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Lateral trunk lean explains variation in dynamic knee joint load in patients with medial compartment knee osteoarthritis

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Summary

Objective: To test the hypothesis that selected gait kinematics, particularly lateral trunk lean, observed in patients with medial compartment knee osteoarthritis explain variation in dynamic knee joint load.

Method: In this cross-sectional observational study, 120 patients with radiographically confirmed varus gonarthrosis underwent three-dimensional gait analysis at their typical walking speed. We used sequential (hierarchical) linear regression to examine the amount of variance in dynamic knee joint load (external knee adduction moment) explained by static lower limb alignment (mechanical axis angle) and gait kinematics determined *a priori* based on their proposed effect on knee load (walking speed, toe-out angle, and lateral trunk lean angle).

Results: Approximately 50% of the variation in the first peak external knee adduction moment was explained by mechanical axis angle (25%), Western Ontario and McMaster Universities Osteoarthritis Index pain score (1%), gait speed (1%), toe-out angle (12%), and lateral trunk lean angle (13%). There was no confounding or interaction with Kellgren and Lawrence grade of severity.

Conclusions: Gait kinematics, particularly lateral trunk lean, explain substantial variation in dynamic knee joint load in patients with medial compartment knee osteoarthritis. While largely ignored in previous gait studies, the effect of lateral trunk lean should be considered in future research evaluating risk factors and interventions for progression of knee osteoarthritis.

Key words: Knee osteoarthritis, Gait, Knee adduction moment, Kinematics.

Introduction

Excessive localized loading within the knee joint during walking has been hypothesized to be an important risk factor for the progression of knee osteoarthritis (OA)[1,2]. Although lower limb malalignment has been established as a risk factor[3] and is consistent with this hypothesis, it is typically measured using static radiographs that may not adequately reflect knee joint loading during locomotion. There is considerable evidence to suggest that quantitative gait analysis provides an appropriate means to measure knee joint load during walking[4–8]. In particular, the external adduction moment about the knee has been demonstrated to be an indirect measure of the load on the medial compartment[9–13] and a risk factor for disease progression[14]. Accordingly, the external knee adduction moment is being used increasingly in the study of knee OA, including its use as an outcome measure in intervention studies[15–22].

Notably, the external knee adduction moment reflects other characteristics of walking (gait kinematics) that would not be detected under static situations and therefore provides additional information that static radiographs cannot[8,17,23]. Although the importance of identifying and controlling for covariates to proposed risk factors and outcomes measures is well-recognized in health research[24], covariates to the external knee adduction moment remain unclear. While strong theoretical rationale for associations between the external knee adduction moment and various gait kinematics has been proposed[2,4,25–28], experimental evidence quantifying these relationships is limited. To better understand potential predictive factors for disease progression, as well as the effect of interventions aimed at decreasing knee joint load, the relationships among the external knee adduction moment and gait kinematics need to be further investigated.

In general, the external knee adduction moment can be described as the product of the frontal plane components of the ground reaction force (GRF) and lever arm[29]. The lever arm magnitude is the orthogonal distance between the knee joint centre of rotation and the GRF line of action vector as it passes from the centre of pressure (COP) under the foot to the vicinity of the centre of mass (COM) of the body (Fig. 1, panel A). As a result, the lever arm magnitude, and therefore the external knee adduction moment, can be readily changed by altering the relative positions of the knee centre, COP, and COM during walking.

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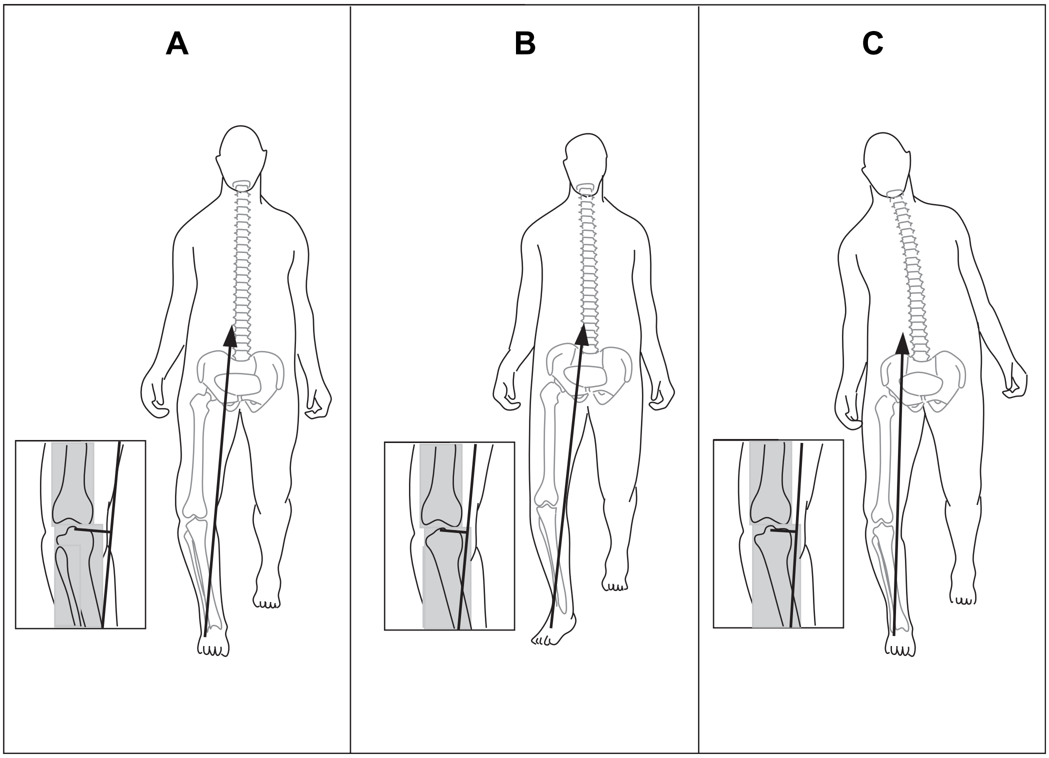

