



# A biomechanical analysis of trunk and pelvis motion during gait in subjects with knee osteoarthritis compared to control subjects

Heather S. Linley<sup>a,\*</sup>, Elizabeth A. Sled<sup>b</sup>, Elsie G. Culham<sup>b</sup>, Kevin J. Deluzio<sup>a</sup>

<sup>a</sup> Department of Mechanical and Materials Engineering, Queen's University, Kingston, Ontario, Canada

<sup>b</sup> School of Rehabilitation Therapy, Queen's University, Kingston, Ontario, Canada

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## ABSTRACT

**Background:** Trunk lean over the stance limb during gait has been linked to a reduction in the knee adduction moment, which is associated with joint loading. We examined differences in knee adduction moments and frontal plane trunk lean during gait between subjects with knee osteoarthritis and a control group of healthy adults.

**Methods:** Gait analysis was performed on 80 subjects (40 osteoarthritis). To define lateral trunk lean two definitions were used. The line connecting the midpoint between two reference points on the pelvis and the midpoint between the acromion processes was projected onto the lab frontal plane and the pelvis frontal plane. Pelvic tilt was also measured in the frontal plane as the angle between the pelvic and lab coordinate systems. Angles were calculated across the stance phase of gait. We analyzed the data, (i) by extracting discrete parameters (mean and peak) waveform values, and (ii) using principal component analysis to extract shape and magnitude differences between the waveforms.

**Findings:** Osteoarthritis subjects had a higher knee adduction moment than the control group ( $\alpha=0.05$ ). Although the discrete parameters for trunk lean did not show differences between groups, principal component analysis did detect characteristic waveform differences between the control and osteoarthritis groups.

**Interpretation:** A thorough biomechanical analysis revealed small differences in the pattern of motion of the pelvis and the trunk between subjects with knee osteoarthritis and control subjects; however these differences were only detectable using principal component analysis.

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## 1. Introduction

The adduction moment at the knee has received significant attention in the study of the pathomechanics of knee osteoarthritis (OA) (Baliunas et al., 2002; Hurwitz et al., 2002). It has been shown to be a good surrogate for evaluating medial compartment loading at the knee joint (Baliunas et al., 2002; Hurwitz et al., 2002, 1998) and is related to the severity and progression of OA (Astefan et al., 2008; Miyazaki et al., 2002).

Gait alterations have been linked with changes in joint loading in the lower limb (Mundermann et al., 2005). By increasing trunk lean in the frontal plane towards the stance limb by about 10 degrees, healthy subjects were able to reduce their first peak knee adduction moments an average of 65% (Mundermann et al., 2008). This finding led to an interest in studying trunk lean in subjects with knee OA as a potential mechanism to reduce the knee adduction moment. Hunt et al. (2008) evaluated knee adduction moments and lateral trunk lean during the

stance phase of gait in participants with medial knee OA. The authors found negative correlations between the knee adduction moment and lateral trunk lean angle at both the first and second adduction moment peaks, however the study lacked a control subject group for comparison. These two studies highlight the necessity for further evaluation of trunk and pelvic motions, including an assessment of analysis strategies. Additionally, while these studies provide evidence of a link between lateral trunk lean and a reduction in the knee adduction moment, there remains a need to compare trunk lean between populations of control and OA subjects. One study assessed trunk lean angles during gait between three populations: unilateral knee OA, bilateral knee OA, and healthy subjects (Tanaka et al., 2008). Trunk lean was measured as a single value at the point of maximum lean over the stance or swing limb, and trunk motion was not separated from pelvis motion. The authors found no differences in trunk lean angle between subject groups, however it is still unclear whether the patterns of motion are different between control and OA populations.

The biomechanical relationship of frontal plane pelvic and thoracic tilt and the hip adduction moment is unclear. Pelvic control in the frontal plane has been attributed to the strength of the hip abductor muscles (MacKinnon and Winter, 1993). In a longitudinal study,

\* Corresponding author.

E-mail address: [linley@me.queensu.ca](mailto:linley@me.queensu.ca) (H.S. Linley).

Chang et al. (2005) evaluated internal hip abduction moments and knee OA progression in subjects with mild medial knee OA. Results of the study showed that a greater internal hip abduction moment at baseline lowered the odds of progression of knee OA. The authors speculated from these results that weakness of the hip abductor muscles causes a drop in the pelvis on the swing side and shifts the centre of mass farther over the swing limb, thereby increasing medial compartment loading in the knee, however they did not measure hip muscle strength or pelvic motion (Chang et al., 2005). Another study compared knee and hip adduction moments between the affected and contralateral limbs in a population of unilateral knee OA subjects (Briem and Snyder-Mackler, 2009). The authors found that the first peak hip adduction moment on the involved side was significantly smaller than that of the contralateral side, but found no significant interlimb differences in peak knee adduction moment. The authors suggested that an explanation for this low hip adduction moment is a lateral trunk lean; however trunk motion was not measured. The findings of Mundermann et al. (2005) add to these previous hypotheses by evaluating the first and second peak external hip adduction moments in subjects with severe and less severe knee OA, each with their own matched control group. The authors found that patients with severe knee OA had significantly lower first and second peak external hip adduction moments compared to matched control subjects, while subjects with less severe knee OA had hip adduction moments throughout stance that were similar to their matched controls. The authors suggested that subjects with less severe knee OA have sufficient hip abductor muscle strength to maintain an altered position of the trunk laterally over the support limb, while those with more severe knee OA lack this muscle strength and drop the pelvis on the contralateral side. They hypothesized that a drop of the pelvis would lead to a lean of the trunk away from the support limb, resulting in a higher first peak knee adduction moment. It should be recognized that neither motion of the pelvis and trunk, nor hip muscle strength was measured in any of these studies.

