

## Chapter 8

# The effectiveness of voluntary modifications of gait pattern to reduce the knee adduction moment



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*Submitted*

## 8.1 Abstract

It has been suggested that gait modifications should be used in the retraining of patients with osteoarthritis (OA) of the knee, in order to reduce the external knee adduction moment (KAdM). This study focused on the effect of walking speed, foot position and trunk sway, on the 3D knee moments. Gait analyses of fourteen young healthy volunteers were performed in a gait laboratory. Subjects walked at three different speeds, with toe-in and toe-out gait, and with medio-lateral trunk sway. Fast walking speed increased the KAdM (17-30%) and knee flexion moment (32%), but a slower walking speed did not decrease the KAdM. Toe-in mainly decreased the KAdM (45%) and the transverse moment (38%) during early stance. Toe-out decreased the KAdM during late stance (56%), but increased the KAdM during early stance and midstance (21-24%), due to decreased endorotation of the hip ( $5-8^\circ$ ) with knee flexion. Trunk sway was effective in decreasing the KAdM during early stance and midstance (31-33%). The main effect of the gait modifications was observed in the KAdM, but changes in sagittal and transverse knee moments and kinematics were also observed. This indicates that, when estimating knee-load, taking only the frontal plane kinetics into consideration may lead to erroneous simplifications. No conclusive beneficial effects were found in any of the gait modifications throughout the entire stance phase. The use of gait alterations in the retraining of patients with knee OA (either medial or lateral) remains questionable.

## 8.2 Introduction

Normal mechanical loading on the knee joint appears to be essential for maintaining healthy articular cartilage. However, abnormal high joint loading leads to local cartilage damage, and therefore increases the risk of osteoarthritis (OA) of the knee, since cartilage has limited self-repair ability [1-3]. Varus/valgus malalignment of the knee joint, or changes in the net external frontal knee moment (i.e. the knee adduction moment (KAdM)), have been found to be important determinants in the onset and progression of knee OA [4-7].

The KAdM reflects the distribution and magnitude of the load transferred through the medial versus the lateral compartment of the tibiofemoral joint [6,8-10]. It is mainly the product of the ground reaction force (GRF) and its lever arm with respect to the knee joint. During normal gait, the load on the medial compartment of the knee is approximately 2.5 times greater than the load on the lateral compartment [3,6,8]. In patients suffering from medial compartment knee OA, an increase in KAdM has been found (up to 20-40%) during gait, compared to asymptomatic subjects [6,11-14].

To decrease the rate of progression of knee OA, it is suggested that mechanical loading on the affected compartment of the knee should be minimized. Different interventions and strategies have been used to reduce the KAdM in patients with compartment knee OA. Such interventions include valgus-bracing, heel wedges [15], quadriceps muscle-strengthening [16] and high tibial osteotomy [17].

Several studies have investigated modifications in the gait pattern, as a strategy to alter the load on the knee joint, and thereby to decrease the pain and further progression of OA [18]. These modifications include a reduction in walking speed, a toe-out foot position, and medio-lateral trunk sway. Patients with knee OA are reported to walk more slowly compared to age-matched controls [19-21], and a correlation between KAdM and walking speed has been reported [22]. A toe-out foot position has been associated with a decrease in KAdM, due to an outward shift in the centre of pressure (CoP), and with less likelihood of OA progression [14,23-27]. Lateral trunk lean has been associated with a decrease in the KAdM during walking, due to a shift in the centre of mass (CoM) of the body [28-30].

While the effects of the various modifications in gait pattern on the frontal plane knee moment (i.e. the KAdM) have been studied before, there is still a lack of studies focusing on the relative magnitude of the influence of these different gait manipulations on the 3D knee joint loading, unbiased by differences in subjects or methodology. Therefore, the aim of this study was to investigate the effect of walking speed, foot positioning and trunk

sway on the 3D knee moment and 3D joint kinematics in a group of young healthy volunteers. We hypothesized that a slower walking speed and medio-lateral trunk sway would decrease the KAdM. Furthermore, we hypothesized that toe-in walking would result in a higher KAdM, and toe-out walking would result in a decrease in the KAdM (caused by a medial or lateral shift in the CoP with respect to the CoM) over the entire stance phase, accompanied by an increase in the external knee flexion moment. Finally, we hypothesized that the greatest decrease in KAdM, without changes in sagittal and transverse knee moments, would result from the medio-lateral trunk sway.