Fig. 1. The external knee adduction moment is primarily the product of the GRF vector in the frontal plane and the frontal plane lever arm distance (inset) between the GRF and knee joint centre (panel A). Increasing the magnitude of the toe-out angle (panel B) will theoretically shift the COP laterally thereby reducing the GRF lever arm distance at the knee, and subsequent knee adduction moment. Leaning the trunk laterally over the stance limb (panel C) will act to displace the COM laterally which will also reduce the lever arm and subsequent knee adduction moment magnitudes.

Toeing-out, resulting in lateral displacement of the COP, is the most commonly proposed kinematic variable suggested to reduce the external knee adduction moment (Fig. 1, panel B)[16,19,25,26,30–32]. Additionally, lateral trunk lean resulting in lateral displacement of the COM towards the stance limb during walking would theoretically result in a lateral shift of the GRF vector and subsequent reduction in lever arm magnitude (Fig. 1, panel C). Although such a gait pattern is observed clinically in some patients with OA and has been proposed by several authors as a mechanism to decrease knee joint load[28,33–35], we are unaware of any previous gait analysis studies that have quantified lateral trunk lean in patients with knee OA, or its relationship with other kinematics, lower limb alignment and knee joint loading. Therefore, the objective of this study was to test the hypothesis that selected gait kinematics, particularly lateral trunk lean, observed in patients with medial compartment knee OA explain variation in dynamic knee joint load.

Materials and methods

PARTICIPANTS

Participants were recruited from patients presenting to a tertiary care centre for treatment of unresolved knee pain localized to the medial compartment. These patients were referred by their primary care physician for consultation with an orthopaedic specialist regarding treatment options. All patients were diagnosed with varus gonarthrosis, defined as $>0°$ varus alignment and OA confined primarily to the medial compartment of the tibiofemoral joint. The diagnosis of OA was based on criteria described by Altman *et al.*[36]. We assessed grade of severity using the Kellgren and Lawrence (KL) scale[37]. Pain was assessed using the Western Ontario and McMaster Universities OA Index (WOMAC) pain domain[38] (transformed to scores out of 100, where higher values indicated less pain). Only patients with no history of lower limb surgery on the side being tested, with the exception of arthroscopic meniscectomy and/or debridement, were included in the study. Participants were also excluded if they were unable to walk independently without the use of a gait aid. All participants signed an informed consent form before testing. The study was approved by the institution's Research Ethics Board for Health Sciences Research Involving Human Subjects.

LOWER LIMB ALIGNMENT

The amount of varus alignment was based on the mechanical axis angle measured on double-limb standing anteroposterior (AP) radiographs. Radiographs taken in double-limb standing have previously been shown to provide valid and reliable measurement of lower limb alignment[39,40]. Patients stood in bipedal stance with the patellae centred over the femoral condyles and feet straight ahead to attain a true AP radiograph[33] and to control for any effects of foot rotation on measures of lower limb alignment[41]. The X-ray beam was centred on the knee at a distance of approximately 2.5 m and beam exposure was determined based on each patient's leg mass.

The joint centres of the hip, knee, and ankle were identified on each radiograph. The centre of the hip was found as the geometric centre of the femoral head using a circular template[42,43], the centre of the knee was identified as the midpoint of the tibial spines extrapolated inferiorly to the surface of the intercondylar eminence[43], and the centre of the ankle was defined as the mid-width of the tibia and fibula at the level of the tibial plafond[33].

The mechanical axis angle of the lower limb was measured on each radiograph and was defined as the included angle formed between a line drawn from the centre of the hip to the centre of the knee and a line drawn from the centre of the ankle to the centre of the knee[44]. Positive values corresponded to valgus alignment while negative values indicated varus alignment of the lower limb.

GAIT ANALYSIS

Patients underwent three-dimensional (3D) gait analysis using an eight-camera motion capture system (Eagle EvaRT; Motion Analysis Corporation, Santa Rosa, CA) synchronized with a single, floor-mounted force platform (Advanced Mechanical Technology Inc., Watertown, MA). Passive-reflective markers were placed on the patient using a 22-marker, modified Helen Hayes marker set[45]. In addition, extra markers were placed bilaterally over the medial knee joint line and medial malleolus during an initial static standing trial on the force platform to determine body mass, marker orientation, and positions of joint centres of rotation for the knee and ankle. These four additional markers were removed prior to gait testing.

During the gait analysis, patients were instructed to walk across the laboratory at their typical walking speed while kinetic (sampled at 1200 Hz) and kinematic data (sampled at 60 Hz) were collected during the middle of several strides. All gait analyses were conducted barefoot to negate the potentially confounding effect of shoe type on walking biomechanics. A total of five trials were obtained for the limb in which the patient reported had the greatest severity of symptoms and pain. Thus, each patient contributed radiographic and gait data from only one limb to the overall analysis.

