Chapter 10

Micromotion of mobile bearing versus posterior stabilized total knee prostheses

Randomized RSA study of 40 knees followed for 2 years

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Abstract

On theoretical grounds mobile bearing total knees should reduce the micromotion of the tibial component relative to the bone.

We used radiostereometric analysis to measure the three-dimensional micromotion in 42 tibial components during 2 years of follow-up. The patients had been randomized as to whether they would receive a mobile bearing (MB) or posterior stabilized (PS) design. We expected that the MB knee would facilitate dissipation of forces from the prosthesis-bone interface by the motion of the bearing and by load sharing with the soft tissues, leading to less micromotion. In the PS designs, limited free rotation caused by the cam-post articulation might cause additional stress at the bone-prosthesis interface.

We found no significant differences between the MB and PS group at the 2-year follow-up evaluation with respect to Knee Society scores and radiographic results. The PS group had a higher variability in subsidence and anterior-posterior tilting of the component than the MB group.

The low variability of the data in the MB knee prosthesis group suggests that this design is more predictable and forgiving with respect to micromotion of the tibial component.
10.1 Introduction

Wear is one of the critical factors limiting the long-term success of total knee prostheses (TKP). Retrieval studies of tibial inserts have shown that low-conformity and the thickness of the polyethylene insert are associated with increased wear (Bartel et al., 1986; Collier et al., 1991; Wright et al., 1992). One design option which deals with this problem is the mobile bearing (MB) insert, combining high conformity with maximized range of motion (Buechel and Pappas, 1990; Jordan et al., 1997). In contrast to this mobile design the fixed bearing posterior stabilized (PS) TKP, which has a high degree of internal constraint and has during some periods been most frequently used (Aglietti et al., 1999; Stern et al., 1992; Li et al., 1999).

The better wear characteristics of the MB prostheses are based on the relatively large contact area of these designs, thus the contact stresses in the polyethylene are lower compared to the PS designs. Furthermore, if the femoral component and tibial components are malaligned and slight orientation differences are present between the components, the fixed-bearing PS insert will be forced in a certain direction during the range of motion. Contrary, differences between component rotation and the actual knee joint will be corrected by a movable bearing. PS implants will show additional polyethylene wear debris (Puloski et al., 2001), while the MB designs result in less wear of the polyethylene (Jones et al., 1999).

Also torque and shear forces in a MB TKP will be better dissipated from the prosthesis-bone interface by the motion of the bearing and by load sharing with the ligaments and other soft tissue structures (Buechel and Pappas, 1990; Goodfellow and O’Connor, 1992; Callaghan et al., 2001). While in the PS design the limited free rotation caused by the cam-post articulation might cause additional stress at the bone-prosthesis interface.

Early micromotion of implants has been shown to be related to implant survival (Grewal et al., 1992). This micromotion can be very accurately assessed using Roentgen Stereophotogrammetric Analysis (RSA). The value of RSA is -besides high accuracy- its predictive value for future prosthesis loosening (Kärrholm et al., 1994; Ryd et al., 1995; Selvik, 1989). The purpose of our study was to examine the amount of three-dimensional micromotion of the tibial component in a randomized RSA study, comparing a PS fixed with a MB design.
10.2 Patients and methods

42 Consecutive primary cemented TKP (33 patients; 21 female, 12 male) were included. During surgery, a randomization scheme selected the patients for either a PS or MB prosthesis. If a patient was operated bi-laterally the randomization scheme was followed starting with the right leg. In each group two patients received in both knees the same implant design. Patients with a deformity of more then 20 degrees in any plane were excluded from the study. The institution's ethics committee approved the study, and the patients gave informed consent.

In each group 21 cemented implants were included. The PS group consisted of 15 knees of patients with rheumatoid arthritis and 6 knees of patients with osteoarthrosis. The MB group consisted of 14 knees of patients with rheumatoid arthritis and 7 knees of patients with osteoarthrosis. The two groups were similar with regard to age (66 SD 12 years), body mass index (27 SD 5 kilogram per square meter), or stage of osteoarthrosis: Ahlbäck 4, (min-max: 3-4) and Larsen 4, (min-max: 3-5) (Ahlback, 1968; Larsen et al., 1977).