## **8.3 Methods**

### **8.3.1 Subjects**

Fourteen healthy young adults participated in the study (8 females, 6 males; age  $23.8 \pm 3.9$  years (mean  $\pm$  standard deviation); body mass  $72.7 \pm 16.3$  kg; height  $1.81 \pm 0.11$  m). The subjects were recruited from the student population of the VU University (Amsterdam, the Netherlands) and from acquaintances of the authors. The Ethical Review Board of the Faculty of Human Movement Science (VU University, Amsterdam, the Netherlands) approved the study protocol and full written informed consent was obtained from all subjects.

### **8.3.2 Procedure**

Gait analyses of the subjects were performed in the gait laboratory of the VU University Medical Centre (Amsterdam, the Netherlands). The subjects walked barefoot on a 10 metre walkway: (i) at three different speeds: self-selected, slow and fast walking speed, (ii) with toe-in and toe-out gait, and (iii) with medio-lateral trunk sway. Slow walking speed was 0.20 m/s slower than the self-selected speed, fast walking speed was 0.20 m/s faster than the self-selected speed. Toe-in, toe-out and medio-lateral trunk sway were performed at the same walking speed as the self-selected speed for normal gait. Walking speed was monitored during each trial with an infrared speedometer.

For the toe-in and toe-out gait conditions the subjects were asked to walk at their self-selected speed with the maximum amount of comfortable foot rotation throughout the entire trial. This maximal foot progression angle (FPA) was determined prior to the measurement, and the subjects were given the opportunity to get used to the new gait

manipulation. To ensure that each subject would start and remain in the correct position, tape was stuck to the floor on either side of the walkway, and the trials were video-taped. For the medio-lateral trunk sway, the subjects were instructed to drop the arm on the ipsilateral side of the stance leg throughout the entire trial [28] while walking at their self-selected gait speed. Prior to the measurement, the subjects were given the opportunity to get used to the new gait condition.

During the measurements, kinematic and kinetic data were collected with an optoelectronic marker system (OptoTrak 3020, Northern Digital Instruments, Waterloo, Canada) and a force plate embedded in the walkway (AMTI OR6-5-1000, Watertown, MA, USA). The movements of the trunk, pelvis, thighs, shanks, and feet were tracked by means of technical clusters of three markers, in combination with virtual anatomical markers [31], at a sample frequency of 100 Hz. Force plate data were collected with a sample frequency of 1000 Hz.

Data on three successful trials (i.e. a foot placement within the outline of the force plate and a correct walking speed) were collected for both the right and the left leg in each gait condition. Prior to the measurements, body mass, height and the circumference of the thighs and shanks were measured.

### **8.3.3 Data analysis**

Optoelectronic marker data and force plate data were analysed with BodyMech ([www.BodyMech.nl](http://www.BodyMech.nl)), custom-made software based on MATLAB (R2009b, MATLAB®, The Mathworks). Anatomical frames, according to Cappozzo et al. [31], were used to calculate the joint kinematics.

The stance phase in the gait cycle was determined with the GRF, measured with the force plate (threshold of 10 N). The kinematic data (i.e. joint and segment angles) and kinetic data (i.e. GRF and knee joint moments) were time normalized to 100% of the stance phase. The external knee moments were expressed with respect to the femur anatomical frame [32] and normalized to body weight (BW in N) and height (H in meters) [33] (%BW\*H). Data on both the right and the left leg were included in the analyses.

The early stance peak (ESP), midstance (MS) and late stance peak (LSP) were determined on the basis of the characteristic shape of the vertical component of the GRF vector. Maximum values for the first and last 50% of the stance phase defined the ESP and the LSP respectively, and MS was defined by the minimum value between ESP and LSP. Based on this, the magnitudes of the frontal (KAdM), sagittal and transverse knee moments were calculated for all gait conditions at 24% (ESP), 50% (MS) and 77% (LSP) of the stance

phase. The impulse in the three planes (i.e. the time integral in %BW\*H\*s) was also calculated over the entire stance phase, because this parameter also involves the effects of stance duration [34].

To determine the FPA in toe-in and toe-out gait, the angle of the long axis of the foot segment in the global coordinate system, relative to the walking direction axis [27], was computed and averaged over the stance phase. The trunk-pelvis angle (TPA) during medio-lateral trunk sway was defined as the medio-lateral flexion angle in the frontal plane between the trunk and the pelvis (i.e. the orientation of the line defined by C7 and T8 with respect to a line perpendicular to the pelvis), and calculated for the ESP and LSP. 3D ankle, knee and hip joint angles were also determined at ESP, MS and LSP. Each parameter was averaged per subject over three trials for each gait condition.

