KNEE JOINT LOADING AND TIBIAL COMPONENT LOOSENING

RSA AND GAIT ANALYSIS IN 45 OSTEOARTHRITIC PATIENTS BEFORE AND AFTER TKA

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We report a prospective study of gait and tibial component migration in 45 patients with osteoarthritis treated by total knee arthroplasty (TKA). Migration was measured over two years using roentgen stereophotogrammetry. We used the previously established threshold of 200 μm migration in the second postoperative year to distinguish two groups: a risk group of 15 patients and a stable group of 28 patients.

We performed gait analysis before operation and at six months and at two years after TKA. On all three occasions we found significant differences between the two groups in the mean sagittal plane moments of the knee joint. The risk group walked with higher peak flexion moments than the stable group. The two groups were not discriminated by any clinical or radiological criteria or other gait characteristics.

The relationship which we have found between gait with increased flexion moments and risk of tibial component loosening warrants further study as regards the aetiology of prosthetic loosening and possible methods of influencing its incidence.


Tibial component loosening may still compromise otherwise good long-term results after total knee arthroplasty (TKA). The cause is not fully understood, but attention has been focused on wear debris particles and on mechanical factors such as initial stability, alignment, tibial bone quality and activity level. Force transmission was regarded as important in the loosening of earlier, constrained prostheses (Wagner and Bourgois 1974; Endoh, Seedhom and Takeda 1978; Goodfellow and O'Connor 1978) and many investigations have studied different designs with varying ability to withstand load. Less attention has been paid to the loads themselves; there is only limited knowledge of the stresses and strains on artificial joints.

We aimed to investigate whether differences in load on the knee during walking would influence tibial component fixation and subsequent loosening. To avoid waiting for long-term follow-up, we used roentgen stereophotogrammetric analysis (RSA) in a prospective series of 45 patients having TKA. RSA can assess prosthetic stability with great accuracy (Ryd 1986; Selvik 1989) and early rapid migration of components has been shown to predict late loosening (Freeman and Plante-Bordeneuve 1994; Kärholm et al 1994; Ryd et al 1995).

We used the reported threshold of migration of 200 μm by two years after surgery to predict a greater risk of future loosening (Ryd et al 1995). We obtained a proportional estimate of the joint load during walking from gait analysis by calculation of knee moments in both sagittal and frontal planes.

PATIENTS AND METHODS

From 1989 to 1992 we selected 45 patients with severe osteoarthritis treated by TKA at the Central Hospital of Västerås, Sweden. The criteria were osteoarthritis in stages 3 to 5 of Ahlbäck's classification (1968) with no previous replacement, osteotomy or fracture of the investigated knee, and age between 60 and 75 years. We excluded patients with any other impairment of the locomotor system, other than varying levels of osteoarthritis of the opposite knee. These had to be accepted because of the low incidence of severe, monoarticular changes.

Patients who fulfilled these criteria were recruited consecutively; all gave informed consent as advised by the
Ethics Committee.

Three types of uncemented tricompartmental prosthesis were used according to a random distribution by sealed envelopes in groups of three, so that there would be 15 patients in each group. The prostheses were the Tricon-M and Tricon stem (both Smith and Nephew, Memphis, Tennessee) and the PCA resurfacing implant (Howmedica, Rutherford, New Jersey). Patellar components were used in all cases. The operations were done in a standard manner by two experienced surgeons who were not involved in the analysis of the results. The three groups showed no significant differences in age, diagnosis, weight, stage of arthritis or alignment.

During the two-year follow-up, two patients died from unrelated causes and their data were excluded. Seven patients had TKA on the opposite knee, four in the PCA group, two in the Tricon-M group and one in the Tricon stem group, all more than six months after the first operation. As we have previously reported (Hilding, Yuan and Ryd 1995), all 43 patients available at the two-year follow-up had good or excellent clinical results, with a mean HSS score of 88 (71 to 94). None showed complete radiolucent zones of 2 mm or more.

Gait analysis. Gait analysis was performed using the Vifor system (Lanshammar 1988). This consists of two synchronised video cameras, a force-plate (Kistler type 9261A until March 1992, and then the improved Kistler type 9284), photo cells and a computer. The force plate is sampled synchronously by the two video cameras and the sagittal and frontal projections of the ground reaction force vector are superimposed on to the two (split-image) video images in real time. The video films show not only the recorded markers, but also the patient, and provide qualitative data (Fig. 1).

To obtain quantitative data, the position of the ground reaction force vector and the centre of the ankle, knee and hip joints during the entire stance phase were digitised offline using a tracker ball and a cross-hair cursor overlaid on to the video image. Joint moments, defined as counteracting the muscular moments, were calculated by multiplication of the joint centre to force vector distance and the ground reaction force amplitude by the program Vifdig (Lanshammar 1991). Gravitational and inertial forces were omitted in the calculations, because of their small contribution to ankle and knee moments at slow walking speeds (Lanshammar 1988).

Gait analysis was undertaken preoperatively and at six months and two years postoperatively. The patients were instructed to walk in their usual manner, at their usual speed, and were wearing their customary, comfortable shoes. Several trials were performed before recording to familiarise the patient and to reach a stable speed of gait. To reduce the risk of interference with normal gait, no instruments were mounted directly on the patients. The centre of the hip joint was marked with tape (greater trochanter laterally; 2 cm distal to the midpoint of the inguinal ligament frontally). The knee and ankle joints were not marked but kept free from clothing.

Recordings were made from a covered force-plate on a 10 m walkway. Because of the camera positions, only one leg was visible on the split video image from both anterior and lateral views simultaneously. This leg was recorded for at least three steps and the other leg for at least six steps, from both walking directions. At least nine steps from each
leg of each patient were recorded and analysed at each measurement session.

The system is two-dimensional and thus cannot take rotation into account. To make the position of the centre of the knee joint independent of internal and external rotation and skin movement, we used the midpoint between the contour lines for digitisation, instead of surface markers on the skin. Frames at 40 ms intervals were utilised from recordings made with 20 ms intervals (50 Hz). One stance phase included from 16 to 24 frames, depending on its duration. The co-ordinate system was laboratory fixed and the sagittal plane of the patient was supposed to coincide with the plane of progression.

In a previous error analysis (Lanshammar 1988), the accuracy (maximum error) of the Vifor system was calculated to be 9 Nm for moments at the knee during midstance in the frontal plane and 11 Nm in the sagittal plane (Lanshammar 1988). The variability of the results was estimated by calculating the standard deviation of two repeat readings on all patients (randomly sampled from the available three to six registered steps), according to:

$$S = \sqrt{\frac{1}{2n} \sum (x_i - y_i)^2}$$

where $x_i$ and $y_i$ represent two different readings on the ith individual. The $s_s$ of the measurements for moments, peak and mean, frontal and sagittal, were all between 1.8 and 5.0 Nm (Table I).

Digitisation was performed by one author, and interobserver reliability was estimated by calculating the $s_o$ of the measurements from two series (with the above formula), by two of the authors. The $s_o$ for knee joint moments was between 1.3 and 4.7 Nm (Table II), generally less for other variables except for those dependent on the position of the hip joint.

The knee joint moments have been analysed previously by the Vifor system in 76 healthy subjects (40 men, 36 women), aged 20 to 60 years (Lanshammar, Turan and Blomgren 1992). In the sagittal plane the mean value was 8.3 Nm extension (sd 15.7) and the peak extension moment was 37.2 Nm (sd 20.2). In the frontal plane the mean adduction moment was 11.8 Nm (sd 7.1) and the peak adduction moment was 27.2 Nm (sd 10.9).

Roentgen stereophotogrammetry. The methods of RSA have been described previously by Selvik (1974,1989) and Ryd (1986). At the TKA operation, four to six 0.8 mm tantalum balls were placed in each of the tibial metaphysis and the plastic part of the tibial component. Subsequent radiographs were taken inside a calibration cage simultaneously in pairs with a 90° angle between them. The pairs of two-dimensional radiographs were later converted to a three-dimensional co-ordinate system by computer, and changes in position of the two sets of markers were calculated by rigid-body kinematics.

Migration of the tibial component can be expressed as angular and translatory movements with six degrees of freedom, or as migration of individual markers. A standard RSA method is to express the motion as the vector-sum of motions of the marker that moved the most, recording magnitude but not direction to give maximum total point motion (MTPM). The RSA results for this group of patients have been published (Hilding et al 1995), and we found no statistically significant differences between the three designs of prosthesis for mean migration (MTPM) at two years; they were 1.3, 1.4 and 1.5 mm.

Statistics. The gait variables from all recorded steps at each measurement session, usually three to six from each leg of each patient, were pooled as averages and analysed statistically by using the mean values calculated for the whole stance phase, the single maximum and the minimum (peak) values. Two-way analysis of variance (repeated-measurements ANOVA) was used to test group differences with repeated (dependent) observations. Chi-squared analysis was used to compare groups with categorical variables, and $p$ values of $<0.05$ were considered significant.

RESULTS

The PCA, the Tricon stem and the Tricon-M groups showed no significant differences in walking speed ($p = 0.6$), stance phase duration ($p = 0.4$), ground reaction force ($p = 0.8$), angular excursions ($p = 0.5$ to 1.0), moments ($p = 0.1$ to 0.9) and moment arm ($p = 0.6$ to 0.9), both in the frontal and the sagittal planes, at any time. Because there were no significant differences between the three different tibial components in these gait parameters,
the previously reported RSA migration (Hilding et al 1995) or in demographical, clinical or radiological parameters, the groups were collapsed for all further analyses.

In most patients with an initial varus deformity, TKA significantly reduced the frontal plane varus angle, the moment arm and the adduction moment. In the five knees with preoperative valgus, there was a corresponding reduction of abduction moment. There were no further changes between six months and two years. The sagittal plane moments (peak flexion moment, peak extension moment, and stance-phase mean moment) did not change significantly after the operation; neither did the moment arms in the sagittal plane.

Preoperatively, the patients walked without fully extending their knees (mean angular extension deficit 5.4° (SEM 1.1)). This extension deficit was reduced after TKA (p = 0.016), to a mean of 3.2° (SEM 0.7) at six months and 3.6° (SEM 0.5) at two years. As shown in Table III walking speed increased from a mean 0.87 m/s preoperatively to 0.97 m/s at six months after the operation (p < 0.001). An additional increase to 1.01 m/s was registered two years after the operation (p = 0.04). The mean duration of the stance phase of the investigated leg was 0.85 s preoperatively, reducing to 0.81 s at six months (p = 0.003) and 0.79 s at two years (Table III), with a corresponding change in the opposite leg. The ground reaction force, both total and divided into components, was increased by the operation with similar levels at six months and at two years.

We used the RSA migration results (Hilding et al 1995) to predict the risk of future loosening, using a previously established threshold of 200 μm MTPM migration during the second postoperative year (Ryd et al 1995). This allowed us to form two groups with different prognoses, one containing 15 patients with ‘unstable’ prostheses, in which a risk of loosening was predicted, and one of 28 patients with stable fixation. As previously reported (Hilding et al 1995), age, diagnosis, stage of arthritis, alignment, weight, clinical score, and the distribution of prosthetic types were not significantly different between the two groups. In addition, there were no differences between the groups at any time in walking speed (p = 0.6; Table III), stance phase duration (p = 0.7; Table III), peak angular excursions (p = 0.2 to 0.5), or total ground reaction force level (p = 0.9).

The frontal plane moments (peak adduction, mean, and peak abduction) showed no statistically significant differences in mean values between the stable and unstable groups on any of the three occasions (p = 0.2 to 1.0; Table IV). Adduction moments were reduced after the operation in both stability groups, but, because of the five valgus knees with predominantly abduction moments, we then considered the absolute values of reduction of moment; using these the unstable group showed a significantly greater postoperative reduction of frontal plane moments (peak moment, p = 0.025; mean moment, p = 0.014).

In the sagittal plane, the unstable group showed a larger peak flexion moment, both before the TKA, at a mean of 26.0 Nm compared with 17.0 Nm in the stable group (p = 0.045), and two years after the operation, at a mean of 22.2 Nm compared with 15.1 Nm (p = 0.007) (Table IV; Fig. 2). At six months there was also a difference in mean values, 24.5 compared with 14.3 but, due to a larger spread, this did not reach statistical significance (p = 0.065) (Table IV; Fig. 2). The statistical significance of the difference in RSA stability was stronger if all three measurement occasions were analysed together as repeated measurements (p = 0.011).

The peak extension moments were not significantly different between the two groups at any time (Table IV). As an indication of the predominant moment pattern in the sagittal plane, we examined the mean moment. Here again, significant differences were found between the two groups (repeated measurements, p = 0.005; Table IV and Fig. 3). Preoperatively, the mean moment was 5.5 Nm flexion in the unstable group compared with a mean moment of 4.8 Nm extension in the stable group (p = 0.034). At six months the mean moment was reduced to 0.5 Nm flexion in the unstable group, while the mean moment in the stable group was 6.2 Nm extension (p = 0.016), with similar levels persisting two years after TKA; the mean moment was 1.7 Nm flexion and 6.4 Nm extension in the two groups respectively (p = 0.002) (Table IV; Fig. 3). The moment arms showed no significant differences between the two groups (p = 0.3 to 0.9). The moments were also normalised.

**Table III.** Walking speed and stance phase duration (involved leg) for all patients (n = 43) and in the two groups with different RSA fixation, unstable (n = 15) or stable (n = 28), in the second postoperative year (mean value and standard error).

<table>
<thead>
<tr>
<th></th>
<th>Preoperative</th>
<th>6 months</th>
<th>2 years</th>
<th>Mean change Pre- to postop</th>
<th>6 mths to 2 yrs</th>
</tr>
</thead>
<tbody>
<tr>
<td>Walking speed (m/s)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>All</td>
<td>0.87 ± 0.03</td>
<td>0.97 ± 0.03</td>
<td>1.01 ± 0.02</td>
<td>0.09 ± 0.03</td>
<td>0.05 ± 0.02</td>
</tr>
<tr>
<td>Unstable</td>
<td>0.90 ± 0.03</td>
<td>0.96 ± 0.05</td>
<td>1.03 ± 0.05</td>
<td>0.06 ± 0.04</td>
<td>0.07 ± 0.05</td>
</tr>
<tr>
<td>Stable</td>
<td>0.85 ± 0.04</td>
<td>0.97 ± 0.03</td>
<td>1.00 ± 0.03</td>
<td>0.13 ± 0.03</td>
<td>0.04 ± 0.02</td>
</tr>
<tr>
<td>Stance phase duration (s)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>All</td>
<td>0.85 ± 0.02</td>
<td>0.81 ± 0.02</td>
<td>0.79 ± 0.01</td>
<td>-0.04 ± 0.02</td>
<td>-0.02 ± 0.02</td>
</tr>
<tr>
<td>Unstable</td>
<td>0.82 ± 0.02</td>
<td>0.82 ± 0.02</td>
<td>0.79 ± 0.02</td>
<td>0.01 ± 0.03</td>
<td>-0.05 ± 0.04</td>
</tr>
<tr>
<td>Stable</td>
<td>0.87 ± 0.03</td>
<td>0.80 ± 0.02</td>
<td>0.79 ± 0.02</td>
<td>-0.07 ± 0.02</td>
<td>-0.01 ± 0.02</td>
</tr>
</tbody>
</table>
to body weight and height to enable comparison with other results in the literature. This gave no new information; the same variables showed statistical significance.

**DISCUSSION**

The external moment, the force tending to bend a joint around an axis, can be measured by gait analysis. This moment has normally to be counteracted by muscular force to maintain balance and thus represents a proportional estimate of the total joint load (Johnson, Leitl and Waugh 1980; Harrington 1983; Andriacchi, Stanwyck and Galante 1986). A detailed load analysis, including load distribution, requires more calculations, if not invasive techniques (Harrington 1983). An increased adduction moment in the frontal plane has been associated with worse clinical results in the treatment of osteoarthritis of the knee by tibial osteotomy (Prodromos, Andriacchi and Galante 1985; Wang et al 1990) and with the recurrence of varus deformity (Johnson et al 1980; Prodromos et al 1985; Catani et al 1992). The consequences of different moment patterns before and after TKA have not been established, but have been taken into consideration regarding the longevity of prostheses (Andriacchi, Galante and Fermier 1982; Strick-
The use of RSA or other precise radiological methods allows a prediction to be made of the risk of tibial component loosening after only two years of follow-up (Freeman and Plante-Bordeneuve 1994; Ryd et al 1995). To date, all components monitored by RSA and later revised for aseptic loosening have been shown to have had rapid early migration, but not all components with rapid early migration have become loose. It is not yet established, therefore, whether all or only a certain proportion of the components in the RSA unstable group will loosen. By combining the risk prediction from RSA data and the measurements of moments from gait analysis, we have shown for the first time a correlation between knee joint loading before and after TKA and future tibial component loosening. We have shown that asymptomatic patients with tibial components in the unstable RSA group are subjected to larger peak flexion moments as well as significantly different mean sagittal moments. The fact that these differences are present not only after TKA but also before the operation, suggests that they cannot be an effect of the surgery. Rather, we interpret these data to indicate that certain gait patterns promote tibial component loosening.

It seems reasonable to assume that the biphasic flexion-extension moment gives rise to corresponding cyclic eccentric loading of the tibial component. If these forces exceed the fixation strength, they will cause a cyclic, rocking motion, as previously shown by Ryd and Toksvig-Larsen (1993). Such early micromotion of the tibial component may be an important factor in the generation of component loosening. This hypothesis is further supported by our findings in a previous study (Hilding et al 1995) in which the unstable group showed larger relative motions between implant and bone when external rotatory forces were applied. Another aspect of this argument is that the fixation strength attained at operation may be insufficient. Theoretically, therefore, an implant may already be loose at the end of an unsuccessful operation.

Other factors which influence moments, such as gait speed, stance-phase duration, weight, components of ground reaction force or even moment-arm length, were found not to differ between the groups, and the significantly different moment levels therefore could not be explained as being secondary to these factors. Another possibility would be to consider these parameters to be less reliable than moments in joint load assessments.

Angular excursions during gait may be expected to influence directly the flexion-extension moments. We could not verify a close relationship because both stable and unstable groups showed similar values of peak flexion-extension angles during stance (curve shapes were not compared). One explanation may be that the level of flexion moment in the knee joint is affected by the muscular activity around the hip, and it is not necessarily accompanied by excessive knee flexion (Winter 1980, 1984; Simon et al 1983; Strickland et al 1983).

The variability of moments may be a matter of concern. Svensson and Weidenhielm (1993) recommended caution in interpreting sagittal plane moments, because of high variability in an examination of ten normal subjects. Kadaba et al (1989) investigated 40 normal subjects and found less variation in sagittal plane moments than in frontal plane moments, and proposed single gait evaluations as reasonable. Winter (1980, 1984) found high variability in both knee and hip joint sagittal moments. Ankle moments were more stable, but when the support moment of all three joints were considered together, the variability was remarkably reduced. To what extent variability depends on methodological shortcomings or biological step-by-step variation is difficult to evaluate. The much smaller variability of the support moment indicates a biological variation to a substantial extent (Winter 1980).

One factor which directly influences knee joint moments and moment arms is the location of the knee joint centre. In kinetic analyses, the femorotibial contact point is the preferred site but the exact localisation is subject to error because it is not static but moves posteriorly during flexion. The digitisation of the joint contours instead of skin markers may reduce this error; skin movement is a well-recognised problem. It is possible that our method generates a location that is too much posterior because of the thicker soft-tissue envelope in the popliteal fossa. If this is so, the correct mean stance phase moment would include more flexion.

Different techniques are used in different gait laboratories. Three-dimensional methods, taking inertial and gravitational components into account, have their advocates but the cost and effort have limited their clinical use. Our method, which is comparatively quick and simple for the patients, was probably a prerequisite to obtain data from all 43 patients on three separate occasions.

Normal sagittal plane moment shows a biphasic flexion-extension curve during stance (Andriacchi et al 1982; Simon et al 1983; Kadaba et al 1989; Brugioni, Andriacchi and Galante 1990; Ramakrishnan et al 1994) but there is considerable inconsistency (Winter 1980). In patients after TKA, abnormal patterns are often found, which are predominantly in either flexion or extension (Andriacchi et al 1982; Simon et al 1983; Strickland et al 1983; Brugioni et al 1990; Ramakrishnan et al 1994). Adduction moments seem to be more clearly influenced by operative changes of alignment, but individual variations in patterns tend to remain (Jefferson and Whittle 1989; Whittle and Jefferson 1989; Wang et al 1990; Weidenhielm 1992; Ramakrishnan et al 1994).

Some researchers have investigated moment levels in preoperative, arthritic, patients. Peak flexion moment was found to be lower in a group of 15 patients who eventually required TKA than in 12 patients who had unicompartmental prostheses; the 28 normal control subjects had the highest flexion moments (Mikosz et al 1994). By contrast Schip-
plein and Andriacchi (1991) found higher magnitudes of both peak flexion moments and adduction moments in a group of 19 arthritic patients who later had high tibial osteotomies, compared with 15 normal subjects. The increased sagittal plane moment was described as a compensatory, stabilising mechanism; it used both flexor and extensor muscles simultaneously to protect against the high adduction moment which tended to open the joint laterally.

The interpretation of causes and compensatory effects in different moment patterns is difficult because there are complex interrelations and conflicting aims. A painful knee may benefit from a reduction of moments, and adaptive mechanisms that decrease these moments can be readily observed in such patients. These include reduced walking speed and stride length (Andriacchi, Ogle and Galante 1977; Andriacchi et al 1982; Jefferson and Whittle 1989; Catani et al 1992; Mikosz et al 1994) and leaning of the trunk over the weight-bearing leg to reduce the moment arm between the centre of gravity and the joint axis (Johnson et al 1980; Harrington 1983; Andriacchi et al 1986). In the frontal plane, external rotation at the ankle reduces the adduction moment by shortening the moment arm (Wang et al 1990). It can be assumed that the capacity of an individual to use these mechanisms determines the amount of knee joint unloading, but the cost may be a reduced walking capacity.

In our study the sagittal plane patterns seemed to be present before the operation and to remain unaffected by TKA, suggesting that some aspects of gait are not influenced by standard arthroplasty. Patients with high magnitudes of flexion moments should perhaps have different operations, or if possible, receive special physiotherapy, aiming at adaptive mechanisms. The possible influence of different types of footwear may be worth attention.

**Conclusions.** We have shown that the association between RSA stability and differences in flexion moments existed before any operation. This suggests that individual gait characteristics may contribute to the loosening of tibial components. Individual differences in joint loading during walking deserve further study, as regards the aetiology of prosthetic loosening and possible methods of reducing its incidence.

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