The Effect of Foot Progression Angle on Knee Joint Compression Force During Walking

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It is unclear how rotations of the lower limb affect the knee joint compression forces during walking. Increases in the frontal plane knee moment have been reported when walking with internally rotated feet and a decrease when walking with externally rotated feet. The aim of this study was to investigate the knee joint compressive forces during walking with internal, external and normal foot rotation and to determine if the frontal plane knee joint moment is an adequate surrogate for the compression forces in the medial and lateral knee joint compartments under such gait modifications. Ten healthy males walked at a fixed speed of 4.5 km/h under three conditions: Normal walking, internally rotated and externally rotated. All gait trials were recorded by six infrared cameras. Net joint moments were calculated by 3D inverse dynamics. The results revealed that the medial knee joint compartment compression force increased during external foot rotation and the lateral knee joint compartment compression force increased during internal foot rotation. The increases in joint loads may be a result of increased knee flexion angles. Further, these data suggest that the frontal plane knee joint moment is not a valid surrogate measure for knee joint compression forces but rather indicates the medial-to-lateral load distribution.

Keywords: walking, compression forces, inverse dynamics, foot progression angle

Several studies have directed attention to the fact that internal and external rotation of the lower limb during walking causes a redistribution of moments and forces in the knee.1-4 It is unclear how this redistribution affects the loading of the knee joint. It has been claimed that internal and external foot rotation reduces the load on the lateral and medial knee compartment, respectively.3,5 Theoretically, external foot rotation would result in a reduction of the lever arm of the ground reaction force that tends to adduct the knee joint.6 Theoretically, the external knee joint moment in the frontal plane (hereafter referred to as frontal plane knee joint moment) tends to adduct the knee, which in turn results in a loading of the medial knee joint compartment.6 In contrast, walking with internal foot rotation has been reported to lead to an increase in the frontal plane knee joint moment.2,3 It has been shown that a greater frontal plane knee joint moment increases the risk of progression of medial tibiofemoral osteoarthritis (OA) in patients with this disease.7 Furthermore, a recent study showed that progression of knee OA was associated with walking with less external foot rotation, which was explained by the anticipated relationship between the frontal plane knee joint moment (as the indirect measure for medial knee joint compartment loading) and the foot progression angle.8 Thus, it has been claimed that the frontal plane knee joint moment can be used as a measure for loading of the medial knee joint compartment.7,9-11 However, this has recently been disputed by Walter et al,12 who showed that a decrease in the frontal plane knee joint moment was not necessarily associated with a decrease in the medial knee contact force measured in vivo on a single subject with an instrumented knee joint. On basis of this a more detailed estimation of the knee compressive forces is needed to clarify the effect of foot rotations on the knee joint loading. Yet, no studies have quantified the loading of the knee compartments during walking with external and internal foot rotation. Accordingly, the aim of the current study was to quantify the knee compressive force during walking with internal, external and normal foot rotation and to determine if the frontal plane knee joint moment is an adequate surrogate for the compression forces in the medial and lateral knee joint compartments under such gait modifications. We hypothesized that internal foot rotation would reduce the loading of the lateral knee compartment, while external foot rotation would reduce the load on the medial knee compartment when compared with normal foot rotation.
Methods

Ten healthy young men participated in the study (mean [range]: 29.7 [25-32] years, 1.83 [1.75–1.91] m, 76.8 [60.5–90.5] kg). Informed consent was obtained from all subjects before participation.

The subjects were asked to walk at three different walking conditions: normal walking (NW), internal foot rotation (IFR) and external foot rotation (EFR). The order of walking modifications was fixed. The subjects were allowed to practice each walking condition for as long as they wanted to get familiar with the procedure and experimental setup. Each walking condition was supervised by visual inspection and corrected until each condition was performed successfully. A successful trial was defined as at least 5° increased foot rotation (ie, either internally or externally) compared with the normal condition. This was evaluated after each trial and if the foot progression angle did not exceed 5° the trial was discarded. The target walking speed was set to 4.5 km/h and controlled in each trial by photocells placed before and after the force platforms. A display showing the walking speed provided immediate feedback and was used to exclude trials that did not meet the specified speed 4.5 km/h ± 10%.

Three-dimensional (3D) gait analyses of the three different walking conditions were performed by using a Vicon MX system (Vicon Motion Systems, Oxford, UK) with six cameras (MX-F20, Vicon Motion Systems) operating at 100 Hz. Two force platforms (AMTI OR 6-5-1000, Watertown, Massachusetts, USA) embedded in the laboratory floor captured ground reaction force (GRF) data at 1500 Hz synchronized with the kinematic data. The 3D orientation of seven body segments of interest (pelvis, left and right thighs, left and right shanks, both feet) was obtained by tracking trajectories according to a common commercially available kinematic model (Plug-In-Gait, Vicon Peak, Oxford, UK) with markers placed bilaterally on the anterior and posterior iliac spines, lateral aspect of the thighs, lateral femoral epicondyles, lateral aspects of the shanks, lateral malleoli, calcanea and second metatarsal heads. The foot progression angle was calculated as the angle between the progression direction of the pelvis segment and the long axis of the foot.