The prosthesis used was the Interax Posterior Stabilized TKP (Stryker-Howmedica, Rutherford, New Jersey, USA) and the Interax Integrated Secure Asymmetric (ISA) TKP, a MB design (Figure 1). The Interax system provides a tibiofemoral articulation surface of the femoral prosthesis, which is spherically shaped in both the frontal and sagittal planes. The base plate of the Interax system is manufactured from cast Vitallium and has a highly polished surface. The articulating surface used in both designs is made of ultra-high-density-molecular-weight polyethylene (UHMWP). The PS design has a fixed articulating surface with a central cam providing medial-lateral stability.

The knee prosthesis with the MB insert permits anterior/posterior sliding and rotation of the inlay on the tibial tray. Two metal pins guide movement of the inlay on the tibial tray, which match the profile of a corresponding groove on the underside of the inlay. Movement of the MB insert is guided by this metal pin, which matches the profile of a corresponding groove on the underside of the inlay. The maximum possible movement of the inlay center is 8.5 mm anterior/posterior and 18 degrees of axial rotation relative to the tibial tray.
During the knee arthroplasty 6 to 8 tantalum markers (Ø = 1 mm: Industrial Techtonics, Ann Arbor, Michigan, USA) were inserted by a specialized instrument into the tibial metaphysis of each patient. The manufacturer inserted two one-millimeter-diameter tantalum markers into a polyethylene wing, which was rigidly connected between the tibial base plate and the stem. A third implant marker was attached at the tip of the stem in a polyethylene notch.

A patellar component made of UHMWP was used in all TKR. The operation was performed through a standard mid-line incision with a medial arthrotomy. Ligament tensioning and balancing of the MB knee group was standardized during surgery using the Monogram Knee Balancer (Stryker-Howmedica, Rutherford, New Jersey, USA). After soft-tissue releases, the prosthesis was implanted according to the manufacturer’s instructions. During surgery the stability of each PS knee was assessed in extension and 90 degrees of flexion. Discrepancies of less than 2 mm were accepted after medial and lateral stressing of the knee.
The cut bone surfaces were mechanically pulse-lavaged with a device manufactured by Zimmer (Warsaw, Indiana, USA) before Palacos bone cement (Schering, Kenilworth, New Jersey, USA) was applied. The tibial base plate and the central stem were cemented in all patients. Post operatively, all of the knees were fitted with a Jones bandage.

The patients were evaluated pre operatively, at 1 week, 3 weeks, 6 weeks, 3 months, 6 months, 1 year, and 2 years post operatively. At each evaluation the clinical status was assessed and radiographs for RSA were exposed. Immediately after the operation, at the 1-year, and at the 2-year follow-up, standard anterior/posterior exposures were taken with the patient standing (Ewald, 1989). Lateral radiographs as well as axial radiographs of the patella were made supine. The femoral tibial angle was determined based on a preoperative and one-year hip-knee-ankle radiograph. 2 patients (one PS and one MB) were lost to follow-up before the final follow-up evaluation of the study due to severe mobility restrictions unrelated to the knee prosthesis (severe rheumatoid arthritis with multiple joint involvement).

The RSA set-up consisted of two synchronized roentgen tubes positioned approximately 1.5 meter above a roentgen cassette (35x43 cm) at a 20° angle to the vertical. Both roentgen tubes simultaneously exposed the roentgen film. A calibration box made of Perspex™ was used to define the three-dimensional (laboratory) coordinate system. For this purpose 26 tantalum 1-mm markers were positioned in the lower plane of the box (fiducial markers). In order to calculate the focus position, 16 1-mm tantalum markers were positioned in the upper plane of the box (control markers).

