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Gait characteristics of patients with knee osteoarthritis

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Abstract

The knee kinematics and kinetics of 139 patients (47 males and 92 females) with Grade II knee osteoarthritis (OA) were measured during level walking, stair ascent and stair descent. There was no significant difference in knee motion between the patients and normal subjects. The patients with knee OA had a significantly reduced internal knee extensor moment compared to normal subjects. This difference reflects the patient's compensation to reduce the knee joint loading. Further, subjects with OA and a higher body mass index have a lower knee extensor moment. The female subjects had significantly greater knee flexion and a greater knee extensor moment. This gender difference may partially explain the increased prevalence of OA in females. Most tests of OA treatments are assessed by criteria that do not reflect functional activities. This study demonstrates that objective gait analysis can be used to document gait adaptations used by patients with knee OA. © 2001 Elsevier Science Ltd. All rights reserved.

Keywords: Knee; Osteoarthritis; Gait; Body mass index; Stairs

1. Introduction

Osteoarthritis (OA) is the most prevalent form of arthritis in the elderly. This disease is particularly disabling when the knees are affected. It is estimated that 9% of men and 18% of women over age 65 have knee OA (Davis et al., 1991). Individuals with knee OA experience pain, stiffness, and decreased range of motion of the joints. These symptoms significantly limit an individual's ability to rise from a chair, stand comfortably, walk, or climb stairs. Ultimately, these limitations lead to a loss of functional independence.

The mechanics of walking on a level surface are well understood and characterized (Bresler and Frankel, 1950; Perry, 1992). Several studies have also analyzed the mechanics of stair walking. Kinematic and kinetic data of the lower limb in the sagittal plane during stair ascent and descent have been described in the sagittal plane (Andriacchi et al., 1980; McFadyen and Winter, 1988), and the frontal plane (Andriacchi et al., 1980; Kowalk et al., 1996; Yu et al., 1996). However, all of these studies have been conducted on able-bodied subjects. Few studies have quantified the gait changes associated with knee OA. One study has shown a reduction in sagittal plane knee motion (Stauffer et al., 1997). Three studies have shown a relationship between the knee adduction moment, OA disease severity and knee alignment (Schnitzer et al., 1993; Weidenhielm et al., 1994; Sharma et al., 1998). All of these studies were conducted on a level surface.

A greater understanding of gait will be useful for quantifying the pathomechanics of patients with knee OA. These patients experience pain and thus may compensate to minimize joint loading and resultant pain. Therefore, the purpose of this study was to analyze the gait characteristics of subjects with knee OA. The hypothesis tested was that the knee kinematics and kinetics would be significantly lower than normal in this population.

2. Methods

This study was performed on 139 adults diagnosed with knee OA. There were 47 males and 92 females in

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the study population. The subjects ranged in age from 30 to 82, and had a mean age of 57 years (± 12.5). Their mean weight was 85 kg (± 17) and their mean height was 167 cm (± 9.7). Subjects were recruited on a volunteer basis to participate in the study. Enrollment was sought by advertisements in local newspapers. Potential subjects were initially contacted by telephone. Those individuals who seemed to meet the inclusion criteria were asked to come for an initial exam. Once it had been determined that the patients met the inclusion/exclusion criteria (Table 1), they were asked to sign an informed consent form approved by the Institutional Review Board at Mayo Clinic.

As a basis for comparison, 20 healthy subjects were also studied. This group consisted of 9 males and 11 females. Subjects ranged in age from 20 to 42 and had a mean age of 30 (± 8). Their mean weight was 75 kg (± 17) and their mean height was 173 cm (± 11). These subjects had no history of knee OA, knee instability, or major lower extremity joint surgery. These individuals had normal strength, full range of motion of the lower extremities, and no neurologic deficits. These subjects were also recruited on a volunteer basis.

The walking conditions studied were those most commonly encountered during activities of daily living (Morlock et al., 2000), namely: level walking, ascending stairs, and descending stairs. All objective gait measurements were collected in the Biomechanics Laboratory. Kinematic parameters were acquired with a computerized motion analysis system utilizing six video cameras (Expertvision-Motion Analysis Corporation, Santa Rosa, CA). The spatial distribution of the cameras was optimized to yield reliable motion data at the hip, knee, and ankle, bilaterally. A set of 21 reflective markers was placed on the body of each subject as described by Kadaba et al. (1989). Markers were placed on the sacrum, bilaterally on the acromion processes, lateral epicondyle of the elbows, center of the dorsum of the wrists, anterior superior iliac spines (ASIS), lateral femoral condyles, lateral malleoli, the space between the first and second metatarsal heads, the heels, and on 10 cm wands placed at mid-thigh and mid-shank. The markers, placed at bony prominences, were used for establishing anatomic coordinate systems for the pelvis, thigh, shank, and foot. The motion analysis system was calibrated prior to each gait analysis. The calibration volume was $2 \times 2 \times 1$ m with a subsequent displacement accuracy to 1 mm and rotational measurement accuracy to 1° (An et al., 1991). One set of data corresponding to the standing position (static data) was recorded in order to calculate the location of the joint centers. Ground reaction load (GRL) data were collected at a sampling rate of 60 samples/s. Data from these force plates was time synchronized with the six motion cameras.