The purpose of this study was to provide a thorough biomechanical analysis of frontal plane trunk and pelvis motion in control and OA subjects. We hypothesized that subjects with knee OA alter their gait as a potential mechanism to reduce the joint loading at the knee by leaning their trunks farther over their stance limb compared to controls. This alteration was thought to be a combination of both pelvic and thoracic tilt. This study will contribute to the development of a thorough understanding of the biomechanics of lateral trunk lean in knee OA patients.

## 2. Methods

### 2.1. Participants

Participants were recruited locally through advertisements and by recommendation of local orthopaedic surgeons. Subjects included those with medial compartment knee OA and a control group of healthy subjects. The study was approved by the University Health Sciences Research Ethics Board and each subject provided informed consent prior to participating. Inclusion criteria for the OA group were age ( $\geq 40$  years), diagnosis of knee OA by a physician, subject-reported pain in the knee joint during most days of the month, and one of either radiographic evidence of medial compartment OA or evidence of medial compartment cartilage loss verified by arthroscopy or magnetic resonance imaging. Participants with more lateral compartment OA than medial compartment OA were excluded from the study. If subjects were found to have bilateral OA, the most affected limb was used as the test limb. Participant exclusion criteria included intra-articular corticosteroid or visco-supplementation injection into either knee within a previous 3 month period, significant co-morbidities including heart disease, stroke, and active cancer treatment, OA caused by other arthritis types, a history of avascular necrosis, previous peri-articular fractures of the knee joint, Paget's disease, villonodular synovitis, joint infection, neuropathic

arthropathy, acromegaly, Wilson's disease, hemochromatosis, gout or recurrent pseudogout, and osteopetrosis (Dieppe et al., 1995).

Control subjects were healthy adults with a negative clinical diagnosis of knee OA, hip OA, and rheumatoid arthritis, and no reports of knee or hip pain or knee trauma based on the same criteria used for the OA subjects. All control subjects were screened radiographically to verify a negative diagnosis of knee OA. Subjects from the OA group were matched with subjects from the control group based on gender and age primarily, followed by an agreement between subject pairs in height. The test leg for each control subject was assigned to correspond with the test leg of the matched OA subject.

Radiographic OA disease severity was graded using the Kellgren and Lawrence (KL) scale while pain, stiffness, and physical function in both the OA and control groups were assessed using the Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC) (Bellamy et al., 1988).

### 2.2. Gait analysis

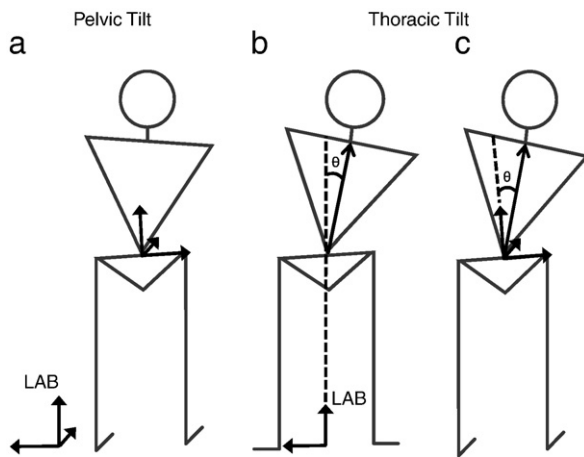
Gait analysis was performed on all subjects. Two Optotrak motion capture cameras (Optotrak 3020, Northern Digital Inc., Waterloo, ON, Canada) were used to track participants as they walked along an 8 m walkway. Three dimensional kinematic data were recorded at a sample rate of 100 Hz. Marker clusters containing infrared light emitting diodes (IREDs) were placed on the dorsum of the foot over the metatarsals, lateral shank, lateral thigh, sacrum, and over the spinous processes of the 7th cervical/1st thoracic vertebrae and were secured with Velcro straps to prevent movement during gait trials. Two AMTI (Advanced Mechanical Technology Inc., Watertown, MA, USA) force platforms embedded in the walkway recorded the subject's ground reaction forces and moments at a sampling rate of 200 Hz. Participants provided and wore their own comfortable walking shoes during testing. Subjects were instructed to walk at a self-selected speed along the walkway. Five trials were obtained for each subject in which the foot of the test limb successfully landed on the force platform, and all markers were visible by the cameras.

The locations of bony landmarks in relation to marker clusters were recorded using a digitizing pointed probe fitted with a cluster of markers. The landmarks were found by palpations performed by a physiotherapist. Landmarks included the first metatarsal heads, the fifth metatarsal heads, the medial and lateral malleoli, the medial and lateral epicondyles of the knee, the greater trochanters, points on the pelvis directly vertical to the greater trochanters at the level of the mid-iliac crest bilaterally, and the acromion processes of the scapulae.

