettelkamp DB and Nasca R (1973). Biomechanics and knee replacement arthroplasty. *Clin Orthop 94:8–14.*

Kilgus DJ, Moreland JR, Finerman GAM, Funahashi TT, Tipton JS (1991). Catastrophic wear of tibial polyethylene inserts. *Clin Orthop 273:223–231.*

Kilgus DJ, Funahashi TT, Campbell PA (1992). Massive femoral osteolysis and early disintegration of a polyethylene–bearing surface of a total knee replacement. *J Bone Joint Surg 74–A:770–774.*

Knahr K, Salzer M, Schmidt W (1990). A radiological analysis of uncemented PCA tibial implants with a follow–up period of 4 – 7 years. *J Arthroplasty 5:131–141.*

Knutson K, Lindstrand A, Lidgren L (1986). Survival of knee arthroplasties. A nation-wide multicentre investigation of 8000 cases. *J Bone Joint Surg 68–B:795–803.*

Krause WR, Krug W, Miller J (1982). Strength of the cement–bone interface. *Clin Orthop 163:290–299.*

Kurosawa H, Walker PS, Abe S, Garg A, Hunter T (1985). Geometry and motion of the knee for implant and orthotic design. *J Biomech 18:487–498.*

Kärrholm J, Selvik G, Elmqvist L–G, Hansson L–I, Jonsson H (1988a). Three-dimensional instability of the anterior cruciate deficient knee. *J Bone Joint Surg 70–B:777–783.*

Kärrholm J, Selvik G, Elmqvist L–G, Hansson L–I (1988b). Active knee motion after cruciate ligament rupture. Stereoradiography. *Acta Orthop Scand 59:158–164.*

Kärrholm J (1989). Roentgen stereophotogrammetry. Review of orthopedic applications. *Acta Orthop Scand 60:491–503.*

Kärrholm J and Snorrason F (1992a). Late migration of porous coated acetabular prostheses fixed with screws. Roentgen stereophotogrammetric analysis. *J Orthop Res, in press.*

Kärrholm J and Snorrason F (1992b). Subsidence, tip and hump micromovements of non–coated ribbed femoral prostheses. *Clin Orthop, in press.*

Lafortune MA and Cavanagh PR (1985): The measurement of normal knee joint motion during walking using intracortical pins. *In* Whittle M, Harris D, eds: *Biomechanical measurement in orthopaedic practice. Oxford Medical Engineering series 5. Clarendon Press Oxford: 234–243.*

Landon GC, Galante JO, Casini J (1985). Essay on total knee arthroplasty. *Clin Orthop 192:69–74.*

Landon GC, Galante JO, Maley MM (1986). Noncemented total knee arthroplasty. *Clin Orthop. 205:49–57.*

Lange DR (1979). The mechanical bonding of methylmetacrylate to cancellous bone. Effect of a hemostatic agent. *J Bone Joint Surg 61–A:254–256.*

Laskin RS (1981). Total Condylar knee replacement in rheumatoid arthritis. A review of one hundred and seventeen knees. *J Bone Joint Surg 63–A:29–35.*

Laskin RS (1988). Tricon–M uncemented total knee arthroplasty. A review of 96 knees followed for longer than 2 years. *J Arthroplasty 3:27–38.*

Laskin RS (1990). Total Condylar knee replacement in patients who have rheumatoid arthritis. A ten–year follow–up study. *J Bone Joint Surg 72–A:529–535.*

Laskin RS (1991). Uncemented total knee replacement: The Tricon–M and Genesis prostheses. *In* Laskin RD ed: *Total Knee Replacement. Springer Verlag, New York, Berlin, Heidelberg: 162–182.*

Lee RW, Volz RG, Sheridan CD (1991). The role of fixation and bone quality on the mechanical stability of tibial knee components. *Clin Orthop 273:177–183.*

Lee J–M, Salvati EA, Betts F, DiCarlo EF, Doty SB, Bullough PG (1992). Size of metallic and polyethylene debris particles in failed cemented total hip replacements. *J Bone Joint Surg: 74–B:380–384.*

Lereim P, Goldie I, Dahlberg E (1974). Hardness of the subchondral bone of the tibial condyles in the normal state and in osteoarthritis and rheumatoid arthritis. *Acta Orthop Scand 45:614–627.*

Levens AS, Inman VT, Blosser JA (1948). Transverse rotation of the segments of the lower extremity in locomotion. *J Bone Joint Surg 30–A:859–872.*

Lew WD and Lewis JL (1982). The effect of knee–prosthesis geometry on cruciate ligament mechanics during flexion. *J Bone Joint Surg 64–A:734–739.*

Lewallen DG, Bryan RS, Peterson LFA (1984). Polycentric total knee arthroplasty. A ten–year follow–up study. *J Bone Joint Surg 66–A:1211–1218.*

Lewold S, Knutson K, Lidgren L (1992). Reduced failure rate in knee prosthetic surgery with improved implantation technique. *Clin Orthop. In press.*

Lidgren L, Bodelind B, Möller J (1987). Bone cement improved by vacuum mixing and chilling. *Acta Orthop Scand 58:27–32.*

Lindstrand A, Stenström A, Lewold S (1992). Multicenter study of unicompartmental knee revision. PCA, Marmor, and St Georg compared in 3,777 cases of arthrosis. *Acta Orthop Scand 63:256–259.*

Lindstrand A and Stenström A (1992). Polyethylene wear of the PCA unicompartmental knee. Prospective 5 (4–8) year study of 120 arthrosis knees. *Acta Orthop Scand 63:260–262.*

Lotke PA and Ecker ML (1977). Influence of positioning of prosthesis in total knee replacement. *J Bone Joint Surg 59–A:77–79.*

Lundberg A, Svensson OK, Németh G, Selvik G (1989). The axis of rotation of the ankle joint. *J Bone Joint Surg 71-B:94-99.*

MacIntosh DL (1958). Hemiarthroplasty of the knee using space occupying prosthesis for painful varus and valgus deformities. *J Bone Joint Surg 40-A:1431.*

Maloney WJ, Jasty M, Rosenberg A, Harris WH (1990). Bone lysis in well-fixed cemented femoral components. *J Bone Joint Surg 72-B:966-970.*

Markolf KL, Bargar WL, Shoemaker SC, Amstutz HC (1981). The role of joint load in knee stability. *J Bone Joint Surg 63-A:570-585.*

McKeever DC (1960). Tibial plateau prosthesis. *Clin Orthop 18:86-95.*

Miller J, Burke DL, Krause W, Ahmed A, Kelebay L, Tremblay G (1979). Improved fixation of knee arthroplasty components by the injection of acrylic cement into cancellous bone surfaces. *J Bone Joint Surg 61-B:515.*

Miller J (1984). Fixation in total knee arthroplasty. *In:* Insall JN ed: *Surgery of the knee. Churchill Livingstone, New York, Edinburgh, London, Melbourne:717-728.*