After a brief orientation session, the subject was asked to walk along a 12 m walkway. The stair ascent and descent was done without using a railing. The stairs were a flight of four, 18-cm high stairs with a 25 cm run. These dimensions were shown to be optimum for stairway design (Irvine et al., 1990). The first and second steps were independently attached to two separate force plates (Kistler Instrument Corp., Amherst, NY; model 9281B) and were structurally independent from the remaining stair structure. With this configuration, the GRLs that occurred during the support phase on the surface of each of these two steps was calculated from the dimension of the step together with the data provided by each attached force plate. The unique attachment of the staircase to the force plates was designed so that the measurement of GRLs on the surface of the staircase would not interfere with the

Table 1
Subject enrollment criteria

Inclusion criteria	*Exclusion criteria*
• 30 years of age or older. • Current symptoms of chronic stable (6-month) pain and stiffness in one or both knees during weight-bearing activities. • Involved joint is primary factor limiting physical or functional activity. • Joint pain with passive range of motion. • Intermittent effusion by history or physical exam. • Radiographic signs of hypertrophic changes, marginal spur formation, subchondral sclerosis or cyst formation, or nonuniform joint space narrowing.	• Subjective complaint of instability/"giving way" by history. • Ligamentous instability > Grade I. • Knee flexion contracture > 5°. • Asymptomatic osteoarthritis of one or both knees, incapacitating arthritis, or inflammatory arthritis. • Major lower extremity joint surgery, e.g., knee arthotomy. • Multiple major joint involvement. • Any condition which severely limits local ambulation, such as amputation or stroke. • Use of gait aids for ambulation. Inability to walk on treadmill without assistive device. • Cannot use step-over-step techniques in either ascending or descending stair conditions. • Dementia or inability to give informed consent.

performance of the subjects. The GRL measurement has been calibrated (Yu et al., 1996). A third force plate (Bertec Corporation, Worthington, OH; model 4060A) positioned immediately before the stairs allowed collection of GRL data on ground level.

The video records were tracked to obtain the three-dimensional (3D) coordinates of the markers using the ADTECH motion analysis software system (AMASS). The 3D marker trajectories were smoothed using a 2-pole, low-pass Butterworth filter implemented with time reversal to induce zero phase lag and set to a cut-off frequency of 7.4 Hz given the 60 Hz sampling frequency (Yu, 1988). The 3D marker coordinates and force plate data were used as input to a commercial software program, OrthoTrak 4.0 (Motion Analysis Corp., Santa Rosa, CA), to calculate the joint kinematics and kinetics. The OrthoTrak 4.0 program was used to create the joint center approximations and segment coordinate systems from the 3D marker trajectories, as well as the subsequent rigid body kinematic/kinetic calculations. Embedded right-hand Cartesian coordinate systems were used in this model to describe the position and orientation of the lower extremity rigid body segments. The unit vector formulations using the marker coordinates for this model were the same as those described by Kadaba et al. (1990), with the exception of the foot. A 3D coordinate system was used for the foot with an anteriorly directed x-axis, y-axis pointing to the body's left side, and a superiorly directed z-axis. With these embedded coordinate systems, the joint angles were determined using the floating-axis or Euler angle convention described by An and Chao (1984). Using the ASIS marker locations and the orientation of the pelvis coordinate system, the hip center location was found (Vaughan et al., 1992). The knee center was found on a vector directed medially from the knee marker and at a distance of one-half of the measured knee width. Similarly, the ankle center was located by a vector directed medially from the lateral malleolus marker at one-half the distance of the measured ankle width. The segmental joint forces and moments were calculated based on these coordinate axes.

A set of custom computer programs was used for data reduction. The gait cycle was the time from foot contact to ipsilateral foot contact. All gait events were expressed as a percentage of the gait cycle, irrespective of the actual time for a stride, to yield a normalized gait cycle. For the stair ascending condition, the gait cycle was defined for the foot strike beginning on the first stair through foot strike on the third stair. For stair descending condition, the gait cycle was defined as the time period from foot contact on the second stair to foot contact on the floor level force plate. The knee joint moments were normalized to body weight and body height and were expressed as net internal moments. Results were averaged from three trials for each gait condition.

The weight and height measures of each subject were used to determine their body mass index (BMI). The BMI was calculated by dividing the weight (in kilograms) by the square of the height (in meters). This index provides a relative measure of obesity (Whitney and Rolfes, 1999). Normal weight is defined as $18 \leqslant \mathrm{BMI} \leqslant 25$. Overweight is defined as $25 \leqslant \mathrm{BMI} \leqslant 30$. People are defined as obese when $30 \leqslant \mathrm{BMI} \leqslant 40$, and severely obese when $\mathrm{BMI} > 40$.