An inverse dynamics approach (Visual 3D, C-Motion, Germantown, MD, USA) was used to calculate frontal plane moments for the knee and hip joints. Segment coordinate systems were defined for both the pelvis and the thorax. Joint moments and pelvic and thoracic tilt angles in the frontal plane were calculated across the stance phase of gait for both subject groups. Individual subject trials were event- and time-normalized to 100% of stance and then averaged at each percent of stance to provide one mean curve per subject. For each of the two subject groups, the mean subject curves were then ensemble averaged to provide a mean curve for each group across the stance phase of gait (MATLAB, The Math Works Inc., Natick, MA, USA).

### 2.3. Pelvic and thoracic tilt definitions

Fig. 1 shows a visual representation of the definition used to describe pelvic tilt and two definitions of thoracic tilt. The angular motion of the pelvis was defined using Cardan angles as previously suggested by Kadaba et al. (1990). Lateral pelvic tilt was calculated as the frontal plane rotation of the pelvis coordinate system in reference to the lab coordinate system. The Cardan angle sequence of sagittal pelvic rotation—lateral pelvic tilt—transverse pelvic tilt was used as recommended by Baker (2001). The angle of thoracic tilt was calculated using



**Fig. 1.** Pelvic and thoracic tilt definitions. (a) Pelvic tilt defined as the lateral motion of the pelvis coordinate system with respect to the lab coordinate system, (b) two dimensional planar rotation of the thorax as a projection onto the frontal plane of the lab coordinate system from the vertical axis (thoracic tilt-lab), (c) two dimensional planar rotation of the thorax as a projection onto the frontal plane of the pelvis coordinate system from the vertical axis (thoracic tilt-pelvis).

a two dimensional planar projection similar to previous studies (Henriksen et al., 2009; Hunt et al., 2008; Tanaka et al., 2008). Thoracic tilt was defined in two ways (Fig. 1b, c): (i) the orientation of the thorax was measured with respect to the coordinate system of the lab (thoracic tilt-lab) and also (ii) with respect to the coordinate system in the pelvis (thoracic tilt-pelvis). While thoracic tilt-lab is more comparable to other studies, thoracic tilt-pelvis allows the analysis of the contributions of motion from both the thorax and the pelvis giving a more thorough understanding of the mechanics of the trunk. The thorax was defined by a line connecting the midpoints between the pelvis landmarks and the acromion processes. The angle of thoracic tilt was first calculated by projecting the distal–proximal line up the thorax onto the frontal plane of the lab coordinate system and tracking it relative to the lab vertical axis. A second definition of thorax tilt used the same distal–proximal thorax line created from the landmarked points, but this time it was monitored throughout stance as a projection onto the frontal plane of the pelvis coordinate system. All angles and moments are reported during the stance phase of gait for the affected limb. Pelvic and thoracic tilt angles are reported as positive for a tilt over the stance limb.

#### 2.4. Statistical analysis

Four discrete parameters were extracted from the knee and hip adduction waveforms as well as the pelvic and thoracic tilt waveforms:

- 1) Means were calculated across the mid portion of stance (20–80% stance).
- 2) Peak values were recorded for both groups in the first phase of stance.
- 3) Peak values were recorded for both groups in the second phase of stance.
- 4) Midstance values (50% stance) for both groups were compared.

Student's *t*-tests were performed to detect differences between the OA and control groups ( $\alpha = 0.05$ ).

#### 2.5. Principal component analysis

In order to detect waveform shape and magnitude differences between subject groups, principal component analysis (PCA) was performed on pelvic tilt, and both definitions of thoracic tilt separately (Deluzio et al., 1997). Each measure began with a matrix of 80 participants by 101 samples representing 0–100% of the stance phase of gait. These data are transformed into a covariance matrix from which the first three

principal components (PCs) are extracted. These PCs represent the majority of the variation in the data. Z-scores were calculated for each individual subject by multiplying the mean removed original data by each separate loading vector. Z-scores were analyzed for significant differences between groups using Student's *t*-tests ( $\alpha = 0.05$ ). Representative subjects from each group with Z-scores in the 5th and 95th percentiles were extracted and labelled either "low PC" (5th percentile), or "high PC" (95th percentile). These representative subjects, one from the control group and one from the OA group, were used to analyze motion pattern differences between groups which are represented as waveform shape differences.

To illustrate features captured by PCs which are found to have significantly different Z-scores between groups, the original waveform data are reconstructed using the loading vectors from that PC alone. This is accomplished by multiplying the Z-scores of each subject by the loading vector of the desired PC.

### 3. Results

#### 3.1. Participants

Eighty subjects participated in the study, split evenly between the control and OA groups (23 females in each group). Thirty-three of the 40 OA subjects had bilateral medial compartment knee OA. KL grades had a median value of 2 for the OA subjects and 0 for the control subjects. Significant differences were found between the two groups in weight, body mass index (BMI), gait speed, cadence, and double limb support time ( $p \leq 0.05$ ), however no differences were found in age, height, or stride length ( $p > 0.05$ ). A summary of these results is found in Table 1.

The WOMAC scores for pain, stiffness, and physical function are graded on scales from 0 to 20, 0 to 8, and 0 to 68 respectively. The OA group was found to have a pain score of 5.55 (2.87), a stiffness score of 3.08 (1.80), and a physical function score of 19.60 (11.77) reported as mean (standard deviation).

#### 3.2. Gait analysis

Hip and knee adduction moments, as well as pelvic tilt and thoracic tilt-pelvis displayed double peak curves (Fig. 2). However, no discernable second peak could be identified for thoracic tilt-lab.