The knee joint reaction forces, sagittal and frontal plane knee joint moment and knee joint flexion/extension angle obtained from the 3D gait analyses were used as input parameters to a statically determinate knee model used to estimate the knee joint compression force.11 The model assessed whether the total knee compression forces were sufficient to balance the net frontal plane knee joint moment, thereby keeping the knee joint closed laterally. The mediolateral position of the tibiofemoral contact point was fixed at 25% of the knee mediolateral joint diameter from the knee joint center, whereas the anteroposterior contact point varied with flexion. As long as the total knee compression forces acting over the contact point resist the frontal plane knee moment, no tension in lateral soft tissue is required; otherwise, appropriate lateral tissue tension is introduced to avoid lateral joint opening.

Lateral soft tissue tension indicates that all joint forces are supported by the medial joint compartment and more than the estimated muscle force is needed to avoid lateral opening of the joint. The total knee compression force was calculated as the vector sum of (a) the intersegmental reaction forces resolved along the long axis of the tibia, (b) the compression components of the active muscle group forces and (c) the axial component of the cruciate ligament tension. The muscles included were the hamstrings, gastrocnemii and quadriceps muscles. The hamstring and gastrocnemius complex constituted a flexor muscle group active when the net sagittal knee joint moment favored the flexors (ie, negative) and the quadriceps muscle represented an extensor muscle group active when the net moment favored extensors (positive). The muscle forces were calculated by combining the net sagittal plane joint moments with the muscle moment arms derived from a third-order polynomial relating the knee joint angle to the muscle moment arms.13 The muscle forces were distributed in accordance with the original model assuming that the knee extensor moment was carried by the quadriceps alone. The knee flexor moment were assumed to be carried by the hamstrings during the first 20% of the stance phase and by the gastrocnemii during the last 80% of the stance phase. These assumptions were based on previous studies of the muscle activity during walking measured by electromyography (EMG) in healthy subjects.11,14

The axial cruciate ligament forces were estimated under the assumption that the cruciates only resist anteroposterior shear forces. The anteroposterior shear forces were calculated as the vector sum of (a) the intersegmental reaction forces resolved along the anteroposterior axis of the tibial plateau and (b) the anteroposterior components of the active muscle group forces. The model differed from the one published previously by Schipplein and Andriacchi11 in that the knee joint diameter was obtained from each subject and used to calculate the position of the tibiofemoral contact point. In the original model, the knee joint diameter was fixed at 80mm for all subjects. No antagonistic contractions were allowed in the model. The total compressive knee joint forces as well as the force distribution between medial and lateral joint compartments were calculated.

The sagittal and frontal plane knee joint moments and the predicted joint compressive forces were presented as a proportion of the body weight. Peak values of the knee joint compressive forces, joint moments and the knee joint angle were measured in the stance phase from five walking trials obtained for each condition and served as input parameters to the statistical analysis.

The knee compression force curve was, in general, characterized by three peaks (Figure 1). However, the first peak occurring at heel strike varied considerably between the three walking conditions (Figure 1) and between subjects and was therefore not included in the statistical analysis. The two other peaks (referred to as 1st and 2nd peak) occurring around 20% and 80% of the stance phase were clearly detectable in all subjects and conditions and thus included in the statistics (Figure 1). In the following,
these knee compression force peaks will be referred to as the first and second peak, respectively. Finally, the peaks of the knee joint moment in the frontal and sagittal plane, the peak knee flexion angle and the average foot progression angle were included in the statistics. All observations were used, that is, no within-subject averaging.

A mixed linear model (PROC MIXED in the SAS system) was applied focusing on the main effects of walking technique (three levels: NW, IFR and EFR). The analyses were adjusted for walking speed. Significance was accepted at $P < .05$.

Results

The subjects were able to walk at 4.5 km/h ±10% during all three walking conditions. However, the walking speed during the IFR condition was significantly lower than the EFR and NW conditions (Table 1). Thus, all further statistical comparisons were adjusted for walking speed.

The foot progression angle differed significantly between the three walking conditions (Table 1). The foot was slightly externally rotated during NW (Table...

![Figure 1](image_url)

**Figure 1** — Estimated knee joint compressive forces (BW) from one representative subject (average of five walking trials in each walking condition: normal walking (solid lines), internal foot rotation (dashed lines) and external foot rotation (dotted lines). The total knee compression force, medial compartment compression force and lateral knee compartment compression force during the stance phase of each walking condition. Peak values from the early stance (first) and late stance (second) phase were extracted for each study subject and used as input parameters to the statistical analysis.
1) and was increased by 26.6° during the EFR condition. The difference between the mean foot progression angle during the NW and the IFR conditions was 27.5° (Table 1).