The SAS Statistical Analysis System (SAS Institute, Inc., Cary, NC) was used for data analysis. The gait parameters for the involved leg of each subject were used. For subjects with bilateral involvement, the mean of the gait parameters from both limbs was used. A repeated measures Analysis of Variance was used to test for significant differences in velocity, joint angles, and gait cycle between the normal subjects and the patients with OA. A repeated measures Analysis of Covariance was used to control for differences in gait velocity when making comparisons for the knee moments. Statistical differences were defined as significant at the $\alpha = 0.05$ level.

3. Results

The subjects with OA walked slower than the normal subjects (Table 2). These differences in walking velocity were statistically significant ($p < 0.01$).

The knee kinematic patterns were different for each of the walking conditions (Fig. 1). During level walking there was an initial loading response (knee flexion) with the greatest knee flexion during swing. During stair ascent the knee started in a flexed position and extended throughout the entire stance phase. Conversely, during the stance phase of stair descent, the knee started at an extended position and flexed throughout the stance cycle reaching a maximum at toe off. For all walking conditions, the maximum knee flexion angle occurred during swing phase.

The knee flexion angle on stairs was greater than on level ground (Fig. 2). The maximum knee flexion angle for the patients with OA did not differ significantly from the flexion angle for able-bodied subjects ($p = 0.12$).

Table 2
Walking velocity (cm/s)[a]

Group	Level	Up stairs	Down stairs
Normal	117 (14)	57 (9)	71 (12)
OA	109 (11)	48 (9)	59 (13)

[a] Data are mean with standard deviations in parentheses.
$p = 0.0001$.

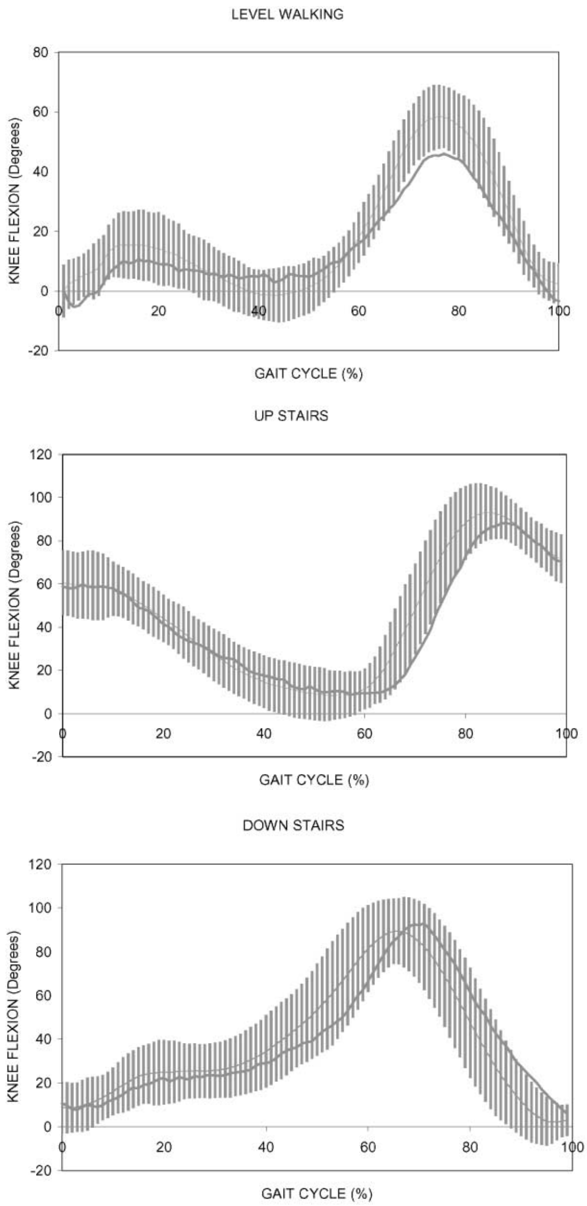

Fig. 1. Knee flexion during level walking, stair ascent, and stair descent for a typical subject in the study. The gait cycle begins with stance. Toe-off occurred at 66% gait cycle. The shaded region represented the normal range, defined as ± 2 standard deviations from the mean.

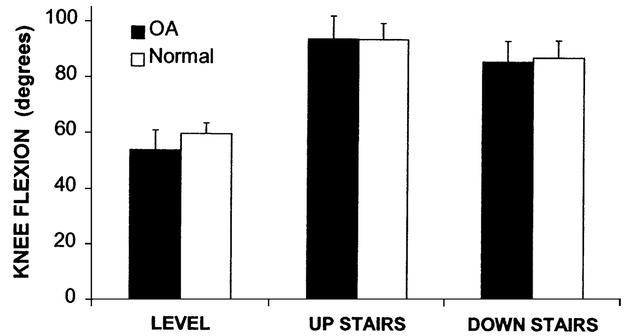

Fig. 2. Maximum knee flexion during gait for patients with OA compared to normal subjects. There is no significant difference in the knee flexion angle ($p = 0.12$).