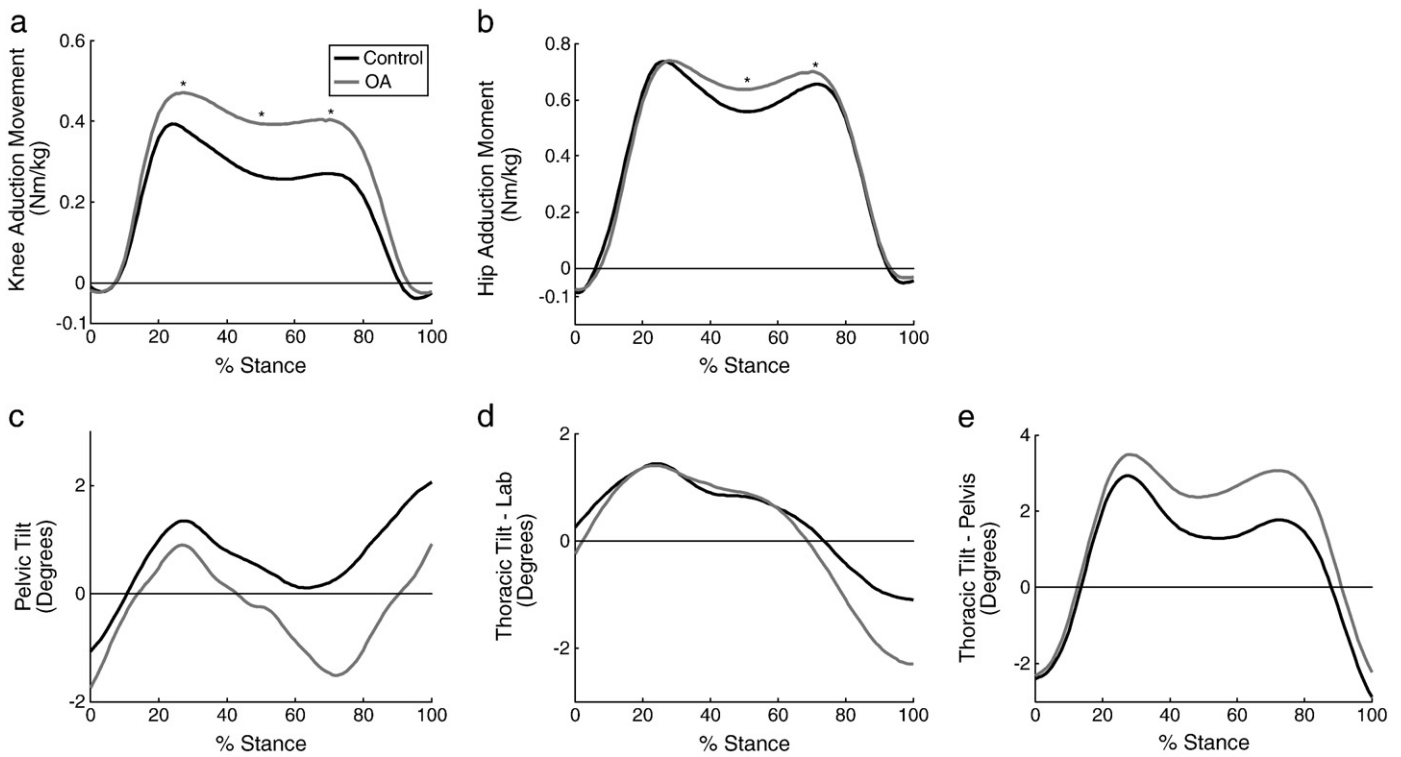
Knee and hip adduction moments were found to be significantly higher for the OA group across the mid portion of stance (20–80% stance), at the midstance (50% stance) point, and in the second peak. The first peak knee adduction moment was also significantly higher for the OA group (Table 2). Most discrete measures of pelvic and thoracic tilt showed the OA group to have greater peak and mean

**Table 1**  
Subject demographics and gait speed.

Variable	OA group (n = 40) mean (SD)	Control group (n = 40) mean (SD)	P-value
Gender	23 Female	23 Female	
Age	63 (10)	64 (9)	0.59
Height	1.73 (0.11)	1.70 (0.09)	0.23
Mass (kg)*	82.3 (20.0)	69.7 (11.0)	0.001
BMI*	27.4 (5.5)	24.0 (3.2)	0.001
Gait speed (m/s)*	1.00 (0.2)	1.12 (0.19)	0.006
Stride length (m)	1.19 (0.16)	1.26 (0.19)	0.08
Radiographic KL score**			
0	0	31	
1	4	7	
2	20	2	
3	9	0	
4	7	0	

\*Statistically significant difference between groups are noted for  $\alpha = 0.05$ .

\*\*Radiographic Kellgren–Lawrence score (grade 0–4) reported as a frequency. BMI = body mass index.



**Fig. 2.** Group mean curves for OA and control groups across the stance phase of gait for the affected limb as indicated for (a) the external knee adduction moment (b) the external hip adduction moment (c) pelvic tilt (d) thoracic tilt-lab (e) thoracic tilt-pelvis. Positive values in (a) and (b) indicate adduction moments. Positive values in (c), (d) and (e) indicate a tilt over the stance limb. Significant differences in discrete parameters (peaks and means) are indicated by \* ( $p < 0.05$ ).

values than the control group, however these differences were not statistically significant.