With a Vidar VXR-12 scanner (Vidar, Lund, Sweden), the radiographs were scanned at 150 dots per inch resolution and eight-bit gray scale resolution. The measurement of marker coordinates in the digitized radiographs, the three-dimensional reconstruction of the marker positions, and the micromotion analysis was done with RSA-CMS (MEDIS, Leiden, The Netherlands). This software package performs the RSA procedure automatically using digitized or digital radiographs (Vrooman et al., 1998; Valstar et al., 2000; Valstar, 2001).

In order to assess the micromotion of the implant with a high accuracy, the bone markers need to be well fixated in the bone. Bone markers were defined unstable.
when they moved more than 0.3 mm with respect to the other bone markers. Unstable markers were excluded automatically from analysis.

The first RSA examination served as the reference baseline. All subsequent evaluations of micromotion were related to the relative position of the prosthesis with respect to the bone at that time of the evaluation. Micromotion of the components was expressed as translation of the center of gravity of the prosthesis makers and rotation of the rigid body defined by the prosthesis markers about this center of gravity. Positive directions for translations along the orthogonal axes were: transverse (medial-lateral), longitudinal (caudal-cranial), and sagittal posterior-anterior. Positive directions for rotations about the coordinate axes were anterior tilt (transverse axis), internal rotation (longitudinal axis), and varus (sagittal axis).

The reproducibility of the RSA measurements was determined by means of double examination of a subgroup of patients (Table 1). Double examination consists of 2 RSA examinations of the same patient exposed within a time interval of about 10 minutes. Because of the short time-interval between these two radiographs, the assumption is made that the implant did not migrate between these two exposures relative to the surrounding bone. By measuring, analyzing, and comparing these two radiographs, the accuracy of the micromotion parameters can be assessed (Ranstam et al., 2000).

Table 1. Precision of the RSA measurements based on double examinations. Presented numbers are the upper limits of the 95%-confidence interval (translations n=22; rotations n=16).

<table>
<thead>
<tr>
<th>Translation (mm)</th>
<th>Rotation (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Trans Long Sag</td>
<td>Trans Long Sag</td>
</tr>
<tr>
<td>0.07 0.06 0.28</td>
<td>0.45 0.42 0.16</td>
</tr>
</tbody>
</table>

10.2.1 Statistics

Mean values and standard deviations were calculated for all variables. For comparison of the median values of the two groups, Mann-Whitney U-test was used. This test was used to compare the micromotion data between the two types of bearing designs at the 2 year follow-up. Levene's test was used to determine whether the
group variances for the micromotion were equal. For all analyses, significance was
determined by a p-value ≤ 0.05.

10.3 Results

10.3.1 Clinical results

There were no significant differences between the two groups in the pre operative
Hospital for Special Surgery Score (HSS), Knee score and Functional score according
to the system of the Knee Society (Table 2), or at the 2-year follow-up evaluation.
The mean pre operative flexion of both groups was 111, (SD 16) degrees (range, 80
to 135 degrees), which increased to 116, (SD 13) degrees (range, 90 to 140 degrees)
at the 2-year follow-up evaluation. The medio-lateral instability scores, based on the
Knee Society are presented in Table 3.

Table 2. HSS, Knee Society Scores and Functional Scores (mean, SD) preoperatively, at the
6-month, 1 year and 2-year follow-up for the MB knee (n=21) and PS group (n=21).

<table>
<thead>
<tr>
<th></th>
<th>HSS score</th>
<th>Knee Society Score</th>
<th>Function score</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>MB</td>
<td>PS</td>
<td>MB</td>
</tr>
<tr>
<td>Pre operative</td>
<td>40 (13)</td>
<td>40 (12)</td>
<td>24 (14)</td>
</tr>
<tr>
<td>6 months</td>
<td>89 (6.2)</td>
<td>76 (23)</td>
<td>88 (14)</td>
</tr>
<tr>
<td>1 year</td>
<td>90 (7.2)</td>
<td>86 (11)</td>
<td>92 (7.2)</td>
</tr>
<tr>
<td>2 years</td>
<td>92 (5.8)</td>
<td>83 (13)</td>
<td>85 (17)</td>
</tr>
</tbody>
</table>

One patient from the MB group (RA treated with corticosteroids and
methotrexate), developed skin necrosis with secondary patella tendon necrosis and
deep infection. After removal of the prosthesis, debridement and treatment with
antibiotics, an arthrodesis was performed.
Table 3. Medio-lateral instability according to the system of the Knee Society. Number of patients scored pre operatively and post operatively.