Mjöberg B, Brismar J, Hansson LI, Pettersson H, Selvik G, Önnerfält R (1985). Definition of endoprosthetic loosening. Comparison of arthrography, scintigraphy and roentgen stereophotogrammetry in prosthetic hips. *Acta Orthop Scand 56:469-473.*

Mjöberg B (1986). Loosening of the cemented hip prosthesis. The importance of heat injury. *Acta Orthop Scand (Suppl 221):1-40.*

Mjöberg B, Rydholm A, Selvik G, Önnerfält R (1987). Low- versus high-viscosity bone cement. Fixation of hip prostheses analyzed by roentgen stereophotogrammetry. *Acta Orthop Scand 58:106-108.*

Mjöberg B and Selvik G (1988). Reduced risk of loosening of hip prostheses with a new cold-curing bone cement. *Acta Orthop Scand 59:343-345.*

Moran CG, Pinder IM, Lees TA, Midwinter MJ (1991). Survivorship analysis of the uncemented Porous-Coated Anatomic knee replacement. *J Bone Joint Surg 73-A:848-857.*

Morberg P (1991). On bone tissue reactions to acrylic cement. *Thesis. University of Gothenburg, Sweden.*

Morberg P and Albrektsson T (1991). Removal torque for bone-cement and titanium screws implanted in rabbits. *Acta Orthop Scand 62:554-556.*

Moreland JR (1988). Mechanisms of failure in total knee arthroplasty. *Clin Orthop 226:49-64.*

Morrison JB (1968). Bioengineering analysis of force actions transmitted by the knee joint. *Bio-Med Eng 3:164-170.*

Morrison JB (1969). Function of the knee joint in various activities. *Bio-Med Eng 4:573-580.*

Morrison JB (1970). The mechanics of the knee joint in relation to normal walking. *J Biomech 3:51–61.*

Morscher EW (1987). Current state of cementless fixation of endoprostheses. *Swiss Med 9(8):29–39.*

Murase K, Crowninshield RD, Pedersen DR, Chang T-S (1983). An analysis of tibial component design in total knee arthroplasty. *J Biomech 16:13–22.*

Nafei A, Nielsen S, Kristensen O, Hvid I (1992). The press-fit Kinemax knee arthroplasty. High failure rate of non-cemented implants. *J Bone Joint Surg 74-B:243–246.*

Natarajan R and Andriacchi TP (1988). The influence of displacement incompatibilities on bone ingrowth in porous tibial components. *34th Annual ORS 13:331.*

Nilsson KG, Brobäck LG, Kärrholm J (1989). Does a stem on the tibial component improve fixation of uncemented Freeman-Samuelson knee prosthesis? *Acta Orthop Scand (Suppl 231):34.*

Nisell R, Németh G, Ohlsén H (1986). Joint forces in extension of the knee. Analysis of a mechanical model. *Acta Orthop Scand 57:41–46.*

Noble PC and Swarts E (1983). Penetration of acrylic bone cements into cancellous bone. *Acta Orthop Scand 54:566–573.*

Nolan JF and Bucknill TM (1992). Aggressive granulomatosis from polyethylene failure in an uncemented knee replacement. *J Bone Joint Surg 74-B:23–24.*

Nordin M and Frankel VH (1989). Biomechanics of the knee. *In* Nordin M, Frankel VH, eds: *Basic biomechanics of the musculosceletal system. 2nd ed. Lea & Fabiger, Philadelphia, London:115–134.*

Nyström L (1990). Algorithms and program system for roentgen stereophotogrammetric analysis. *Thesis. Institute of Information Processing, University of Umeå, Sweden.*

Nyström L and Söderkvist I (1990). Kinerr. Methods for detecting and correcting systemic errors in roentgen stereophotogrammetric analysis. *Technical report UMINF-184.90, Institute of Information Processing, University of Umeå:1–34.*

Pauwels F (1980). Biomechanics of the locomotor apparatus. *Springer Verlag, Berlin.*

Peters PC, Engh GA, Dwyer KA (1991). Aggressive endosteal erosion following cementless total knee arthroplasty. In: *Transactions of the AAOS 58th Annual Meeting:201.*

Pilliar RM, Lee JM, Maniatopoulos C (1986). Observations on the effect of movement on bone ingrowth into porous-surfaced implants. *Clin Orthop 208:108–113.*

Ragnarsson JI and Kärrholm J (1992). Factors influencing postoperative movement in displaced femoral neck fractures: Evaluation by conventional radiography and stereoradiography. *J Orthop Trauma 6:152–158.*

Ranawat CS and Shine JJ (1973). Duo-condylar total knee arthroplasty. *Clin Orthop 94:185–195.*

Ranawat CS, Jordan L, Ramsay Straub L (1973). Mac Intosh hemiarthroplasty in rheumatoid knee. *Acta Orthop Belgica 39:102–112.*

Ranawat CS, Johansson NA, Rimnac CM, Wright TM, Schwartz RE (1986). Retrieval analysis of porous-coated components for total knee arthroplasty. *Clin Orthop 209:244–248.*

Ranawat CS and Boachie-Adjei O (1988). Survivorship analysis and results of Total Condylar knee arthroplasty. Eight- to 11- year follow-up period. *Clin Orthop 226:6–13.*

Rand JA and Bryan RS (1982). Revision after total knee arthroplasty. *Orthop Clin North Am 13:201–212.*

Rand JA and Coventry MB (1988). Ten-year evaluation of Geometric total knee arthroplasty. *Clin Orthop 232:168–173.*

Rand JA and Ilstrup DM (1991). Survivorship analysis of total knee arthroplasty. Cumulative rates of survival of 9200 total knee arthroplasties. *J Bone Joint Surg 73-A:397–409.*

Reuleaux F (1876). The kinematics of machinery: Outline of a theory of machines. *London, Macmillan.*

Reilly D, Walker PS, Ben-Dov M, Ewald FC (1982). Effects of tibial components on load transfer in the upper tibia. *Clin Orthop 165:273–282.*

Riley D and Woodyard JE (1985). Long-term results of Geomedic total knee replacement. *J Bone Joint Surg 67-B:548–550.*

Ritter MA, Campbell E, Faris PM, Keating EM (1989). Long-term survival analysis of the posterior cruciate condylar total knee arthroplasty. A 10-year evaluation. *J Arthroplasty 4:293–96.*

Rittman N, Kettelkamp DB, Pryor P, Schwartzkopf GL, Hillberry B (1981). Analysis of patterns of knee motion walking for four types of total knee implants. *Clin Orthop 155:111–117.*

Rorabeck CH, Bourne RB, Nott L (1988). The cemented Kinematic-II and the non-cemented Porous-Coated Anatomic prostheses for total knee replacement. A prospective evaluation. *J Bone Joint Surg 70-A:483–490.*

Rosenberg AG, Andriacchi TP, Barden R, Galante JO (1988). Patellar component failure in cementless total knee arthroplasty. *Clin Orthop 236:106–114.*

