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Knee biomechanics of moderate OA patients measured during gait at a self-selected and fast walking speed

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Abstract

Osteoarthritis (OA) is a chronic disorder resulting in degenerative changes to the knee joint. Three-dimensional gait analysis provides a unique method of measuring knee dynamics during activities of daily living such as walking. The purpose of this study was to identify biomechanical features characterizing the gait of patients with mild-to-moderate knee OA and to determine if the biomechanical differences become more pronounced as the locomotor system is stressed by walking faster. Principal component analysis was used to compare the gait patterns of a moderate knee OA group ($n = 41$) and a control group ($n = 43$). The subjects walked at their self-selected speed as well as at 150% of that speed. The two subject groups did not differ in knee joint angles, stride length, and stride time or walking speed. Differences in the magnitude and shape of the knee joint moment waveforms were found between the two groups. The OA group had larger adduction moment magnitudes during stance and this higher magnitude was sustained for a longer portion of the gait cycle. The OA group also had a reduced flexion moment and a reduced external rotation moment during early stance. Increasing speed was associated with an increase in the magnitude of all joint moments. The fast walks did not, however, increase or bring out any biomechanical differences between the OA and control groups that did not exist at the self-selected walks.

Keywords: Principal component analysis (PCA); Moment; Three-dimensional gait; Knee dynamics; Osteoarthritis (OA)

1. Introduction

Gait analysis is a very useful tool for quantitatively describing the functional differences associated with knee osteoarthritis (OA). Understanding gait characteristics of individuals early in the OA disease process provides potential for developing non-surgical interventions to reduce pain and slow the disease progression; however, the majority of knee OA gait studies have focused on the later stages of the disease. Individuals with knee OA have been shown to walk with slower velocities, slower cadence, greater stance ratios, smaller stride lengths, and greater stance durations (Andriacchi et al., 1977; Baliunas et al., 2002; Kaufman et al., 2001; Teixeira and Olney, 1996). OA knee joints have been shown to have less range of motion throughout the stance phase of the gait cycle (Schnitzer et al., 1993; Stauffer et al., 1977). The net external adduction moment of the knee is significantly higher in patients with OA compared to healthy individuals (Baliunas et al., 2002; Hurwitz et al., 2002) and it is believed that this greater adduction moment corresponds to an increased knee load on the medial compartment relative to the lateral compartment (Schnitzer et al., 1993) and supports the concept that increased medial compartment loads play a role in the development of knee OA. Bone distribution between the medial and lateral compartments of the proximal tibia has also been correlated to the external knee adduction moment (Hurwitz et al., 1998). Baliunas et al. (2002) found that the peak external knee flexion

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moment during mid-stance was significantly lower in an OA population, when compared to a normal population.

While most OA gait studies have been able to identify biomechanical features characterizing populations with varying degrees of knee OA, no study to our knowledge has investigated the overall magnitude and shape of the kinematics and kinetics waveforms in a homogeneous subject group with mild-to-moderate knee OA. As well, no study has addressed how individuals with mild-to-moderate knee OA respond mechanically when their locomotor system is stressed while walking fast.

Principal component analysis (PCA), developed by Hotelling (1933), is a multivariate statistical technique that is effective in analyzing the overall magnitude, shape, and temporal pattern of kinematics, kinetics, and EMG waveforms (Astephen and Deluzio, 2005; Deluzio et al., 1997, 1999; Hubley-Kozey et al., 2006; Raptopoulos et al., 2006; Wootten et al., 1990) as well as other movement tasks (Hubley-Kozey and Smits, 1998; Hubley-Kozey and Vezina, 2002; Wrigley et al., 2005). This technique uses inter-subject variation throughout entire waveforms to identify features that can be interpreted biomechanically and assessed for differences between subject groups and/or tasks. In a review article of gait data analysis techniques, it was stated that PCA has the practical ability of providing a powerful understanding into both healthy and pathological human gait (Chau, 2001).