The total compression forces were significantly higher during both the IFR and EFR than during NW (Table 1, Figure 1). This was the case for both 1st and 2nd peak, corresponding to early and late stance. In both early and late stance, the compression force in the medial compartment was significantly higher during the EFR condition than both IFR and NW conditions (Table 1). The medial compression force was significantly higher in early stance during NW than IFR, while during the late stance phase there were no differences between NW and IFR (Table 1). In relation to NW, the knee joint moment in the frontal plane in early stance increased significantly by 72% during the EFR condition (\(P < .0001\)), while it decreased significantly by 75% (\(P < .0001\)) during the IFR condition (Figure 2). In late stance, the knee joint moment in the frontal plane was significantly lower during both the EFR (38%) and IFR (52%) conditions than during the NW condition (\(P < .0001\)) (Figure 2).

In general, the knee extensor moment was characterized by extensor dominance during both the EFR and IFR conditions compared with the NW condition (Figure 2). In the early as well as late stance the knee extensor moment was on average 50% higher during the EFR and IFR conditions than the NW condition (Figure 2).

The knee was significantly more flexed during the whole stance phase during the EFR and IFR conditions than the NW condition (Figure 2). During early stance foot rotation significantly increased the peak knee flexion angle by 12° (\(P < .0001\)) (Figure 2).

### Table 1 Knee joint compression forces (bw), foot progression angle (°) and walking speed (km/h) during normal walking (NW), and walking with internal foot rotation (IFR) and external foot rotation (EFR). Means (SD).

<table>
<thead>
<tr>
<th></th>
<th>NW</th>
<th>IFR</th>
<th>EFR</th>
<th>(P)-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Overall compression</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1st peak</td>
<td>4.0 (0.8)†‡</td>
<td>5.0 (1.1)</td>
<td>4.8 (0.8)</td>
<td>&lt;.0001</td>
</tr>
<tr>
<td>2nd peak</td>
<td>2.3 (0.5)†‡</td>
<td>3.0 (1.1)</td>
<td>3.0 (0.6)</td>
<td>&lt;.0001</td>
</tr>
<tr>
<td>Medial compartment compression</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1st peak</td>
<td>2.7 (0.6)†‡</td>
<td>2.3 (1.3)§</td>
<td>3.7 (0.5)</td>
<td>&lt;.0001</td>
</tr>
<tr>
<td>2nd peak</td>
<td>1.8 (0.3)†</td>
<td>1.8 (0.9)§</td>
<td>2.0 (0.3)</td>
<td>.0124</td>
</tr>
<tr>
<td>Lateral compartment compression</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1st peak</td>
<td>1.2 (0.4)†</td>
<td>2.7 (0.9)§</td>
<td>1.2 (0.8)</td>
<td>&lt;.0001</td>
</tr>
<tr>
<td>2nd peak</td>
<td>0.7 (0.3)†‡</td>
<td>1.4 (0.4)§</td>
<td>1.1 (0.5)</td>
<td>&lt;.0001</td>
</tr>
<tr>
<td>Foot progression angle</td>
<td>−11.1 (3.8)†‡</td>
<td>16.4 (8.7)§</td>
<td>−37.7 (8.0)</td>
<td>&lt;.0001</td>
</tr>
<tr>
<td>Walking speed</td>
<td>4.5 (0.2)‡</td>
<td>4.4 (0.2)§</td>
<td>4.5 (0.2)</td>
<td>.0029</td>
</tr>
</tbody>
</table>

Statistically significance located between: ‡ NW vs. IFR, † NW vs. EFR, § IFR vs. EFR.

### Discussion

The main finding of the current study was that knee joint compression force increased significantly during walking with either internal or external foot rotation. Furthermore, the results revealed that the medial knee joint compartment compression force increased during EFR and the lateral knee joint compartment compression force increased during IFR. This was in contrast to our initial hypothesis and to several earlier studies that have concluded that the compression force in the medial knee joint compartment was reduced when walking with external foot rotation and that the compression force in the lateral knee joint compartment was reduced when walking with an internally rotated foot.\(^1,13,15,16\) However, the underlying results behind these conclusions may be insufficient. Previous studies have used the frontal plane knee joint moment as an indirect measure of the compression forces in the knee.\(^3,13,15-17\) This study used a statically determinate knee model\(^11\) to calculate the compression forces in the knee joint, which revealed that the frontal plane knee joint moment was an insufficient proxy for knee joint compression forces. Rotations of the lower extremity influence the compression forces in the knee joint significantly.\(^12,18\) Scrutinizing the agreement between the frontal plane knee joint moment and the knee joint compression forces shows that in early stance the peak frontal plane knee joint moment during external foot rotation was higher than during internal foot rotation (Figure 2). However, there were no significant differences in the total compression forces in early stance between IFR and EFR (Table 1). In late stance the peak frontal knee joint moment during normal walking was approximately twice as large as the peak values of both...
external and internal foot rotation (Figure 2), but the total compression forces were significantly higher during both external and internal foot rotation than normal walking (Table 1). Furthermore, a high frontal plane knee joint moment resulted in a reduction in the compression force of the lateral knee joint compartment while it increased the compression force of the medial knee joint compartment (Figure 3). This confirmed that the frontal plane knee joint moment is not a valid proxy for knee joint compression forces during walking with internally or externally rotated feet when assessed using the statically determinate knee model that originally stated the importance of the frontal plane moments in medial-lateral compartment load distribution.