Rosenberg AG, Barden RM, Galante JO (1990). Cemented and ingrowth fixation of the Miller-Galante prosthesis. Clinical and roentgenographic comparison after three- to six-year follow-up studies. *Clin Orthop 260:71–79.*

Rosenthall L, Lepanto L, Raymond F (1987). Radiophosphate uptake in asymptomatic knee arthroplasty. *J Nucl Med 28:1546–1549.*

Rozing PM, Bohne WH, Insall JN (1982). Bone scanning for the evaluation of knee prosthesis. *Acta Orthop Scand 53:291–294.*

Ryd L (1985). Micromotion in knee arthroplasty. A roentgen stereophotogrammetric analysis of four concepts of prosthetic fixation. *Thesis. University of Lund, Sweden.*

Ryd L (1986). Micromotion in knee arthroplasty. A roentgen stereophotogrammetric analysis of tibial component fixation. *Acta Orthop Scand (Suppl 220):1–80.*

Ryd L, Albrektsson BEJ, Herberts P, Lindstrand A, Selvik G (1988). Micromotion of noncemented Freeman–Samuelson knee prostheses in gonarthrosis. *Clin Orthop 229:205–212.*

Ryd L and Linder L (1989). On the correlation between micromotion and histology of the bone–cement interface: Report of three cases of knee arthroplasty followed by roentgen stereophotogrammetric analysis. *J Arthroplasty 4:303–309.*

Ryd L, Lindstrand A, Stenström A, Selvik G (1990). Porous coated anatomic tricompartmental tibial components. The relationship between prosthetic position and micromotion. *Clin Orthop 251:189–197.*

Ryd L, Toksvig-Larsen S, Stenström A, Carlsson L, Herberts P, Selvik G (1991). The stability of tibial components in the postoperative phase. *37th Annual ORS 16:559*

Ryd L (1992). Roentgen stereophotogrammetric analysis of prosthetic fixation in the hip and knee joint. *Clin Orthop 276:56–65.*

Rydberg J (1983). Preoperativ undersökningsteknik vid gonarthros. *Proc Swed Soc Radiol 16:11–13.*

Röstlund TV (1990).On the development of a new arthroplasty. With special emphasis on the gliding elements in the knee. *Thesis. University of Gothenburg, Sweden.*

Saha S and Pal S (1984). Mechanical properties of bone cement: A review. *J Biomed Mater Res 18:435–462.*

Sandborn PM, Cook SD, Spires WP, Kester MA (1988). Tissue response to porous-coated implants lacking initial bone apposition. *J Arthroplasty 3:337–346.*

Santavirta S, Hoikka V, Eskola A, Konttinen YT, Paavilainen T, Tallroth K (1990). Aggressive granulomatous lesions in cementless total hip arthroplasty. *J Bone Joint Surg 72–B:980–984.*

Schneider R, Freiberger RH, Ghelman B, Ranawat CS (1982). Radiologic evaluation of painful joint prostheses. *Clin Orthop 170:156–168.*

Schneider R and Soudry M (1986). Radiographic and scintigraphic evaluation of total knee arthroplasty. *Clin Orthop 205:108–120.*

Scuderi GR, Insall JN, Windsor RE, Moran MC (1989). Survivorship of cemented knee replacements. *J Bone Joint Surg 71–B:798–803.*

Seireg A and Arvikar RJ (1975). The prediction of muscular load sharing and joint forces in the lower extremities during walking. *J Biomech 8:89–102.*

Selvik G (1974, 1989). Roentgen–stereophotogrammetric method for the study of the kinematics of the skeletal system. *Thesis. University of Lund, Sweden.* Reprint *Acta Orthop Scand (Suppl 232):1–51.*

Shiers LGP (1954). Arthroplasty of the knee. Preliminary report of a new method. *J Bone Joint Surg 36–B:553.*

Shimagaki H, Bechtold JE, Sherman RE, Gustilo RB (1990). Stability of initial fixation of the tibial component in cementless total knee arthroplasty. *J Orthop Res 8:64–71.*

Shoemaker SC, Markolf KL, Finerman GAM (1982). *In vitro* stability of the implanted Total Condylar prosthesis. *J Bone Joint Surg 64–A:1201–1213.*

Sledge CB and Walker PS (1984). Total knee arthroplasty in rheumatoid arthritis. *Clin Orthop 182:127–136.*

Smillie IS (1978). Biomechanics of rotation: mechanisms of injuries of the menisci: and sequelae. *In:* Smillie IS: *Injuries of the knee joint. Churchill Livingstone, Edinburgh, London, New York:71–82*

Smith BE, Askew MJ, Gradisar IA, Gradisar JS, Lew MM (1992). The effect of patient weight on the functional outcome of total knee arthroplasty. *Clin Orthop 276:237–244.*

Snorrason F, Kärrholm J, Löwenhielm G, Hietala S–O, Hansson LI (1989). Poor fixation of the Mittelmeier hip prosthesis. A clinical, radiographic, and scintimetric evaluation. *Acta Orthop Scand 60:81–85.*

Snorrason F and Kärrholm J (1990). Primary migration of fully–threaded acetabular prostheses. A reontgen stereophotogrammetric analysis. *J Bone Joint Surg 72–B:647–652.*

Soudan K, van Audekercke R, Martens M (1979). Methods, difficulties and inaccuracies in the study of human joint kinematics and pathokinematics by the instant axis concept. Example: the knee joint. *J Biomech 12:27–33.*

Soudan K, van Audekercke R, Martens M (1980). Exact knowledge of knee motion kinematics as a basis for Well–designed knee prostheses: the instant axis concept. *Acta Orthop Belg 46:757–765.*

Soudry M, Walker PS, Reilly DT, Kurosawa H, Sledge CB (1986). Effects of total knee replacement design on femoral–tibial contact conditions. *J Arthroplasty 1:35–45.*

Stern SH and Insall JN (1990). Total knee arthroplasty in obese patients. *J Bone Joint Surg 72–A:1400–1404.*

Strickland AB, Chan KH, Andriacchi TP, Miller J (1988). The initial fixation of porous coated tibial components evaluated by the study of rigid body motion under static load. *34th Annual ORS 13:476*

Stürup J, Kramhöft M, Nimb L, Steen–Jensen J (1992). The effects of polymerization heat and toxicity from PMMA bone cement and the effects of a new MMA/DMA/IBMA bone cement. *Acta Orthop Scand (Suppl 248):27.*

Sumner DR, Jacobs JJ, Turner TM, Urban RM, Galante JO (1989). The amount and distribution of bone ingrowth in tibial components retrieved from human patients. *35th Annual ORS 14:375.*

Söballe K, Hansen ES, Rasmussen HB, Jörgensen PH, Bünger C (1990a). Hydroxyapatite coating provides a stronger fibrous anchorage of implants with controlled micromotion. *36th Annual ORS 15:206.*

