Hip and knee joint rotations differ between patients with medial and lateral knee osteoarthritis.

Gait analysis of 30 patients and 15 controls.

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Summary

The motions and moments in the hip and knee in female patients on waiting list for knee prosthesis surgery with medial (n=15) or lateral (n=15) osteoarthritis (OA) were compared with a control group (n=15). We hypothesized that not only the kinematics and kinetics of the knee but also of the hip would differ between patients with medial and lateral OA of the knee.

Results: At midstance cases with lateral OA showed slightly (2°) more maximal (peak) adduction (p=0.015) of the hip joint and cases with medial OA 7° more abduction (p<0.001) than did controls. In patients with lateral OA the femur was positioned in about 7° more external rotation (max value, p=0.001), but did not differ between medial OA and controls (p ≥ 0.8). There were a tendency to higher internal rotation moment in lateral OA compared to controls (p=0.021). The maximum values of the internal knee abduction moments were 52 % higher in medial OA (p=0.005) and 63 % less in lateral OA (p<0.001) compared to controls. Cases with medial OA had 9° more, whereas those with lateral OA had 6° less external tibial rotation than controls (medial vs. lateral OA: p=0.001).

Conclusions: We found an association between presence of lateral OA of the knee and the biomechanics of the hip joint. It remains to be evaluated if the changed biomechanics of the hip joint is a reason for development of lateral OA or an observation, which is a result of this disease.

Key words: Knee, Hip, Osteoarthritis, Biomechanics, Gait analysis
Introduction

Anatomical and biomechanical factors are important for the development or progression of knee osteoarthritis (OA). According to some studies (3, 23) there is a positive correlation between the knee adduction moment and presence of medial knee osteoarthritis as well as progression of the disease (15). A reduced adduction moment also seems to be important for favorable outcome after high tibial osteotomy (17).

Previous studies have to a certain extent shown a relation between the degree of knee deformity and the forces acting at the knee. Weidenhielm et al (27) found a weak correlation between the peak adduction moment and the Hip-Knee-Ankle (HKA)-angle (r=0.32, p<0.05) in patients with medial OA, clinically stable joint and comparatively mild deformity (Ahlbäck Grades 1 to 3). They also observed a moderately high correlation between the adduction moment and HKA-angle in Mid stance (r=0.46, p<0.001). These authors reported that one fifth of the medio-lateral knee load in Mid stance could be explained by the varus deformity of these knees with osteoarthritis. Hurwitz et al (12) found that the radiographic measures of OA severity in the medial compartment were predictive of peak adduction moments (r=0.43 to 0.48, p<0.001). The authors emphasized the need for a dynamic evaluation of the knee joint loading. They observed that the mechanical axis accounted for 50% of the peak adduction moments; other factors only increased the ability to predict the peak knee adduction moments by 10-18%.

In early osteoarthritis varus alignment has been found to imply a 4-fold increased risk of progression during the subsequent 18 months (4, 25). The corresponding risk increase of valgus alignment in lateral OA has been estimated at 2 to 5 times (25). In knees with moderate osteoarthritis (Kellgren and Lawrence grade 3), a 10-fold increased risk of progression has been postulated regardless presence of varus or valgus deformity (4).

Development of knee osteoarthritis with varus or valgus malalignment of the joint can be expected to involve changes or alterations of motions not only in the knee but also in the hip joint. There are also speculations that medial and lateral osteoarthritis have different origin (5, 6). One possible etiology or reason for a more rapid progress of either type of knee OA could be alterations of the pelvic and hip anatomy. Such variations could include femoral offset, presence of acetabular dysplasia and pelvic width. To study this field further there is a need for basic research regarding any differences in gait patterns between patients with medial and lateral OA. Previous gait analyses in patients with knee OA have mainly focused on the knee, however, recently a study had shown that patients with more severe medial knee OA had lower hip adduction moments compared with their matched control subjects (16). To the authors’ knowledge, no study has been specifically directed towards the kinematics of the knee in patients with lateral OA.

