

## Effects of severity of degeneration on gait patterns in patients with medial knee osteoarthritis

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### Abstract

This study tested the hypothesis that patients with mild and severe medial knee osteoarthritis (OA) adopt different compensatory gait patterns to unload the diseased knee, in not only the frontal plane but also the sagittal plane. Fifteen patients with mild and 15 with severe bilateral medial knee OA, and 15 normal controls walked while the kinematic and kinetic data were measured. Compared to the normal group, both OA groups had significantly greater pelvic anterior tilt, swing-pelvis list, smaller standing knee abduction, as well as smaller standing hip flexor and knee extensor moments during stance. The severe group also had greater hip abduction, knee extension and ankle plantarflexion. The mild group successfully reduced the extensor moment and maintained normal abductor moment at the diseased knee mainly through listing and anterior tilting the pelvis. With extra compensatory changes at other joints and increased hip abductor moment, the severe group successfully reduced the knee extensor moment but failed to reduce the abductor moment. These results suggest that, apart from training of the knee muscles, training of the hip muscles and pelvic control are essential in the rehabilitative intervention of patients with knee OA, especially for more severe patients.

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**Keywords:** Knee osteoarthritis; Gait; Pelvis-leg apparatus; Joint kinematics; Joint kinetics

### 1. Introduction

Osteoarthritis (OA), the most common form of arthritis, affects an increasing proportion of the population [1], with the knee being the most commonly affected joint in the lower limb [2]. Changes related to OA are more frequently observed in the medial compartment with a varus alignment and are associated with an increased external knee adduction moment during gait [3]. Much research effort has thus been devoted to the study of the clinical relevance of this

moment, such as disease severity [4], disease progression [5] and surgical outcome [6]. However, since walking is a three-dimensional, multi-joint movement, knowledge of how the lower limb joints work together in all three planes to compensate for the changes at the diseased knee may be helpful for the management of these patients and for the evaluation of treatment outcomes.

Many of the previous studies on OA knees during level walking have focused on reporting the variables of the knee joint [7–12]. Biomechanical changes in the frontal plane are important because varus alignment is frequently accompanied by knee OA [5,8,9,13]. Greater knee abductor moments were found in patients with knee OA, increasing with increased severity of the diseased knee [4]. Angles and moments of the knee in the sagittal plane were also found to be affected by this disease [7,14]. Reduced flexion angle,

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reduced flexor but increased internal rotator moments at the diseased knee were found in the OA group [7]. Since the knee joint does not function alone, especially during weight-bearing activities, changes in the knee may affect its adjacent joints, namely the hip and ankle, and vice versa.

A limited number of studies have reported that biomechanical changes caused by knee OA were found not only at the knee but also at the hip and ankle [15–17]. Chang et al. [18] found that the likelihood of medial OA progression was reduced by the increased hip abductor moments. However, they did not provide data on the knee. Increased pelvic motion [16], toe-out angles [3] and hip extension eccentric powers [15] were also found in patients with knee OA during level walking. Mundermann et al. [19] reported the loads at the joints of the lower extremities secondary to knee OA of varied severity. The changes of the joint moments in the frontal plane were noted and discussion of the compensatory strategy was focused in that plane too. While rapid shift of body weight from the contralateral limb to the support limb was suggested to be a potential strategy in reducing the frontal load at the knee, which was related to the altered position of the pelvis and trunk, the pelvic motions were not measured and only sagittal angles of the lower limb joints were reported. Potential compensatory strategies in the sagittal plane, including altered motion of the pelvis, in reducing the knee joint load were not noted. It seems that a more complete description of the compensatory gait patterns of the pelvis-leg apparatus, in not only the frontal plane but also the sagittal plane, in an attempt to unload the diseased knee in patients with knee OA is necessary for a better rehabilitative intervention of these patients.

Several potential factors may affect the observed compensatory gait patterns in patients with knee OA. Walking speed, severity of knee OA and its involved compartment were reported to affect the biomechanical changes during level walking. Reduced knee extensor moments were found in patients with knee OA in some studies [20], while others found that the knee extensor moments did not differ statistically between the OA and healthy groups [19,21]. These different results may have been caused by the different walking speeds between the subject groups. Increased walking speed was reported to increase stride length and stride rate [22,23], as well as peak knee flexion during the loading response phase and swing phase [24]. The lower limb joint moments also increased with walking speed, especially in the sagittal plane [25]. In addition, severity of OA and the involved compartment may affect the performance of level walking. However, most of the previous studies included a variety of patients with knee OA in the subject group [7,16,20], which did not allow for any differentiation of the biomechanical changes within different groups with knee OA. Hence, removing the influence of these factors by differentiating the type and severity of knee OA is necessary.

The purpose of the study was to investigate the kinematic and kinetic changes of the pelvis-leg apparatus in patients with mild and severe bilateral medial knee OA as compared

to their matched healthy subjects during level walking. It was hypothesized that patients with mild and severe medial knee OA adopt different compensatory gait patterns to unload the diseased knee, in not only the frontal plane but also the sagittal plane.