External moments about the knee were calculated from the kinematic and kinetic data using commercial software (Orthotrak 6.0; Motion Analysis Corporation, Santa Rosa, CA)[46] and custom post processing and data reduction techniques. Each lower limb segment (foot, shank, and thigh) was modelled as a rigid body with a local coordinate system that coincided with anatomically relevant axes. Inertial properties of each limb segment were approximated anthropometrically and translations and rotations of each segment were reported relative to neutral positions defined during the initial standing static trial.

Walking speed was calculated as the average walking speed between successive foot contacts of the tested limb. The toe-out angle was calculated as the angle between a line drawn between the centre of the ankle and the head of the second metatarsal and the forward progression of the body. Positive values corresponded to toeing-out while negative values corresponded to toeing-in. The lateral trunk lean angle was calculated as the angle of a line drawn from the midpoint of the anterior superior iliac spines (ASISs) to the midpoint of the anterior tips of the acromion processes with respect to vertical (Fig. 2). Positive angles corresponded to a shift in the body's COM over the stance limb while negative angles corresponded to a shift in the body's COM to the swing limb. Lastly, pelvic obliquity was calculated for each trial as the angle between a line drawn between the ASIS markers with respect to the horizontal. The maximum and minimum pelvic obliquity angles for each trial were identified to assess the possible role of the pelvis in the calculation of the lateral trunk lean angle. Positive pelvic obliquity angles corresponded to a rise of the pelvis on the swing limb, while negative values represented a pelvic drop on the swing limb.

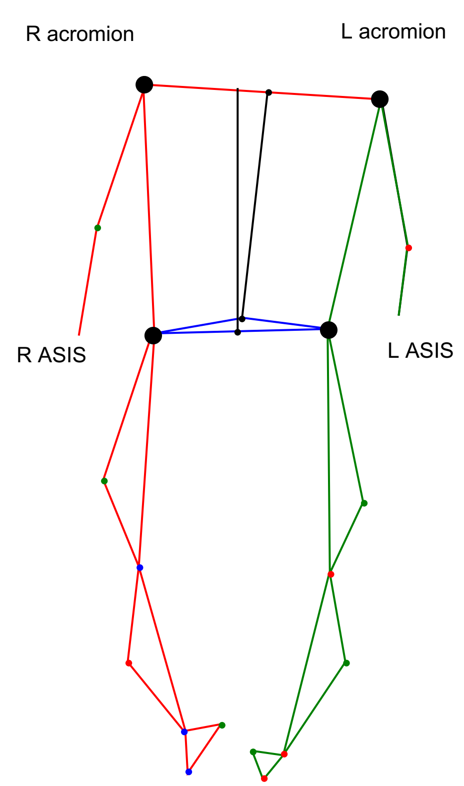

Fig. 2. The lateral trunk lean angle was calculated as the angle from vertical of a line connecting the midpoints of the acromion processes and the midpoints of the ASISs.

The overall peak magnitudes of the external knee adduction moment in the first and second halves of stance (i.e., first and second knee adduction moment peaks) were identified from its waveform and were normalized to body size (%BW × ht). The first and second knee adduction moment peaks were identified using an algorithm that employed a moving window to examine knee adduction moment values. Specifically, local peaks in the external knee adduction moment waveform were identified if they were immediately preceded and followed by lesser values. In addition, to ensure a true peak had occurred, local peaks were recorded only if they were preceded by a minimum of five continuously ascending values and followed by a minimum of five continuously descending values. The first and second knee adduction moment peaks were then identified as the largest of these local peaks in the first and second halves of stance, respectively. The percentages of stance where these peaks occurred were also identified. In the event that no identifiable peak occurred in a given half of stance, no knee adduction moment value for that half of stance was reported.

The gait kinematic variables investigated were quantified as their magnitudes at the point in stance coinciding with the first and second peak external knee adduction moments, as well as their peak magnitudes regardless of point in stance. If no identifiable knee adduction moment peak occurred for a given half of stance, no corresponding kinematic data were identified. Average values for all variables were calculated by averaging across the five trials for each participant. Intraclass correlation coefficient (ICC) estimates of the reliability of the mechanical axis angle ($ICC_{2,1} = 0.98$)[39] and peak knee adduction moment ($ICC_{2,1} = 0.86$)[47] measurements have been previously reported. Reliability of the speed ($ICC_{2,1} = 0.92$), toe-out angle ($ICC_{2,1} = 0.69$), and trunk lean angle ($ICC_{2,1} = 0.91$) measurements were evaluated on a subset of 15 patients from the present study who returned for a second test session at least 24 h after, and within 1 week, of the first test session.

STATISTICAL ANALYSIS

We used scatterplots and Pearson correlation coefficients to examine the bivariate relationships among the external knee adduction moment, mechanical axis angle, WOMAC pain score, gait speed, toe-out angle and lateral trunk lean angle. We used sequential (hierarchical) linear regression to create forced entry models evaluating the amount of variance in the external knee adduction moment explained by mechanical axis angle, WOMAC pain domain score, gait speed, toe-out angle, and lateral trunk lean angle.

The potential impact of KL grade on these models was examined by testing for confounding and interaction (effect modification)[48]. We first created two subgroups based on KL grades 1 or 2 and 3 or 4. We tested for confounding by comparing the beta coefficients for the toe-out and trunk lean variables before and after removal of the KL subgroup variable from the full model. Operational confounding was defined as a change in the beta coefficient of more than 10%[48]. We then created interaction terms between the KL subgroup and toe-out angle (KL × toe-out), and KL subgroup and trunk lean angle (KL × trunk lean). We tested for effect modification by entering the interaction terms into the model and tested for their significance[48].