### 3.3. Principal component analysis

PCA detects shape and magnitude differences in the waveform data. Since each PC is independent of the others, we can examine them

**Table 2**  
Discrete parameter values and comparisons between the two subject groups.

Group	Means (SD)		P-value
	Control	Osteoarthritis	
<b>Knee adduction moment (Nm/kg)</b>			
20–80% Stance*	0.30 (0.10)	0.41 (0.14)	$p < 0.001$
First peak*	0.41 (0.11)	0.52 (0.15)	$p < 0.001$
Midstance (50%)*	0.26 (0.10)	0.39 (0.15)	$p < 0.001$
Second peak*	0.29 (0.11)	0.43 (0.15)	$p < 0.001$
<b>Hip adduction moment (Nm/kg)</b>			
20–80% Stance*	0.63 (0.08)	0.67 (0.10)	$p = 0.03$
First peak	0.77 (0.11)	0.79 (0.12)	$p = 0.37$
Midstance (50%)*	0.56 (0.10)	0.64 (0.11)	$p = 0.002$
Second peak*	0.67 (0.10)	0.72 (0.12)	$p = 0.02$
<b>Pelvic tilt (degrees)</b>			
20–80% Stance	0.6 (3.9)	−0.3 (3.7)	$p = 0.26$
First peak	2.6 (4.0)	1.9 (4.1)	$p = 0.46$
Midstance (50%)	0.5 (4.0)	−0.2 (3.9)	$p = 0.41$
Second peak	−0.4 (3.8)	−1.9 (3.9)	$p = 0.09$
<b>Thoracic tilt-lab (degrees)</b>			
20–80% Stance	0.7 (1.7)	0.7 (2.0)	$p = 0.84$
First Peak	1.9 (1.6)	1.9 (2.0)	$p = 0.92$
Midstance (50%)	0.8 (1.8)	0.9 (2.2)	$p = 0.88$
<b>Thoracic tilt-pelvis (degrees)</b>			
20–80% Stance	1.8 (5.4)	2.8 (3.3)	$p = 0.32$
First peak	3.2 (5.5)	4.1 (3.4)	$p = 0.42$
Midstance (50%)	1.3 (5.3)	2.4 (3.3)	$p = 0.29$
Second peak	1.9 (5.3)	3.3 (3.5)	$p = 0.18$

\*Statistically significant difference between groups are noted for  $\alpha = 0.05$ .

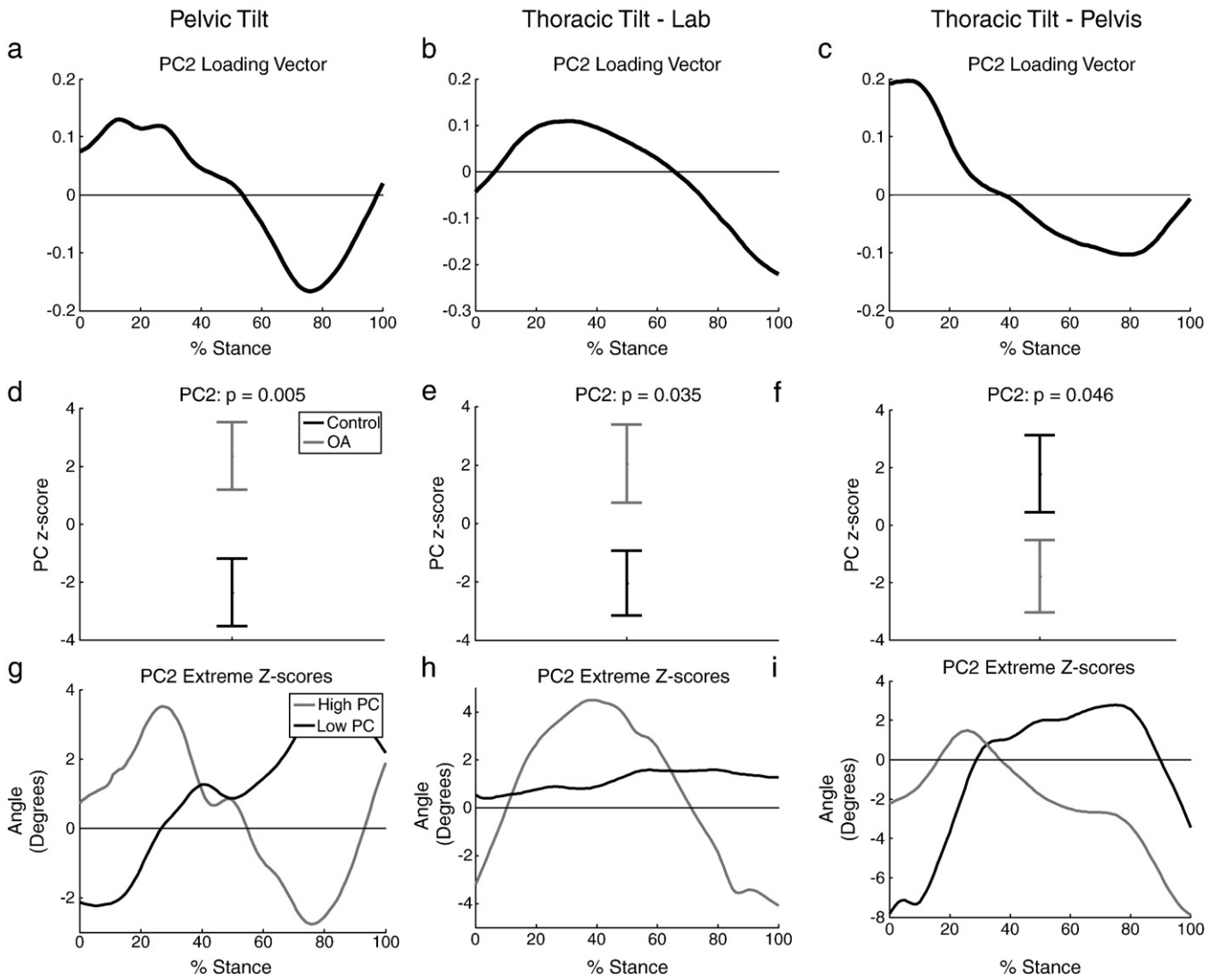
individually for differences between groups. In the model of pelvic tilt and both models of thoracic tilt over 97% of the total variation in the data was captured in the first three PCs.

The first PC captured the largest source of variation in the data. The loading vectors for PC1 had positive coefficients throughout the entire stance cycle for all models of pelvic and thoracic tilt. As seen in previous studies, this describes an overall magnitude feature in the data (Deluzio and Astephen, 2007). Therefore, PC1 captured a measure of the average angle throughout stance. This angle is a measure of an individual subject's neutral posture. The magnitude variation in the data is related to measuring absolute segment angles, which vary from person to person. Therefore, the largest source of variation in the waveform data is related to differences in neutral posture between subjects and is unrelated to the actual motion pattern or shape of the waveforms. In all three models, there was no difference in PC1 between the control and OA groups. Due to the independence of PCs, PC1 provides a unique and objective means of removing the variation due to neutral position.

The second PC described a difference in waveform shape and revealed statistically significant differences between the control and OA groups in pelvic tilt and both definitions of thoracic tilt (Fig. 3). The loading vectors for all models displayed positive values in early stance and negative values in late stance. This captures a range of motion in pelvic and thoracic tilt. Comparing the subjects with high and low Z-scores showed that a high PC2 score corresponds to an increased difference in motion between the early and late stages of stance. The OA subjects demonstrated an increased range of motion during stance for both pelvic tilt and thoracic tilt-lab ( $p < 0.05$ ). The control subjects had an increased range of motion during stance for thoracic tilt-pelvis ( $p < 0.05$ ).

The third PC also describes the waveform shape but does not show significant differences between the control and OA groups in any of the three models. PC3 captures the waveform's positive and negative peaks.

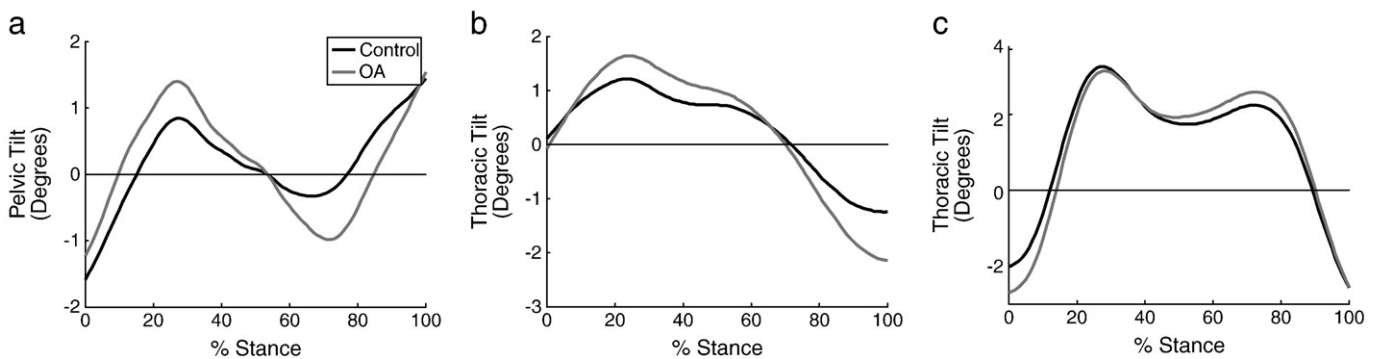
Another way of illustrating the features captured by the second PC is by reconstructing the original waveforms. The visual representation



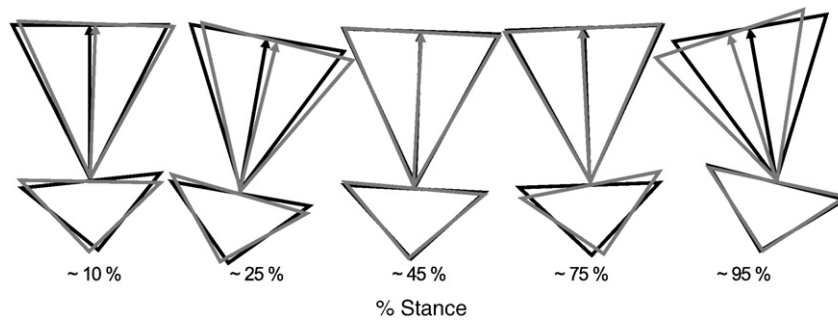
**Fig. 3.** PCA loading vectors for (a) pelvic tilt, (b) thoracic tilt-lab, (c) thoracic tilt-pelvis. PCA interaction plots for (d) pelvic tilt, (e) thoracic tilt-lab, (f) thoracic tilt-pelvis. PCA high and low Z-scores for (g) pelvic tilt, (h) thoracic tilt-lab, (i) thoracic tilt-pelvis.

of pelvic and thoracic tilt is based entirely on the reconstructed data having removed the bias due to the variation in neutral position between subjects (Fig. 4). Fig. 5 shows that subjects with OA began the stance phase with their pelvis raised slightly less on the stance side compared to control subjects. The pelvis then dropped on the stance side in both groups and the thorax tilted over the stance

limb. At approximately 75% of the stance phase both groups dropped their pelvis on the swing side; however, this was more pronounced in the OA group. Near the end of stance the pelvis was raised on the swing side for both groups, and the thorax was tilted over the swing side, but this thoracic tilt was larger for the OA subjects.



**Fig. 4.** Reconstructed data using PC2 for (a) pelvic tilt, (b) thoracic tilt-lab, and (c) thoracic tilt-pelvis. A positive angle reflects a tilt over the stance limb.



**Fig. 5.** Visual representation of the variation in motion of the thorax and the pelvis throughout the stance cycle in the control (black) and OA (grey) subjects. Each visualization represents the rear view with the stance limb on the right, and the swing limb on the left. Motions are exaggerated beyond actual angle values for effective visualization.

#### 4. Discussion

A biomechanical analysis of trunk lean and its relationship with knee mechanics was conducted. Knee and hip adduction moments were significantly higher in the OA group compared to controls, while discrete measures found no differences between groups in pelvic or thoracic tilt. However, PCA detected significant group differences in range of motion in pelvic and thoracic tilt. Small differences in motion pattern exist in pelvic and thoracic tilt between subjects with knee OA and control subjects. The hypothesis that subjects with knee OA alter their gait by leaning their trunks farther over their stance limb compared to controls is not explicitly supported by these results. Instead, we found that the pattern of motion between the two groups varies slightly during gait.

The knee adduction moment of OA subjects was found to be higher than control subjects throughout the stance phase of gait as measured by all discrete parameters. These observations are consistent with previous studies (Astefhen et al., 2008; Mundermann et al., 2005). It has been reported that the overall magnitude of the adduction moment captured by PC1 is not affected by speed differences between groups (Landry et al., 2007).

The external hip adduction moment was found to be higher throughout stance phase in all discrete parameters except the first peak. This agrees with Huang et al. (2008) who found that subjects with severe OA displayed significantly higher mid-stance and second peak internal hip adduction moments compared to controls. Alternatively, two studies have reported lower external hip adduction moments in OA subject groups compared to controls (Astefhen et al., 2008; Mundermann et al., 2005). Astefhen et al. (2008) reported a significant difference in the first peak hip adduction moment between the control group and both the moderate and the severe OA group, with the controls having a higher peak than both OA groups ( $p < 0.002$ ). Similarly, Mundermann et al. (2005) reported that severe OA subjects had significantly lower hip adduction moments in the first and second peaks of stance compared to the control groups. Interestingly, although their less severe OA subjects did show slightly higher hip adduction moments for both peaks during stance compared to controls, the differences were not statistically significant. Some theories have been put forth to explain the results of these data including speculation about hip abductor muscle strength (Chang et al., 2005; Mundermann et al., 2005). However, none of the reviewed studies have collected hip abductor muscle strength data in order to justify these suggestions. Yamada et al. (2001) found no difference in isometric hip abductor strength between OA and control groups.

The equivocal results of these studies and lack of explanatory data suggest that other factors may affect the hip adduction moment throughout stance such as gait speed, BMI, and disease severity. In fact, a recent study found that gait velocity and subject mass were the main contributors to the frontal plane hip moment, while hip abductor strength and gluteus medius muscle activation explained only a small amount of variability in the hip adduction moment waveform during midstance (Rutherford and Hubble-Kozey, 2009).

Definitions of pelvic tilt vary significantly between studies. It has been suggested that an optimal sequence of rotations for the pelvis is axial rotation, lateral tilt, and transverse tilt in order to correspond to conventionally defined anatomical terms (Baker, 2001). Baker demonstrated that measures of lateral pelvic tilt correlate better to relative hip height using the proposed sequence rather than the conventional transverse tilt, lateral tilt, and axial rotation sequence. We chose this proposed sequence in order to best interpret our results clinically. Our lateral pelvic tilt data show a progressive increase in tilt over the stance limb reaching the first peak at approximately 25% of stance phase. The second peak shows a tilt over the swing side which occurs at about 75% of stance. These results are similar to the findings of Hunt et al. (2008) who reported a peak rise of the pelvis on the swing side at 23.1% stance in 120 OA patients, and a peak drop of the pelvis on the swing side at 81.5% stance. Our lateral pelvic tilt data did not compare well in waveform shape to studies by Huang et al. (2008), Crosbie et al. (1997a,b) or Whittle and Levine (1999) which all used different sequences of rotation to define lateral pelvic tilt. However, it is known that the selected sequence of rotations highly influences the values of the Cardan angles (Baker, 2001).

The difference between control and OA subjects is in the range of motion through the stance phase of gait. When the bias due to the variation in neutral position between subjects is removed (Fig. 4) there is a vertical shift in the overall mean curve for the group. This difference is highlighted for example in the pelvic tilt curves before and after removing the position bias (Figs. 2c and 4a). While the waveforms retain their overall shape, the removal of the neutral position bias shifts the control curve down and the OA curve up relative to one another.