## 2. Materials and methods

### 2.1. Subjects

Forty-five subjects, namely fifteen normal controls and thirty patients with bilateral OA in the medial compartment of the knee, participated in this study after giving their written informed consent following the guidelines of the Institutional Human Research Ethics Committee. OA patients recruited by an orthopedic surgeon (TMW) at National Taiwan University Hospital were classified based on Kellgren–Lawrence (K/L) grades for both knees. Fifteen patients (9 female, 6 male, age:  $63.1 \pm 11.9$  years, height:  $161.9 \pm 6.5$  cm, mass:  $68.4 \pm 10.3$  kg) with grade 1 or 2 K/L OA were classified as the mild OA group, and 15 patients (2 male, 13 female, age:  $63.1 \pm 8.2$  years, height:  $156.0 \pm 8.9$  cm, mass:  $64.0 \pm 8.5$  kg) with grade 3 or 4 K/L OA were placed in the severe OA group. OA patients were not included if they had other neuromusculoskeletal diseases which might affect gait and/or cognitive function, such as rheumatoid arthritis, neuropathic arthropathy, stroke or fracture. If the subjects had received an intra-articular corticosteroid injection in the previous 2 months, or were planning for a total knee replacement in the next year, they were also excluded from the study. A group of 15 control subjects was matched to the OA group by age, height, and mass (9 female, 6 male, age:  $63.2 \pm 9.9$  years, height:  $159.3 \pm 8.1$  cm, mass:  $60.5 \pm 8.4$  kg). They were free from any musculoskeletal, cardiovascular, or neurological pathology, or any other disorders that might have affected the activity.

For the subjects with OA, the knee pain was assessed using 100-mm visual analog scales (VAS; 0 = no pain, 100 = worst imaginable pain) while their physical function evaluated using the short form 36 health survey questionnaire (SF-36) with eight multi-item variables. For each variable item, scores were transformed into a scale from 0 (worst possible health state) to 100 (best possible health state).

### 2.2. Gait analysis

In a gait laboratory, each subject walked at a self-selected, comfortable pace on an 8-m walkway. Twenty-eight markers were used to track the motion of the pelvis-leg segments [26]. Three-dimensional (3D) marker trajectories were measured using a 7-camera motion analysis system (Vicon 512, Oxford Metrics Group, UK) at a sampling rate of 120 Hz. After system calibration, the residual error was less than 1 mm. Data on the 3D ground reaction force (GRF) vector were collected at a frequency of 1080 Hz using two for-

Table 1  
Means and standard deviations (S.D.) of the spatiotemporal gait parameters in healthy subjects, and patients with mild and severe knee osteoarthritis

	Group						<i>p</i> -Value
	Control		Mild OA		Severe OA		
	Mean	S.D.	Mean	S.D.	Mean	S.D.	
Speed (m/s)	0.89	0.32	0.94	0.15	0.83	0.18	0.4237
Cadence (steps/min)	106.1	17.1	103.4	11.1	99.1	12.6	0.3774
Stride length (%LL)	1.22	0.44	1.28	0.11	1.25	0.25	0.839
Step width (m)	0.09	0.03	0.08	0.03	0.10	0.06	0.5661

ceplates (50.8 cm × 46.2 cm, OR-6-7-1000, AMTI, USA). Three successful gait cycles for each limb were recorded for each subject.

### 2.3. Data analysis

The pelvis-leg apparatus was modeled as a 7-link system, with each link embedded with a coordinate system with the positive *x*-axis directed anteriorly, the positive *y*-axis superiorly and the positive *z*-axis to the right. A cardanic rotation sequence (*Z–X–Y*) was used to describe the rotational movements of each joint [27]. Pelvic list was calculated as the angle between the *z*-axis (inter-ASIS axis) and horizontal while pelvic tilt as the angle in the *x–y* (sagittal) plane between the *y*-axis and vertical. In order to minimize the errors owing to skin movement artifacts, the pose of each segment was determined in terms of its transformation matrix by minimizing the marker array deformation from its reference shape in a least squares sense [28]. With the measured GRF and kinematic data, inverse dynamics were used to calculate the intersegmental forces and moments at the lower limb joints. Inertial properties for each body segment, namely segment mass, center of mass and moment of inertia, were obtained using Dempster's coefficients [29]. All the calculated joint moments were normalized to body weight and leg length (distance between ipsilateral ASIS and medial malleolus). For all the calculated variables, peak values and their values at the

beginning and end of single leg stance (SLS) were extracted for subsequent statistical analysis. Gait speed, cadence, step width and stride length were also calculated. Stride length was normalized by lower limb length (LL).