During level walking the patients with OA had 6° less peak knee motion than normal subjects (54 ± 7 vs. 60 ± 4, $p < 0.01$). The knee flexion between the two groups differed by less than two degrees when using the stairs. There was no significant difference in the time of maximum knee flexion for all three walking conditions ($p = 0.44$).

The knee kinetic pattern for the patients with OA was similar to the normal subjects (Fig. 3). During level walking, the knee experienced a flexor moment initially, followed by an extensor moment to provide knee stability. During the last half of stance a flexor moment existed. During stair ascent, there was an initial internal knee extensor moment required to initiate the upward displacement of the body. During late stance, the knee moment changed to an internal flexor moment. Conversely, during stair descent the initial knee moment was an internal flexor moment that changed to an internal knee extensor moment during most of stance. The peak knee extension moment during stair descent occurred during late stance, in contrast to the peak extension moment occurring during early stance for level walking and stair ascent.

The patients with OA have a reduced knee peak extension moment compared to the normal subjects (Fig. 4). The reduced moment was statistically significant ($p = 0.02$). The timing of the maximum knee extension moment did not differ significantly between the two groups ($p = 0.07$). Similarly, the knee flexion angle at the peak moment was not significantly different between the two groups ($p = 0.64$). A peak moment occurred earlier in stance during stair ascent (19 ± 7 and 32 ± 23% gait cycle for normal and OA, respectively) than for level walking (48 ± 19 and 44 ± 19% gait cycle for normal and OA, respectively) or stair descent (51 ± 6 and 54 ± 4% gait cycle for normal and OA, respectively). The knee flexion angle at the peak moment was closer to extension for level walking than for stair walking. The knee flexion angle at the peak moment during level walking was 20 ± 5 and 21 ± 9 for normal and OA, respectively. During stair ascent, the knee flexion angle was 54 ± 6 and 57 ± 14 for normal and OA, respectively. During stair descent, the knee flexion angle at the maximum moment was $48 \pm 9°$ and $44 \pm 19°$ for normal and OA, respectively.

The other knee moments were also compared between subjects with OA and normal subjects (Table 3). The varus moment was significantly increased in the patients with OA ($p = 0.02$). The knee was slightly more extended when the peak varus moment was generated in the subjects with OA ($p = 0.03$). The flexion moment

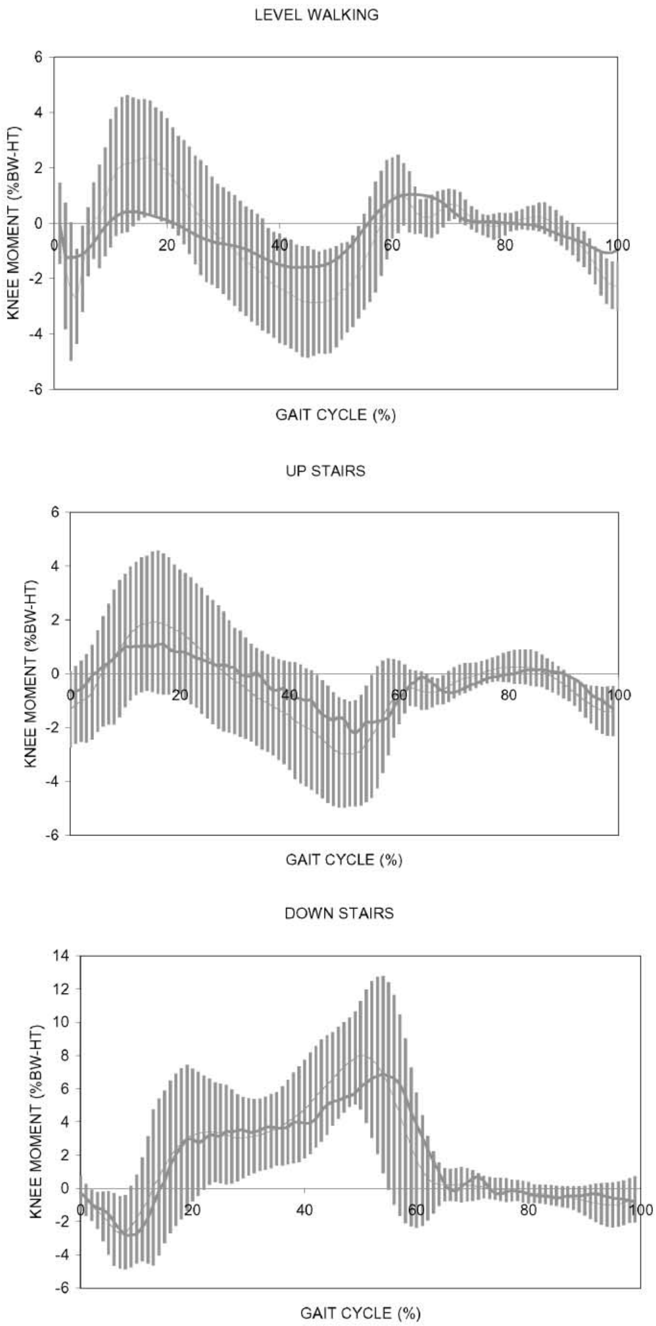

Fig. 3. Sagittal plane moments for a typical subject during level walking, stair ascent, and stair descent. A positive value is an internal knee extensor moment. The gait cycle begins with stance. Toe-off occurred at 66% gait cycle. The shaded region represented the normal range, defined as ± 2 standard deviations from the mean.