We also conducted a number of sensitivity analyses by repeating these regression procedures while excluding those patients with body mass index (BMI) ≥ 35 ($n = 18$), those with KL grade 1 ($n = 19$), and those with suspected anterior cruciate ligament insufficiency based on physical examination (JRG) ($n = 15$). We also repeated the regression analyses while using the previously described different methods of quantifying peak external knee adduction moment, toe-out angle and lateral trunk lean angle. Regression diagnostics were conducted on all models using residual analysis to ensure that the equations satisfied the assumptions for linear modelling. All statistical analyses were performed using the Statistical Package for the Social Sciences (SPSS v. 15; SPSS Inc., Chicago, IL).

Results

One hundred and twenty participants (60 females and 62 right knees) were tested from November 2002 to September 2006. Data from two patients with BMI over 50 kg/m^2 were excluded due to difficulty in locating anatomical landmarks for proper marker placement and the likelihood of marker movement during data collection. Four patients were missing WOMAC pain data. Complete data from 114 patients (55 females and 57 right knees) were used in the primary analysis. Patient demographic and clinical characteristics are provided in Table I. Reported means suggested that the present sample was similar to previous investigations of patients with varus gonarthrosis[7,18,20,49]. The characteristics of the present sample varied widely, especially for age and body mass which is consistent with the target patient population. Quartiles indicated that patients were generally

Table I

Participants' demographic and clinical characteristics. Possible WOMAC pain scores range from 0 to 100 with lower scores indicating more pain. Possible KL grades of radiographic OA severity range from 0 to 4, with higher values indicating more disease severity (n = 114)

	Mean (SD)	Min, max	Quartiles
Age (yr)	45.5 (0.8)	21, 76	39, 45, 53
Mass (kg)	86.4 (19.7)	43.2, 141.5	71.0, 87.1, 100.1
Height (m)	1.71 (0.10)	1.47, 1.96	1.63, 1.71, 1.78
BMI	29.4 (5.8)	18.0, 48.8	25.0, 28.0, 33.6
WOMAC pain	56.9 (20.6)	0, 100	45, 55, 70
Mechanical axis angle	−7.4 (3.6)	−0.6, −16.4	−4.2, −7.1, −10.1
KL grade in medial compartment of knee			
0	0		
1	19		
2	30		
3	25		
4	40		

middle aged, overweight and had moderate-to-large amounts of varus malalignment. There were several patients with KL grades 2, 3, and 4. The relatively high number of patients with grade 4 changes should be noted and reflects the fact that these patients presented to a tertiary care centre.

Ensemble averages (i.e., waveforms obtained by averaging across subjects at each percent of stance such that data were time-normalized to 100% of stance) for the external knee adduction moment, lateral trunk lean angle, and toe-out angle are shown in Fig. 3. On average, the knee adduction moment exhibited the typical double peak waveform[50] with peaks occurring at approximately 31 and 76% of stance. One patient did not exhibit a definitive external knee adduction moment peak value for the first half of stance. Instead, the knee adduction moment waveforms for all five trials exhibited gradual increases until about 30% of stance, then levelled off until midstance, before rising to a peak at 75% of stance. Six patients did not exhibit an identifiable peak knee adduction moment value in the second half of stance. Instead, values tended to taper off after reaching the peak at around 35% of stance. This is similar to what has been reported in some patients by previous authors[30]. The amount of lateral trunk lean varied throughout stance, exhibiting a sinusoidal-like pattern as the COM migrated between alternating stance limbs. Pelvic obliquity angles indicated that participants exhibited small amounts of contralateral pelvic drop (2.67°) or rise (3.07°) during stance, suggesting that lateral trunk lean was achieved primarily by leaning the shoulders and trunk over the stance limb.

Summary statistics for the knee adduction moment, mechanical axis angle, pelvic obliquity angles and gait kinematics are provided in Table II. Participants exhibited less toe-out (7.71° vs 8.82°) and more trunk lean (3.11° vs 1.90°) at the time of the first external knee adduction moment peak compared to the second peak. Correlation coefficients describing the relationships between the first peak knee adduction moment ($n = 113$), mechanical axis angle, WOMAC pain domain score, gait speed and gait kinematics occurring at the time of the first peak knee adduction moment are provided in Table III(a), while the same relationships with the second peak knee adduction moment ($n = 108$) are shown in Table III(b). Correlation coefficients among peak values for toe-out and trunk lean regardless of time in stance were similar to those in Table III(a) and (b) and are not reported. There was a significant positive correlation between the mechanical axis angle