PCA as an analysis technique has distinctive advantages over traditional waveform analysis methods that use discrete parameters (i.e., peak values or ranges) for biomechanical comparisons. Defining discrete parameters is subjective (Chau, 2001) and the parameters can be difficult to extract from all waveforms, particularly in pathological cases. PCA has the advantage of: (i) objectively identifying biomechanical features based on the variation in the data, (ii) data from the entire gait cycle are used, and (iii) temporal information is also considered (Astephen and Deluzio, 2005; Chau, 2001; Deluzio et al., 1997, 1999; Deluzio and Astephen, 2006; Jones and Rice, 1992; Wootten et al., 1990). A recent study has also provided support for the increased sensitivity of PCA over traditional analysis techniques in detecting differences in kinematics and kinetics data (Wrigley et al., 2005).

This study set out to compare knee joint angle and moment waveform patterns between mid-to- moderate OA patients and control subjects using PCA. Subjects were monitored while walking at a self-selected and fast walking speed and the hypothesis was that biomechanical differences between the two groups would be more pronounced during the fast walks when the locomotor system was stressed.

2. Methods

2.1. OA patients

Forty-one OA patients were recruited from the Orthopedic and Sports Medicine Clinic of Nova Scotia. All OA patients were:

(i) symptomatic,
(ii) on a waiting list for or had undergone exploratory knee arthroscopy at least 1 year prior,
(iii) able to walk at 150% of their normal walking speed,
(iv) assessed radiographically with a Kellgren–Lawrence (KL) score between 1 and 3 in their most affected knee.

2.2. Control subjects

Forty-three asymptomatic subjects over the age of 35 years were recruited from the general population and they needed to have no history or evidence of arthritis or surgery to the lower limbs. Subjects were excluded from either group if they had any other form of arthritis, gout, neuromuscular disorders, or prior history of stroke or cardiovascular disease.

2.3. Gait analysis

Informed consent, in accordance with Dalhousie University Ethics Review Board, was obtained from all subjects prior to undergoing a complete gait analysis of the lower limb. The three-dimensional motion of the body was described using bone embedded coordinate systems established from tracking 16 infrared emitting diode markers during the walking trials and eight virtual markers identified during a standing calibration trial. The infrared markers tracked during the motion included marker triads on the pelvis, thigh, shank, and foot and individual markers on the greater trochanter, lateral epicondyle, lateral malleolus, and shoulder. Virtual markers, used to establish the anatomical coordinate systems in the pelvis, thigh, shank, and foot included the right and left anterior superior iliac spines, medial epicondyle, fibular head, tibia tubercle, medial malleolus, second metatarsal, and heel (Cappozzo et al., 1997). The infrared markers were captured at 100 Hz using 2 Optotrak 3020 optoelectronic motion analysis sensors (Northern Digital, Incorporation, Waterloo, ON, CA). An AMTI force platform (Advanced Mechanical Technology, Incorporation, Watertown, MA, USA) was used to collect the ground reaction forces and moments at 1000 Hz.

The position and orientation of each segment was estimated using a least-squares optimization routine (Challis, 1995). Joint motion was described using the conventions proposed by Grood and Suntay (1983), with flexion/extension occurring about the medial/lateral axis of the thigh, internal/external rotation occurring about the distal/proximal or long axis of the shank, and adduction/abduction occurring about the floating axis (intermediate axis perpendicular to both the flexion/extension and internal/external rotation axis). Net external inter-segmental joint reaction moments were calculated for the gait cycle using an inverse dynamics model implemented in MATLAB (The MathWorks, Natick, MA, USA), which combined marker positional data, ground reaction force and moment data, and limb segment mass/inertia properties (Clauser et al., 1969). Gait variables included stride characteristics (stride length, stride time, stance time, stance percentage, and speed), knee flexion/extension angle and moment, knee adduction/abduction moment, and knee internal/external axial rotation moment. We did not examine knee adduction/abduction and internal/external rotation angles because their relative values during gait are of a similar magnitude to that of the measurement error associated with kinematic cross-talk and skin motion (Kadaba et al., 1990; Piazza and Cavanagh, 2000; Ramsey and Wretenberg, 1999; Reinschmidt et al., 1997).

During the gait analysis, all subjects were instructed to walk along the walkway 5 times at a self-selected walking speed. Using two infrared light timing gates controlled by LabVIEW (National Instruments Corporation, Austin, TX, USA), an average self-selected walking velocity was calculated for the 5 trials. An additional 5 walking trials at a fast speed ($\pm$10% of the 150% self-selected walking velocity) were also collected.