In a previous study (27) we observed that the pelvic and hip anatomy differed patients with medial and lateral OA of the knee. In this study we aimed to map out if these differences had any effect on the kinematics and kinetics of the hip and the knee. In addition a control group without any known disease with influence on the musculoskeletal system was studied. Because of female dominance among patients with lateral OA, we restricted our study to only embrace female gender. Observations of shorter femoral neck and head-shaft distance on radiographs (27) could also indicate increased external rotation of the hip compared to normals, with less hip abduction and decreased internal abduction moment. In patients with medial OA, who had increased femoral offset compared to controls, we did not expect to find any corresponding changes. We hypothesized that not only the kinematics and kinetics of the
knee but also of the hip would differ between patients with medial and lateral OA of the knee.

**Material**

(Table 1). 15 women with lateral (unilateral/bilateral: 9/6), and 15 women with medial (unilateral/bilateral: 4/11; Fisher’s Exact Test: p=0.14) osteoarthritis of the knee were identified on the waiting list for knee prosthesis surgery. The Ahlbäck grading differed between 1-5 or 1-4 within each group (Table 1, p=0.09). Patients with any history of inflammatory arthritis or multiple joint disease, previous major knee surgery (e.g. osteotomy of the knee, fracture of the tibia or femur, knee prosthesis) or hip surgery (hip prosthesis) were excluded. Radiographic examinations of the pelvis or standing HKA radiographs were not available. Fifteen healthy women without any history of knee/hip pain or knee/hip trauma acted as controls (Table 1). These controls did not undergo any radiographic examination. They were either relatives or friends to the authors to obtain a firm confirmation of absence of any previous joint symptoms.

There were no differences in age (p=0.5), height (p=0.3) or weight (p=0.1) between the groups, but the BMI tended to be somewhat higher in the group with medial OA (p=0.05, Table 1).

**Methods**

Six infrared cameras recording at 240 Hz (MCU 240 Qualisys ®Medical AB, Göteborg, Sweden) were used. The camera system was calibrated to a measurable volume of 9.2 m³ (2 m x 2 m x 2.3 m). Twenty-one retro reflective spherical markers were attached to the skin over bony landmarks (acromion, 12th thoracic vertebra, 2 markers on sacrum, anterior superior iliac spine, greater trochanter, lateral knee joint line, proximal to the superior border of the patella, tibial tubercle, heel, lateral malleolus and between the second and third metatarsals). Markers were placed bilaterally. The position of each marker was detected of the system inside the limit of 0.2 mm. This marker set was a modification of Helen Hayes marker set (8). It has been used previously by our group (19-21). The positions of the markers were partly chosen to limit the effect of subcutaneous fat thickness.

The room coordinate system was defined with one axis pointing in the direction of walking and parallel to the floor. Its direction was thus based on the first and the last “frame” (i.e. the first position in the gait cycle, to the last one). The next axis was perpendicular to the first one and the floor.

The relations between the marker positions on the skin surface and the joint centers were based up on data published by Vaughan et. al. (26). Thus, the joint centers of interest were mathematically reconstructed based on marker positions. The obtained laboratory or technical coordinate system was transferred to a clinical coordinate system, which was aligned to palpable or indirectly identified skeletal landmarks within each body segment, i.e. hip, thigh, shank and foot (7).

This implied that each segment had its individual coordinate system. The marker set attached to the pelvis (sacrum and 2 markers on each of the two anterior superior iliac spines) was used to calculate the hip joint centers, as well as to align the coordinate system to the pelvis. The markers on the shank were used for detection of the knee joint as well as the ankle joints. The remaining body segments were defined using the computed joint centers as landmarks and as follows; the foot segment was identified using the heel and toe markers and the centre of the ankle joint. The anterior-posterior axis of the foot segment was pointing from the heel marker towards the toe marker. The shank segment was defined by the knee and ankle joint centers (determined by the marker on the tibial tuberosity and on the lateral knee joint line, ankle joint and the marker placed on the tibial tuberosity) with the
longitudinal axis pointing (positive direction) from the ankle to knee joint.