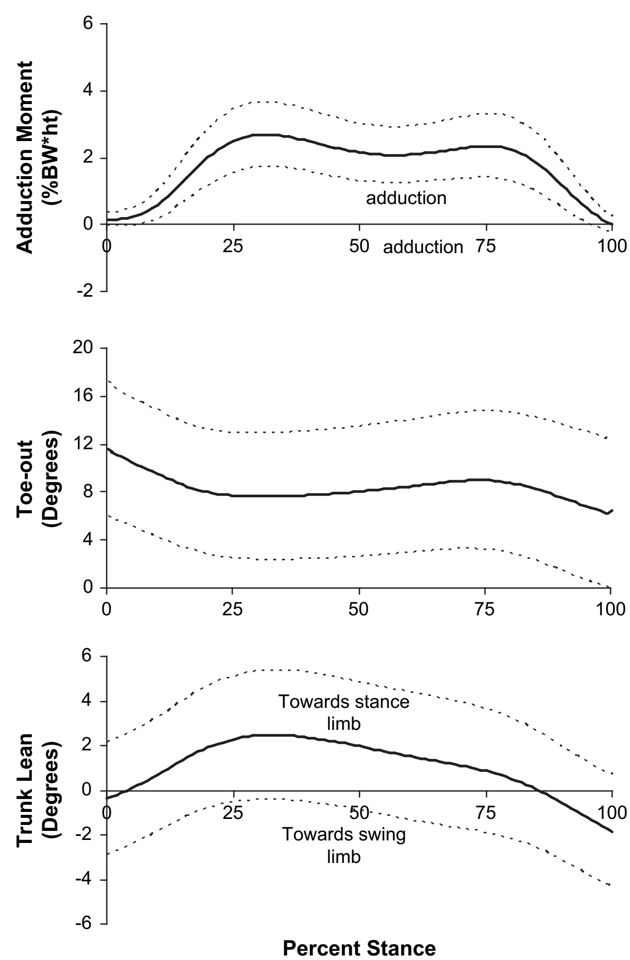

Fig. 3. Group ensemble averages for the knee adduction moment (top), toe-out angle (middle), and lateral trunk lean angle (bottom). Solid lines indicate the group ensemble averages while the dotted lines correspond to ±1 standard deviation (SD). Positive values correspond to an external knee adduction moment, trunk lean towards the stance limb, and out-toeing of the foot.

Table II

Participants' gait data including the position of the stance cycle in which they occurred as well as 25th, 50th, and 75th percent quartiles for each variable. Positive trunk lean angles correspond to a leaning of the trunk over the stance limb. Note that data pertaining to the first (n = 113) and second (n = 108) peak knee adduction moments and corresponding gait kinematics only include those with definitive waveform peaks

$n = 114$	Mean (SD)	% Stance	Min, max	Quartiles
Speed (m/s)	1.06 (0.16)		0.65, 1.53	0.94, 1.08, 1.17
Knee adduction moment (*N m*)				
First peak	2.91 (0.92)	32.5 (5.5)	0.84, 5.92	2.21, 2.83, 3.52
Second peak	2.59 (0.86)	75.9 (4.2)	0.65, 5.03	2.01, 2.49, 3.24
Toe-out angle (°)				
Overall peak	12.17 (5.17)	15.2 (21.4)	1.83, 26.79	8.25, 12.14, 15.61
Value at first peak knee adduction moment	7.71 (5.18)		−2.98, 22.95	4.21, 8.12, 10.80
Value at second peak knee adduction moment	8.82 (5.59)		−3.61, 25.13	4.84, 8.86, 11.97
Lateral trunk lean angle (°)				
Overall peak	3.62 (2.67)	37.6 (14.4)	−2.15, 10.45	1.75, 3.37, 5.84
Value at first peak knee adduction moment	3.11 (2.64)		−2.28, 8.92	1.05, 2.76, 5.36
Value at second peak knee adduction moment	1.90 (2.79)		−4.89, 8.92	0.49, 1.90, 3.94
Pelvic obliquity (°)				
Peak pelvic rise	3.07 (2.48)	23.1 (10.4)	−3.09, 9.79	1.42, 3.22, 4.55
Peak pelvic drop	−2.67 (2.35)	81.5 (24.3)	−7.62, 4.10	−4.23, −2.45, −1.33
Range	5.73 (2.17)		2.16, 11.54	4.06, 5.47, 7.14

and the first ($r = 0.51$) and second peak ($r = 0.61$) knee adduction moments. In contrast, there were significant negative correlations between the first and second peak knee adduction moments and toe-out ($r = -0.31$, -0.26) and lateral trunk lean ($r = -0.39$, -0.33) occurring at the time of these knee adduction moments. Lateral trunk lean measures were also significantly correlated to WOMAC pain scores ($r = -0.18$, -0.21). Representative data from two female patients with similar demographic and clinical characteristics are shown in Fig. 4 to highlight the differences in knee adduction moment magnitudes based on changes in trunk lean and toe-out angulation.

Residual analyses indicated that all data were consistent with the assumptions of linear regression. Excessive multicollinearity was not observed in any model (variance inflation factor (VIF) < 1.1, condition index < 30). Approximately 50% of the variation in the first peak external knee adduction moment was explained by mechanical axis angle (25%), WOMAC pain score (1%), gait speed (1%), toe-out angle (12%), and lateral trunk lean angle (13%). Beta coefficients for the toe-out and trunk lean variables changed by less than 10% after adding the KL subgroup variable revealing no evidence of confounding. Addition of the KL × toe-out or KL × trunk lean interaction terms did not significantly ($P > 0.6$) contribute to the model. The relationships between toe-out angle and first peak knee adduction moment, and lateral trunk lean and first peak knee adduction moment, are plotted for both KL subgroups in Fig. 5. Approximately 60% of the variation in the second peak external knee adduction moment was explained by mechanical axis angle (38%), WOMAC pain score (2%), gait speed (0%), toe-out angle (11%), and lateral trunk lean angle (7%). Again, beta coefficients for the toe-out and trunk lean variables changed by less than 10% after adding the KL subgroup variable, and the addition of the KL × toe-out or KL × trunk lean interaction terms did not significantly ($P > 0.5$) contribute to the model. Model summaries for the regression analyses, including the KL subgroup variable, are shown in Table IV(a) and (b). Results from all of the repeated regression analyses were consistent with the above results. For vexample, the total amount of variance in the external knee adduction moment explained by the