Söballe K, Stender-Hansen E, Brockstedt-Rasmussen H, Möger-Pedersen C, Bünger C (1990b). Hydroxyapatite coating enhances fixation of porous coated implants. A comparison in dogs between press fit and noninterference fit. *Acta Orthop Scand 61:299-306.*

Söballe K, Brockstedt-Rasmussen H, Stender-Hansen E, Bünger C (1992a). Hydroxyapatite coating modifies implant membrane formation. Controlled micromotion studied in dogs. *Acta Orthop Scand 63:128-140.*

Söballe K, Hansen ES, Brockstedt Rasmussen H, Jörgensen PH, Bünger C (1992b). Tissue ingrowth titanium and hydroxyapatite-coated implants during stable and unstable mechanical conditions. *J Orthop Res 10:285-299.*

Söderkvist I (1990). Some numerical methods for kinematical analysis. *Thesis. Institute of Information Processing, University of Umeå, Sweden.*

Tew M and Waugh W (1982). Estimating the survival time of knee replacements. *J Bone Joint Surg 64-B:579-582.*

Tew M and Waugh W (1985). Tibiofemoral alignment and the results of knee replacement. *J Bone Joint Surg 67-B:551-556.*

Tew M, Waugh W, Forster IW (1985). Comparing the results of different types of knee replacement. A method proposed and applied. *J Bone Joint Surg 67-B:775-779.*

Thornhill TS, Dalziel RW, Sledge CB (1982). Alternatives to arthrodesis for the failed total knee arthroplasty. *Clin Orthop 170:131-140.*

Tibrewall SB, Grant KA, Goodfellow JW (1984). The radiolucent line beneath the tibial components of the Oxford meniscal knee. *J Bone Joint Surg 66-B:523-528.*

Toksvig-Larsen S and Ryd L (1991). Surface flatness after bone cutting. A cadaver study of tibial condyles. *Acta Orthop Scand 62:15-18.*

Toksvig-Larsen S, Franzen H, Ryd L (1991). Cement interface temperature in hip arthroplasty. *Acta Orthop Scand 62:102-105.*

Toksvig-Larsen S (1992). On bone cutting. *Thesis. University of Lund, Sweden.*

Townley CO (1964). Articular-plate replacement arthroplasty for the knee joint. *Clin Orthop 36:77-85.*

Townley CO (1985). The anatomic total knee resurfacing arthroplasty. *Clin Orthop 192:82-96.*

Townley CO (1988). Total knee arthroplasty. A personal retrospective and prospective review. *Clin Orthop 236:8-22.*

Townsend MA, Izak M, Jackson RW (1977). Total motion knee goniometry. *J Biomech 10:183-193.*

Turner TM, Sumner DR, Urban RM, Rivero DP, Galante JO (1986). A comparative study of porous coatings in a weight-bearing total hip-arthroplasty model. *J Bone Joint Surg 68-A:1396-1409.*

Uematsu O, Hsu HP, Kelley KM, Ewald FC, Walker PS (1987). Radiographic study of kinematic total knee arthroplasty. *J Arthroplasty 2:317-326.*

Vince KG, Insall JN, Kelly MA (1989). The Total Condylar prosthesis. 10- to 12-year results of a cemented knee replacement. *J Bone Joint Surg 71-B:793-797.*

Volz RG, Nisbet JK, Lee RW, McMurtry MG (1988). The mechanical stability of various noncemented tibial components. *Clin Orthop 226:38-42.*

Walker PS, Shoji H, Erkman MJ (1972). The rotational axis of the knee and its significance to prosthesis design. *Clin Orthop 89:160-170.*

Walker PS and Shoji H (1973). Development of a stabilizing knee prosthesis employing physiologic principles. *Clin Orthop 94:222-233.*

Walker PS, Wang C-J, Masse Y (1975). Joint laxity as a criterion for the design of condylar knee prostheses In: *Total Knee Replacement 22-29, New York Mechanical Engineering Publications:22-29.*

Walker PS, Hsu H-P, Zimmerman RA (1990). A comparative study of uncemented tibial components. *J Arthroplasty 5:245-253.*

Walldius B (1957). Arthroplasty of the knee using an endoprosthesis. *Acta Orthop Scand (Suppl 24):1-112.*

Walldius B (1960). Arthroplasty of the knee using an endoprosthesis. 8 years' experience. *Acta Orthop Scand 30:137-148.*

Waugh TR, Smith RC, Orofino CF, Anzel SM (1973). Total knee replacement. Operative technique and preliminary results. *Clin Orthop 94:196-201.*

Weidenhielm L (1992). Knee osteoarthrosis. Aspects on clinical symptoms, corrective surgery, leg alignment and knee joint load. *Thesis. Karolinska Institutet, Stockholm, Sweden.*

Werner F, Foster D, Murray DG (1978). The influence of design on the transmission of torque across knee prostheses. *J Bone Joint Surg 60-A:342-348.*

Windsor RE, Insall JN, Sculco TP (1986). Bone grafting of tibial defects in primary and revision total knee arthroplasty. *Clin Orthop 205:132-137.*

Windsor RE, Scuderi GR, Moran MC, Insall JN (1989). Mechanisms of failure of the femoral and tibial components in total knee arthroplasty. *Clin Orthop 248:15-20.*

Woltring HJ, Huiskes R, De Lange A (1985). Finite centroid and helical axis estimation from noisy landmark measurements in the study of human joint kinematics. *J Biomech 18:379-389.*

Wright TM, Rimnac CM, Stulberg SD, Mintz L, Tsao AK, Klein RW, McCrae C (1992). Wear of polyethylene in total joint replacements. Observations from retrieved PCA knee implants. *Clin Orthop 276:126-134.*

Young HH (1963). Use of a hinged vitallium prosthesis for arthroplasty of the knee. *J Bone Joint Surg 45-A:1627-1642.*

Appendix

Clinical, radiographic, and roentgen stereophotogrammetric data in the total material.

Stereo nr:	Case number
Side:	R = right L = left
Gender:	F = female M = male
Fixation:	C = cemented UC = uncemented
Diagnosis:	OA = osteoarthrosis RA = rheumatoid arthritis SFOA = arthrosis secondary to fracture
HSS:	Hospital for Special Surgery knee score
HKA:	Hip Knee Ankle angle, leg alignment (< 180° varus; > 180° valgus alignment)
Alignm. a–d:	Alignment of the femoral (a,b) and the tibial component (c,d). See Fig. 4.