All gait waveforms were time normalized by having the gait cycle described with 101 data points ranging from 0% (first heel strike) to 100% (second heel strike) of the gait cycle. The magnitudes for all moment waveforms were normalized to body mass (N m/kg).

2.4. Data analysis

All stride characteristics and gait waveform measures were ensemble averaged and grouped into one of four groups: (1) control subjects at self-selected walk, (2) OA patients at self-selected walk, (3) control subjects at fast walk, and (4) OA patients at fast walk.

PCA was applied to the gait waveforms to test for differences in the shape of the waveforms (Deluzio et al., 1997). For each waveform measure, a matrix, X, was created that included the self-selected and fast walking ensemble average waveforms, for both the OA patients and control subjects. The covariance matrix of these data (X) was used to generate eigenvectors, U, or principal components (PCs). Each PC represented an independent waveform feature based on the variability in the original waveform data set, with the first extracted PC corresponding to the largest source of variation, the second PC corresponding to the second largest source of variation and so forth. The PCs can be considered analogous to the sinusoids used in Fourier analysis where the sinusoids represent a new basis for the original waveform data. The PCs in this study formed a new orthogonal, independent basis and the PC Z-scores represented the coefficients measuring the degree to which a particular feature was present in each waveform. The Z-scores were calculated as $Z = (X - \overline{X})U$, with X being the original variables and U being the transformed variables or PCs. The Z-scores are a measure of the distance a subject's waveform is from the mean for a given PC, with high Z-score subjects having an original waveform that is well correlated with that PC. The maximum and minimum waveforms and the corresponding PC were used together to interpret the feature of variation that the PC was describing (Jones and Rice, 1992).

2.5. Statistical analysis

Student t-tests were used to detect statistical differences ($\alpha = 0.05$) between the OA patient and control subject groups for the anthropometric measures of age, height, weight, and body mass index (BMI = mass/height^2). Z-scores and stride characteristics were analyzed using a 2-factor repeated-measures ANOVA model with the repeated measure being speed ($\alpha = 0.05$). Post hoc pairwise comparisons were performed for significant interactions using a Tukey adjustment ($p < 0.05$). All statistical tests were performed in MATLAB and SAS (SAS Institute Inc., Cary, NC, USA).

Table 1
OA and control demographics, data are mean (standard deviation)

	Height (m)	Weight (kg)	Age (year)	BMI (kg/m^2)
Control ($N = 43$)	1.71 (0.09)	72.5 (13.5)	50.7 (10.2)	24.8 (3.9)
OA ($N = 41$)	1.74 (0.10)	91.8 (16.0)	58.2 (8.3)	30.3 (4.5)
p value	0.18	**<0.001**	**<0.001**	**<0.001**

Bold indicates statistical difference ($p < 0.05$) between control and OA groups.

3. Results

Of the 41 OA patients in the study, 5 had a KL score of 1, 22 had a score of 2, and 14 had a score of 3. The OA patients were older and heavier than the control subjects (Table 1). The BMI for the OA patients was also larger than that of the control subjects ($p < 0.001$).

No group differences between the OA patients and control subjects were found in stride length, stride time, and walking speed, for either the self-selected or fast walks (Table 2). The OA patient group spent a larger portion of the gait cycle (1%) in stance compared to the control subject group at both speeds ($p < 0.05$). The faster walks resulted in an overall increase in stride length and a decrease in stride time, stance time, and stance percentage compared to the self-selected walks for both groups. The speed of the fast walks for the OA patients was on average 131% faster than the speed of the self-selected walks. The corresponding increase in speed for the control subjects was 137% and this was not statistically different from the OA patients. Stride time and stance time were the only measures to have interaction effects; however, post hoc pair-wise comparisons did not detect group differences at either speed.

The OA patients had a larger overall adduction moment magnitude during stance than the control subjects, for both the self-selected and fast walks ($p = 0.01$) (Table 3 and Fig. 1). PC1 (Fig. 1B) captured this overall moment magnitude during stance, with a high Z-score corresponding to a large adduction moment magnitude (Fig. 1C). The interactions plot in Fig. 2A shows the magnitude of the adduction moment was not affected by speed ($p = 0.62$).

The shape of the adduction moment waveform was also important. PC2 captured the magnitude of the adduction moment during mid-late stance relative to the magnitude during early stance (Fig. 1B). Higher PC2 Z-scores corresponded with higher adduction moments during mid-late stance (Fig. 1D). Both groups had higher adduction moments during mid-late stance (PC2 Z-scores) at the self-selected speed than the fast speed (Fig. 2B,

Table 2
OA and control stride characteristics for self-selected and fast walking speeds. Data are mean (standard deviation)

	Stride length (m)	Stride time (sec)	Stance time (sec)	Stance percentage (%)	Speed (m/sec)
Control (self-selected walk)	1.45 (0.12)	1.07 (0.10)	0.67 (0.07)	62.6 (1.7)	1.38 (0.19)
Control (fast walk)	1.66 (0.13)	0.93 (0.10)	0.56 (0.07)	60.6 (1.6)	1.81 (0.24)
OA (self-selected walk)	1.42 (0.13)	1.10 (0.11)	0.70 (0.08)	63.7 (1.4)	1.30 (0.19)
OA (fast walk)	1.66 (0.16)	0.94 (0.09)	0.58 (0.06)	61.4 (1.4)	1.78 (0.25)
Group	0.58	0.25	0.10	**<0.05**	0.26
Speed	**<0.001**	**<0.001**	**<0.001**	**<0.001**	**<0.001**
Interaction	0.06	**<0.05**	**<0.05**	0.07	0.13

Group, speed and interaction effect p values obtained from a 2-factor repeated measures ANOVA model with speed being the repeated measure. Bold indicates statistical difference ($p < 0.05$).

Table 3
P values for group, speed and interaction effects and description of the waveform feature that each significant PC explains

Gait measure	PC	Group effect	Speed effect	Interaction effect	Biomechanical feature described
Ad/abduct Moment	1	**0.01**	0.62	0.65	Overall ad moment magnitude during stance
	2	**0.03**	**<0.001**	**0.01**	Ratio of moment in mid-late stance vs. early stance
Flex/ext Moment	1	**0.04**	**<0.001**	0.14	Max flex moment magnitude during early stance
	2	0.27	**<0.001**	0.25	Overall amplitude during stance
Int/ext rot moment	1	**0.03**	**<0.01**	0.46	Max ext. rot moment magnitude during early stance
	2	0.40	**<0.01**	0.57	Max int. rot moment magnitude during mid-late stance
Flex/ext angle	1	0.07	**<0.001**	0.53	Magnitude shift during entire gait cycle
	2	0.06	**<0.001**	0.42	Phase shift in waveforms

Group, speed and interaction effect *p* values obtained from a 2-factor repeated measures ANOVA model with speed being the repeated measure. Bold indicates statistical difference ($p<0.05$).

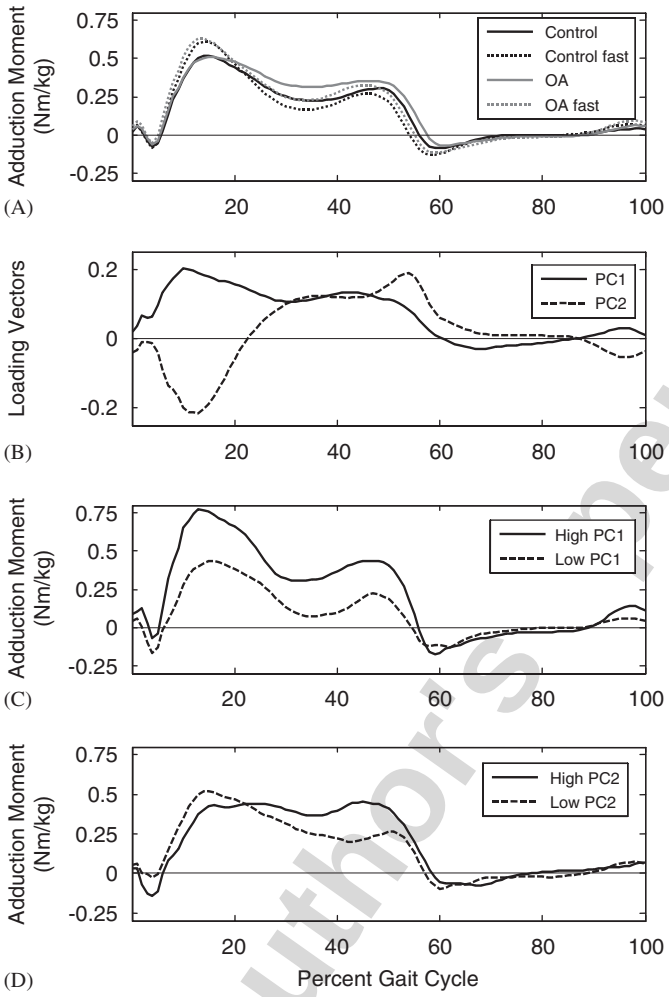

Fig. 1. (A) Mean adduction/abduction moment waveforms of OA patients and control subjects at self-selected and fast walks for the entire gait cycle. (B) PC1 capturing overall adduction moment magnitude during stance and PC2 capturing the relative magnitude of the adduction moment in mid-late stance relative to early stance. (C,D) The 5th and 95th percentile waveforms based on the individual *Z*-scores for PC1 and PC2.

$p<0.001$). The moderate OA group had higher adduction moments than the controls during mid-late stance only at the self-selected speed (Fig. 2B, $p = 0.01$).

The first two PCs for the flexion/extension moment waveforms captured group and speed differences in early stance and speed differences throughout stance (Table 3). PC1 captured the flexion moment magnitude during the first 25% of the gait cycle (Fig. 3). The OA patients had smaller flexion moment magnitudes (low *Z*-scores) than the control subjects, at both speeds ($p = 0.04$). The fast walks also resulted in larger flexion moment magnitudes (high *Z*-scores) compared to the self-selected walks, for both groups ($p<0.001$) (Fig. 2C). PC2 for the flexion/extension moment captured an overall amplitude difference during stance (Fig. 3). There was no difference between the two groups ($p = 0.27$); however, increasing speed was associated with an increase in the overall amplitude of the flexion moment ($p<0.001$) (Fig. 2D).

The first two internal/external rotation moment PCs had group and speed differences during early stance and speed differences during late stance (Table 3). PC1 captured the magnitude of the moment tending to externally rotate the tibia during early stance (5–20%) in Fig. 4B. OA patients had lower *Z*-scores, or a smaller moment tending to externally rotate the tibia than the control group ($p = 0.03$) (Fig. 4). Increasing speed was associated with an increase in the magnitude of this moment for both groups ($p<0.01$) (Fig. 2E). PC2 captured the magnitude of the moment tending to internally rotate the tibia during mid-late stance (Fig. 4B). There was no difference between the groups; however, increasing gait speed resulted in an increase in the magnitude of this moment for both groups ($p<0.01$) (Table 3).

The final waveform analyzed was flexion/extension angle and a speed effect was detected for both PC1 ($p<0.001$) and PC2 ($p<0.001$). PC1 captured the overall magnitude of the flexion angle throughout the gait cycle. The flexion angle, during both stance and swing were larger for the fast walks than the self-selected walks. PC2 captured a phase shift in the angle waveforms and the shift was most pronounced during swing and late stance. The fast walks resulted in an earlier rise to maximum knee flexion during swing and earlier fall towards full knee extension prior to heel strike.

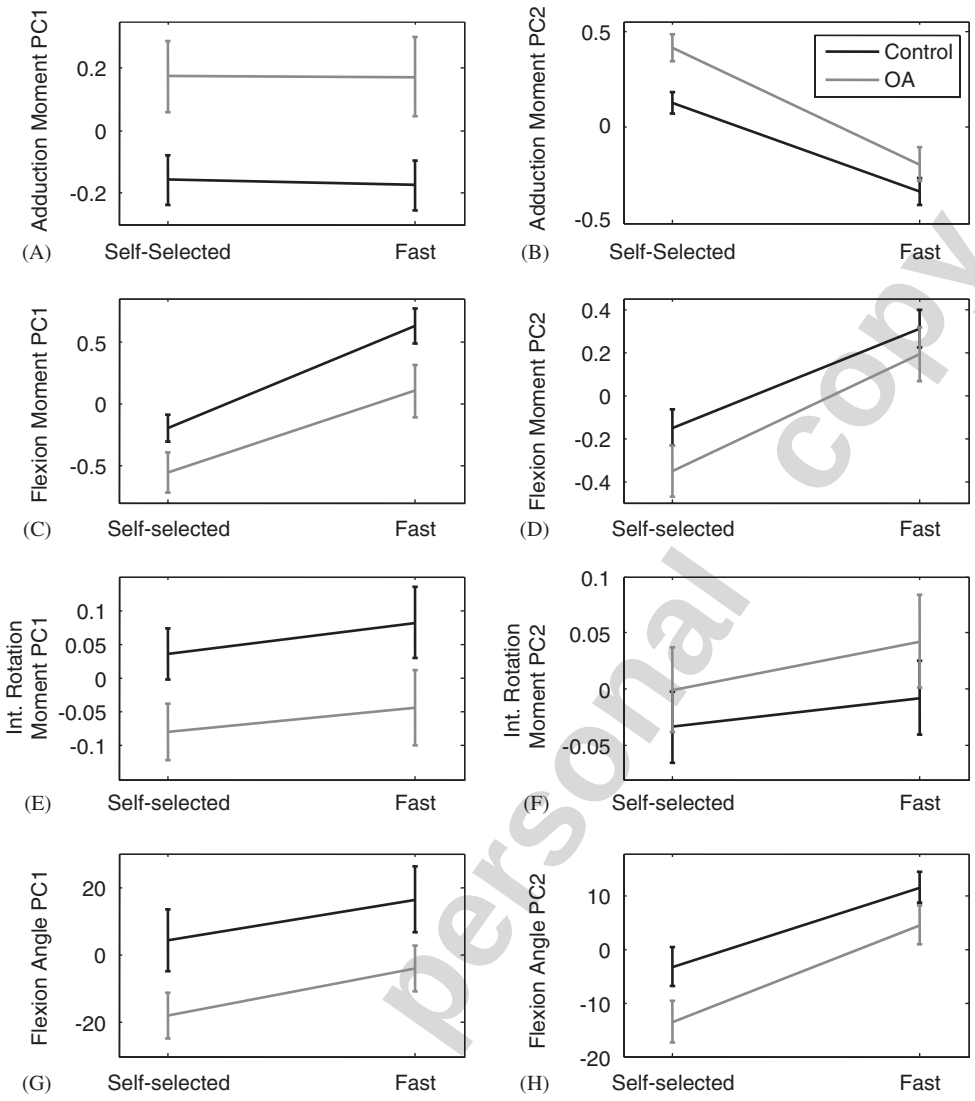

Fig. 2. *Z*-score means with standard error of the means for the OA patients and control subjects, at both the self-selected and fast walks. Each *Z*-score plot corresponds to one of the eight PCs (listed above) that were used to describe group, speed, and interaction effects in the angle and three moment waveforms.

4. Discussion

This study set out to identify biomechanical features characterizing the gait of patients with mild-to-moderate knee OA and to determine if the biomechanical differences become more pronounced as the locomotor system is stressed by walking faster. We found that moderate OA patients walked with similar stride characteristics and knee joint kinematics as the control subjects. The OA patients had higher adduction moments throughout stance and smaller flexion and external rotation moments during early stance. Increasing walking speed did not amplify the biomechanical differences between the OA patients and control group.

The OA patients were slightly older and heavier than those with asymptomatic knees. The difference in BMI between groups is consistent with the literature for subjects of similar ages with and without OA (Childs et al., 2004; Kaufman et al., 2001).

There were no differences between the moderate OA patients and the controls for walking speed or knee flexion angle. This is in contrast to the majority of studies of more severe OA patients, who have reported that OA patients tend to walk at slower speeds (Kaufman et al., 2001; Mattsson et al., 1990; Teixeira and Olney, 1996) and with reduced knee flexion (Andriacchi et al., 1982; Schnitzer et al., 1993). Our findings do agree with Mundermann et al. (2004) who reported similar walking speeds between moderate OA patients and an age-matched control group. The lack of difference in the descriptive measures (i.e., gait speed) with differences in the causal measures (i.e., kinetics) is important as it suggests that changes in knee joint

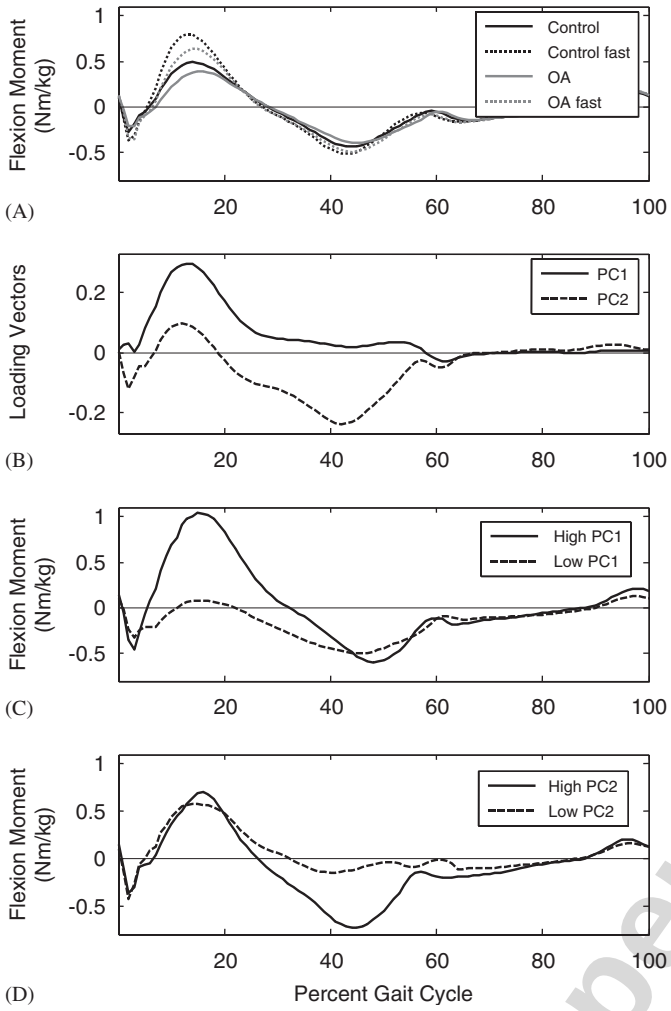

Fig. 3. (A) Mean flexion/extension moment waveforms of OA patients and control subjects at self-selected and fast walks for the entire gait cycle. (B) PC1 capturing the maximum flexion moment in early stance and PC2 capturing the overall amplitude of the flexion/extension moment during stance. (C,D) The 5th and 95th percentile waveforms based on the individual Z-scores for PC1 and PC2.

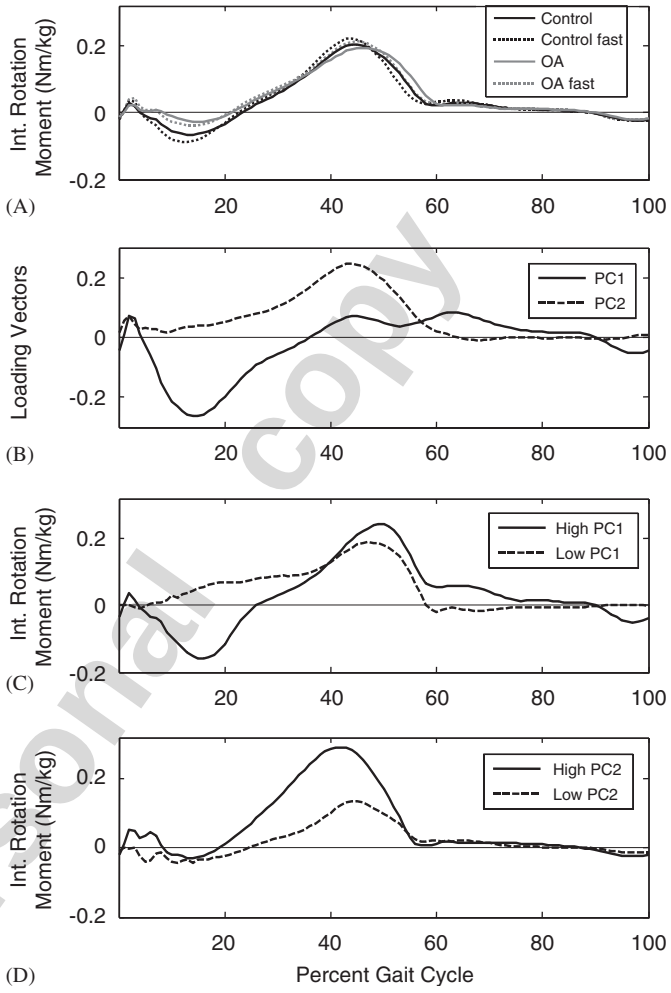

Fig. 4. (A) Mean internal/external rotation moment waveforms of OA patients and control subjects at self-selected and fast walks for the entire gait cycle. (B) PC1 capturing the maximum external rotation moment magnitude during early stance and PC2 capturing the maximum internal rotation magnitude during late stance. (C,D) The 5th and 95th percentile waveforms based on the individual Z-scores for PC1 and PC2.