### **8.3.4 Statistical analysis**

The statistical analysis was performed in SPSS (Version 15.0). The means of the data from the three trials for each leg of each subject were included in the analysis. Prior to the analysis, the data were checked for normal distribution and assumptions of sphericity. Repeated measures analysis of variance (ANOVA, SPSS Software Version 15.0) was used to determine whether the 3D knee moment magnitudes at ESP, MS and LSP, the 3D knee moment impulse, and the 3D ankle, knee and hip kinematics for each gait condition (speed, toe-in/out and trunk sway) differed significantly from the magnitudes for normal gait at self-selected walking speed. The differences were considered to be significant at  $P < 0.05$ .

## **8.4 Results**

Figure 8.1 shows the average of the 3D knee moments of all trials for all subjects and all walking conditions. Table 8.1 shows the KAdM at ESP, MS, LSP and the impulse for all walking conditions, compared to the normal gait at self-selected speed. Table 8.2 presents the significant differences in the sagittal and transverse knee moments.

### **8.4.1 Walking speed**

There was a 15% decrease in walking speed for slow walking and a 14% increase for fast walking, compared to self-selected walking speed. At slow walking speed, there was an

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increase in KAdM magnitude in midstance (+40%) and an increase in KAdM impulse over the entire stance phase (+28%). However, there was no difference in the KAdM magnitude in early and late stance. At ESP, the knee flexion moment and the moment in the transverse plane were decreased, accompanied by a decrease in knee flexion ( $4^\circ$ ) and hip flexion ( $6^\circ$ ). The vertical GRF was decreased at ESP (5%) and LSP (5%) and increased at MS (14%).

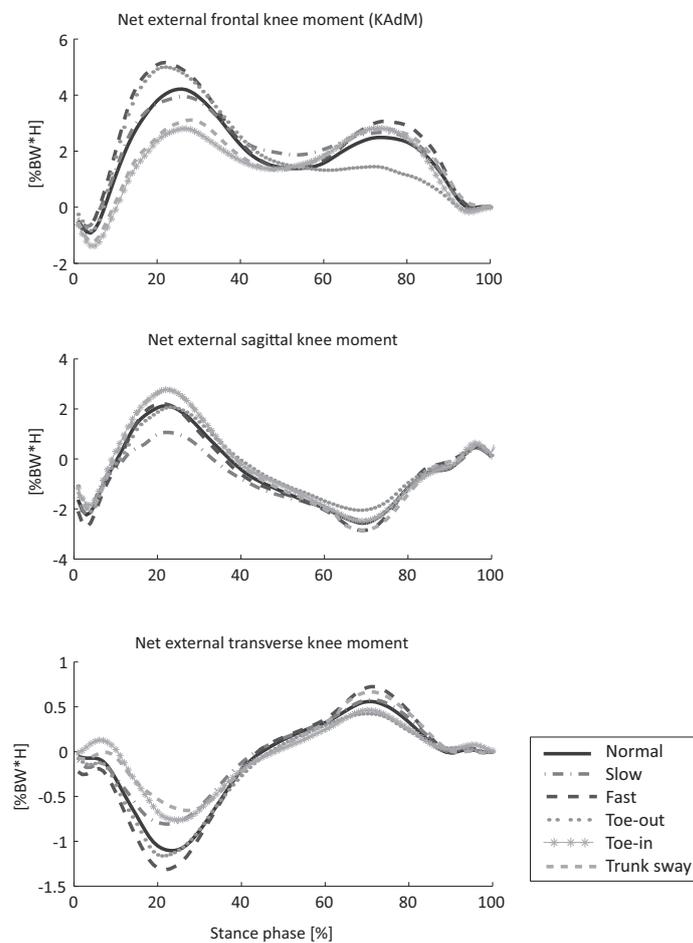


Figure 8.1. Average frontal, sagittal and transverse knee moments of 14 healthy subjects of normal gait at self-selected walking speed (dark solid line), of gait at slow walking speed (dot-dashed line) and fast walking speed (black dashed line), of toe-out gait (dotted line) and toe-in gait (asterisk line), and of trunk sway gait (grey dashed line).

**Table 8.1. The KAdM at different gait modifications of 14 healthy subjects**

<i>KAdM</i>		normal	manipulation	difference	
		<i>mean±SE</i>	<i>mean±SE</i>	<i>mean (%)</i>	<i>P</i>
<b>Slow</b>					
<i>Walking speed</i>	<i>[m/s]</i>	1.37±0.03	1.17±0.04	-0.20 (15%)	< 0.001
<i>KAdM (+ad)</i>	<i>ESP</i>	3.68±0.25	3.61±0.33	n.a.	0.760
	<i>MS</i>	1.56±0.13	2.18±0.19	+0.62 (40%)	< 0.001
	<i>LSP</i>	1.96±0.30	2.09±0.33	n.a.	0.170
	<i>Impulse</i>	1.19±0.11	1.52±0.15	+0.33 (28%)	0.008
<b>Fast</b>					
<i>Walking speed</i>	<i>[m/s]</i>	1.39±0.03	1.59±0.03	+0.20 (14%)	< 0.001
<i>KAdM (+ad)</i>	<i>ESP</i>	3.76±0.27	4.87±0.43	+1.11 (30%)	0.006
	<i>MS</i>	1.56±0.15	1.71±0.18	n.a.	0.463
	<i>LSP</i>	2.00±0.34	2.44±0.35	+0.44 (22%)	< 0.001
	<i>Impulse</i>	1.23±0.12	1.44±0.13	+0.21 (17%)	0.019
<b>Toe-in</b>					
<i>FPA (+lat)</i>	<i>[°]</i>	8.71±1.74	-10.0±3.54	-19°	< 0.001
<i>KAdM (+ad)</i>	<i>ESP</i>	3.71±0.23	2.03±0.44	-1.68 (45%)	< 0.001
	<i>MS</i>	1.44±0.13	1.29±0.14	-0.15 (10%)	0.048
	<i>LSP</i>	2.02±0.28	2.03±0.38	n.a.	0.974
	<i>Impulse</i>	1.14±0.09	0.80±0.12	-0.34 (30%)	0.001
<b>Toe-out</b>					
<i>FPA (+lat)</i>	<i>[°]</i>	8.32±1.62	24.2±1.49	+16°	< 0.001
<i>KAdM (+ad)</i>	<i>ESP</i>	3.78±0.22	4.70±0.30	+0.92 (24%)	< 0.001
	<i>MS</i>	1.42±0.11	1.72±0.19	+0.30 (21%)	0.022
	<i>LSP</i>	2.05±0.25	0.91±0.23	-1.14 (56%)	< 0.001
	<i>Impulse</i>	1.12±0.09	1.11±0.09	n.a.	0.776
<b>Trunk sway</b>					
<i>TPA (+lat)</i>	<i>ESP</i>	3.76±0.62	17.9±2.05	+14°	< 0.001
	<i>LSP</i>	3.48±0.61	-2.85±1.76	-6°	0.013
<i>KAdM (+ad)</i>	<i>ESP</i>	3.68±0.25	2.55±0.28	-1.13 (31%)	< 0.001
	<i>MS</i>	1.56±0.13	1.05±0.22	-0.51 (33%)	0.010
	<i>LSP</i>	1.96±0.30	2.09±0.33	n.a.	0.370
	<i>Impulse</i>	1.13±0.11	0.91±0.15	-0.22 (19%)	0.008

*KAdM* = Knee Adduction Moment (%BodyWeight\*Height, i.e. [Nm/(N\*m)\*100])

*ESP* = Early Stance Peak; *MS* = Midstance; *LSP* = Late Stance Peak

*FPA* = foot-progression angle; *TPA* = trunk-pelvis angle

*SE* = standard error of the mean

*n.a.* = not applicable (if *P*>0.05)

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**Table 8.2. The external sagittal and transverse knee moments at different gait modifications of 14 healthy subjects (for  $P < 0.05$  compared to normal gait)**

Sagittal and transverse knee moment			normal	manipulation	difference	
			mean±SE	mean±SE	mean	P
<b>Slow</b>						
Sagittal knee moment (+flex)	ESP	[%BW*H]	1.87±0.46	0.88±0.51	-0.99	0.001
	Impulse	[%BW*H*s]	-0.29±0.20	-0.49±0.23	-0.20	0.025
Transverse knee moment (+exo)	ESP	[%BW*H]	-1.04±0.07	-0.86±0.09	+0.18	0.029
<b>Fast</b>						
Sagittal knee moment (+flex)	ESP	[%BW*H]	1.58±0.43	2.09±0.32	+0.51	0.043
Transverse knee moment (+exo)	ESP	[%BW*H]	-1.08±0.07	-1.40±0.10	-0.32	0.005
	LSP	[%BW*H]	0.41±0.06	0.52±0.05	+0.11	0.002
<b>Toe-in</b>						
Transverse knee moment (+exo)	ESP	[%BW*H]	-1.08±0.07	-0.67±0.10	+0.41	0.002
	MS	[%BW*H]	0.03±0.05	-0.05±0.05	-0.08	0.020
	Impulse	[%BW*H*s]	-0.09±0.02	-0.03±0.02	+0.06	0.012
<b>Toe-out</b>						
Transverse knee moment (+exo)	ESP	[%BW*H]	-1.07±0.07	-1.18±0.08	-0.11	0.023
	MS	[%BW*H]	0.04±0.04	-0.04±0.05	-0.08	0.003
	LSP	[%BW*H]	0.40±0.05	0.29±0.05	-0.11	0.003
	Impulse	[%BW*H*s]	-0.08±0.02	-0.12±0.02	-0.04	0.002
<b>Trunk sway</b>						
Transverse knee moment (+exo)	ESP	[%BW*H]	-1.04±0.07	-0.62±0.10	+0.42	0.002
	Impulse	[%BW*H*s]	-0.08±0.03	-0.03±0.03	+0.05	0.050

Joint moments are expressed in %BodyWeight\*Height, i.e. [Nm/(N\*m)\*100]

ESP = Early Stance Peak; MS = Midstance; LSP = Late Stance Peak

SE = standard error of the mean

Only significant differences are shown ( $P < 0.05$ )

Fast walking speed caused an increase in KAdM impulse (+17%) and KAdM magnitude during early (+30%) and late stance (+22%), but not in midstance. At ESP, the external knee flexion moment and transverse moment were increased, but there was no significant difference in joint kinematics. At LSP, the knee moment in the transverse plane was increased. The vertical GRF was increased at ESP (7%) and LSP (3%) and decreased at MS (11%).

### 8.4.2 Foot progression angle

FPA's differed significantly between the three conditions (normal gait 8° exorotation; toe-in 10° endorotation; toe-out 24° exorotation). Walking speeds in toe-in and toe-out gait did not differ significantly from self-selected normal walking speed.

Toe-in resulted in a reduced KAdM magnitude in early (-45%) and midstance (-10%), and a reduced KAdM impulse (-30%). However, no difference was found in late stance. In the frontal plane, ankle adduction (inversion) increased over the entire stance phase (3-7°), knee abduction (valgus) decreased at ESP and MS (2-3°), and the hip adduction decreased at MS and LSP (3-4°). The transverse knee moment decreased at ESP and increased at MS. In the transverse plane, there was an increase in endorotation (5-8°) in ankle and hip joint angles, and a decrease in exorotation (5-6°) in knee joint angles during the entire stance phase. The external knee flexion moment increased at ESP in toe-in gait, however this difference was not significant, due to high standard deviations. In the sagittal joint angles there was a slight increase in hip flexion at ESP and MS (2-3°), and also in knee flexion at MS (2°), and a decrease in ankle dorsal flexion (3°).

In toe-out there was no difference in the KAdM impulse over the entire stance phase, but the KAdM magnitude increased at ESP (+24%) and MS (+21%) and decreased at LSP (-56%). In frontal plane kinematics there was a decrease in hip adduction (3-5°) and an increase in knee abduction (valgus) (2-3°) at ESP and MS. The knee moment in the transverse plane increased at ESP and at LSP. Ankle endorotation decreased at LSP (3°), hip endorotation decreased at ESP and MS (5-8°), and hip exorotation increased at LSP (5°). In the knee, there was an increased exorotation (6-8°) during the entire stance phase. At MS, the knee extension moment decreased (0.40 %BW\*H), but this difference did not reach significance, due to high standard deviations. There was no significant difference in the sagittal knee moment at ESP and LSP. Sagittal joint angles only showed a decrease of 3° in hip flexion at ESP and an increase of 2° in ankle dorsal flexion at LSP.

### 8.4.3 Medio-lateral trunk sway

In trunk sway the TPA increased with 14° at ESP, compared to normal gait (i.e. increased lateral flexion towards the stance leg). At LSP, the TPA decreased with 6°. The trunk sway resulted in a decrease in the KAdM at ESP (-31%) and MS (-33%), and a decrease in the KAdM impulse (-19%). The KAdM magnitude did not differ at LSP. Hip adduction decreased (4°) at ESP and hip extension increased (4°) at LSP. At ESP, there was a significant decrease in the transverse knee moment. The knee flexion moment and knee

flexion angle both increased at ESP (moment: 1.15 %BW\*H; angle 4°), however these differences did not reach significance, due to high standard deviations.

## 8.5 Discussion

### 8.5.1 Walking speed

The results of the present study showed increases in KAdM in early and late stance in the fast walking condition, and an increase in the KAdM in midstance in the slow walking condition. These were caused by an increase in the magnitude of the vertical GRF. It is well-known that a faster walking speed results in an increase in the vertical GRF in early and late stance, and in a decrease in the vertical GRF in midstance [35].

A slower walking speed resulted in a decrease in knee and hip flexion, as well as in the knee flexion moment during early stance, which is consistent with reports in the literature [20,35,36]. No changes were observed in the KAdM in early or late stance at slow walking speed, while the vertical GRF showed a small reduction of 5%. At slower walking speed, the vertical GRF becomes almost as flat as in a static standing trial (i.e. bodyweight). Therefore, a further decreasing in the walking speed would likely not result in any further decrease in the GRF peaks or KAdM.

The KAdM impulse increased at both slow and fast walking speed, which is not consistent with the findings of Robbins et al. [34]. They only found an increase in impulse in slow walking speed. However, in our study the impulse in fast walking speed increased, despite the decrease in stance duration. This can be explained by the increase in KAdM magnitude in early and late stance. Walking speed also influenced the transverse knee moments. Landry et al. [36] also observed an increase in the transverse knee moments with an increase in faster walking speed.

The slow walking speed in this study is still faster than the walking speeds that are often reported for patients with OA [20]. However, in a study carried out by Mündermann et al. [22], patients with knee OA had higher self-selected normal walking speed, compared to the slow walking speed in our study. They concluded that patients with less severe knee OA can reduce the KAdM by reducing their walking speed, although this is highly patient-specific, and may not be effective for patients with more severe knee OA, due to increased varus alignment or a slower self-selected walking speed. The results of our study show that a faster walking speed has a negative effect on medial knee-load throughout stance.

This may explain why increased walking speeds are avoided by OA patients. However, the results also suggest that a slower walking speed than the self-selected speed is unlikely to result in a further decrease in knee-loading.

### **8.5.2 Foot progression angle**

In contrast to our hypotheses, a toe-in position decreased the KAdM in early and midstance. In toe-out, the KAdM decreased in late stance, but increased during early and midstance. In the knee flexion/extension moments there was no significant difference in toe-in and toe-out, although a small increase was observed in early stance in toe-in, and in midstance in toe-out. Furthermore, there was a change in transverse knee moments and joint rotations.

Lynn et al. [24,25] also reported a decrease in the magnitude of the KAdM in toe-out during late stance. Jenkyn et al. [27] reported a decrease in the KAdM in toe-out throughout the entire stance phase, and an increase in the knee flexion moment in early stance. Lin et al. [37] reported an increase in the KAdM in toe-in during the entire stance phase.

Although toe-out has been associated with a decrease in KAdM and less likelihood of progression of medial knee OA [23-27,37,38], the results of our study show a considerable increase in medial knee-load in early and midstance. Toe-out is only effective in reducing the KAdM in late stance, due to a more lateral CoP. This result is line with the findings of recent systematic review on gait modifications which showed that toe-out consistently reduced the late stance KAdM [18]. In early stance the CoP is still positioned under the heel. Hence, toe-out gait does not change the CoP, compared to normal gait. An increased in KAdM in early stance is the result of a decrease in endorotation of the hip. Hip rotation causes the knee to point more laterally. In combination with knee flexion during stance, the lever arm of the GRF vector increases (i.e. when the GRF is located at the medial side of the knee). Similarly, in the toe-in position, the hip is more in endorotation, and in combination with knee flexion, this results in a decrease in the KAdM in early stance (Figure 8.2). When knee malalignment is present, these effects may even be amplified. However, when walking at a slower walking speed the effect will be less prominent, due to a decrease in knee flexion. A pure rotation of the foot, without hip rotation, would not result in a change in KAdM during early stance, however this posture seems to be hard to achieve during gait.

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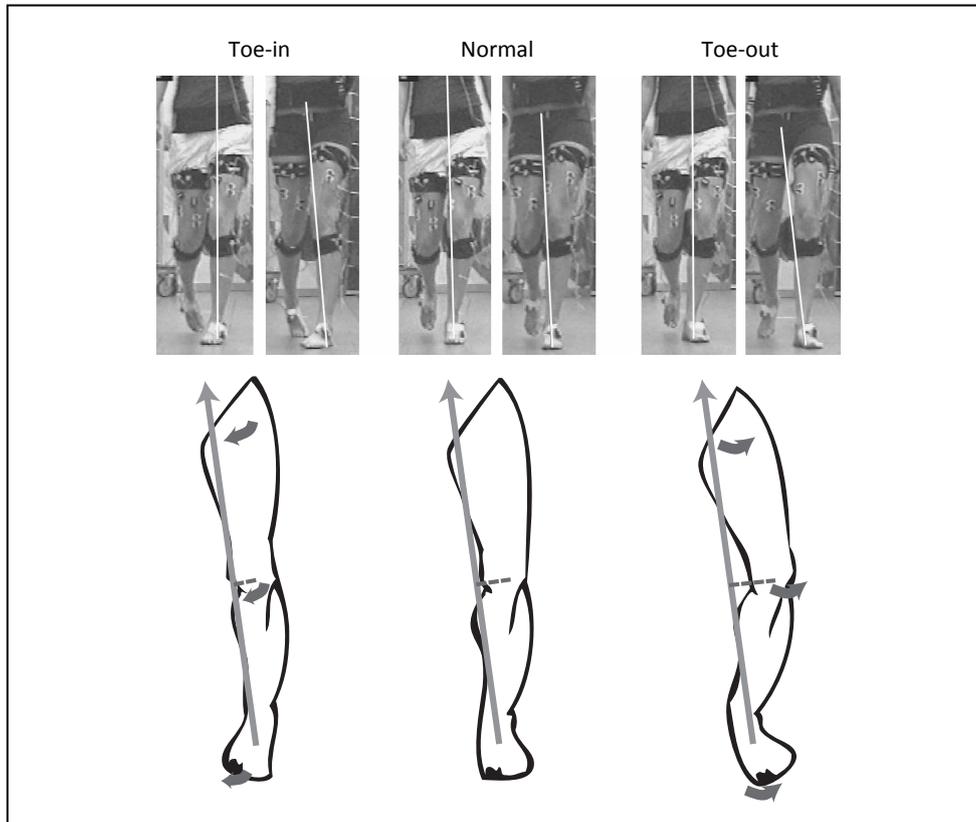


Figure 8.2. Example of two subjects and schematic drawing of the frontal view of Toe-in, Normal and Toe-out gait during early stance. Hip endorotation decreases with increasing toe-out angle, causing an increase in the lever arm of the GRF vector to the knee joint, which results in an increased KAdM (amplified by an increase in knee flexion).

### 8.5.3 Medio-lateral trunk sway

Medio-lateral trunk sway explains a substantial variation in dynamic knee joint loading in patients with medial compartment knee OA [29]. In healthy subjects, Mündermann et al. [28] reported even a 65% reduction in KAdM during early stance with increased trunk sway ( $10 \pm 5^\circ$ ), a 57% reduction in early stance hip adduction moment, and also an increase in knee flexion ( $3^\circ$ ). However, they reported no significant differences in lateral GRF.

As expected, the results of our study show that trunk sway decreased the KAdM in early stance and midstance. However, no difference in the KAdM was found during late stance.

This confirms the finding by Mundermann et al. [28] that demonstrated a large reduction in early stance, but no change during late stance [18]. In early stance, there was a lateral flexion of the trunk-pelvis angle towards the stance leg, which resulted in a lateral shift in CoM. Consequently, the KAdM decreased, due to a more lateral direction of the GRF vector. In late stance, the trunk already moved towards the contra-lateral side to prepare for the stance phase of the current swing leg (Figure 8.3). This explains why there was no decrease in the KAdM during late stance. We hypothesized that there would be no changes in the sagittal and transverse planes with trunk sway, but in early stance the knee moment in the transverse plane decreased and the knee flexion moment increased.

#### 8.5.4 Clinical implications

Gait modifications cause obvious changes in knee joint loading and kinematics. Therefore, when evaluating patients with knee OA (e.g. to assess progression or to evaluate treatment), changes in walking speed, foot positioning and trunk sway need to be taken into consideration. Patients might change their gait in an attempt to change the knee loading.

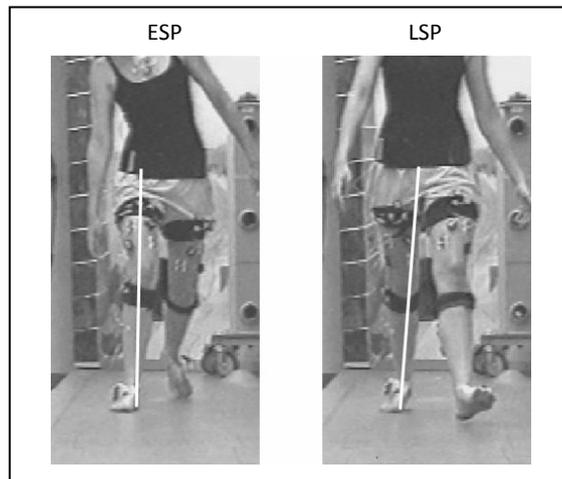


Figure 8.3. Medio-lateral trunk sway during early and late stance. At early stance (ESP) the trunk is laterally flexed towards the stance leg, at late stance (LSP) the trunk moves towards the contra lateral side

However, the benefits of these gait modifications in the retraining of knee OA patients as a simple and inexpensive treatment strategy [18,39] are still unclear. Although the KAdM is generally accepted as an indicator of load distribution in the knee joint, some remarks must be made. At micro-level, cartilage damage is the result of local stresses and strains, caused by joint loading and influenced by the cartilage volume and the collagen fibril-network morphology [40]. Several studies have reported a significant positive correlation of the KAdM with tibia bone distribution, although no association has been found with cartilage volume [41-44]. In addition to the magnitude of the load, the duration and rate of loading may also play a crucial role in cartilage damage [45,46]. It is still unknown which factor is the most important in the onset and progression of cartilage degeneration.

Hence, it is not clear which parameter of the KAdM is the most important as a clinical outcome measure for knee OA. Consequently, it is difficult to conclude which gait modification is the most effective to reduce knee-load, since none of the manipulations had an unambiguous 'positive' effect on the KAdM (i.e. a decrease in the KAdM during the entire stance phase, which would have been beneficial for patients with medial knee compartment OA).

Many studies that have investigated the KAdM in patients with knee OA have used the magnitude of the first peak of the stance phase as an outcome measure. In normal gait this is the highest peak [18]. It is reported to be a predictor of the presence, severity and rate of progression of medial compartment knee OA [3,6,13]. In that case, trunk sway would be the most effective modification, but it might be detrimental with regard to energy cost [30], walking speed and back-load. Moreover, in a recent study no correspondence of this first peak was found with medial knee contact forces, measured with a knee implant [47]. In another study, based on musculoskeletal modelling, significant differences were only found between healthy controls and patients with moderate OA in the second peak of knee contact force [48]. Although the relationship of the KAdM with the internal knee contact forces needs further investigation, the findings of these studies may suggest that the second peak of the KAdM might be important for the evaluation of knee-load.

The second peak has been reported to be influenced by in-toeing and out-toeing [24,27]. In our study, toe-out gait decreased the KAdM significantly during late stance, but toe-in gait had the greatest effect during early stance and on the impulse, which reflects both the duration and the magnitude of the load. Unwanted effects of a change in foot positioning, such as slower walking speed [25] (which will increase load duration and impulse), energy cost, and stability, must also be taken into consideration.

In the present study, we did not evaluate loading rate, which can be calculated from the slope of the KAdM in loading response. Slower walking speed will probably be most effective in reducing the loading rate, although it resulted in the highest increase in impulse.

Furthermore, gait modifications showed changes in the sagittal and transverse knee moments and kinematics. A recent study, based on advanced musculoskeletal modelling [48], reported that a reduction in total knee contact force during late stance, caused by a slower walking speed, was the result of a reduction in gastrocnemius muscle force. The gastrocnemius is responsible for propulsion of the body in forward direction at toe-off. Apparently, not only the frontal plane kinetics, but also the sagittal plane kinetics should be taken into consideration to estimate knee-load from the external joint moments. Taking only the frontal plane kinetics into consideration may cause erroneous simplifications. This questions the accuracy of the KAdM as the sole indicator of knee-load [27,47].

A limitation of the current study is that the effects of the gait modifications were studied in a group of young healthy subjects. Biomechanically the effects are expected to be similar in an elderly population, however knee malalignment, slower walking velocity or overweight may alter the effects in a knee OA population. Furthermore, it should be noted that a 'positive' effect on knee-load for patients with medial compartment knee OA might be negative for patients with lateral compartment knee OA and totally different for patients with patellofemoral OA. The influence of the alterations in gait on efficiency, energy cost, muscle activity (e.g. hamstrings [25]), changes in other kinematics and kinetics (e.g. ankle, hip, and back), and on long-term clinical relevant changes in disease progression should be the subject of comprehensive evaluation in future OA research.

## 8.6 Conclusion

Walking speed, a toe-in/out foot position and medio-lateral trunk sway significantly affect knee joint kinetics. The main effect is observed in the KAdM. However, no conclusive beneficial effect on the KAdM of any of the gait modifications was found throughout the entire stance phase. Furthermore, all gait modifications resulted in changes in sagittal and transverse knee moments and kinematics. This indicates that, when estimating knee-load, taking only the frontal plane kinetics into consideration may result in erroneous simplifications. Therefore, the use of gait alterations in the retraining of patients with knee OA (either medial or lateral) remains questionable.

## 8.7 References

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