Patient	1	2		3	4	5	6
Stereo nr	950	951	970	952	953	954	955
Side	R	R	L	R	R	R	R
Gender	F	F	F	F	F	F	M
Age	67	62	63	68	67	62	65
Prosthesis	Tricon-M	Tricon-M	Tricon-M	Tricon-M	Tricon-M	Tricon-M	Tricon-M
Fixation	C	UC	UC	UC	C	C	C
Diagnosis	SFOA	OA	OA	OA	OA	RA	RA
Weight	85	62	62	78	94	85	70
HSS pre			50			67	39
6m			58	71	71	90	71
12m	87	76	69	84	89	97	84
24m	90	71	70	92	96	93	91
5 y	86		78	84	94	61	68
HKA pre	171	200	195	190	188	187	170
post	174	193	188	184	187	180	176
Alignm. a	6	16	11	11	7	7	8
b	4	31	6	8	6	10	7
c	83	86	88	88	90	87	86
d	90	88	89	83	80	88	97
Paper	I, V, VII	I	I, V	I, V	I, V, VII	I, IV	I, IV, V
Migration:							
MTPM 6w							
3m			0,36			0,59	0,68
6m	1,61	0,89	0,9			1,3	0,86
12m	1,74		0,71		1,02	1,68	0,79
24m	2,31	1,47	1,02		1,36	1,64	1
5 y	3,03		1,67		2,3	2,6	1,72
Migration 2 y:							
Rotation X	−1,58	−0,11	0,49	−0,11	0,71	0,25	0,43
Y	−1,9	−1,34	0,3	−1,29	0,58	0,74	0,23
Z	−0,46	1,19	0,41	−0,11	0,67	−0,73	0,71
Migration 5 y:							
Rotation X	−1,27	could not be	0,08	−0,92	1,35	0,75	−0,18
Y	−3,34	analyzed	1,66	−1,05	1,45	1,44	0,9
Z	−1,55		0,28	−0,3	−0,82	−1,28	1,16

REVISION:							
cause:						Patella loosening	
months postop:						43	
what done:						Patella. rev	
prosthesis:						Tricon-P	

Patient	7		8	9	10	11	12
Stereo nr	956	651	957	958	959	960	961
Side	L	R	R	L	L	R	L
Gender	F	F	F	F	F	F	F
Age	76	78	73	66	71	78	78
Prosthesis	Tricon-M	Miller-G	Tricon-M	Tricon-M	Tricon-M	Tricon-M	Tricon-M
Fixation	UC	UC	UC	C	C	C	C
Diagnosis	OA	OA	OA	OA	RA	OA	OA
Weight	70	70	53	74	60	68	68
HSS pre	54	29	58	64	58	15	56
6m	87	80	88	78	65	73	79
12m	92	81	96	86		75	80
24m	92	77	95			86	
5 y	80		92			76	
HKA pre	188	203	169	200	188	155	195
post	182	180	179	178	182	188	188
Alignm. a	7	9	10	12	10	9	12
b	7	4	7	5	2	7	-2
c	84	88	92	83	83	87	86
d	86	75	91	90	89	89	88
Paper	I, IV	III, VI	I, IV, V	I, IV	I, IV	I, IV	I, IV
Migration:							
MTPM 6w		1,09					
3m	0,84	0,75	0,71	1,69	0,26	0,47	0,2
6m	0,85	0,6	0,72	2,16	0,26	0,51	0,25
12m	1,62	1,02	0,64	2,24	0,23	1,07	0,46
24m	0,98	0,44	0,64		0,35	1,68	
5 y	1,87		0,62			1,66	
Migration 2 y:							
Rotation X	−1,62	0,45	0,03		−0,53	−1,57	
Y	0,73	−0,09	−0,64		0,15	0,07	
Z	−0,34	0	−0,43		−0,21	2,22	
Migration 5 y:							
Rotation X	−1,5		0,73			−1,57	
Y	1,14		0,07			0,07	
Z	0,82		−0,42			2,22	
REVISION:							
cause:				Patella+tibia loos.			
months postop:				21			
what done:				Patella+tibia rev			
prosthesis:				Tricon-M stem			
					Died 36 mo postop		Died 15 mo postop

Patient	13		14	15	16	17	18
Stereo nr	963	671	964	965	966	967	968
Side	L	R	R	R	R	L	R
Gender	F	F	F	F	F	F	F
Age	62	64	64	72	67	36	76
Prosthesis	Tricon-M	Miller-G	Tricon-M	Tricon-M	Tricon-M	Tricon-M	Tricon-M
Fixation	C	C	UC	C	UC	UC	C
Diagnosis	OA	OA	OA	RA	OA	RA	OA
Weight	87	87	85	65	80	58	72
HSS pre	69	48	55	41	57	64	58
6m	82	83	65	75		55	72
12m	86	85	75	78	88	69	64
24m	87	78	74	86	92	74	65
5 y	95		29 before rev		94	73	
HKA pre	165	173	167	180	165	183	185
post	185	178	174	180	180	182	178
Alignm. a	10	7	8	13	7	12	11
b	-4	7	2	12	7	1	6
c	89	87	81	85	90	86	83
d	84	80	85	86	87	95	94
Paper	I, IV, V	III	I, IV	I, IV, V	I, IV, V	I, IV	I, IV
Migration:							
MTPM 6w	1,45		0,5	0,33	0,79	0,53	0,32
3m	1,97	0,24		0,89	1,1	1,02	0,48
6m	2,62	0,31	1,43	0,4	1,02	1,21	0,87
12m	2,84	0,65	1,83	0,96	1,4	1,38	2,33
24m	2,43	0,7	2,88	1	1,39	1,51	3,89
5 years	2,87				1,38	1,55	
Migration 2 y:							
Rotation X	-3,51	0,35	-3,12	-0,36	-0,83	-0,79	-2,43
Y	-2,29	-0,63	-1,83	0,37	-1,5	-0,94	-1,86
Z	-0,6	0,27	3,14	-0,31	0,03	-1,37	2,73
Migration 5 y:							
Rotation X	-4,16				-0,72	-0,75	
Y	-2,15				-1,64	-1,11	
Z	-1,31				-0,09	-1,42	

	13		14	15	16	17	18
REVISION:							
cause:			Tibia component fracture				
months postop:			59				
what done:			Total revision				
prosthesis:			MG rev.				
							Died 37 mo postop

Patient	19	20	21		22	23	24
Stereo nr	969	971	972	684	973	974	975
Side	L	R	R	L	R	R	L
Gender	F	M	F	F	F	F	M
Age	54	70	60	62	72	67	55
Prosthesis	Tricon-M	Tricon-M	Tricon-M	Miller-G	Tricon-M	Tricon-M	Tricon-M
Fixation	UC	UC	C	UC*	C	C	C
Diagnosis	RA	OA	OA	OA	RA	OA	RA
Weight	50	93	85	85			83
HSS pre	25	58	41	50	32	74	48
6m	87	96	82	85			77
12m	77	94	96	89		84	78
24m	90	96	95	97		87	
5 y	90	96	94				
HKA pre	186	165	175	173	190	191	180
post	183	184	785	182	180	180	182
Alignm. a	9	8	9	6			9
b	0	6	2	3			6
c	86	88	90	90			87
d	88	90	93	82			93
Paper	I, IV	I, V	I, IV	III	I	I	I, IV
Migration:							
MTPM 6w	0,71	1,01	0,13	1,28			0,13
3m	0,73	0,68	0,25	1,01			0,54
6m	1,39	0,79	0,49	0,95	0,81	0,34	0,59
12m	0,83	0,45	0,28	0,73	1,6	0,35	0,61
24m	0,98	0,68	0,45	0,84		0,52	
5 y	1,17	0,54	0,91				
Migration 2 y:							
Rotation X	0,88	-0,57	-0,27	0,53		-0,93	
Y	-0,91	-0,19	0,11	-0,63		-0,17	
Z	0	0,29	0,32	0,86		-0,02	
Migration 5 y:							
Rotation X	1,5	0,21	1,09				
Y	-0,77	0,17	0,41				
Z	-0,02	0,01	-0,09				
				* Incorrectly operated without cement. Discovered after 2-year follow-up			
REVISION:							
cause:							
months postop:							
what done:							
prosthesis:							
							Died 18 mo postop