and rotational moments did not display any significant differences between the two groups.

Gender differences were identified in the patients with OA. The female subjects had a significantly greater ($p = 0.0001$) peak knee flexion (Fig. 5a) while there was no significant difference in the time of peak knee flexion. The difference in peak knee flexion is most likely due to a significant difference ($p = 0.001$) in height between the female and male subjects, 162 ± 6 cm vs. 177 ± 8 cm, respectively. However, there was no significant difference ($p = 0.35$) in walking velocity between the female and male OA subjects (Table 4). The female OA subjects generated significantly greater ($p = 0.01$) peak knee extension moments (Fig. 5b). The normal female subjects also generated greater knee extensor moments than the normal male subjects, but the difference was less significant ($p = 0.07$). For both the normal and OA subjects, the females generated greater external rotation moments than the male subjects, with the exception of the level walking condition for the OA subjects ($p \leqslant 0.03$). There were no other significant differences for the OA subjects (Table 5) or the normal subjects.

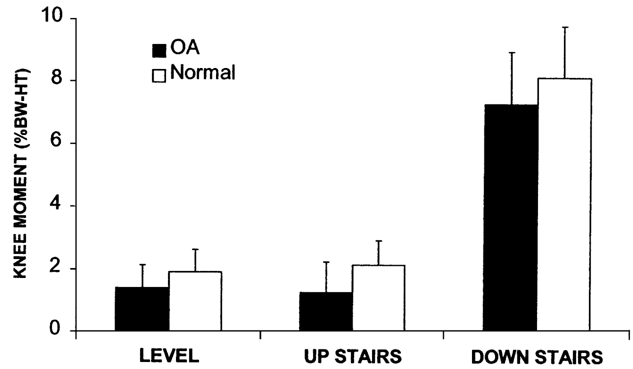

Fig. 4. Maximum knee internal extension moment during level walking, stair ascent and stair descent. The patients with osteoarthritis have a lower moment for all walking conditions. The difference was statistically significant ($p = 0.02$).

All the subjects in this study could be considered to be at normal weight or overweight. None of the subjects in this study had a BMI index <21. There were 17.3% of the subjects that could be considered of normal weight. An additional 38.8% of the subjects could be considered overweight. Over one-third of the subjects (37.4%) could be considered obese. Finally, 6.5% of the subjects could be considered severely obese. As the BMI increased, the knee extension moment decreased (Fig. 6). The decline in knee extension moment was nearly identical for all ground conditions. Thus, individuals with increased BMI demonstrated a greater compensation to reduce the joint loading by reducing the knee extension moment.

4. Discussion

Walking is a common functional activity of daily living. This study provides meaningful information on gait adaptations used by patients with knee OA. These adaptations provide pain relief from the dynamic joint loading encountered during gait. Subjects with OA exhibited significantly lower knee extensor moments

Table 3
Gait characteristics of patients with OA compared to normal subjects[a]

	Level walking		Upstairs		Downstairs		p-level
	OA	Normal	OA	Normal	OA	Normal	
Flexion moment							
Peak value (%BW-HT)	2.49 (0.81)	3.10 (0.75)	3.08 (1.06)	3.03 (0.62)	3.25 (1.16)	2.28 (0.55)	0.38
Knee flexion (degrees)	2 (6)	1 (4)	13 (15)	12 (13)	8 (8)	11 (4)	0.74
Gait cycle (%)	52 (18)	56 (11)	55 (11)	52 (13)	19 (22)	19 (25)	0.81
Valgus moment							
Peak value (%BW-HT)	2.23 (0.82)	2.60 (0.76)	1.52 (1.03)	1.53 (0.60)	2.50 (2.00)	3.08 (1.76)	0.41
Knee flexion (degrees)	9 (8)	11 (9)	33 (20)	33 (20)	35 (19)	41 (16)	0.44
Gait cycle (%)	43 (16)	46 (12)	41 (19)	41 (16)	52 (19)	48 (16)	0.94
Varus moment							
Peak value (%BW-HT)	0.39 (0.28)	0.36 (0.36)	1.03 (0.73)	0.93 (0.41)	1.69 (1.46)	0.75 (0.37)	0.02
Knee flexion (degrees)	32 (15)	35 (13)	39 (23)	49 (14)	45 (22)	54 (23)	0.03
Gait cycle (%)	45 (26)	49 (23)	61 (23)	55 (27)	50 (21)	56 (21)	0.67
External rotation moment							
Peak value (%BW-HT)	0.59 (0.29)	0.70 (0.28)	0.87 (0.87)	0.60 (0.35)	2.24 (0.77)	2.27 (0.50)	0.20
Knee flexion (degrees)	13 (9)	16 (6)	33 (20)	29 (13)	36 (13)	44 (11)	0.40
Gait cycle (%)	32 (15)	32 (7)	41 (20)	48 (14)	44 (16)	51 (5)	0.03
Internal rotation moment							
Peak value (%BW-HT)	1.39 (0.43)	1.60 (0.31)	1.94 (1.05)	2.36 (0.78)	0.90 (0.59)	0.46 (0.28)	0.87
Knee flexion (degrees)	5 (6)	4 (4)	27 (15)	33 (11)	22 (19)	22 (15)	0.48
Gait cycle (%)	66 (10)	68 (1)	35 (14)	31 (8)	31 (23)	24 (18)	0.15