The combined motions of the pelvis and thorax showed both segments having more motion compared to the lab coordinate system in the OA group; however, when the thorax was projected on the pelvis coordinate system, the control subjects showed a greater range of motion throughout stance phase. This indicates that the OA subjects moved the trunk and pelvis more as a single unit, whereas the control group balanced motions between the two segments. Hunt et al. (2008) suggested that due to the small range of motion in the pelvis, lateral thoracic tilt is primarily achieved by leaning the trunk and shoulders (Hunt et al., 2008). However, we found that the thorax displayed an equally small range of lateral tilt motion as the pelvis. These results lead us to believe that lateral trunk lean is a combination of both pelvic and thoracic tilt. Even a small angular tilt could move the mass of the trunk sufficiently to alter the loading in the knee, which is consistent with suggestions by Chang et al. (2005), however contrary to Chang et al. (2005) we did not find that a drop in the pelvis on the swing side shifts the body's mass towards the swing limb. Instead we found during very late stance that an increased thoracic tilt over the swing limb is coupled with a pelvic tilt over the stance limb. The thoracic tilt over the swing limb is more pronounced in the OA group. This leads us to believe that even small angle differences in the pelvic and thoracic patterns of motion between groups are relevant.

Differences in thoracic and pelvic motions between control and OA groups are found around 25% of stance, and then again toward late stance, and very late stance (Figs. 4 and 5). The points of 25% and 75% of the stance phase of gait correspond approximately to the points of first and second peaks for the knee and hip adduction moments. It is notable that these two points in the stance cycle correspond to the times at which pelvic motion differs most between the control and OA groups.

A standard definition for tracking the lateral motion in the thorax has not been proposed. Some studies have tracked the thorax at two positions along the spine relative to each other and the pelvis (Crosbie et al., 1997b). Others have defined a virtual line up the centre of the body using the midpoints between landmarked points on the pelvis and shoulders and tracked this line relative to the vertical axis (Hunt et al., 2008). Tanaka et al. (2008) monitored the lateral bend in the thorax by tracking a line between C7 and S1 of the spine relative to a line perpendicular to the floor. Henriksen et al. (2009) defined trunk lean in two ways, similar to us: as the angle between the trunk and pelvis segments, and as the angle between the vertical trunk axis and the global vertical axis. Our OA group data for thoracic tilt-lab correspond well to that reported by both Hunt et al. (2008) and Crosbie et al. (1997b). Whittle and Levine (1999) defined lateral bend similarly to our model of thoracic tilt-pelvis, however, they recorded the motion of the lower portion of the thorax at the thoracolumbar junction of the spine. Our thoracic tilt angles and our pattern of motion match well with the model of Whittle and Levine (1999). Using our thoracic tilt-pelvis definition involves measuring data from two tracked rigid bodies. While the disadvantage of this definition is that it increases our measurement error, it has the advantage of allowing us to analyze whether the thorax and pelvis are moving together or separately, and how much motion is attributed to each segment. We found evidence that the control group does move their thorax separately from the pelvis more than the OA group, indicating the importance of this thoracic tilt definition. Ultimately it would be ideal to track the motion of the centre of gravity of the full upper body.

Differences in footwear and progression angle between subjects may be seen as a limitation of this study due to the confounding effects they may have on pelvic and trunk motion. However, leaving these variables uncontrolled allowed each subject to walk as naturally as possible. Subjects were asked to walk at a self-selected speed in order to obtain the best representation of their true gait patterns. Control subjects were found to have a significantly higher walking speed than OA subjects. Frontal plane pelvic tilt has been shown to increase significantly with an increase in gait speed (Crosbie et al., 1997a). We found that the OA group, which ambulated at a slower gait speed, displayed increased lateral tilt in the pelvis throughout stance compared to the control group. This suggests that if the two groups had equivalent gait speed, the difference between the OA and control groups would have been larger. Therefore the difference in pelvic motion found between the two subject groups is not due to gait speed differences.

The radiographic data indicated that these subjects represent a moderate OA group with a median KL score of 2 on a scale from 0 to 4 and individual subject scores varied widely within that range (scores of 1–4). Subjects with differences in disease severity have been shown to display gait pattern differences (Asthephen et al., 2008). These results may not be representative of the patterns found in subjects with very mild or very severe knee OA. Although two control subjects were found to have KL scores of 2 and seven subjects were found to have scores of 1, these subjects did not have any knee pain or limitations, and had not suffered any previous knee injury. Thus, they had a negative clinical diagnosis of knee OA and were included in our control set.

In summary, a thorough biomechanical analysis of the relationship between thoracic and pelvic tilt and knee and hip adduction moments has been presented. Subjects with OA in the medial

compartment of the knee displayed greater knee and hip adduction moments as well as altered gait patterns in both the thorax and the pelvis throughout the stance cycle of gait compared to controls. Knee OA subjects also tended to display a slightly increased range of motion in their trunk and pelvis, and tended to move these two segments more as a single unit compared to control subjects; however this difference between groups was only detectable using PCA. In recognizing our limitations to conclusively discuss causality in our data, we recommend future studies be designed to test the effects of controlled pelvic and thoracic tilt on hip and knee adduction moments in subjects with knee OA.

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