The results of this study support the results from Walter et al., who found that the frontal plane knee joint moment did not correspond directly to the loading of the medial knee joint compartment. Walter et al. studied a single patient with an instrumented force-measuring knee implant; thus, the results from this may be regarded as a
The results suggest that the present model, despite of the lack of allowance for co-contractions, used to predict compression forces in the medial knee joint compartment are in agreement with the knee contact forces measured in vivo by Walter et al.\textsuperscript{12}

Gou et al\textsuperscript{1} suggested that external foot rotation could be used as a therapeutic modality in patients with mild-to-moderate knee OA to reduce medial knee joint compartment loading. However, the present results do not indicate that foot rotation leads to reduced loading of the medial knee joint compartment. During external foot rotation the loading of the lateral knee joint compartment showed the same level as when walking normally, but the medial knee joint compartment was loaded significantly more than during both internal foot rotation and normal walking. When walking with internal foot rotation the loading of the medial knee joint compartment was significantly lower than during both external foot rotation and normal walking. On basis of this there is no obvious advantage of using foot rotation as a modality to unload the medial knee joint.

In the current study the subjects walked with more flexed knees in the rotated conditions than in normal walking. Increased knee flexion may explain the increased knee joint compressions observed during the rotated conditions,\textsuperscript{19} as the sagittal plane knee moment is a significant contributor to the joint load in the knee joint model used. The subjects were only instructed to walk with external and internal foot rotation but no instructions regarding the knee flexion angle, hip rotation, trunk position or step lengths were given. The effects of these parameters on the knee joint compression forces are beyond the scope of this study. Thus future investigations are needed to reveal this.

It is not possible to keep the same degree of knee flexion, hip rotation and trunk position as during normal walking when walking with large internal and external foot rotation. In contrast to the present findings, Lin et al\textsuperscript{2} reported that the knee joint flexion angle was increased during normal walking compared with walking with internal and external foot rotation. However, Lin et al\textsuperscript{2} did not report the walking velocities, which may explain kinematic differences. Walking velocities may also be the reason for the difference between the present results and the results from Lynn et al\textsuperscript{3} where subjects walked slower (∼4.0 km/h) than in the current study (Table 1). Earlier studies have established that velocity is a main contributor to the magnitude of the knee joint moments.\textsuperscript{20,21}

There are some limitations to this study. The order of walking was fixed. This could potentially induce bias since one gait type may affect the following, especially when internal and external rotation is wanted. The impact of this bias is unknown, but we assume that it has not affected the main conclusions, since the foot progression angle was controlled after each trial and the single trial was discarded if the angle did not exceed 5 degrees from the average normal walking angle.

\textbf{Figure 3} — The effect of changes in the frontal plane moment on the distribution of compressive forces in the medial (solid dots) and lateral (open triangles) joint compartments. Averaged (n = 10) first peak values of the frontal plane knee joint moment from the three gait conditions (NW, IFR and EFR) on the x-axis. On the y-axis are the lateral and medial compressive forces expressed as percentage of the total compression force (first peak).
Further, we only included 10 subjects and they were all healthy young males, and since OA primarily affects women above 50 years of age the transferability to the OA population may be questionable. Furthermore, the validity of the statically determinate knee model has not yet been tested in patients with instrumented knees and is therefore unknown. The model does not account for any co-contractions, leading to a possible underestimation of the compression forces. The knee flexion angles in IFR and EFR gaits could be speculated to reduce the co-contraction contribution as higher knee flexion angles would reduce the compressive component of the hamstring forces.

In conclusion, this study showed that the total estimated knee joint compression forces increased when walking with external or internal foot rotation. More specifically, the estimated medial knee joint compartment loading increased when walking with external foot rotation while the estimated lateral knee joint compartment load increased when walking with internal foot rotation. The increases in joint loads may be a result of increased knee flexion angles. Furthermore these data suggest that the frontal plane knee joint moment is not a valid surrogate measure for knee joint compression forces.

References