The center of the knee joint, the hip joint and the marker placed on the superior border of patella, determined the thigh segment. The thigh segment had its longitudinal axis pointing from the knee joint centre to the hip joint centre (positive direction). The anterior-posterior axis was perpendicular to the longitudinal one and passed the suprapatellar marker (positive anterior direction). The transverse axis was perpendicular to these axes with positive medial direction. Corrections are made for marker offset, i.e. the distance from the centers of the markers to the skin surface. Zero degree position (start position) is defined when the patient is standing upright. In this position the coordinate systems are fixed to the different segments before any motions are initiated. Those are presented as relative motions of the distal segment using the proximal one as fixed reference. The description of moments is done accordingly.

Two force plates (Kistler 9281C, Kistler Instruments AG, Winterthur, Switzerland) were used to record ground reaction forces during level walking. Eight piezoelectric sensors in each force plate recorded forces 3-dimensionally.

The ground reaction forces were calculated from the input of the eight channels of each force plate. The standard equations provided from Kistler™ were used to compute force vectors, the moment about the normal axis of the force plate surface, and the location of the ground reaction force vector. The force vectors were then transferred from the internal coordinate system of each force plate into the global coordinate system for calculations of joint moments.

The joint moments were calculated using inverse dynamics approach. The mass of each segment is calculated as percentage of body weight. The moments over each joint were expressed as applied to the distal segment of each joint. The data from the six cameras and the forces were recorded synchronically.

Recordings of motion were achieved with the software, QtracC® version 2.51 (Qualisys Medical AB, Göteborg, Sweden). Reconstruction from 2-dimensional into three-dimensional data was made with QtracV® version 2.60 (Qualisys Medical AB, Göteborg, Sweden). QGait 2.0® (Qualisys Medical AB, Göteborg, Sweden) was used for calculations of rotations in relation to the three cardinal axes.

The patients and the healthy controls were asked to walk without targeting on the force plates at self-selected speed. Several trials preceded the actual measurements to define a start line, which facilitated stepping on both force plates according to the step length of the individual patient. Patients had the opportunity to test the walkway several times before recording was done. When patients felt that they were familiar with the situation and were able to perform the test without targeting the force plates, two measurements were completed. The most representative measurement was chosen for further analysis. This selection was made visually by comparison between the walk tests prior to recording and the two recorded tests. In this study we evaluated flexion/extension, adduction/abduction and external/internal rotation of the hip joint (thigh vs. pelvis) and knee joint (shank vs. thigh). The 3 dimensional angles were calculated according to the Euler-angle method described by Kadaba et. al.(13) and Davis et. al. (8). The result of the angle calculations was expressed as motion of the distal segment relative a fixed proximal segment (shank vs. fixed thigh and thigh vs. fixed pelvis). The transverse axis was directed medially, the longitudinal proximally and the sagittal axis anteriorly.

Internal moments are presented. Flexion, internal rotation and adduction have positive values. Moments were normalized to body weight (Nm/BW). The time between two consecutive heel-strikes was normalized for inter-subject comparison. Data were transferred
into SPSS 11.5 (SPSS Inc. Chicago, Illinois, USA) for statistical analysis.

Information from both right and left side were collected. Each patient contributed with one knee, the side with symptoms or in cases with bilateral symptoms the one with most symptoms. When comparing left and right side in controls we found no differences. In this study only the right side was used.

Presentation of results

The results are presented as maximum values for each parameter when the diagrams showed unambiguous maxima and minima of angles and moments in the three different groups analyzed. The hip adduction/abduction, knee rotation and hip rotation angles showed more variability concerning the occurrence of maximum and minimum values when related to a standardized gait cycle (Figures 1, 3 and 4). For the two former variables we instead used the recorded values at midstance for statistical analysis, when the knee is extended to maintain hip and knee stability. Even if there are minimum differences in ratio stand/swing phase, we decided to standardize the term midstance to occur when approximately 40% of the gait cycle had elapsed. Filtering of data was made by using sliding average, with a window of 25 frames at recording speed of 240 frames per second.

Statistics

Non-parametric tests were used. The two groups of patients with medial and lateral OA and the control group were analyzed using non-parametric ANOVA (Kruskal-Wallis test). If this test revealed presence of a difference, further analysis between the groups were done using Mann-Whitney test. To reduce the risk of spuriously occurring significances, the significance level was set at p < 0.025.

Results

(Table 1 - 4) Walking speed and stride length

There were a tendency to reduced speed in the two patient groups (p=0.027). The stride lengths were shorter in medial OA (p=0.001) and lateral OA (p=0.02) compared to controls, but these parameters did not differ between the two patient groups (p=0.8/0.7).

<table>
<thead>
<tr>
<th></th>
<th>Controls (n=15)</th>
<th>Medial OA (n=15)</th>
<th>Lateral OA (n=15)</th>
<th>p-value a</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (year)</td>
<td>69 (60–86)</td>
<td>70 (47–79)</td>
<td>70 (61–83)</td>
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</tr>
<tr>
<td>Height (cm)</td>
<td>163 (150–180)</td>
<td>164 (148–166)</td>
<td>166 (151–178)</td>
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<tr>
<td>Weight (kg)</td>
<td>70.5 (49.8–76.9)</td>
<td>77.8 (51–99.8)</td>
<td>69 (47.3–101)</td>
<td>0.1</td>
</tr>
<tr>
<td>BMI (kg/m²)</td>
<td>26.5 (21–30.1)</td>
<td>29.6 (20.2–40.5)</td>
<td>27.4 (19.4–35.2)</td>
<td>0.05</td>
</tr>
<tr>
<td>Cadence (stride/min)</td>
<td>111 (87–140)</td>
<td>102 (56–124)</td>
<td>102 (33–123)</td>
<td>0.2</td>
</tr>
<tr>
<td>Speed (m/s)</td>
<td>1.10 (0.8–1.7)</td>
<td>0.9 (0.4–1.3)</td>
<td>0.8 (0.2–1.2)</td>
<td>0.027</td>
</tr>
<tr>
<td>Stride length (m)</td>
<td>1.2 (1.0–1.4)</td>
<td>1.1 (0.7–1.2)</td>
<td>1.1 (0.5–1.4)</td>
<td>0.004</td>
</tr>
<tr>
<td>Ahlgbäck (Grade)</td>
<td>-</td>
<td>3 (1–5)</td>
<td>1 (1–4)</td>
<td>0.09</td>
</tr>
</tbody>
</table>

a Kruskal-Wallis between three and Mann-Whitney between two groups
Flexion/extension

Hip.

Both patient groups showed smaller median value of the maximum hip extension angle (medial OA 5° and lateral OA 11° less, respectively) than did the controls (medial: p=0.023; lateral: p=0.008). The corresponding flexion angle did not differ.

In the early swing phase where there is a peak flexion moment, this moment was higher in controls compared to patients with medial OA (p=0.004). A similar tendency was found in cases with lateral OA (p=0.026).

Knee.

Patients with lateral OA showed decreased flexion (p=0.006), whereas patients with
medial OA did not (p=0.033). Both groups walked with less maximum knee extension than observed in controls (p=0.004). The two patient groups had diminished knee extension moments (p<0.001), but their flexion moments did not differ from normal (p=).

Adduction/abduction

Hip.

At midstance cases with lateral OA had 2º more adduction of their hip joints than had controls (p=0.015). Compared to controls, cases with medial OA displayed 7º more abduction (p<0.001) at midstance but at the end of the cycle (past 60%) the pattern became closer to normal. (Figure 1)

Patients with OA walked with reduced abduction moments throughout stance. The maximum values (the most negative value) were

### Table 4. Statistical overview for all parameters in the total study population. Mann-Whitney U Test was used for comparison between the individual groups in pairs. Two digits are used for values below 0.05.

<table>
<thead>
<tr>
<th></th>
<th>Kruskal-Wallis Test</th>
<th>Mann-Whitney U Test</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Medial OA Controls</td>
<td>Medial OA Controls</td>
</tr>
<tr>
<td></td>
<td>Lateral OA Controls</td>
<td>Lateral OA Controls</td>
</tr>
<tr>
<td></td>
<td>(n=15)</td>
<td>(n=15)</td>
</tr>
<tr>
<td><strong>Hip</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Maximum motion</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Flexion</td>
<td>0.8</td>
<td>0.6</td>
</tr>
<tr>
<td>Extension</td>
<td>0.014</td>
<td>0.023</td>
</tr>
<tr>
<td>Internal rotation</td>
<td>0.1</td>
<td>0.8</td>
</tr>
<tr>
<td>External rotation</td>
<td>0.027</td>
<td>1.0</td>
</tr>
<tr>
<td>Adduction</td>
<td>&lt;0.001</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Maximum moment</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Flexion</td>
<td>0.014</td>
<td>0.004</td>
</tr>
<tr>
<td>Extension</td>
<td>0.6</td>
<td>0.8</td>
</tr>
<tr>
<td>Adduction</td>
<td>0.2</td>
<td>0.9</td>
</tr>
<tr>
<td>Abduction</td>
<td>0.014</td>
<td>0.007</td>
</tr>
<tr>
<td>Internal rotation</td>
<td>0.035</td>
<td>1.0</td>
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<tr>
<td>Maximum moment</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Flexion</td>
<td>1.0</td>
<td>0.9</td>
</tr>
<tr>
<td>Extension</td>
<td>&lt;0.001</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Adduction</td>
<td>0.021</td>
<td>0.030</td>
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<tr>
<td>Abduction</td>
<td>&lt;0.001</td>
<td>0.009</td>
</tr>
<tr>
<td>Internal rotation</td>
<td>0.08</td>
<td>0.7</td>
</tr>
<tr>
<td>Maximum moment</td>
<td>0.3</td>
<td>0.2</td>
</tr>
</tbody>
</table>

*40% of gait cycle (midstance)

Figure 1. Mean value of hip or relative femoral adduction/abduction angles related to a standardized gait cycle in women with medial or lateral OA and in the controls (Normal). Data represent mean values based on division of the Gait Cycle into 200 intervals for each patient, independent of the individual time period for each cycle. In the figures the mean value for each of the 3 groups (n=3x15) at each time interval is presented. Thus, each curve is based on 200 subsequent mean values. No further filtering of data was done.
reduced by 13 % (medial, p=0.007) and 8 % (lateral, p=0.019). The maximum adduction moment did not differ.

Knee.
In controls the maximum adduction angle reached 4° and the maximal abduction angle 3°. In medial OA the knee was in 8° more adduction (max value) through the gait cycle (p=0.002), and in lateral OA in 8° more abduction (max value) compared to controls (p<0.001).
The maximum (the most negative value) values of knee abduction moments were 52 % higher in medial OA (p=0.009) and 63 % less in lateral OA (p<0.001) than observed in the controls. (Figure 2)

Rotations
Hip.
In patients with lateral OA the femur was positioned in about 7° more external rotation (maximum value, p=0.011), but seemed to otherwise follow the variations observed in normal hip during the gait cycle. Patients with medial OA did not differ from controls (p=0.8). (Figure 3)
The outward rotation moments were small (-0.2 to 0.1 Nm/Kg) without any obvious differences (p≥ 0.035), but with a tendency to higher internal rotation moment in lateral OA compared to controls (p=0.021).

Knee.
At midstance the amount of internal/external tibial rotation differed between the groups (p=0.004). Cases with medial OA had 9° more external tibial rotation than controls and cases with lateral OA 6° less. These variations did not reach significance (p>0.06), but comparison between the OA groups revealed a significant difference (p=0.001). (Figure 4) The moments were small and did not differ (p≥ 0.08).

Discussion
Increased knowledge about any relation between lateral osteoarthritis of the knee and anatomical and biomechanical changes of the hip may be helpful in clinical diagnosis and treatment. Such information could also be of value to prevent or delay the onset of the disease. Our findings are, however, so far not firm enough to establish any certain association. Should future studies confirm such an association it seems likely that surgical changes of the hip anatomy such as osteotomy or and change of offset by insertion of a total hip replacement could influence the biomechanics of the knee in such a way that progression and even development of knee osteoarthritis will become more or less likely.
Our hypothesis that patients with lateral OA of the knee showed hip kinematics and kinetics different from those with medial OA could be verified. Both the hip and knee angles and moments differed between patients with medial and lateral OA of the knee during gait. In the knee the difference in abduction/adduction angles and moments were expected, whereas the diverging pattern of femoral and tibial rotations around the longitudinal axis was not. Our recordings of tibial and femoral rotations should however, be interpreted cautiously. Even if the technical resolution of our gait analysis is high, errors caused by limited spread of the retroreflective markers in the transverse plane and soft tissue motions have to be considered. Soft tissue motion is another problem and especially for markers placed at certain anatomical positions such as the superior border of the patella. This marker is subjected to skin movement during flexion/extension of the knee. It is, however, not used for detecting movement in transverse plane (i.e. the z-axis) but is used to record movement in the two other planes where skin movement has a reasonably small dignity.

We tried to obtain as equal groups as possible concerning patient related variables. None of the patients had hip symptoms. Patients with medial OA tended to be heavier than controls and mainly therefore also a tendency towards increased BMI. One study (24) found that the severity of the OA increased with higher BMI in cases with medial but not in those with lateral OA. The BMI did, however, not significantly differ between the two study groups. Bilaterality was also more common in the lateral group. Since these differences did not reach significance their possible influence has to be studied in larger materials.

Walking speed and stride length may influence the gait parameters and especially the moments in the sagittal plane (2, 10, 14, 26). Draganich et al (9) studied healthy adults in one younger and one older group, who walked at three different speeds. They found that the hip external rotation moment increased with 26% in early stance and the hip internal rotation moment with 27 % in late stance with increased speed. This observation is not directly applicable to our study since Draganich et al had their patients to pass an obstacle. Nonetheless, the differences in flexion and extension moments between patients and controls observed by us could partly be an effect of different walking speed. We think that the angular velocity has a higher influence on the hip than on the knee joint and also on the knee when compared to the ankle joint. The mass of the shank is comparatively low and will have less effect on the moment of the knee. The mass of the thigh is bigger which means that with higher walking speed the subject has to increase the extension moment at the end of the swing phase. This will influence the moment over the hip joint. In our study patients walked at self-selected speed, which probably was influenced by functional limitations and perhaps also some pain in individual patients. Standardization of the velocity parameter in these patients is, if possible, probably very difficult and may represent a walking pattern which the patient never uses except from in that particular study. The walking speed did
not differ between the groups and the two patient groups showed about equal median values. To what extent any by us undetected difference in motions and moments could be related to different walking speed cannot be evaluated in our study. The difference in walking speed was, however, numerically very small and the walking speed in the two patient groups was about equal. Thus, we do not think that the observed differences between lateral and medial OA could be related to patient demographics or other confounders even if this possibility never can be ruled out.

Many previous studies have found increased maximal adduction moment in knees with medial OA (1, 3, 15, 23, 27). Most of these studies report that they contrary to us measured the external moments, which are in the opposite direction. The part of the body weight, which influences the moment is also included in the internal moment, which make our computations not straight transferable to external moments. Thus, even if a strict comparison with these studies not is possible due to methodological differences our findings seem to support these previous observations.

Hurwitz et al (1) found that patients with unilateral osteoarthritis of the hip walked with decreased hip moments at extension, adduction, internal and external rotation. The knee adduction moment on the affected side was also significantly smaller than observed in the normal group. This was interpreted as a way to adapt to pain. Increasing adduction moments about the knee joint in patients with OA of the knee and varus alignment following NSAID treatment has been thought to be an effect of pain relief (22).

The two patient groups showed reduced hip extension but this reduction was more pronounced in cases with lateral OA. There could be several explanations. The relative external rotation of the femur in this group may require an associated pelvic rotation of the close to full extension of the hip. This rotation might be avoided by less hip extension.

Presence of hip disease may be another reason. In a previous study of patients with lateral knee OA (28) we observed that as many as half of the cases also had OA of the hip, which will impair the extension of the hip.

The external rotation of the femur was larger in patients with lateral OA both compared to controls and the group with medial OA. The tibial internal rotation at midstance (40% of gait cycle) was larger in cases with lateral OA compared to medial OA. This rotation might be an effect of the abnormal motion pattern in the hip. Rotational malalignment of the femur may be transferred to the knee where tibia will rotate internally to neutralize lower leg and foot alignment. Because of this, the medial femoral condyle may contact the tibia more anteriorly or the lateral femoral condyle more posteriorly, probably resulting in alterations of knee joint motions and tibiofemoral contact patterns during parts of or the arc of motion. If this femoral “abnormality” has been present since childhood, one could suspect that it influenced the development of the bones during growth. A more posterior contact on the tibia of the lateral femoral condyle might also explain findings of posterior location of the wear in lateral OA (29).

The relative tibio-femoral rotations throughout the arc of flexion-extension in knees with degenerative disease are not completely known. In a previous radiostereometric evaluation Saari et al (18) studied the kinematics of the knee during active extension and weight-bearing. From 50º to 20º of flexion they observed reduced internal tibial rotation in knees with varying degree of medial OA. Closer to full extension and past 50º of flexion, tibial rotation may become closer to normal. If so the femur in more pronounced knee flexion may articulate against the most anterior part of the tibial joint area, which in joints with less advanced degenerative changes still has some remaining cartilage. It could also be that the soft tissue envelop during a longer period maintain a more normal function in flexion past 50º than closer to full extension.
The cause for increased external femoral rotation in patients with lateral knee OA is unknown. Changed anatomy of the hip region could be one reason. In a previous radiographic study (28), we found that patients with lateral knee OA had a broader pelvis than controls and patients with medial knee OA. Lateral OA was also associated with a steeper angle of the femoral neck (coxa valga) on the anterior-posterior view, which could be an effect of more anteversion or increased external femoral rotation as observed in this dynamic study. This femoral deformity, which might compensate for a wider pelvis, might have developed during growth because of muscular imbalance and/or acetabular dysplasia. Prospective studies over long time periods are, however, necessary to evaluate this hypothesis and if the observed deviations of femoral and tibial rotation in cases with lateral OA primarily are caused by the disease of knee or not.

Our hypothesis about changed kinematics and kinetics in the hip could be verified. It is, however important to realize that retrospective studies can provide ideas about cause-relationships between predisposing factors and the development of a specific disease, but can never provide a definite answer. So far there are only vague indications that lateral and medial OA of the knee have different origin. This observation suggests that different factors will influence the progression of the disease depending on location. In the present study we did not only observe expected differences of knee angles and moments in the frontal plane, but also recorded diverging patterns of femoral and tibial rotations. These changes may be explained by a specific anatomy of the pelvis and hip observed in cases with lateral OA (28), but this hypothesis needs verification in further studies.

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References