Patient	25	26	27	28	29	30	31
Stereo nr	976	977	978	979	980	981	982
Side	R	R	R	R	R	L	R
Gender	F	F	M	F	F	M	F
Age	72	69	68	72	77	74	62
Prosthesis	Tricon-M	Tricon-M	Tricon-M	Tricon-M	Tricon-M	Tricon-M	Tricon-M
Fixation	C	C	C	C	UC	UC	UC
Diagnosis	OA	RA	OA	OA	OA	RA	RA
Weight			95	73	70	70	60
HSS pre			43	40	50	13	51
6m			70	83	79	63	64
12m	83		87	87	83	75	70
24m	93		97	86	82	70	81
5 y			96	83			
HKA pre	160	189	166	169	164	180	184
post	182	182	180	190	184	180	187
Alignm. a	9		10	10	7	7	9
b	1		10	14	8	2	10
c	91		85	88	88	88	90
d	88		97	91	97	94	87
Paper	I	I	I, IV	I, IV, V	I, IV	I, IV	I, IV
Migration:							
MTPM 6w			0,57	0,71			0,41
3m			0,49	1,27	0,63	0,64	0,99
6m	0,12	0,62	0,45	0,23	1	0,71	1,16
12m	0,3	0,6	0,55	0,28	0,99	0,71	1,18
24m	0,39	0,67	0,48	0,49	2,21	0,57	1
5 y			0,82	0,92			
Migration 2 y:							
Rotation X	-0,78	-0,6	0,09	0,29	0,87	-0,23	-1,54
Y	-0,02	-0,08	-0,07	-0,35	-1,76	-0,17	-1,01
Z	-0,03	-0,47	-0,17	0,15	-0,13	-0,36	0,03
Migration 5 y:							
Rotation X			0,31	0,8			
Y			0,26	-0,41			
Z			-0,21	-0,23			

REVISION:
cause:
months postop:
what done:
prosthesis:

Patient	32		33	34	35	36	37
Stereo nr	983	351	984	985	986	987	988
Side	L	R	L	R	R	R	L
Gender	M	M	F	F	F	F	F
Age	57	58	74	71	73	62	77
Prosthesis	Tricon-M	Tricon Stem	Tricon-M	Tricon-M	Tricon-M	Tricon-M	Tricon-M
Fixation	C	C	C	UC	C	C	UC
Diagnosis	RA	RA	RA	OA	RA	OA	RA
Weight	58	58	57	83	58	73	58
HSS pre	36	60	23	52	48	44	42
6m	77	75	69	78	85	73	84
12m	85	86	77	87	86	81	93
24m	89	91	82	88	93	91	93
5 y							
HKA pre	168	177	188	196	160	168	187
post	182	183	189	189	175	182	183
Alignm. a	7	9	12	9	11	9	10
b	6	4	2	-1	5	2	0
c	86	90	90	85	87	88	88
d	95	93	93	88	87	87	93
Paper	I	II	I, IV, V, VII	I, IV, VII	I, IV, VII	I, IV	I, IV
Migration:							
MTPM 6w	0,42	0,31	0,29	0,32	0,29	0,51	0,11
3m	0,47	0,32	0,25	0,42	0,69	1,46	0,31
6m	0,45	0,37	0,32	0,67	1,22	1,59	0,52
12m	0,53	0,36	0,87	0,61	2,02	1,67	0,41
24m	0,65	0,47	0,87	1,42	2,11	1,41	0,2
5 y							
Migration 2 y:							
Rotation X		0,13	-0,47	-0,69	0	-1,97	-0,16
Y		0,14	-0,75	-0,74	-0,81	0,62	0,04
Z		0,24	-0,78	-1,63	-1,99	0,13	0,09
Migration 5 y:							
Rotation X							
Y							
Z							

REVISION:
cause:
months postop:
what done:
prosthesis:

Patient	38	39		40	41	42
Stereo nr	989	990	363	991	992	993
Side	R	L	R	L	R	R
Gender	F	F	F	F	M	F
Age	71	79	81	78	75	68
Prosthesis	Tricon-M	Tricon-M	Tricon Stem	Tricon-M	Tricon-M	Tricon-M
Fixation	UC	UC	UC	C	UC	C
Diagnosis	RA	RA	RA	RA	SFOA	SFOA
Weight	56	60	60	67	90	60
HSS pre	42	37	46	39	48	34
6m	81	72	82	82	84	87
12m	91	87	89	84	89	93
24m	91	90	91	84	92	86
5 y						
HKA pre	183	174	175	180	163	167
post	186	185	182	179	184	180
Alignm. a	11	10	11	8	10	6
b	5	7	-1	5	9	3
c	93	91	91	85	89	89
d	90	90	90	92	92	99
Paper	I, IV	I, IV	II	I, IV	I, IV	I, IV, VII
Migration:						
MTPM 6w	0,55	0,44		1,58	1,07	0,28
3m	0,75	0,78	0,76	2,59	1,6	0,19
6m	1,36	0,73	1,11	3,88	2,29	0,37
12m	1,88	1,02	1,18	4	2,32	0,48
24m	1,53	1,42	0,89	3,58	2,18	0,57
5 y						
Migration 2 y:						
Rotation X	0,66	0,54	-1,35	-4,49	0,08	-0,02
Y	0,18	-0,92	-0,32	-2,76	-2,9	-0,09
Z	0,71	-0,64	-0,03	-0,75	-0,17	-0,6
Migration 5 y:						
Rotation X						
Y						
Z						

REVISION:
cause:
months postop:
what done:
prosthesis:

Patient	43		44	45	46	47	
Stereo nr	350	358	352	353	354	355	370
Side	L	R	R	R	R	R	L
Gender	F	F	M	F	F	F	F
Age	72	74	64	41	73	59	62
Prosthesis	Tricon Stem	Tricon-Stem	Tricon Stem	Tricon Stem	Tricon Stem	Tricon Stem	Tricon Stem
Fixation	UC	UC	UC	C	C	UC	UC
Diagnosis	RA	RA	RA	RA	RA	RA	RA
Weight	63	63	88	40	62	82	82
HSS pre	29	35	19	23	35	15	38
6m		45	68	84	78	46	72
12m	53	47	74	93	92	57	75
24m	54	48	94	96	92	65	
5 y							
HKA pre	180	176	200	190	191	178	183
post	182	183	178	187	180	185	189
Alignm. a	4	11	9	7	9	7	11
b	6	2	0	4	0	0	5
c	85	90	84	92	89	90	90
d	92	95	90	92	91	88	90
Paper	II	II	II	II	II	II	II
Migration:							
MTPM 6w	0,92		0,43	0,25	0,45		0,46
3m	1,49	1,01	1,03	0,62	1		0,55
6m	2,19	1,02	1,22	0,43	0,79		0,37
12m	3,74	1,23	2,2	0,39	1,58		0,3
24m	4,35	1,14	2,7	0,47	0,58		
5 y							
Migration 2 y:							
Rotation X	-5,37	-0,25	-1,42	0,31	-0,39		-0,45
Y	-1,3	-0,79	0,08	-0,01	0		0,08
Z	-0,78	0,23	-0,52	-0,13	0,08		0,09
Migration 5 y:							
Rotation X							
Y							
Z							

REVISION:
cause:
months postop:
what done:
prosthesis:

Patient	48	49	50	51	52	53	54
Stereo nr	356	357	359	360	361	362	364
Side	R	R	L	R	L	R	L
Gender	M	F	M	F	F	F	F
Age	68	65	59	53	46	64	74
Prosthesis	Tricon Stem	Tricon Stem	Tricon Stem	Tricon Stem	Tricon Stem	Tricon Stem	Tricon Stem
Fixation	C	C	C	C	C	C	UC
Diagnosis	RA	RA	RA	RA	RA	RA	RA
Weight	80	60	85	70	56	79	57
HSS pre	46	48	29	44	34	29	23
6m	64	89	76	94	88	76	56
12m	85	95	91	93	88	79	70
24m	80	97	84	91	88	81	85
5 y							
HKA pre	167	187	195	182	198	168	188
post	182	182	182	180	183	177	180
Alignm. a	11	8	10	7	10	9	8
b	5	0	8	6	4	5	2
c	88	89	88	87	91	86	88
d	92	93	93	90	86	90	92
Paper	II	II	II	II	II	II	II
Migration:							
MTPM 6w	0,4				0,35	0,32	0,15
3m		0,55	0,39		0,51	0,29	0,25
6m	0,55	0,66	0,98	1,27	0,64	0,38	0,28
12m	0,94	0,79	1,11	1,53	0,86	0,47	0,42
24m	0,69	0,78	1,07	1,19	0,77	0,67	0,54
5 y							
Migration 2 y:							
Rotation X	−0,68	−0,79	−1,51	0,39	−0,67	0,2	0,09
Y	−0,45	0,32	−0,37	1,08	0,21	0,03	−0,42
Z	0,35	−0,08	−0,13	−0,43	0,69	−0,32	0,16
Migration 5 y:							
Rotation X							
Y							
Z							

REVISION:
cause:
months postop:
what done:
prosthesis:

Patient	55	56	57	58	59	60
Stereo nr	365	366	367	368	369	650
Side	L	R	L	L	R	L
Gender	F	F	F	M	F	F
Age	64	65	71	80	72	73
Prosthesis	Tricon Stem	Tricon Stem	Tricon Stem	Tricon Stem	Tricon Stem	Miller-G
Fixation	UC	UC	C	C	C	C
Diagnosis	RA	RA	RA	RA	RA	OA
Weight	61	75	65	70	45	55
HSS pre	40	35	52	37	25	65
6m	85	87	88	79	71	92
12m	86	86	94		84	92
24m						95
5 y						
HKA pre	174	181	202	167	170	164
post	182	182	187	176	178	175
Alignm. a	10	7	9	10	7	9
b	4	1	2	2	2	0
c	87	89	89	86	87	86
d	92	90	92	93	92	82
Paper	II	II	II	II	II	III, VI, VII
Migration:						
MTPM 6w	1,14	0,71	0,17	0,14	0,15	0,3
3m	1,33	0,77	0,21	0,34	0,25	0,63
6m	1,61	1,01	0,32	0,27	0,41	0,5
12m	2,59	1,07	0,43			0,66
24m						0,69
5 y						
Migration 2 y:						
Rotation X	1,52	-0,99	0,02			-0,28
Y	-2,15	0,13	-0,08			-0,48
Z	-0,43	0,03	0,15			-0,17
Migration 5 y:						
Rotation X						
Y						
Z						

REVISION:						
cause:						
months postop:						
what done:						
prosthesis:						
				Died 12 mo postop		

Patient	61		62	63	64	
Stereo nr	652	663	653	654	655	678
Side	R	L	L	R	R	L
Gender	F	F	F	F	F	F
Age	74	74	78	72	70	70
Prosthesis	Miller-G	Miller-G	Miller- G	Miller-G	Miller-G	Miller-G
Fixation	C	C	C	C	C	C
Diagnosis	OA	OA	OA	OA	OA	OA
Weight	74	74	76	76	75	75
HSS pre	44	64	31	17	28	31
6m	66	83	74	86	62	61
12m	83	94	82	86	73	70
24m	94	94	85	90	85	79
5 y						
HKA pre	164	168	160	166	202	163
post	180	182	185	174	190	178
Alignm. a	9	10	12	8	6	9
b	6	0	-2	0	1	2
c	88	90	88	85	94	92
d	79	79	79	81	83	77
Paper	III, VI, VII	III, VI, VII	III	III, VI	III, VI	III
Migration:						
MTPM 6w	0,27	0,47	0,65		0,44	
3m	0,45	0,26	0,69	0,35	0,44	0,18
6m	0,87	0,24	1,23	0,51	0,37	0,17
12m		0,28	1,24	0,73	0,34	0,27
24m	0,91	0,53	1,21	0,89	0,44	0,27
5 y						
Migration 2 y:						
Rotation X	-2,14	-0,42	-1,44	0,38	0,07	-0,25
Y	-0,6	0,2	-0,4	-0,46	-0,07	-0,08
Z	0,25	0,53	0,26	0,46	-0,17	-0,05
Migration 5 y:						
Rotation X						
Y						
Z						

REVISION:
cause:
months postop:
what done:
prosthesis:

Patient	65		66	67	68	69	
Stereo nr	656	677	657	658	659	660	681
Side	R	L	R	L	R	R	L
Gender	F	F	M	F	F	F	F
Age	70	71	70	70	71	75	76
Prosthesis	Miller-G	Miller-G	Miller-G	Miller-G	Miller-G	Miller- G	Miller-G
Fixation	UC	UC	UC	C	C	UC	UC
Diagnosis	OA	OA	OA	SFOA	OA	OA	OA
Weight	72	72	110	70	69	62	62
HSS pre	44	38	49	44	47	51	37
6m	79	79	63	61	59	87	90
12m	88	83	80	69	59	86	92
24m	94	91	83	72	58	95	93
5 y							
HKA pre	172	173	165	180	199	194	203
post	175	179	180	184	184	186	180
Alignm. a	8	9	11	7	11	10	7
b	2	0	0	3	9	0	1
c	87	88	85	90	90	90	90
d	80	79	78	82	79	85	77
Paper	III, VI, VII	III, VII	III	III, VI	III	III, VI, VII	III, VII
Migration:							
MTPM 6w	0,35		0,43	0,12		0,64	0,39
3m	0,26	0,27	0,49	0,3	0,22	1,15	0,4
6m	0,31	0,45	0,33	0,32	0,89	1,67	0,58
12m	0,7	0,32	0,39	0,22	0,84	1	0,75
24m	0,81	0,6	0,5	0,39	0,82	1,19	0,21
5 y							
Migration 2 y:							
Rotation X	–0,5	–0,09	0,5	0,03	–0,48	–0,87	0,24
Y	–0,38	–0,08	0,11	–0,22	0,08	–0,6	0,12
Z	0,51	0,68	0,06	–0,3	0,49	0,66	–0,04
Migration 5 y:							
Rotation X							
Y							
Z							

REVISION:
cause:
months postop:
what done:
prosthesis:

Patient	**70**	**71**	**72**	**73**	**74**	**75**	**76**
Stereo nr	661	662	664	665	667	668	669
Side	R	L	R	R	R	L	L
Gender	M	F	F	F	F	F	F
Age	69	63	73	71	75	68	71
Prosthesis	Miller-G	Miller- G	Miller-G	Miller-G	Miller-G	Miller-G	Miller-G
Fixation	UC	C	UC	C	UC	C	UC
Diagnosis	OA	OA	OA	OA	OA	OA	OA
Weight	75	58	75	60	91	76	87
HSS pre	51	37	50	46	33	34	59
6m	67	69	78	78	83	58	75
12m	83	88	87	86	85		81
24m	84	92	86	97	85		82
5 y							
HKA pre	170	170	191	186	199	192	164
post	180	182	177	177	180	187	178
Alignm. a	9	9	7	6	7	10	10
b	5	1	0	0	5	3	0
c	85	89	89	85	87	88	89
d	75	82	78	83	81	84	81
Paper	III, VI, VII	III	III	III	III	III	III
Migration:							
MTPM 6w	0,54	0,48	1,46	0,22	0,26	0,38	
3m	0,72	0,2	1,38	0,23	0,39	0,36	0,9
6m	0,99	0,75	1,66	0,16	0,5	0,28	
12m	0,98	0,43	1,74	0,66	0,37		1,14
24m	0,91	0,47	1,77	0,5	0,45		0,87
5 y							
Migration 2 y:							
Rotation X	−0,25	−0,73	−1,26	0,02	−0,73		−0,41
Y	0,14	0,39	−0,88	0	−0,22		−0,49
Z	−0,99	−0,22	−0,94	−0,61	−0,18		1,31
Migration 5 y:							
Rotation X							
Y							
Z							

REVISION:							
cause:						Instability	
months postop:						9	
what done:						Tibia revision	
prosthesis:						MG tibial insert	

Patient	77		78	79	80	81	82
Stereo nr	670	683	672	673	674	675	676
Side	R	L	L	R	R	R	R
Gender	M	M	F	F	F	F	F
Age	76	77	76	70	78	75	59
Prosthesis	Miller-G	Miller-G	Miller-G	Miller-G	Miller-G	Miller-G	Miller-G
Fixation	UC	UC	C	C	UC	UC	UC
Diagnosis	OA	OA	SFOA	OA	OA	OA	SFOA
Weight	64	64	54	72	77	68	70
HSS pre	46	44	26	51	61	43	51
6m	85	76	67	86	79	64	73
12m	88	87	69	96	87	82	83
24m	96	87	76	96	89	89	93
5 y							
HKA pre	160	168	200	193	166	171	184
post	180	180	180	181	174	174	185
Alignm. a	7	9	8	8	9	8	7
b	1	0	3	0	0	5	2
c	90	85	90	84	84	84	87
d	76	72	84	75	83	80	76
Paper	III, VII	III, VII	III	III	III	III	III
Migration:							
MTPM 6w		0,3					
3m	1,25	0,4	0,45	0,29	0,21	1,14	0,56
6m	1,35	0,61	0,39	0,41	0,27	1,33	0,37
12m	0,94	0,8	0,27	0,27	0,24	1,5	0,34
24m	0,98	0,59	0,6	0,66	0,38	1,53	0,55
5 y							
Migration 2 y:							
Rotation X	-0,63	-0,79	-0,11	-1,25	0,02	-1,89	-0,14
Y	0,28	-0,15	0,65	0,05	-0,31	0,35	-0,16
Z	0,61	-0,35	0,4	-0,05	0,24	0,68	0,65
Migration 5 y:							
Rotation X							
Y							
Z							

REVISION:
cause:
months postop:
what done:
prosthesis:

Patient	83	84	85	86	87	88	89
Stereo nr	679	680	682	450	451	452	454
Side	R	R	L	R	L	R	L
Gender	F	F	F	F	M	F	M
Age	63	77	84	76	68	66	66
Prosthesis	Miller-G	Miller-G	Miller-G	LCS	LCS	LCS	LCS
Fixation	C	UC	C	C	C	C	C
Diagnosis	OA	Osteonecros	OA	OA	OA	OA	OA
Weight	78	78	65				
HSS pre	40	60	61				
6m	91	61	90				
12m	96	58	92	85	88	84	96
24m	96	82	92				
5 y							
HKA pre	171	186	157				
post	175	193	178	180		180	174
Alignm. a	8	14	11	11	10	4	8
b	4	7	2	9	9	11	4
c	85	87	87	87	86	91	83
d	84	72	83	80	81	77	82
Paper	III	III	III	VI	VI	VI	VI
Migration:							
MTPM 6w			0,25				
3m	0,25	0,59	0,26				
6m	0,29	0,56	0,4				
12m	0,29	0,56	0,43				
24m	0,45	0,64	0,96				
5 y							
Migration 2 y:							
Rotation X	−0,34	−0,87	0,13				
Y	−0,25	−0,18	−0,78				
Z	0,27	0,31	−0,04				
Migration 5 y:							
Rotation X							
Y							
Z							

REVISION:			
cause:		Instability+malalignment	
months postop:		12	
what done:		Femur revision	
prosthesis:		MG rev.	

Patient		90
Stereo nr		455
Side		R
Gender		F
Age		74
Prosthesis		LCS
Fixation		C
Diagnosis		OA
HSS	**pre**	
	6m	
	12m	88
	24m	
	5 y	
HKA	**pre**	
	post	184
Alignm.	**a**	9
	b	11
	c	88
	d	85
Paper		VI
Migration:		
MTPM	**6w**	
	3m	
	6m	
	12m	
	24m	
	5 y	
Migration 2 y:		
Rotation	**X**	
	Y	
	Z	
Migration 5 y:		
Rotation	**X**	
	Y	
	Z	

REVISION:
cause:
months postop:
what done:
prosthesis: