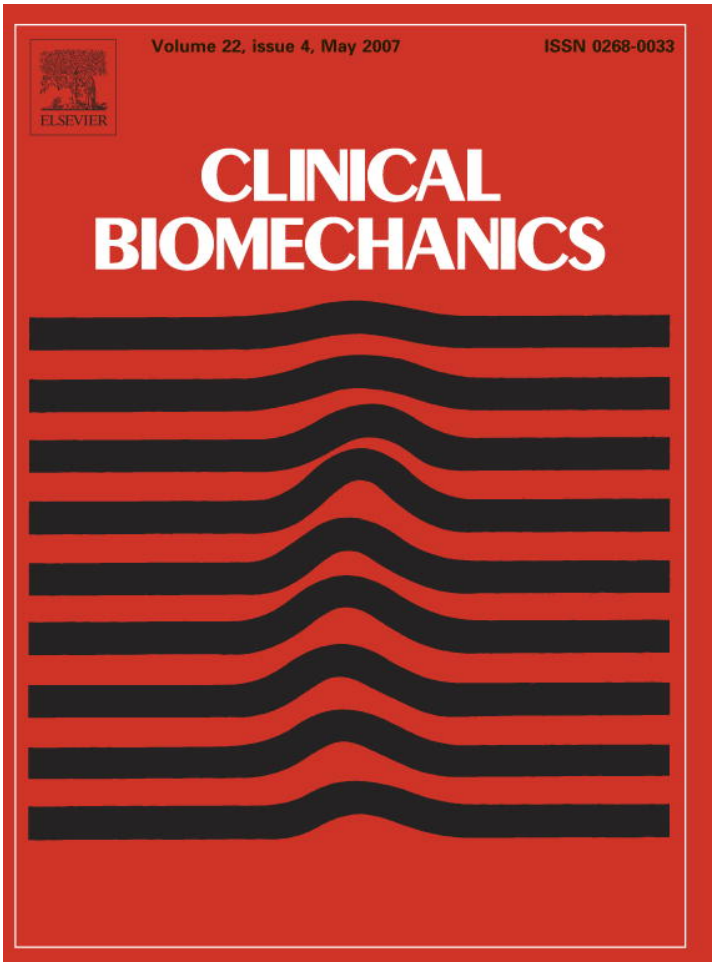

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Gender differences exist in osteoarthritic gait

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Abstract

Background. Knee osteoarthritis is 2–3 times more prevalent in females than males. Biomechanical differences in gait may play a role in this gender predisposition. The purpose of this study was to determine if there are gender-based biomechanical differences in the gait patterns of people with knee osteoarthritis.

Methods. Three-dimensional gait analysis was performed on healthy (18 males and 24 females) subjects and patients with moderate knee osteoarthritis (24 males and 15 females). Kinematics and kinetics at the hip, knee and ankle were calculated. Variables including anthropometrics, stride characteristics, strength, pain, stiffness, function and radiographic disease severity were also quantified. Multivariate statistical techniques and analysis of variance were used to test for main disease effects, main gender effects and disease vs. gender interactions.

Findings. A significant interaction effect between gender and disease was found in the knee flexion angle and the knee moments in the sagittal, frontal and transverse planes. In each of these measures the females exhibited different biomechanics with osteoarthritis, while the osteoarthritic males maintained the same biomechanics as healthy males. This interaction between gender and osteoarthritis was not associated with differences in anthropometrics, stride characteristics, strength, pain, stiffness, function or radiographic disease severity between the populations.

Interpretation. This study has found gait pattern differences between the genders in the osteoarthritic patients that were not apparent in the healthy subjects. This suggests that the biomechanics associated with knee osteoarthritis are gender dependent. Therefore, gender specific design of biomechanical interventions to slow the progression of osteoarthritis should be explored.

Keywords: Gait; Osteoarthritis; Biomechanics; Pattern recognition

1. Introduction

Osteoarthritis (OA) is a dynamic, progressive disease causing significant disability and loss of function. Knee OA is 2–3 times more prevalent in females than males (Buckwalter and Lappin, 2000) and females have two times the risk of developing bilateral knee OA (March and Bagga, 2004). Intrinsic differences in strength (Cureton et al., 1988; Drinkwater, 1988), quadriceps angle (Horton and Hall, 1989; Hsu et al., 1990), joint laxity (Bridges et al., 1992) and muscle activation patterns (White et al., 2003) exist between males and females. These factors affect the biomechanical environment of the lower limb and may cause biomechanical differences in the way males and females walk that contribute to the higher prevalence of OA in females.

In healthy non-osteoarthritic subjects, similar joint kinetics at the knee (Hurwitz et al., 1998; Kerrigan et al., 2000) and equivocal results for joint kinematics and stride

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characteristics (Cho et al., 2004; Kerrigan et al., 1998) have been reported when comparing gait patterns between males and females.

Differences in gait patterns between healthy and osteoarthritic populations have been well documented. Male and female OA patients walk with an increased knee adduction moment (Baliunas et al., 2002; Hurwitz et al., 1998), decreased knee flexion moment (Kaufman et al., 2001), decreased knee flexion angle (Childs et al., 2004) and at a slower velocity than healthy controls (Andriacchi et al., 1977; Kaufman et al., 2001).

The effect of gender differences in an OA population on biomechanics remains unclear. The few studies that have differentiated gait patterns between OA males and females have reported conflicting results. Some studies indicate males exhibit gait patterns more consistent with OA gait than females (Kaufman et al., 2001; Weidenhielm et al., 1994); however, results concerning differences in the adduction moment have been conflicting (Wada et al., 2001; Weidenhielm et al., 1994). Weidenhielm et al. (1994) reported a greater peak and midstance knee adduction moment in OA males compared to OA females, while Wada et al. (2001) reported no difference in the peak knee adduction moment between OA males and OA females. Furthermore, the descriptions of confounding clinical variables, including disease severity have varied between studies. A comprehensive study on a well-described patient population in the early stage of the disease process is lacking.

Therefore, the purpose of this study was to determine if gender differences in an OA population affect biomechanics and if these differences can be explained by differences in anthropometrics, strength, or disease severity. The hypothesis was that biomechanical differences exist between healthy females and OA females that do not exist between healthy males and OA males and that these biomechanical differences are not affected by differences in anthropometrics, strength or disease severity. If biomechanical differences can be detected between males and females in the early stage of the disease, future research can focus on gender specific modification of the biomechanical environment to slow the progression of the disease.

2. Methods

2.1. Subjects

Forty-two healthy subjects (18 males and 24 females) and 39 OA subjects (24 males and 15 females) were included in the study. Healthy subjects were recruited through postings on Dalhousie University campus and they had to be over the age of 35 years, have no history or evidence of arthritis or surgery to the lower limbs. OA subjects were recruited from the waiting list for exploratory knee arthroscopy of the Orthopaedic and Sports Medicine Clinic of Nova Scotia. They were diagnosed clinically and radiographically with moderate knee osteoarthritis indicated by a Kellgren–Lawrence (KL) score between 1 and 3 (Kellgren and Lawrence, 1957). Subjects were excluded from either group if they had any major surgery or trauma to the lower limb, neuromuscular disorders, other forms of arthritis, gout or history of stroke or cardiovascular disease. Informed consent, in accordance with Dalhousie University Ethics Review Board, was obtained from all subjects upon arrival to the laboratory.

2.2. Data collection

2.2.1. Gait analysis

Each subject completed a three-dimensional gait analysis. Subjects wore spandex shorts and a t-shirt to allow placement of light emitting markers on bony landmarks on the shoulder, greater trochanter, lateral epicondyle and lateral malleolus of their affected side (randomly chosen for controls). In addition, a cluster of three infrared markers was fixed to the pelvis, thigh, shank and foot of the same leg. A 1 s calibration trial with the subject standing in a neutral position was collected to use as a reference position. The following virtual points were digitized in the neutral position to aid in the construction of anatomical coordinate systems: right anterior superior iliac spine (RASIS), left anterior superior iliac spine (LASIS), medial epicondyle, fibular head, tibial tuberosity, medial malleolus, second metatarsal head and calcaneus (Cappozzo et al., 1995). The subjects completed five walking trials at a self-selected speed in their regular walking shoes. Marker coordinates were collected at 100 Hz using the Optotrak motion analysis system (National Digital Inc., Waterloo, ON, Canada). A force platform (Advanced Medical Technology Inc., Watertown, MA, US) aligned to the global coordinates of the Optotrak system was used to record ground reaction forces and moments of force for one complete stride of the test leg (1000 Hz). The ground reaction force and lateral malleolus marker were used to define one complete gait cycle, from initial heel strike of the test leg on the force plate to the second heel strike with the same leg (Hreljac and Marshall, 2000). All gait measures were time normalized to 100% of the subject's gait cycle and ensemble averaged for each subject. Subject age, height, weight, foot width, calf and thigh circumference were measured. Height and weight were used to calculate body mass index (BMI). Anthropometrics were used to estimate the inertial properties of the body segments (Clausser, 1969).

Joint kinematics and kinetics were calculated using custom analysis software in Matlab (Mathworks, Natick, MA, US). Markers clusters on the pelvis, thigh, shank and foot were used to model each segment of the lower limb as a rigid body defined by a right handed Cartesian coordinate system. The position and orientation of each segment was estimated using a least squares optimization routine (Challis, 1995). Three-dimensional joint angles were calculated using the joint coordinate system (Grood and Suntay, 1983). At the knee, the flexion/extension axis, or primary

axis, is defined by a bone embedded mediolateral vector between the two epicondyles. The internal/external rotation axis, or secondary axis is a bone embedded axis in the tibia defined by a distal proximal vector from the midpoint of the two malleoli to the midpoint of the tibial plateau. Abduction/adduction is described about an axis orthogonal to the two bone embedded axes. At the ankle, the flexion/extension axis is defined by a mediolateral vector between the two malleoli, the internal/external rotation axis is defined by a distal proximal vector between the second metatarsal head and the calcaneus and the abduction/adduction axis is orthogonal to these two axes, directed anteriorly and inferiorly from the ankle joint. At the hip, the flexion/extension axis is defined by a mediolateral vector between the right and left anterior superior iliac spines, the internal/external rotation axis is defined by a distal proximal vector from the midpoint between the malleoli to the hip joint center and the adduction/abduction axis is the common perpendicular, directed anteriorly from the hip joint. At each joint the motion of the distal segment is described relative to a fixed proximal segment and flexion, adduction and internal rotation are positive.

An inverse dynamics model combined the kinematic data, ground reaction forces, moments of force and estimated inertial properties to calculate resultant joint moments at the hip, knee and ankle (Braune and Fischer, 1895). Joint moments were calculated using a Cartesian coordinate system at each segment and were subsequently transformed into the joint coordinate system to attain anatomical meaning. Joint moments were normalized to body mass (kg) (Costigan et al., 1992) and expressed as external moments.

2.2.2. Strength

Two maximum voluntary isometric contraction exercises were performed after the gait analysis. Seated resisted (1) knee extension (45°) and (2) knee flexion (55°) were each performed twice and held for 3-s using a Cybex Isokinetic Dynamometer (Cybex Inc., Lumex, NY). Torque output for each trial was subdivided into 500 ms windows and averaged within each window. The maximum averaged window was defined as the peak torque for that trial. The maximum torque across the two trials for each condition was used for further analysis.

2.2.3. Health outcome questionnaires

Each subject completed two health outcome questionnaires, the SF-36 (Ware et al., 2000) general health questionnaire and the Western Ontario & McMaster University Osteoarthritis Index (WOMAC) (Bellamy et al., 1988). For the SF-36 higher scores are indicative of a better health status. For the WOMAC higher scores indicate greater impairment.

2.2.4. Radiographic grading and reliability

Anterior posterior knee radiographs for all OA patients were scored by 3 orthopaedic surgeons according to the KL global grading scale (Kellgren and Lawrence, 1957) and the Scott feature-based scoring system (Scott et al., 1993). The feature-based criteria were also used to score the patellofemoral joint using lateral view radiographs. Intraclass correlation coefficients were used to test the interrater reliability of each score. Correlation coefficients <0.39 indicate poor reliability, 0.4–0.74 indicate fair to good reliability and >0.75 indicate excellent reliability (Fleiss, 1986). The reliable scores from one rater were randomly chosen for further analysis.

2.3. Data reduction

Principal component analysis was applied to the waveforms for each gait measure for the OA and healthy groups together (Deluzio et al., 1997). Each gait measure was represented by an $n \times p$ matrix X, where $n = 81$ subjects and $p = 101$ variables, each variable representing a percent of the gait cycle. The eigenvectors, U, or principal components (PCs) extracted from the covariance matrix (S) of X are an orthogonal representation of the original variables that describe features of variation within the original waveforms. The PCs are extracted hierarchically according to the amount of variation they explain. A PC is extracted for each of the original variables; however, the majority of the variation is typically described in the first few PCs. The number of PCs to retain, k, to reduce the data and adequately describe the variation was determined using parallel analysis (Jackson, 1991). This technique involves creating an $n \times p$ matrix, X_R containing standardized, normally distributed random numbers. Principal component analysis is applied to X_R. The PCs extracted from the original data matrix, X, that explain a greater amount of variation than the PCs extracted from X_R are retained. For each retained PC, a score was calculated for each subject, $Z = (X - \overline{X})U$, where X is the original variables and U is the transformed variables. PC scores indicate the distance of the subject's waveform from the mean for a given PC. The maximum and minimum waveforms and the eigenvector are used to interpret the feature of variation that the PC is describing (Jones and Rice, 1992). Subjects with high PC scores have a waveform closer to the maximum described by that PC, while subjects with low PC scores have waveforms closer to the minimum described by that PC.

2.4. Statistical analysis

A two-way ANOVA was used to test for differences in PC scores between gender and disease for all PCs retained using parallel analysis. Tukey's post-hoc honestly significant difference test was applied to the PCs with a significant interaction effect. Anthropometrics, stride characteristics, strength, and health outcome data were also analyzed using a 2-way anova and Tukey's post-hoc analysis. A Fischer's Exact test was used to test for differences in radiographic scores between OA males and females. Post-hoc testing was used to identify differences at each level of the

radiographic scores. Significant P-values were adjusted for the number of comparisons using Bonferonni's adjustment. Statistical tests were conducted in Matlab (Mathworks, Natick, MA, US) and Statistical Packages for Social Sciences 9.0 (SPSS Inc., Chicago, IL, US) using a significance level of $P < 0.05$.

3. Results

3.1. Anthropometrics

Main effect differences were found for both gender and disease (Table 1). Males were significantly taller and heavier than females ($P < 0.01$) OA patients were significantly older ($P < 0.05$), heavier and had a higher BMI than the healthy subjects ($P < 0.01$). There was no interaction effect between gender and disease for age, height, weight or BMI.

3.2. Gait analysis

3.2.1. Stride characteristics

Males had a significantly longer stride length and stance time than females ($P < 0.01$). Healthy subjects walked faster, with a longer stride length and a shorter stance time than OA subjects ($P < 0.05$). There was no interaction effect between gender and disease for any stride characteristics (Table 1).

3.2.2. Kinematics and kinetics

Significant gender effects were found in the internal rotation angle, adduction moment and flexion moment at the hip and the internal rotation moment at the knee (Table 2). Female subjects walked with significantly less hip internal rotation, a smaller flexion moment and a larger adduction moment than male subjects ($P < 0.01$). Females also walked with a smaller internal/external rotation moment at the knee than males ($P = 0.04$).

Significant disease effects were found in gait measures at the hip, knee and ankle (Table 2). OA subjects walked with a smaller internal/external rotation moment ($P < .01$) and a more internally rotated hip ($P = 0.01$) than healthy subjects. At the knee, OA subjects walked with a larger knee adduction moment than healthy subjects, or greater load on the medial compartment during the stance phase ($P = 0.05$) and a smaller knee flexion angle throughout the entire gait cycle ($P = 0.02$) compared to healthy subjects. At the ankle, OA subjects were less plantar flexed at toe-off ($P < 0.01$) and generated a smaller ankle flexion moment during stance ($P < 0.01$).

A significant interaction effect between gender and disease was found in the knee flexion angle and the knee moments in the sagittal, frontal and transverse planes (Table 2). In each of these measures the females exhibited different biomechanics with OA, while the OA males maintained the same biomechanics as healthy males.

The mean waveforms for the knee flexion moment show that OA females produce less flexion/extension moment throughout the stance phase of the gait cycle (Fig. 1a). This amplitude difference was captured by the second eigenvector or PC2 (Fig. 1b). Flexion moment waveforms associated with high and low PC2 scores differ in the extent of knee flexion moment observed in early stance relative to the extension moment observed in late stance (Fig. 1c). Therefore, PC2 captures the overall amplitude of the flexion

Table 1
Anthropometric, stride characteristic, strength, and health outcome data for normal and OA males and females

	Normal				OA				Gender effect	Disease effect	Interaction effect
	Male	$n = 18$	Female	$n = 24$	Male	$n = 24$	Female	$n = 15$			
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	P-value	P-value	P-value
Anthropometrics											
Age (years)	52.2	(10.1)	48.7	(10.3)	55.1	(13.8)	57.0	(11.2)	0.76	0.04	0.30
Height (m)	1.8	(0.1)	1.7	(0.1)	1.8	(0.1)	1.7	(0.1)	<0.01	0.86	0.79
Weight (kg)	78.9	(13.5)	66.2	(9.7)	94.3	(16.6)	85.5	(14.5)	<0.01	<0.01	0.53
BMI (m^2/kg)	24.7	(3.2)	24.4	(3.6)	29.7	(4.6)	31.5	(5.2)	0.44	<0.01	0.28
Stride characteristics											
Speed (m/s^2)	1.6	(0.2)	1.4	(0.2)	1.3	(0.2)	1.3	(0.3)	0.62	0.02	0.57
Stride length (m)	1.5	(0.1)	1.4	(0.1)	1.4	(0.1)	1.3	(0.2)	<0.01	0.01	0.78
Stance time (s)	0.7	(0.1)	0.6	(0.1)	0.7	(0.1)	0.7	(0.1)	<0.01	0.03	0.45
Strength											
Knee extension 45° (Nm)	131.2	(39.4)	96.7	(29.3)	125.1	(37.2)	77.6	(28.6)	<0.01	0.11	0.41
Knee flexion 55° (Nm)	77.1	(29.0)	43.0	(15.5)	57.4	(21.6)	33.6	(15.8)	<0.01	<0.01	0.28
Health outcomes											
WOMAC											
Pain	0.2	(0.6)	0.1	(0.4)	7.3	(3.7)	7.7	(4.0)	0.79	<0.01	0.67
Stiffness	0.1	(0.3)	0.2	(0.8)	3.2	(1.6)	4.2	(1.7)	0.05	<0.01	0.08
Function	1.3	(3.2)	0.7	(2.5)	23.0	(12.2)	21.9	(10.7)	0.63	<0.01	0.89
Total	1.7	(3.9)	1.0	(3.7)	33.5	(16.6)	33.7	(14.9)	0.93	<0.01	0.85

A 2-way ANOVA was used to test for main gender effects, main disease effects and gender by disease interactions ($P < .05$).

Table 2
Interpretation of significant PCs

Gait measure	Interpretation of significant PCs		Gender effect *P*-value	Disease effect *P*-value	Interaction effect *P*-value
	PC	Interpretation			
Hip					
Internal rotation angle	PC1	Magnitude stance and swing	<0.01	<0.01	0.89
Adduction moment	PC2	Magnitude during stance	<0.01	0.39	0.82
Flexion moment	PC1	Magnitude during stance	<0.01	0.06	0.65
Internal rotation moment	PC2	Amplitude during stance	0.98	<0.01	0.09
Knee					
Flexion angle	PC1	Magnitude stance and swing	0.60	0.02	0.72
	PC4	Range of motion during stance	<0.01	0.02	0.03
Adduction moment	PC1	Magnitude during stance	0.28	0.05	0.80
	PC2	Relative magnitude of first and second peak	0.47	<0.01	0.06
Flexion moment	PC2	Amplitude during stance	<0.01	<0.01	0.05
Internal rotation moment	PC1	External rotation moment (early stance)	0.14	0.06	0.05
	PC2	Internal rotation moment (late stance)	0.04	0.70	0.31
Ankle					
Flexion angle	PC2	Magnitude at toe-off	0.67	<0.01	0.94
Adduction moment	PC2	Amplitude during stance	0.75	<0.01	<0.01
Flexion moment	PC1	Magnitude during stance	0.07	<0.01	0.16

A 2-way ANOVA was used to test for main gender effects, main disease effects and gender by disease interactions in z scores ($P < .05$).

moment during stance. The ANOVA of the PC2 scores and the post-hoc pairwise comparisons shown in the interaction plot (Fig. 2e) revealed that OA females had a significantly lower flexion moment amplitude than the healthy females, OA males and healthy males ($P < 0.05$).

The fourth eigenvector, PC4, for the knee flexion angle, quantified the overall range of motion during the stance phase of the gait cycle. That is, higher flexion angle PC4 z scores reflected a greater range of motion during stance. This is illustrated by the difference between flexion angle waveforms corresponding to high and low z scores (Fig. 2b). The interaction plot (Fig. 2a) and the pairwise comparisons revealed the interaction between gender and OA were because gender differences existed in subjects with OA but not in the healthy subjects. Specifically, females with OA exhibited less range of motion during the stance phase than healthy females or males ($P < 0.05$). This pattern was also seen in knee internal rotation moment data. There was an interaction effect in the first PC of the internal rotation moment (Fig. 2g and h). OA females had significantly less *external* rotation moment at the knee than the healthy females ($P < 0.05$), corresponding to less internal/external rotation moment.

We also observed a feature of the knee adduction moment that was different between healthy and OA females but was similar between healthy and OA males. PC2 of the adduction moment captured a relative difference between the first and second peaks of the adduction moment (Fig. 2d). The OA females had lower PC2 scores than the healthy females ($P < 0.05$), indicating a smaller first peak relative to the second peak (Fig. 2c). The healthy females exhibited a higher initial peak adduction moment, however this decreased during the second half of the stance phase; where as, the OA females sustained the adduction moment throughout more of the stance phase.

This biomechanical pattern was not only present at the knee but also at the ankle. Females with OA exhibited different biomechanics for certain gait measures than healthy subjects, while OA males maintained biomechanics closer to healthy males. PC2 for the ankle adduction moment, or toe-in moment, also captured an amplitude difference during the stance phase. PC scores indicated OA females began generating an ankle toe-in moment earlier in the stance phase than healthy females and healthy males, but generated less total amplitude throughout the stance phase (Fig. 3). OA males also generated more toe-in moment than healthy females.

3.3. Strength

Males had significantly greater hamstring and quadriceps strength than females ($P < 0.01$). Healthy subjects had greater hamstring strength than OA subjects ($P < 0.01$). There was no interaction effect for either strength measures (Table 1).

3.4. Health outcome questionnaires

3.4.1. WOMAC

Females reported greater stiffness than males ($P < 0.05$). OA subjects reported more pain, stiffness and loss of function than the healthy subjects ($P < 0.01$). Total WOMAC scores and WOMAC subscales for pain and function did not reveal an interaction effect (Table 1).

3.4.2. SF-36

Healthy subjects scored significantly higher than OA subjects ($P < 0.05$) for all SF-36 subscales except role emotional and mental health, indicating a better health status. There was no main gender effect or gender by disease inter-

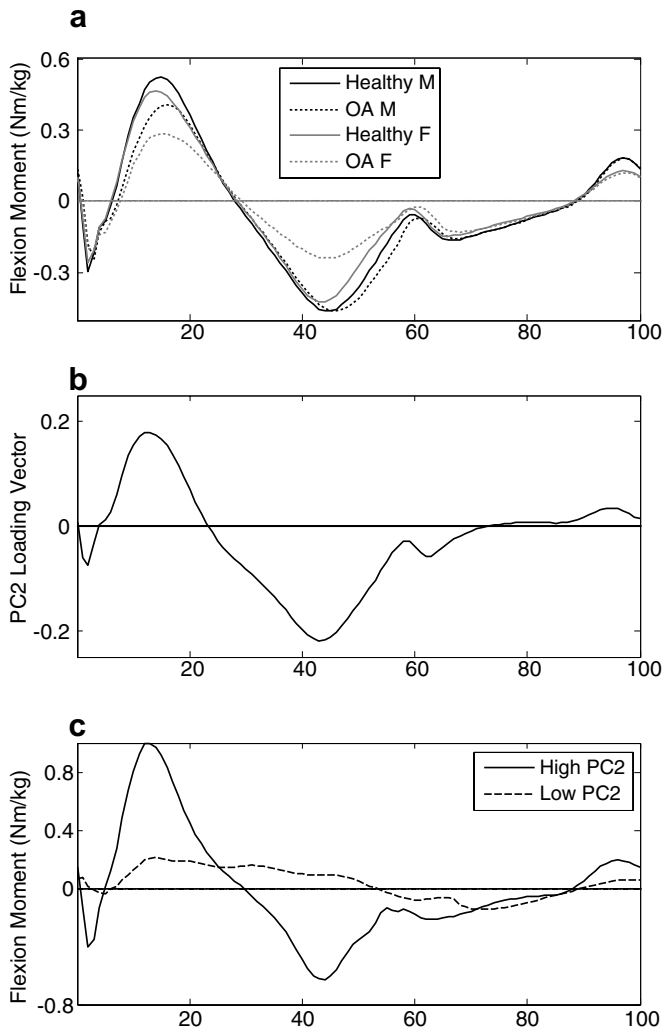

Fig. 1. Analysis of knee flexion-extension moment. (a) Average knee flexion waveforms for each of the four groups: males in black, females in gray, healthy solid lines, and OA dotted lines. (b) The loading vector for PC2 (second eigenvector). (c) Flexion moment waveforms corresponding to high and low PC2 scores demonstrate the difference in the biphasic shape, or amplitude, captured by PC2.

action for any of the SF-36 subscales including mental health, social function, physical function, vitality, general health, bodily pain, role physical or role emotional.

3.5. Radiographs

3.5.1. Reliability

All individual radiographic features except sclerosis and joint space narrowing of the patellofemoral joint had intraclass correlation coefficients between 0.47 and 0.71, indicating fair to good reliability (Table 3). The global KL score was also reliable across the three scorers; the intraclass correlation coefficient was 0.59, with a 95% confidence interval of 0.42–0.74.

3.5.2. OA males vs. OA females

There was a significant difference between OA males and OA females in two measures, the KL score ($P = 0.05$) and sharpening of tibial spines ($P = 0.02$). The distribution of KL scores is shown in Fig. 4. There was a greater number of OA females with a KL score of 1 ($P = 0.02$), indicating a less severe stage of the disease. There was no difference in the number of OA males and females with a KL score of grade 2 or 3. Significantly more OA males had the osteoarthritic feature of sharpened tibial spines.

4. Discussion

This study suggests that moderate knee OA is associated with differences in certain gait biomechanics in females that do not exist in males. This interaction between gender and OA was not associated with differences in anthropometrics, stride characteristics, strength, pain, stiffness, function or radiographic disease severity between the populations. OA females generate less torque at the knee and ankle and less range of motion at the knee, while OA males maintain mechanics closer to normal.

The population studied was a representative sample of moderate OA and healthy subjects. This is evidenced by the higher BMI, slower velocity and shorter stride length of the OA population relative to the healthy subjects. In the literature, OA gait has also been characterized by a greater load on the medial compartment of the knee (Baliunas et al., 2002; Hurwitz et al., 1998) and a decreased knee flexion moment (Kaufman et al., 2001). Our data supports these findings. We also found differences between OA and healthy subjects at the hip and ankle, however, comparison of these results to existing literature is difficult because the focus of most previous studies has been limited to the knee. Differences in anthropometrics (Cho et al., 2004; Kerrigan et al., 1998, 2000), stride characteristics (Kerrigan et al., 1998) and walking biomechanics (Hurwitz et al., 1998; Kerrigan et al., 2000) between males and females were also consistent with the literature. The longer stride length for males than females is likely due to the height difference. The gender difference in stride length disappeared when stride length was normalized to height. Biomechanics were similar between the two groups. Differences that were detected were primarily at the hip joint. These main gender and disease effects confirmed that our subject population was consistent with previous work.

The focus of this study was the interaction of gender and OA on gait biomechanics. We found that OA is associated with changes in certain gait biomechanics in females that do not occur males. As described above, both males and females with OA generate a larger magnitude knee adduction moment throughout stance compared to healthy subjects, however only OA females were associated with a change in the shape of the adduction moment waveform. The ratio between the first and second peaks was smaller in OA females than in healthy females, indicating OA females maintain the load on the medial compartment of the knee throughout the stance phase. Most studies have focused on the peak adduction moment (Baliunas et al., 2002; Hurwitz et al., 1998; Kaufman et al., 2001; Wada

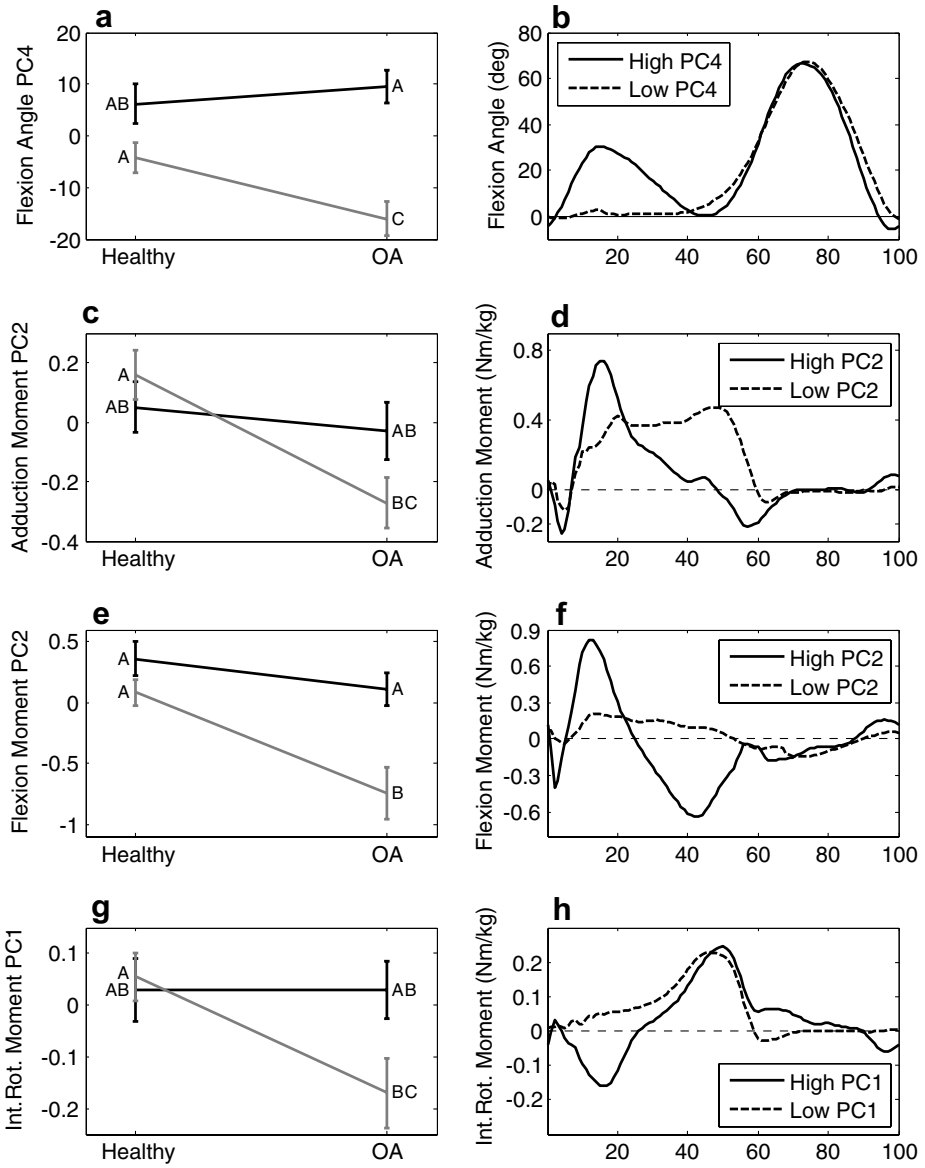

Fig. 2. Interaction effect in knee gait measures. (a, c, e, g) PC score means with standard error of the means for the males (black) and the females (gray). Tukey post-hoc analysis was used for the pairwise comparisons that are indicated on the graphs. Significant differences (from the Tukey post-hoc tests) ($P < 0.05$) are shown by different letters. (b, d, f, h) Gait waveforms corresponding to high and low PC scores demonstrate the interpretation of each of the PCs.

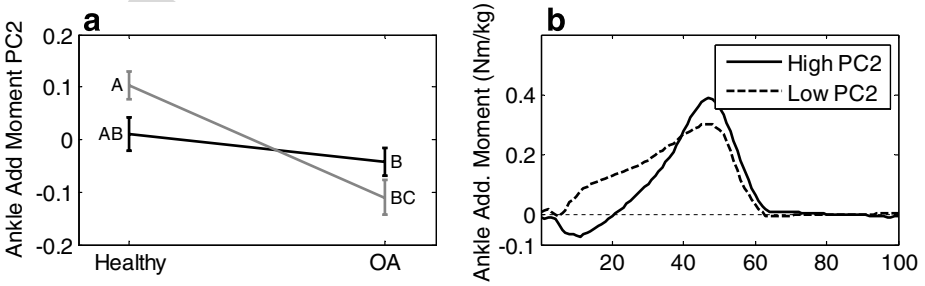

Fig. 3. Interaction effect in ankle gait measure. (a) PC score means with standard error of the means demonstrate the gender by disease interaction. Significant differences (from the Tukey post-hoc tests) are indicated by different letters on the graph. (b) Gait waveforms corresponding to high and low PC2 scores.

et al., 2001; Wang et al., 1990); however, few have explored both peaks, or the relationship between the two peaks. We found that the relationship between the two peaks is, in fact, important in characterizing male and female, healthy

Table 3
Reliability and differences in radiographic features

Radiographic features	Reliability		Fischer's exact test
	ICC[a]	95%CI	*P*-value
Joint space narrowing medial	**0.71**	(**0.57 0.83**)	0.43
Joint space narrowing lateral	**0.47**	(**0.28 0.65**)	0.63
Joint space narrowing PF	0.49	(0.19 0.70)	0.91
Osteophytes medial	**0.60**	(**0.42 0.75**)	0.75
Osteophytes lateral	**0.63**	(**0.46 0.77**)	0.40
Osteophytes PF	**0.61**	(**0.40 0.77**)	0.28
Sclerosis TF	0.28	(0.09 0.50)	0.09
Sclerosis PF	0.25	(0.06 0.46)	0.07
Tibial spines	**0.53**	(**0.35 0.70**)	0.02
Chondrocalcinosis	**0.59**	(**0.41 0.74**)	0.28
Kellgren Lawrence global	**0.59**	(**0.42 0.74**)	0.05

Inter rater reliability for Scott feature-based and Kellgren Lawrence global radiographic scoring. Differences in scores between OA males and OA females were tested using a Fisher's exact test ($P < .05$). Reliable scores are shown in bold.

[a] Intraclass Correlations Coefficient (ICC).

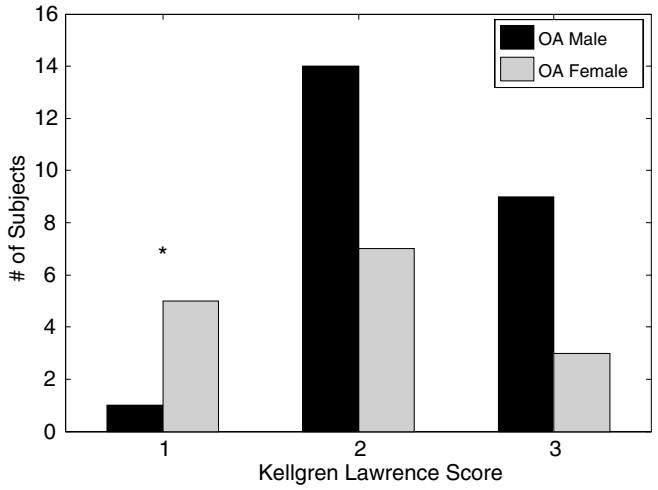

Fig. 4. Distribution of KL global radiographic scores. Radiographic data for the OA subjects revealed that significantly more OA females had a KL score of 1 than OA males ($P < .02$). * indicates significant difference.

and OA gait. Weidenhielm et al. (1994) found the mid-stance adduction moment to be more important for differentiating between OA and healthy subjects than the peak adduction moment. A lower peak adduction moment has also been reported in OA subjects compared to normal subjects, however the OA subjects in both studies walked significantly slower than the normal subjects (Kaufman et al., 2001; Weidenhielm et al., 1994).

OA females also changed their biomechanics in the sagittal and transverse planes at the knee, and in the frontal plane at the ankle compared to healthy females. OA females generated a smaller flexion/extension moment and less range of motion at the knee, biomechanics which are consistent with those documented in OA patients (Childs et al., 2004; Kaufman et al., 2001; Messier et al., 1992). We also found less internal/external rotation moment at the knee and less toe-in moment at the ankle in OA females compared to healthy females. A smaller toe-in moment has been reported in the literature as a mechanism to decrease the load on the medial compartment of the knee in OA patients (Wang et al., 1990; Andrews et al., 1996).

We investigated whether the differences found in the kinetic and kinematic variables were confounded with the group differences in height, weight, BMI, walking speed, and age. Analysis of covariance techniques were used to incorporate these concomitant variables into the analysis of the features (PCs) extracted from the kinematic and kinetic gait variables. The analysis of covariance did not change the results of the two-way ANOVAs described above. Therefore, the differences in gait kinematics, and kinetics existed even after accounting for the confounding variables. We also examined the effect of normalization of joint moments on our results. Some authors normalize joint moments to percent (body mass * height) (Hurwitz et al., 1998); whereas other authors normalize joint moments to body mass (Costigan et al., 1992), as we have done. Normalizing to percent (body mass * height) did not change our results.

We investigated whether these biomechanical differences were associated with strength, or measures of disease severity. These were not affected by the combination of disease and gender; therefore, they do not appear to be responsible for differences in gait. Not only was there no interaction effect in any of these variables, but when these variables were put into an analysis of covariance the kinetic and kinematic differences remained the same. This supports our findings that the interactions in gait parameters were not due to clinical measures of OA.

These clinical variables, in addition to radiographic evidence of OA, were used to define disease severity. In the literature, disease severity is often reported based on radiographic evidence, most commonly using the KL global radiographic score and measures of joint space narrowing (Miyazaki et al., 2002; Wada et al., 2001). While we have shown radiographic disease severity can be assessed reliably across different surgeons, other research has shown a poor correlation between radiographic OA and self-reported pain (Hannan et al., 2000). We collected the clinical variables described above, plus global and feature radiographic scores in order to better capture the stage of disease. The only indication of a difference in disease severity across these parameters was in global radiographic scores. Significantly more females had a KL score of 1, indicating a less severe disease state. Despite the fact that OA females may be in an earlier stage of the disease they exhibited biomechanics more consistent with OA gait than their male counterparts.

It is unknown whether the gender specific biomechanical differences we observed are contributing factors that help explain why more females have knee OA, or are a measure of the different biomechanical effect of knee OA on females. OA females may be altering their biomechanics either

earlier in the disease process or to a greater extent than their male counterparts as a mechanism to alleviate pain. Specifically, a decreased flexion/extension moment has been referred to as quadriceps avoidance and may be an attempt to decrease the forces through the knee joint and ultimately alleviate pain (Andriacchi et al., 1982). Limiting the internal rotation moment at the knee may also eliminate pressure on painful areas, and decreasing the amount of knee flexion limits the area that potentially degenerative articular cartilage and menisci must articulate. In the frontal plane, females may be maintaining load on the medial compartment of the knee throughout stance to alleviate pain associated with a high medial load just after heel strike. Biomechanical strategies at the ankle joint may aid in achieving these changes at the knee. This interjoint interaction has been reported by Wang et al. (1990).

Pain avoidance mechanisms have been proposed in the past to explain gait adaptations associated with OA (Kaufman et al., 2001; Schnitzer et al., 1993); however, the important finding from this study is that females exhibit certain changes with OA that do not occur in males. These changes may be related to differences in the way males and females react to pain, or to intrinsic differences in neuromuscular function between males and females that were not quantified in this study. There are some reports of differences in neuromuscular function between males and females. For example, higher coactivation of hamstring and quadriceps muscles have not only been found in females compared to males (White et al., 2003), but also in an OA population compared to a healthy population (Childs et al., 2004). Furthermore, healthy females have been reported to have lower absolute quadriceps muscle activation levels compared to males, and this was attributed to a lower percentage of fast twitch muscle fibres, especially in the vastus lateralis, (Bilodeau et al., 2003). Females were also shown to have a delayed time to generate maximal hamstring strength compared to their male counterparts (Huston and Wojtys, 1996) and decreased quadriceps muscle strength (White et al., 2003). These neuromuscular differences between males and females may be partially responsible for the differences we observed in the biomechanics of people with knee OA.

It remains unclear from this study whether OA causes these specific biomechanical differences in females or if they are an effect of the OA process. To our knowledge, only three other studies have reported on gender differences in OA gait (Kaufman et al., 2001; Wada et al., 2001; Weidenhielm et al., 1994), however none of these have explored the interaction between gender and disease. To the author's knowledge this 2-way ANOVA design has been the most comprehensive method to date to explore this cause/effect relationship between gender biomechanics and OA, however long-term follow-up studies are required. Furthermore, the analysis of covariance techniques confirmed that the gait differences found in this study cannot be explained by these clinical variables and are therefore due to other mechanism not captured in this study.

These results have implications for conservative and surgical treatments for OA. Shoe orthotics, knee braces and total knee replacements are currently designed to modify the biomechanical environment of the knee. This study has shown that in the presence of OA females walk significantly different, with respect to certain gait biomechanics, than healthy females. These differences were not found in males. Therefore, gender specific design of biomechanical interventions to slow the progression of OA should be explored. Furthermore, stratifying biomechanical relationships by gender is instrumental in order to properly characterize the disease process.

References

Andrews, M., Noyes, F.R., Hewett, T.E., Andriacchi, T.P., 1996. Lower Limb Alignment and Foot Angle are Related to Stance Phase Knee Adduction in Normal Subjects: A Critical Analysis of the Reliability of Gait Analysis Data, pp. 289–295.

Andriacchi, T.P., Ogle, J.A., Galante, J.O., 1977. Walking Speed as a Basis for Normal and Abnormal Gait Measurements, pp. 261–268.

Andriacchi, T.P., Galante, J.O., Fermier, R.W., 1982. The influence of Total Knee-replacement Design on Walking and Stair-climbing, pp. 1328–1335.

Baliunas, A.J., Hurwitz, D.E., Ryals, A.B., Karrar, A., Case, J.P., Block, J.A., Andriacchi, T.P., 2002. Increased knee joint loads during walking are present in subjects with knee osteoarthritis. Osteoarthr. Cartilage 10, 573–579.

Bellamy, N., Buchanan, W.W., Goldsmith, C.H., Campbell, J., Stritt, L., 1988. Validation Study of WOMAC: A HealthStatus Instrument for Measuring Clinically Important Patient-relevant Outcomes Following Total Hip or Knee Arthroplasty in Osteoarthritis, pp. 95–108.

Bilodeau, M., Schindler-Ivens, S., Williams, D.M., Chandran, R., Sharma, S.S., 2003. EMG frequency content changes with increasing force and during fatigue in the quadriceps femoris muscle of men and women. J. Electromyogr. Kinesiol. 13, 83–92.

Braune, W., Fischer, O., 1895. Der Gang des Menschen [The human gait]. B.G. Teubner, Leipzig.

Bridges, A.J., Smith, E., Reid, J., 1992. Joint hypermobility in adults referred to rheumatology clinics. Ann. Rheum. Dis. 51, 793–796.

Buckwalter, J.A., Lappin, D.R., 2000. The disproportionate impact of chronic arthralgia and arthritis among women. Clin. Orthop. Relat. Res., 159–168.

Cappozzo, A., Catani, F., Della Croce, U., Leardini, A., 1995. Position and orientation of bones during movement: anatomical frame definition and determination. Clin. Biomech. 10, 171–178.

Challis, J.H., 1995. A procedure for determining rigid body transformation parameters. J. Biomech. 28, 733–737.

Childs, J.D., Sparto, P.J., Fitzgerald, G.K., Bizzini, M., Irrgang, J.J., 2004. Alterations in lower extremity movement and muscle activation patterns in individuals with knee osteoarthritis. Clin. Biomech. 19, 44–49.

Cho, S.H., Park, J.M., Kwon, O.Y., 2004. Gender differences in three dimensional gait analysis data from 98 healthy Korean adults. Clin. Biomech. 19, 145–152.

Clausser, E.C.M., 1969. Weight, volume and centre of mass of segments of the human body. AMRL Technical Report, Wright Patterson Air Force Base, Ohio (National Technical Information Service No. AD-710-622).

Costigan, P.A., Wyss, U.P., Deluzio, K.J., Li, J., 1992. A Semi-automatic 3D Knee Motion Assessment System, pp. 343–350.

Cureton, K.J., Collins, M.A., Hill, D.W., McElhannon Jr., F.M., 1988. Muscle hypertrophy in men and women. Med. Sci. Sports Exer. 20, 338–344.

Deluzio, K.J., Wyss, U.P., Zee, B., Costigan, P.A., Sorbie, C., 1997. Principal component models of knee kinematics and kinetics: normal vs pathological gait patterns. J. Human Movement Sci. 16, 201–217.

Drinkwater, B., 1988. Training of female athletes. In: HG, A.K., Dirix, T.K. (Eds.), The Olympic Book of Sports Medicine. Blackwell Scientific Publications, Oxford, pp. 309–327.

Fleiss, J.L., 1986. Reliability of measurement. In: The Design and Analysis of Clinical Experiments. John Wiley & Sons, Toronto, pp. 1–32.

Grood, E.S., Suntay, W.J., 1983. A Joint Coordinate System for the Clinical Description of Three-Dimensional Motions: Application to the Knee, pp. 136–144.

Hannan, M.T., Felson, D.T., Pincus, T., 2000. Analysis of the discordance between radiographic changes and knee pain in osteoarthritis of the knee. J. Rheumatol. 27, 1513–1517.

Horton, M.G., Hall, T.L., 1989. Quadriceps femoris muscle angle: normal values and relationships with gender and selected skeletal measures. Phys. Ther. 69, 897–901.

Hreljac, A., Marshall, R.N., 2000. Algorithms to determine event timing during normal walking using kinematic data. J. Biomech. 33, 783–786.

Hsu, R.W.W., Himeno, S., Coventry, M.B., Chao, E.Y.S., 1990. Normal Axial Alignment of the Lower Extremity and Load Bearing Distribution at the Knee, pp. 215–227.

Hurwitz, D.E., Sumner, D.R., Andriacchi, T.P., Sugar, D.A., 1998. Dynamic knee loads during gait predict proximal tibial bone distribution. J. Biomech. 31, 423–430.

Huston, L.J., Wojtys, E.M., 1996. Neuromuscular performance characteristics in elite female athletes. Am. J. Sports Med. 24, 427–436.

Jackson, J.E., 1991. A User's Guide to Principal Components. John Wiley & Sons, New York.

Jones, M.C., Rice, J.A., 1992. Displaying the Important Features of Large Collections of Similar Curves, pp. 140–145.

Kaufman, K.R., Hughes, C., Morrey, B.F., Morrey, M., An, K., 2001. Gait characteristics of patients with knee osteoarthritis. J. Biomech. 34, 907–915.

Kellgren, J.H., Lawrence, J.S., 1957. Radiological Assessment of Osteo-arthrosis, pp. 494–501.

Kerrigan, D.C., Todd, M.K., Della Croce, U., 1998. Gender differences in joint biomechanics during walking. Am. J. Phys. Med. Rehabil. 77, 2–7.

Kerrigan, D.C., Riley, P.O., Nieto, T.J., della, C.U., 2000. Knee joint torques: a comparison between women and men during barefoot walking. Arch. Phys. Med. Rehabil. 81, 1162–1165.

March, L.M., Bagga, H., 2004. Epidemiology of osteoarthritis in Australia. Med. J. Australia 180, S6–S10.

Messier, S.P., Loeser, R.F., Hoover, J.L., Semble, E.L., Wise, C.M., 1992. Osteoarthritis of the knee: effects on gait, strength, and flexibility (published erratum appears in Arch. Phys. Med. Rehabil., 1992 Mar, 73 (3), 252). Arch. Phys. Med. Rehabil. 73, 29–36.

Miyazaki, T., Wada, M., Kawahara, H., Sato, M., Baba, H., Shimada, S., 2002. Dynamic load at baseline can predict radiographic disease progression in medial compartment knee osteoarthritis. Ann. Rheum. Dis. 61, 617–622.

Schnitzer, T.J., Popovich, J.M., Andersson, B.J., Andriacchi, T.P., 1993. Effect of Piroxicam on Gait in Patients with Osteoarthritis of the Knee, pp. 1207–1213.

Scott, W.W., Lethbridg-Cejku, M., Reichle, R., Wigley, F.M., Tobin, J.D., Hochberg, M.C., 1993. Reliability of Grading Scales for Individual Radiographic Features of Osteoarthritis of the Knee, pp. 497–501.

Wada, M., Maezawa, Y., Baba, H., Shimada, S., Sasaki, S., Nose, Y., 2001. Relationships among bone mineral densities, static alignment and dynamic load in patients with medial compartment knee osteoarthritis. Rheumatology (Oxford) 40, 499–505.

Wang, J.-W., Kuo, K.N., Andriacchi, T.P., 1990. The Influence of Walking Mechanics and Time on the Results of Proximal Tibial Osteotomy, pp. 905–913.

Ware, J.E., Snow, K.K., Kosinski, M., 2000. SF-36 Health Survey: Manual and Interpretation Guide. Quality Metric Inc., Lincoln, RI.

Weidenhielm, L., Svensson, O.K., Brostrom, L.A., Mattsson, E., 1994. Adduction moment of the knee compared to radiological and clinical parameters in moderate medical osteoarthrosis of the knee. Ann. Chir. Gynaecol. 83, 236–242.

White, K.K., Lee, S.S., Cutuk, A., Hargens, A.R., Pedowitz, R.A., 2003. EMG power spectra of intercollegiate athletes and anterior cruciate ligament injury risk in females. Med. Sci. Sports Exer. 35, 371–376.