<table>
<thead>
<tr>
<th>Category</th>
<th>MBK Pre</th>
<th>MBK Post</th>
<th>PS Pre</th>
<th>PS Post</th>
</tr>
</thead>
<tbody>
<tr>
<td>Category I (&lt;5°)</td>
<td>2</td>
<td>16</td>
<td>6</td>
<td>17</td>
</tr>
<tr>
<td>Category II (6°- 9°)</td>
<td>6</td>
<td>2</td>
<td>3</td>
<td>3</td>
</tr>
<tr>
<td>Category III (10°- 14°)</td>
<td>6</td>
<td>1</td>
<td>2</td>
<td>–</td>
</tr>
<tr>
<td>Category IV (&gt;14°)</td>
<td>2</td>
<td>–</td>
<td>5</td>
<td>–</td>
</tr>
</tbody>
</table>

10.3.2 Radiographic results

Routine radiographs of the knee revealed no radiolucent lines of 2 mm or more around the tibial, femoral or patellar component in either of the two groups at the two-year follow-up evaluation. In 2 knees from the MB group and 3 knees from the PS group, non-progressive radiolucent lines of 1 mm were observed in the first two medial zones of the anterior-posterior radiograph according to the Knee Society. In one PS knee, the lateral radiograph revealed a radiolucent line of 1 mm in the posterior zone and in another PS knee a radiolucent line of 1 mm in the anterior zone was discovered.

Table 4. Radiographic results (mean (SD) for the MB knee (n=21) and PS (n=21) tibial components at the 2-year follow-up evaluation.

<table>
<thead>
<tr>
<th>Component</th>
<th>MB (°) Mean (SD)</th>
<th>PS (°) Mean (SD)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tibial angle</td>
<td>87 (2.2)</td>
<td>87 (1.9)</td>
</tr>
<tr>
<td>Tibial slope</td>
<td>86 (3.2)</td>
<td>85 (3.1)</td>
</tr>
<tr>
<td>Femoral flexion (AP)</td>
<td>94 (2.3)</td>
<td>95 (2.6)</td>
</tr>
<tr>
<td>Femoral flexion (LAT)</td>
<td>3.0 (2.6)</td>
<td>2.6 (3.0)</td>
</tr>
</tbody>
</table>

No significant differences in the radiographic results were observed between the MB and PS group. The mean pre operative femoral-tibial angle was 181 (SD 8.4) (range, 152 to 192) degrees. At the postoperative evaluation the mean flexion contracture for the MB group was 1.3, (SD 3.3) degrees (range, 5 to 10)
compared to 0.0, (SD 3.3) (range, -5 to 10) degrees for the PS group. At the two-year follow-up evaluation, the femoral-tibial angle was 178, (SD 3.2) (range, 176 to 186) degrees. 4 MB knee prostheses and 1 PS prosthesis had a varus deformity of more than three degrees at the two-year follow-up evaluation (Table 4).

10.3.3 RSA results

The micromotion data of the tibial components is presented in Table 5 and Table 6. In some cases problems occurred with respect to the marking of the tibia. In 3 cases (2 MB, 1 PS), the bone was either marked with less than three markers, or the markers were positioned so that they were occluded by the component. These cases were excluded from the analysis. In 9 components (6 PS, 3 MB) one of the markers, either at the medial or lateral side of the tibial tray, was occluded. As a result only the translations of these specific tibial components could be calculated (Table 7).