Table III

Pearson product moment correlations (95% confidence intervals) among knee adduction moment, mechanical axis angle, average walking speed, and values for toe-out angle and lateral trunk lean angle occurring at the time of the knee adduction moment

	Adduction moment	Mechanical axis angle	Speed	Toe-out	Trunk lean
(a) First peak knee adduction moment (n = 113)					
Mechanical axis angle	0.51* (0.36, 0.63)				
Speed	0.05 (−0.14, 0.23)	−0.17 (−0.34, 0.02)			
Toe-out	−0.31* (−0.47, −0.13)	0.11 (−0.08, 0.29)	−0.12 (−0.30, 0.07)		
Trunk lean	−0.39* (−0.54, −0.22)	0.08 (−0.11, 0.26)	−0.12 (−0.30, 0.07)	0.21* (0.03, 0.38)	
WOMAC pain	0.07 (−0.12, 0.25)	−0.09 (−0.27, 0.10)	0.23* (0.05, 0.40)	−0.11 (−0.29, 0.08)	−0.18* (−0.35, −0.00)
(b) Second peak knee adduction moment (n = 108)					
Mechanical axis angle	0.61* (0.48, 0.72)				
Speed	−0.07 (−0.26, 0.12)	−0.18 (−0.36, 0.01)			
Toe-out	−0.26* (−0.43, −0.07)	0.13 (−0.06, 0.31)	−0.11 (−0.29, 0.08)		
Trunk lean	−0.33* (−0.49, −0.15)	0.01 (−0.18, 0.20)	−0.24* (−0.41, −0.05)	0.12 (−0.07, 0.30)	
WOMAC pain	0.11 (−0.08, 0.29)	−0.10 (−0.28, 0.09)	0.19* (0.00, 0.37)	−0.12 (−0.27, 0.07)	−0.21* (−0.38, −0.02)

*Denotes significantly different from zero at $P < 0.05$.

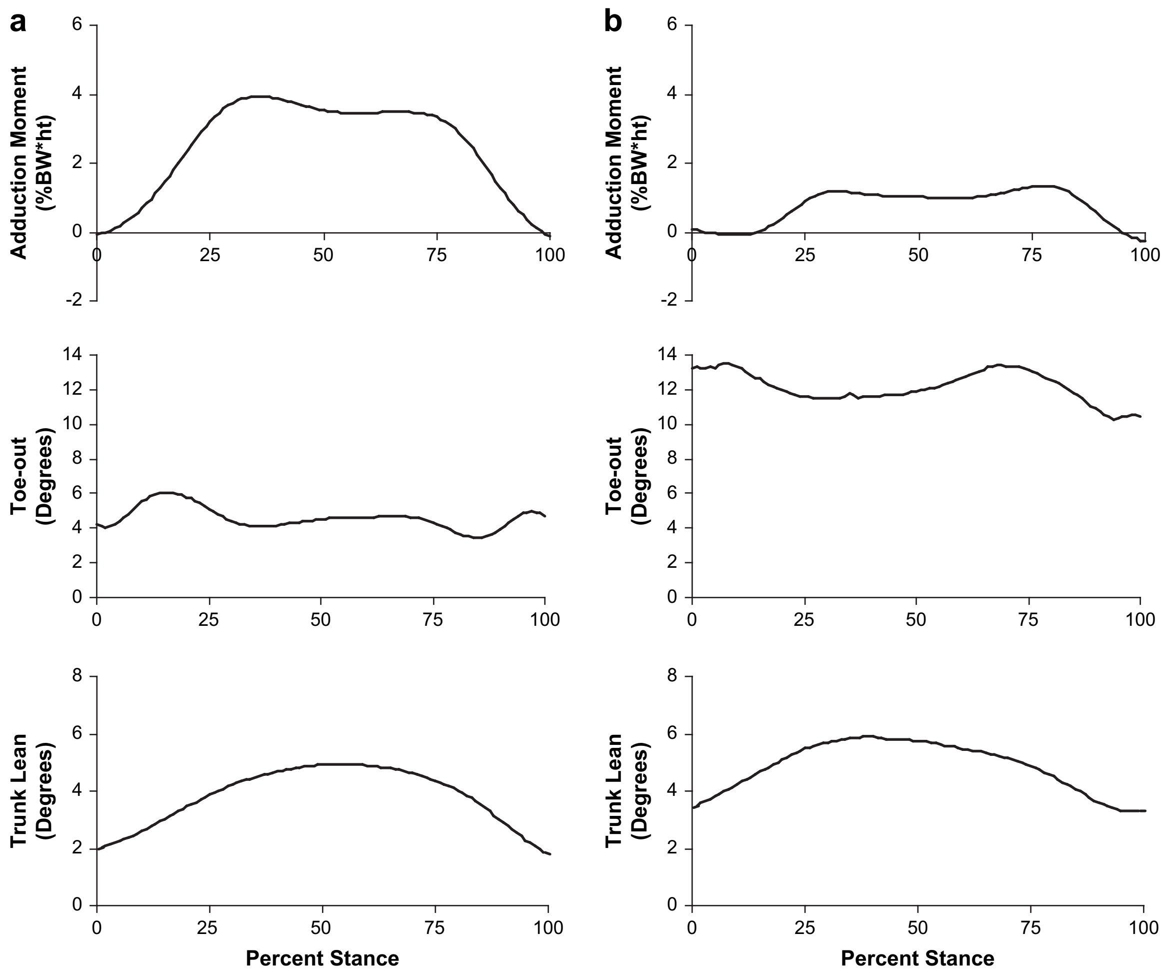

Fig. 4. Representative data from two female subjects indicating the knee adduction moment (top) toe-out angle (middle), lateral trunk lean angle (bottom), and from a single trial. Subject (a) was 50 years old, had a BMI of 26.8, a mechanical axis angle of −4.5°, a KL grade of 3, a WOMAC pain score of 45/100, and walked at 0.83 m/s for this trial. Subject (b) was 44 years old, had a BMI of 24.0, a mechanical axis angle of −4.2°, a KL grade of 3, a WOMAC pain score of 45/100, and walked at 0.90 m/s for this trial.

predictor variables ranged from approximately 50 to 60% and the independent contributions from toe-out angle and lateral trunk lean ranged from 10 to 20%.

Discussion

The present findings illustrate that patients with medial compartment knee OA walk with varying amounts of lower limb rotation and trunk lean that explain substantial variance in knee joint loading. While few patients exhibited small amounts of toe-in and lateral trunk lean towards the swing limb, the vast majority of patients walked with substantial toe-out and lateral trunk lean towards the stance limb (Figs. 3 and 5). Although authors have described the relationships among several individual characteristics of gait and dynamic knee joint loading[4,16,25,27,28,30], we are unaware of previous studies that have quantified the multivariate associations among speed, toe-out, and lateral trunk lean. The substantial amount of variance in the knee adduction moment explained by lateral trunk lean has not been previously reported and may be partially responsible for the wide range of results of previous studies evaluating the relationships and changes in dynamic knee joint load.

Authors have previously reported the negative correlation between toe-out angle and external knee adduction moment[16,25,30,51], and Chang *et al.*[31] have recently demonstrated a role for toe-out gait in disease progression. The magnitude of the relationship between toe-out angle and the knee adduction moment observed in the present study was similar to that of previous studies[25,30]. The magnitude of lateral trunk lean, however, had the highest correlation with the first and second peak knee adduction moments among the kinematic variables investigated in the present study. While controlling for disease severity, mechanical axis angle, pain and gait speed, toe-out explained 12% of the variance in the first peak knee adduction moment [Table IV(a)]. While controlling for these same variables, (i.e., toe-out excluded) trunk lean explained 18% of the variance (not reported in Table IV). Importantly, even when controlling for these same variables plus toe-out, lateral trunk lean still explained 13% of the variance in knee adduction moment [Table IV(a)]. These findings emphasize that lateral trunk lean should be considered in future gait studies of patients with knee OA, including the potential for trunk lean to affect disease progression.

Although not directly quantified, lateral trunk lean has been previously proposed as a mechanism to reduce joint

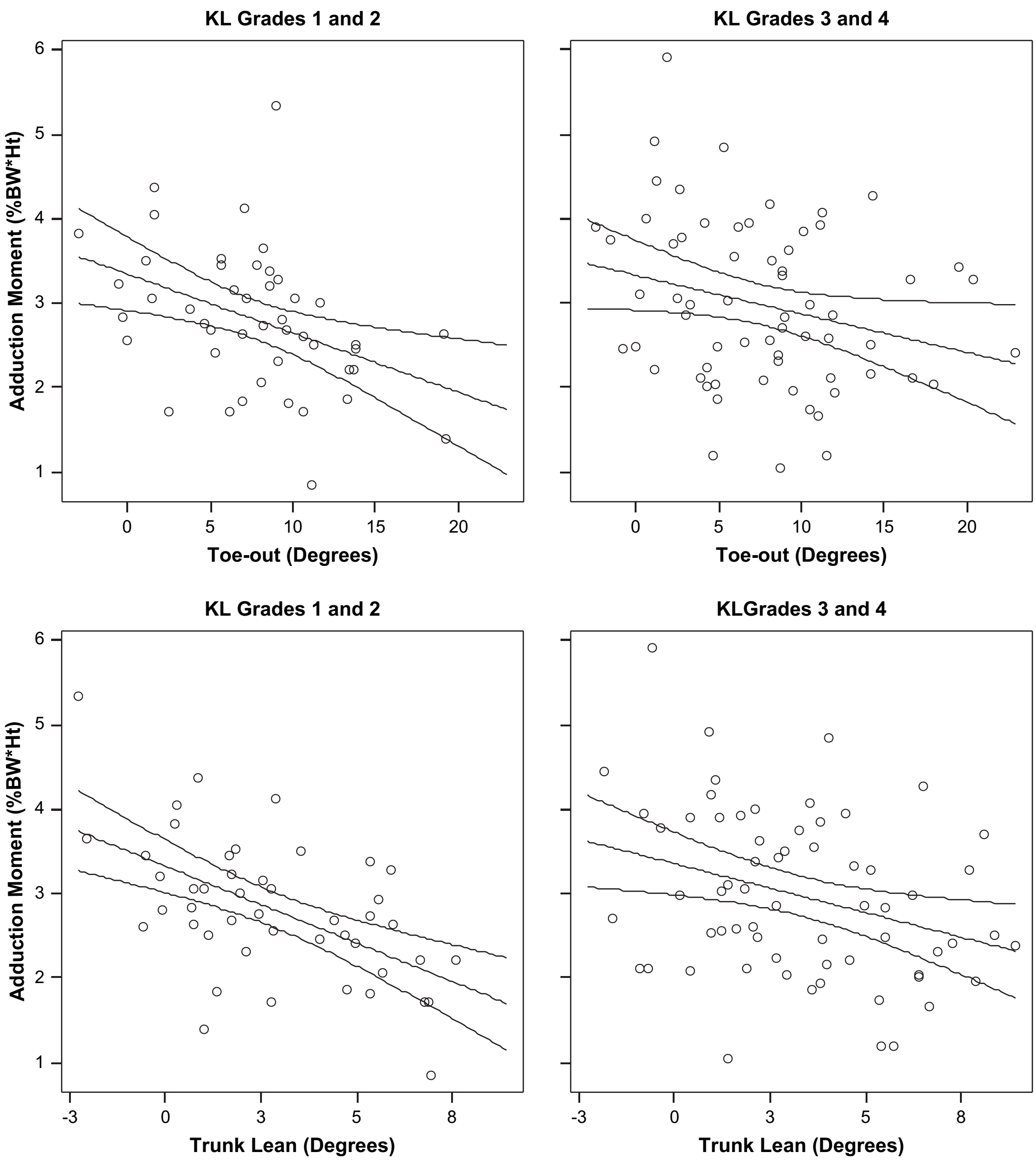

Fig. 5. Scatterplots illustrating the relationships between the first peak external knee adduction moment and toe-out angle and lateral trunk lean angle for patients with KL grades 1 or 2 and KL grades 3 or 4. Linear regression with 95% mean prediction intervals is shown.

loading in patients with knee OA[2,28,33,35]. Mundermann *et al.*[28] investigated lower limb gait biomechanics in 42 patients with bilateral knee OA and found a rapid increase in GRF magnitude combined with increased external hip abduction moments following heel strike. These authors stated that this mechanism was suggestive of a shift of the body's weight towards the stance limb, which is consistent with the present findings regarding trunk lean. In addition, our pelvic obliquity data suggest that patients in the present sample derived trunk lean primarily from movement of the upper body and not as a result of pelvic contributions. As hip kinetics have been implicated in the progression in knee OA[52], further research on the potential role of hip biomechanics on knee joint loading and OA should also consider the effect of trunk lean.

Previous authors have further postulated that alterations in gait kinematics are secondary, compensatory changes adopted by some patients to lessen the load on the knee after the onset of painful OA[16,25,27,28,30,53], and there is some evidence from cross-sectional studies comparing individuals with and without OA to support this theory[28,30]. Although the significant correlation is low (Table III), the observation that patients with greater pain (i.e., lower WOMAC score) exhibited greater trunk lean is also consistent with this theory. In the present study, however, the relationship between the knee adduction moment and toe-out or lateral trunk lean was not significantly modified by KL grade (Fig. 5). This finding highlights both the potential difficulties in measuring disease progression and the limitations in making inferences about secondary, compensatory changes in gait from cross-sectional study designs. Chang *et al.*[31] recently reported results from a longitudinal study that suggest patients with knee OA are more likely to experience tibiofemoral joint space narrowing if they walk with smaller amounts of toe-out. Clearly,

Table IV
Summary of regression models for predicting the knee adduction moment

Model	R	R^2	Adjusted R^2	R^2 change	P
(a) First peak knee adduction moment (n = 113)					
KL grade	0.09	0.01	−0.00	0.01	0.35
KL grade + MAA	0.51	0.26	0.24	0.25	<0.01
KL grade + MAA + pain	0.52	0.27	0.25	0.01	0.18
KL grade + MAA + pain + speed	0.53	0.28	0.26	0.01	0.17
KL grade + MAA + pain + speed + toe-out	0.64	0.40	0.38	0.12	<0.01
KL grade + MAA + pain + speed + toe-out + lean	0.73	0.53	0.50	0.13	<0.01
(b) Second peak knee adduction moment (n = 108)					
KL grade	0.05	0.00	−0.00	0.00	0.61
KL grade + MAA	0.62	0.39	0.37	0.38	<0.01
KL grade + MAA + pain	0.64	0.41	0.39	0.02	0.05
KL grade + MAA + pain + speed	0.64	0.41	0.39	0.00	0.99
KL grade + MAA + pain + speed + toe-out	0.72	0.52	0.50	0.11	<0.01
KL grade + MAA + pain + speed + toe-out + lean	0.77	0.59	0.57	0.07	<0.01

MAA = mechanical axis angle.

more longitudinal research is needed to better understand the potential combined role of various gait kinematics to the pathophysiology of knee OA.

Although walking speed has been previously reported to be correlated to the peak external knee adduction moment[27], it did not explain a significant proportion of variance in the knee adduction moment in the present sample. However, the ability of an independent variable to explain the variability in the dependent variable is minimal unless it too demonstrates appreciable variability. In our study the majority of patients walked with a similar speed limited this variable's ability to predict outcome. It should also be noted that even the largest contributor to the explained variance in external knee adduction moment, the mechanical axis angle, only contributed approximately 25−35%. Even with the addition of gait kinematics investigated in the present study, the total amount of explained variance was only approximately 50−60%. Further research is needed to identify other potential contributors to knee joint loading.

Results from the present study show that lateral trunk lean towards the stance limb explains a substantial and independent portion of variance in the external knee adduction moment during gait. These findings suggest that lateral trunk lean should be considered in future gait studies evaluating risk factors and interventions for patients with knee OA.

Conflict of interest

No authors had any financial or personal relationships with other people or organizations that could inappropriately influence this work.

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