[a] Data are the mean with standard deviations in parentheses.

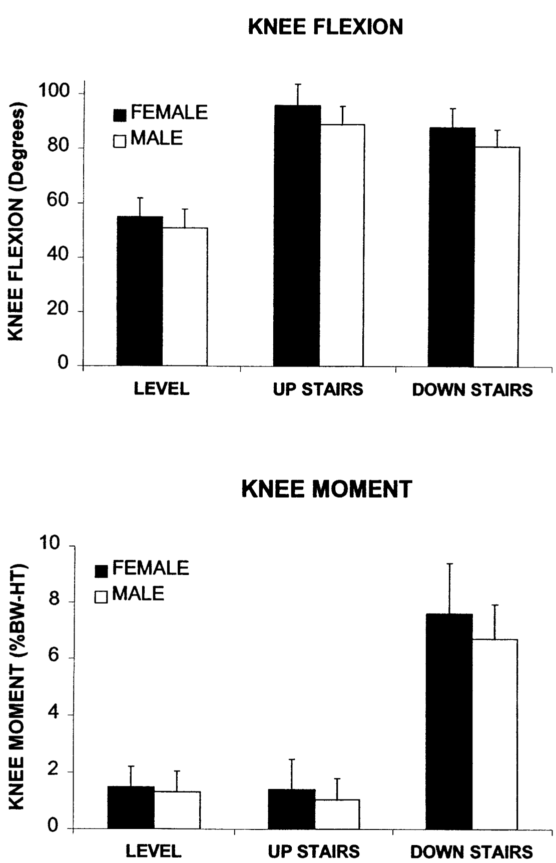

Fig. 5. Gender differences in knee mechanics during gait for patients with OA. The female subjects had significantly greater knee flexion ($p = 0.0001$) and significantly greater knee moments ($p = 0.01$) for all walking conditions.

Table 4
Gender differences in walking velocity for subjects with OA[a]

Gender	Level	Up Stairs	Down Stairs
Female	112 (11)	48 (9)	58 (14)
Male	103 (9)	50 (8)	60 (11)

[a] Velocity is reported in cm/s.
Data are mean with standard deviations in parentheses.
$p = 0.35$.

during gait. The contact forces in the knee joint are proportional to the net external reaction moment. A large internal moment, needed to balance a large external moment, will produce a large contact force. The subjects with OA attempted to minimize their pain by reducing the knee extensor moment. Thus, these reductions represent the intent by the subjects to reduce their pain by minimizing knee joint loading. Further, it can be noted that the highest extension moment occurred while descending stairs. Thus, stair descent is a more stressful activity. The results of this study agree with other studies that show the demands of stair walking produce larger external moments (Andriacchi et al., 1980; Kowalk et al., 1996; Yu et al., 1997) and a 12–25% increase in knee loading (Morrison, 1969). However, these other studies have all been performed on able-bodied individuals.

The patients with OA did not demonstrate a reduction in knee range-of-motion during either stair ascent or stair descent. This may be due to the early stage of OA studied. Further, it is important to note that

Table 5
Gender comparison of gait characteristics in patients with OA[a]

	Level walking		Up stairs		Down stairs		p-level
	Female	Male	Female	Male	Female	Male	
Extension moment							
Peak value (%BW-HT)	1.47 (0.73)	1.31 (0.73)	1.40 (1.07)	1.04 (0.75)	7.61 (1.81)	6.72 (1.22)	0.01
Knee flexion (degrees)	21 (9)	21 (7)	57 (15)	56 (11)	48 (9)	43 (8)	0.16
Gait cycle (%)	42 (19)	47 (19)	32 (24)	33 (23)	55 (3)	54 (4)	0.25
Flexion moment							
Peak value (%BW-HT)	2.51 (0.87)	2.46 (0.66)	2.94 (1.10)	3.37 (0.92)	3.33 (1.20)	3.14 (1.05)	0.74
Knee flexion (degrees)	3 (6)	1 (5)	15 (17)	8 (10)	9 (9)	6 (4)	<0.01
Gait cycle (%)	52 (18)	54 (19)	55 (12)	54 (7)	20 (23)	18 (18)	0.53
Valgus moment							
Peak value (%BW-HT)	2.21 (0.85)	2.29 (0.77)	1.43 (1.06)	1.69 (0.97)	2.75 (2.22)	1.96 (1.32)	0.57
Knee flexion (degrees)	10 (8)	6 (8)	35 (20)	29 (20)	39 (19)	27 (16)	<0.01
Gait cycle (%)	42 (15)	45 (17)	42 (18)	41 (20)	53 (16)	50 (25)	0.76
Varus moment							
Peak value (%BW-HT)	0.40 (0.29)	0.39 (0.26)	1.16 (0.78)	0.77 (0.53)	1.73 (1.49)	1.62 (1.44)	0.18
Knee flexion (degrees)	30 (16)	35 (13)	38 (24)	41 (22)	44 (23)	47 (21)	0.18
Gait cycle (%)	50 (23)	49 (23)	61 (23)	62 (24)	49 (21)	53 (19)	0.09
External rotation moment							
Peak value (%BW-HT)	0.55 (0.28)	0.64 (0.32)	0.99 (0.95)	0.60 (0.59)	2.33 (0.83)	2.10 (0.63)	0.02
Knee flexion (degrees)	14 (10)	11 (9)	35 (20)	29 (18)	37 (14)	36 (12)	0.05
Gait cycle (%)	34 (17)	30 (12)	39 (20)	44 (21)	42 (16)	46 (15)	0.67
Internal rotation moment							
Peak value (%BW-HT)	1.33 (0.44)	1.55 (0.36)	1.84 (0.99)	2.15 (1.14)	0.96 (0.58)	0.78 (0.60)	0.08
Knee flexion (degrees)	6 (6)	3 (6)	27 (15)	27 (15)	24 (19)	18 (17)	0.06
Gait cycle (%)	65 (11)	68 (9)	36 (15)	33 (13)	33 (23)	28 (24)	0.40

[a] Data are the mean with standard deviations in parentheses.

the peak knee moment during stair ascent and descent occurred at a flexion angle of about 50° whereas during level walking the largest moment occurred when the knee was near full extension (~20°).

Female gender is a significant risk factor for OA. A longitudinal study of knee OA showed that women have a 1.8 times greater risk of developing OA than men (Felson et al., 1997). Arthritis is more prevalent among women than among men at all ages (Davis et al., 1991). These gender differences are most prominent when OA affects the knee. For all grades of radiographic severity of OA, more women than men report knee pain. Among those over age 65 years, the prevalence of symptomatic knee arthritis in women is twice the rate in men (Davis et al., 1991). The exact etiology for this difference in prevalence is unknown. The female subjects had significantly greater knee extension moments than their male counterparts. This increased knee loading may be partially responsible for the increased prevalence of OA in females.

Studies of knee mechanics during gait have been conducted in normal subjects. The knee motion reported in this study agrees with a previous report by Kadaba (Kadaba et al., 1990). The knee kinematics during stair ascent for the normal subjects in the present study is similar to the values reported by Livingstone et al. (1991) and greater than the knee flexion reported by Andriacchi (Andriacchi et al., 1980). Stair dimension and subject height are important factors in determining the angular motion of the knee during tasks of stair ascent and descent (Livingstone et al., 1991). A possible reason for this difference could be that although the stairs were slightly higher, the subjects were also taller in the study conducted by Andriacchi (Andriacchi et al., 1980). The knee moments during level walking are similar to previous reports (Eng and Winter, 1995; Hurwitz et al., 1998; Kerrigan et al., 1998; Baliunas et al., 2000). The sagittal plane knee moments in this study are smaller in magnitude than previous reports during stair ascent but similar during stair descent (Andriacchi et al., 1980; McFadyen and Winter, 1988; Kowalk et al., 1996). These differences are most likely due to stair height differences. For similar reasons, the knee valgus moment was less in the present study than reported by others (Andriacchi et al., 1980; Kowalk et al., 1996). The knee rotational moment was greater during stair ascent and lower during stair descent than previously reported by Andriacchi (Andriacchi et al., 1980). So, in general, there is good agreement with previously published values on normal subjects during level walking. During stair walking, the moments reported in this study are somewhat lower due to a difference in step height.

The knee mechanics for patients with knee OA have not been studied as extensively. Previous studies have examined knee mechanics during level walking only

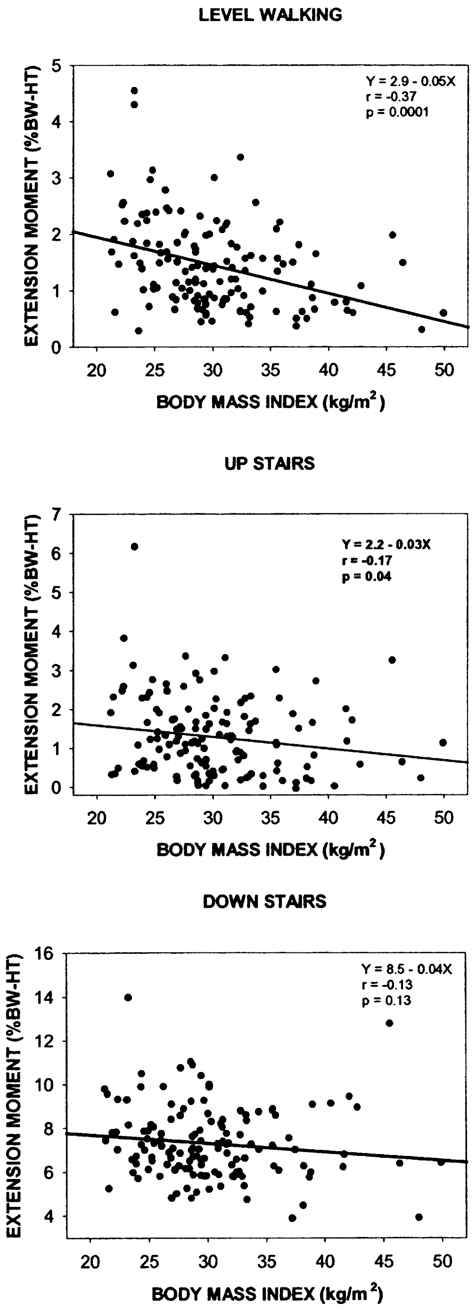

Fig. 6. Relationship between knee extension moment and body mass index. Subjects with an increased body mass index protected their knee joint by decreasing the knee extensor moment. This gait pattern was consistent for all walking condition.

(Schnitzer et al., 1993; Sharma et al., 1998; Baliunas et al., 2000; McNichols et al., 2000). The knee extensor moment reported in this study is within the range reported by previous investigators (Schnitzer et al., 1993; Baliunas et al., 2000). The knee valgus moment reported in this study is similar to reports by Sharma and Hurwitz (1998) and McNichols et al. (2000), but is slightly less than previous reports by Schnitzer et al. (1993) and Baliunas et al. (2000). An explanation for these differences may be the stage of osteoarthritis studied, the BMI of the subjects, or differences in the choice of the abduction/adduction axis (Kowalk et al., 1996). This is the first study to report knee mechanics during stair ascent and descent, and thus, no comparison could be made.

Studies have emphasized the importance of dynamic knee joint loads using the external knee adduction moment. The external knee adduction moment has been shown to correlate with bone distribution between the medial and lateral compartments of the proximal tibia (Hurwitz et al., 1998). A greater adduction moment corresponds to increased load on the medial compartment relative to that of the lateral compartment (Schipplein and Andriacchi, 1991). Further, studies of subjects with knee OA have shown that surgical outcome (Prodromos et al., 1985; Wang et al., 1990), pain relief (Schnitzer et al., 1993), and radiographic disease severity (Sharma and Hurwitz, 1998) are related to the peak external knee adduction moment. The peak external knee adduction moment in these and other studies is equivalent to the knee valgus moment in this study. The present study did not demonstrate a significant difference in the knee valgus moment between normal subjects and patients with OA. Moreover, the patients with OA had lower valgus moments than the normal subjects. An explanation for this difference may be the knee compartment and the stage of OA studied, BMI of the subjects, the gait velocity, or differences in the biomechanical model used.

OA has been regarded as a "wear and tear" or "degenerative" condition. Overweight is a strong risk factor for disabling knee OA (Manninen et al., 1996; McAlindon et al., 1996). Several studies have demonstrated that individuals with an increased BMI are at an increased risk of developing knee OA (Schouten et al., 1992; Hochberg et al., 1995; Manninen et al., 1996; McAlindon et al., 1996; Felson et al., 1997). The subject population of this study confirms these previous studies, since most of the subject population (82%) would be considered overweight. However, the definition of overweight does not necessarily reflect increased body fat. A BMI >30 is almost always associated with excessive bodyweight except in body builders and other highly muscular people (Bray and York, 1992). Nonetheless, this study demonstrated that individuals with OA and an increased BMI have gait patterns that deviated farther from normal. Knee loading is related to body mass, and thus, biomechanical stresses are magnified in overweight subjects. The overweight subjects modified their gait pattern to minimize joint loading.

Subjects with OA compensate to reduce the knee extensor moment, and consequently the knee joint loading. By analyzing a patient's gait pattern, it is possible to define relevant quantitative criteria for judging the locomotor handicap caused by OA. This data can be used to assess the therapeutic effectiveness of nonsurgical interventions. Nonsurgical treatments for OA which are effective will result in a reduction in symptomatic pain and increased knee loading as indicated by an increased knee extensor moment.

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