kinematics and stride characteristics may be the result of the OA process; while the kinetic differences we observed may be associated with the pathomechanics of the disease.

The overall adduction moment magnitude throughout stance was greater for the OA patients at both gait speeds and we are the first to report changes in the shape of the adduction moment associated with moderate OA. Most previous studies have compared only the peak value of the adduction moment (Baliunas et al., 2002; Hurwitz et al., 1998; Kaufman et al., 2001) and when we tested for a difference in the peak value of the adduction moment, no group difference was found (group: $p = 0.69$; speed: $p < 0.001$; group × speed: $p = 0.21$). Kaufman et al. (2001) also reported no difference in the peak adduction moment between moderate OA patients and controls; however, this was possibly due to the slower speed of their OA patients. The problem of comparing peak knee adduction moments without controlling for walking speed has been raised by others (Messier, 1994; Mundermann et al., 2004). Other studies have selected gait trials at a speed closest to 1 m/s (Baliunas et al., 2002; Hurwitz et al., 2002; Sharma et al., 1998) to deal with this issue, but the adduction moment measured at this speed may not reflect the mechanical loads that normally occur outside the laboratory. Our results agree with Mundermann et al. (2004) who reported similar peak adduction moments in moderate OA and age-matched control groups walking at similar gait speeds. Our analysis of the peak adduction moment revealed higher peaks at the faster walking speed ($p < 0.001$), indicating that the peak adduction moment was affected by speed but not by moderate knee OA. The overall magnitude of the adduction moment (PC1), however, was not affected by speed but was higher for the moderate OA group. Using PCA, we have identified gait characteristics differentiating OA and control subjects that may not be identified using more traditional analysis techniques.

Furthermore, these findings are important to understanding the pathomechanics of OA since a longer duration adduction moment in addition to a higher magnitude will have an effect on cumulative joint loading.

The OA patients had a smaller flexion moment during early stance than the control subjects. This difference existed for both the self-selected and fast walks and corresponded very closely with the portion of the gait cycle where the maximum knee flexion moment occurred. Smaller knee flexion moments have been reported in OA patients (Baliunas et al., 2002) and have been associated with quadriceps or pain avoidance gait (Kaufman et al., 2001).

To our knowledge, only one other study has described differences in internal/external rotation moments between OA patients and control subjects (Gok et al., 2002). These authors reported that the OA patients had higher net intrinsic knee rotation moments, which is contrary to our findings. However, their peak knee rotation moment corresponds to the peak during the late stance phase, which was captured by PC2 of the internal/external rotation moment. The moment in the transverse plane of the knee has received very little attention in the literature and it is unknown if differences in this moment are related to the pathomechanics of the disease.

Increasing the speed for both subject groups was associated with differences in the magnitude and amplitude of the gait waveforms, consistent with previous data (Andriacchi et al., 1977; Mundermann et al., 2004). It was hypothesized that by having the two subject groups undergo a fast walk, thereby increasing the stress on their locomotor system; differences between the two groups from the self-selected speed would become larger or more pronounced. While our data did not support this hypothesis and stressing the locomotor system with a different task may be more appropriate, the faster walks did provide valuable knowledge into how knee loading is altered from a self-selected walk in both a moderate knee OA and control subject group.

In conclusion, this study has identified biomechanical features characterizing gait in a moderate knee OA population at two different speeds. PCA was able to detect changes from controls in the shape of the gait waveforms that have not been identified previously with other analysis techniques. While the clinical importance of these features need further study, these results are important as they identify differences in knee joint kinetics for those in the mild-to-moderate stage of the disease. These features may be used in future studies to help design interventions to slow disease progression.

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