No significant differences in translations and rotations were found between the MB group and the PS group. At the 2-year follow-up evaluation, the PS components had subsided 0.10, (SD 0.18) (range, -0.01 to 0.59) mm; the mobile bearing components had subsided 0.06, (SD 0.09) (range, -0.01 to 0.20) mm. The mean anterior-posterior tilting of the component at two-year follow-up was for MB components -0.09, (SD 0.27) (range, -0.39 to 0.35) degrees and for the PS components -0.06, (SD 0.86) (range, -1.42 to 1.75) degrees. 6 patients had a rotation of more than 0.30° about this axis. Three of these patients also had the highest body mass index (BMI) of the PS group. The PS group showed a significant higher variability in subsidence (Levene's test, p = 0.04) and rotation about the transverse axis (Levene's test, p = 0.05) compared to the MB group. This indicates that some PS components showed a relatively high micromotion (e.g. maximum anterior tilt or posterior tilt) compared to the components in the MB group.

All tibial components with radiolucent lines of one millimeter rotated about the longitudinal axis 0.4 degrees or more. Components without radiolucent lines rotated less.
### Table 5. Mean translations (95% CI, Lower Bound – Upper Bound) of the PS and MB tibial components during 2-year follow-up.

<table>
<thead>
<tr>
<th></th>
<th>Lateral-Medial</th>
<th>Caudal-Cranial</th>
<th>Posterior-Anterior</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>PS</td>
<td>MB</td>
<td>PS</td>
</tr>
<tr>
<td>3 weeks</td>
<td>-0.00 (-0.05 – 0.04)</td>
<td>0.06 (0.01 – 0.103)</td>
<td>0.01 (-0.037 – 0.06)</td>
</tr>
<tr>
<td>6 weeks</td>
<td>0.02 (-0.03 – 0.07)</td>
<td>0.01 (-0.03 – 0.05)</td>
<td>-0.03 (-0.01 – 0.07)</td>
</tr>
<tr>
<td>3 months</td>
<td>-0.02 (-0.08 – 0.04)</td>
<td>0.02 (0.00 – 0.04)</td>
<td>0.02 (-0.03 – 0.07)</td>
</tr>
<tr>
<td>6 months</td>
<td>-0.02 (-0.07 – 0.03)</td>
<td>0.00 (-0.08 – 0.08)</td>
<td>-0.04 (-0.04 – 0.02)</td>
</tr>
<tr>
<td>12 months</td>
<td>-0.05 (-0.11 – 0.00)</td>
<td>-0.01 (-0.07 – 0.05)</td>
<td>0.05 (-0.04 – 0.14)</td>
</tr>
<tr>
<td>24 months</td>
<td>-0.05 (-0.10 – 0.01)</td>
<td>-0.02 (-0.10 – 0.05)</td>
<td>0.09 (-0.02 – 0.20)</td>
</tr>
</tbody>
</table>

### Table 6. Mean rotations (95% CI, Lower Bound – Upper Bound) of the PS and MB tibial components during 2-year follow-up.

<table>
<thead>
<tr>
<th></th>
<th>Anterior tilt</th>
<th>Internal rotation</th>
<th>Varus rotation</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>PS</td>
<td>MB</td>
<td>PS</td>
</tr>
<tr>
<td>3 weeks</td>
<td>-0.03 (-0.40 – 0.33)</td>
<td>0.06 (-0.28 – 0.39)</td>
<td>-0.06 (-0.21 – 0.01)</td>
</tr>
<tr>
<td>6 weeks</td>
<td>0.14 (-0.21 – 0.48)</td>
<td>0.05 (-0.23 – 0.33)</td>
<td>-0.00 (-0.19 – 0.18)</td>
</tr>
<tr>
<td>3 months</td>
<td>0.09 (-0.23 – 0.42)</td>
<td>-0.02 (-0.18 – 0.14)</td>
<td>0.10 (-0.10 – 0.29)</td>
</tr>
<tr>
<td>6 months</td>
<td>0.08 (-0.31 – 0.46)</td>
<td>-0.08 (-0.31 – 0.16)</td>
<td>0.02 (-0.19 – 0.22)</td>
</tr>
<tr>
<td>12 months</td>
<td>-0.16 (-0.49 – 0.75)</td>
<td>-0.07 (-0.29 – 0.15)</td>
<td>-0.02 (-0.27 – 0.24)</td>
</tr>
<tr>
<td>24 months</td>
<td>-0.06 (-0.60 – 0.49)</td>
<td>-0.09 (-0.30 – 0.12)</td>
<td>-0.01 (-0.28 – 0.26)</td>
</tr>
</tbody>
</table>
Table 7. Number of observations that could be used for migration calculation for each follow-up examination.

<table>
<thead>
<tr>
<th></th>
<th>MB Translations</th>
<th>MB Rotations</th>
<th>PS Translations</th>
<th>PS Rotations</th>
</tr>
</thead>
<tbody>
<tr>
<td>3 weeks</td>
<td>19</td>
<td>10</td>
<td>20</td>
<td>11</td>
</tr>
<tr>
<td>6 weeks</td>
<td>19</td>
<td>13</td>
<td>20</td>
<td>12</td>
</tr>
<tr>
<td>3 months</td>
<td>19</td>
<td>16</td>
<td>20</td>
<td>13</td>
</tr>
<tr>
<td>6 months</td>
<td>19</td>
<td>14</td>
<td>20</td>
<td>15</td>
</tr>
<tr>
<td>1 year</td>
<td>19</td>
<td>16</td>
<td>20</td>
<td>14</td>
</tr>
<tr>
<td>2 years</td>
<td>19</td>
<td>16</td>
<td>20</td>
<td>15</td>
</tr>
</tbody>
</table>

10.4 Discussion

Wear causes aseptic loosening of the prosthesis due to the osteolytic activity around the prosthesis induced by wear particles and fluid pressure (Schmalzried et al., 1992; Santavirta et al., 1993; Vis, 1997). Because of the relatively small contact area of fixed bearing designs, the contact stresses in the polyethylene can be high leading to excessive wear. Especially for patients with a high BMI the contact pressure on the posterior part of the tibial tray increases rapidly during flexion, due to a posterior translation of the femoral-tibial contact point (Stiehl et al., 1999) which can explain the observed anterior-posterior tilting of the component in this study (Stukenborg et al., 2002). However, in MB designs wear is also observed between the polyethylene bearing and the tibial tray. The mechanical solutions proposed to restrict bearing motion to a unidirectional motion by rotating platforms or to entrap meniscal bearings in curved tracks (Lewandowski et al., 1997), may induce erratic movements with polyethylene on metal impingement. A prerequisite for a good load distribution in mobile bearing knees is a continuous motion over time of the polyethylene insert during flexion. However, the mobile bearing may be encapsulated by soft-tissue after a period of time or may have erratic movements (Hartford et al., 2001). Consequently,
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the advantage of the mobility of the polyethylene insert – which should prevent excessive wear – is cancelled out, and could induce even more wear.

In cam-post articulation of PS implants an additional source of polyethylene wear debris is present. The stabilizing cam acts as a contact guide in limiting tibial subluxation and ensuring appropriate femoral rollback. Therefore the most polyethylene wear will occur over the posterior surface of the post (Li et al., 1999). Pre operative function and general condition of the patient are important factors determining the post operative function after TKA (Nelissen et al., 1995). The preoperative function score was only slightly higher for the MB group compared to the PS group. Although in the beginning of the follow-up the Knee Society function score was higher for the MB group, at the two-years evaluation no differences were observed.

Nelissen et al. (1998), showed a subsidence rate after two-years of cemented fixed bearing tibial components of the Interax total knee prosthesis of -0.05 mm ± 0.11 mm, which is comparable with the subsidence in the results of the MB prostheses (0.08 mm ± 0.09 mm). In the former study a low-conformity, posterior cruciate retaining TKP was used, allowing more rotation than the Interax PS design. Apparently a low conformity design as well as a higher conformity MB design give less variability in the subsidence data compared to the PS design (Nelissen et al., 1998).

Three clinical factors could explain the variability in the micromotion between the two groups: the bone-cement interface, ligament balancing and alignment of the components.

A good bone-cement interface implies a stable cemented prosthesis. Interface motions stresses the bone/cement, which will induce loosening of the prosthesis. Especially the variability in the tilting of the components was large for the PS knees. In high flexion, the posterior cam of the femoral component slides against the polyethylene post and pushes the tibial component into an anterior tilt. Inducible displacements caused by this cam-post mechanism leads to increased interface motions (Uvehammer and Kärrholm, 2001).

The second clinical factor causing variability in TKP micromotion data is improper ligament balancing. Some knees might be more lax in extension and tight in flexion causing high contact pressures posterior of the tibial tray or at the
cam-post contact point (Uvehammer and Kärholm, 2001). Patients with a flexion contracture stress the posterior part of the tibial tray continuously resulting in a rotation of the component about the transverse axis (i.e. anterior tilt of the tibial tray). However, in this study no significant difference in flexion contracture was noted between groups. Finally, postoperative collateral medio-lateral stability scores were similar comparing the MB and the PS knee prostheses. However, for the MB prostheses, load inequality between the medial and lateral condyles resulting from an adduction moment during gait (Schipplien and Andriacchi, 1991; Yu et al., 1997) or improper balancing of the knee (Vedi et al., 1999) is expected to be less at the tibial component due to the mobility of the bearing. If alignment can be restored to normal anatomy, load sharing gives positive remodeling at the bone level (Wolff’s Law) and at the soft tissue level (Roux’s Law) (Malone et al., 2002). Appropriate tensioning of the soft tissues will not only enhance stress distribution reducing bone-implant interface stresses, but will also provide appropriate knee stability and improved function. There is a low margin of error allowed during surgery. Even small errors could result in increased wear due to high contact stresses and could undermine theoretical benefits of prosthesis designs. The mobile bearing design is more forgiving, by correcting slight rotation differences between the actual anatomy and the TKP, but also correcting for slight rotation differences between the femoral and tibial component.

A third clinical factor affecting micromotion is alignment of the TKP. Residual varus deformity following TKP has been shown to create a substantial load imbalance, resulting in the type of stress eventually leading to tibial component loosening (Schipplien and Andriacchi, 1991; Stiehl et al., 1999; Perillo-Marcone et al., 2000). In a finite element analysis, the contact stresses in a 3 degrees varus knee increased by 45% in comparison to a neutral alignment of the component (Liau et al., 2002). In the MB group, 4 prostheses had a varus deformity of more than 3 degrees. These components might be at risk for high contact stresses in the medial compartment of the bearing resulting in abnormal kinematics and a high polyethylene wear rate. However, no increased micromotion (i.e. rotation of the tibial component about the sagittal axis) was found in these prostheses during the two-year follow-up. Stresses are dissipated away through the movable insert of the MB design.
We found no evidence of periprosthetic osteolysis on the plain radiographs when defining 2 mm of radiolucency as a risk of radiologic loosening (Pilliar et al., 1986; Søballe et al., 1993; Reading et al., 1999). When traditional radiographs are used for assessment, the rate of early loosening is underestimated. The PS components with a partial radiolucency of one millimeter also showed a substantial rotation about the longitudinal axis (i.e. external rotation). Rotations about this axis are hard to discriminate in traditional radiographs. The rotations about the longitudinal axis of the PS components can be explained by the femoral component bouncing against the cam during flexion. This cam prevents internal and external rotation of the tibia during flexion or walking (Lafortune et al., 1992). Puloski et al. (2001) observed surface damage on the tibial posts of PS tibial components caused by cam-post articulation. Damage on the tibial posts will result in long-term failure of the PS knees through excessive wear, delamination, or even lead to fatigue fracture of the tibial component.

The low variability of the data in the MB knee prosthesis group suggests that implantation of a MB design is more predictable and forgiving with respect to micromotion of the tibia component in the bone after two-year follow-up compared to fixed bearing PS designs.
Chapter 10

References