### 2.4. Statistical analysis

For all the calculated variables, one-way analysis of variance (ANOVA) was performed between patients with mild knee OA, severe knee OA, and matched control subjects. *Post hoc* Bonferroni tests were utilized in conjunction with the ANOVAs. All significance levels were set at  $\alpha = 0.05$ . SPSS version 11.0 (SPSS Inc., Chicago, USA) was used for all statistical analyses.

## 3. Results

Comparisons between the OA and healthy control groups showed no significant differences in the gait speed, cadence, step widths and stride lengths (Table 1). VAS and SF-36 scores were not significantly different between the OA groups, except for the physical functioning and vitality scores in the SF-36 (Table 2).

Around heel-strike, a more extended knee, plantarflexed ankle and abducted hip were found in the severe OA group compared to the healthy subjects. Similar results were also

Table 2  
Means and standard deviations (S.D.) for visual analog scales (VAS) and eight variables of the SF-36 in patients with mild and severe knee osteoarthritis (\**p* < 0.05)

Variable	Group				<i>p</i> -Value
	Mild OA		Severe OA		
	Mean	S.D.	Mean	S.D.	
Knee pain (VAS scale)	41.7	15.7	46.1	17.1	0.476
SF-36 subscales					
Physical functioning	72.0	14.5	56.3	18.3	0.023*
Social functioning	64.6	31.0	66.7	19.5	0.846
Role limitations					
Physical	60.0	35.1	41.7	36.2	0.170
Emotional	78.1	23.5	58.5	34.9	0.230
Mental health	79.0	20.1	70.7	16.1	0.221
Energy/vitality	81.7	11.9	72.3	11.8	0.040*
Pain	60.5	17.9	56.3	17.2	0.511
General health perceptions	67.1	12.1	60.5	16.3	0.434

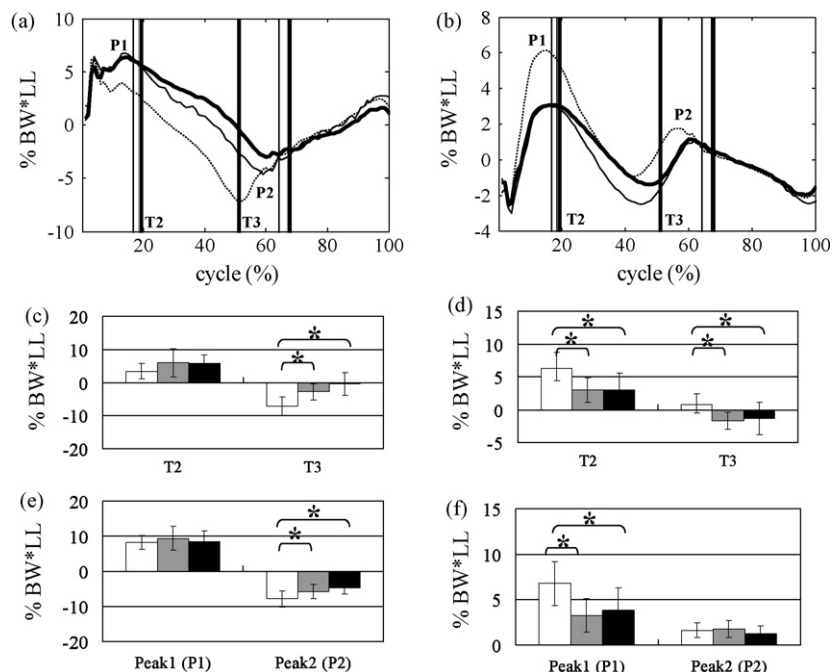


Fig. 1. Sagittal patterns, peaks during the early stance and late stance, values at the beginning and the end of the single leg stance of hip and knee moments between the control (dashed lines or white bars), mild OA (thin lines or grey bars), and severe OA (thick lines or black bars) groups ( $*p < 0.05$ ). Positive values indicate extensor moments, while negative values indicate flexor moments. (T2: beginning of single leg stance; T3: end of single leg stance). (a) Hip extensor/flexor moment, (b) knee extensor/flexor moment, (c) hip extensor moment, (d) knee extensor moment, (e) peak hip moment and (f) peak knee extensor moment.

found in patients with mild OA, but were not statistically significant (Table 3). Smaller knee abduction during the SLS was found in both OA groups. This difference was more pronounced in patients with severe knee OA (Table 3). In addition, during the the swing phase, the angles of pelvis list were significantly greater in OA group than in healthy subjects. Significantly greater pelvic anterior tilt was found only in the severe group at the beginning of SLS but in both OA groups at the end. All transverse plane angles at the ankle, knee, and hip were similar between all groups.

The sagittal joint moments were different between the patients groups and matched control subjects, while these were not affected by the severity of the disease. All patients with knee OA had greater hip extensor moments during SLS (Fig. 1), while smaller hip flexor moments were found at toe-off ( $p < 0.025$ ) as compared with the control group (Fig. 1). They had smaller peak knee extensor moments during the early stance phase, approximately 48% and 56%, respectively, of those in the healthy subjects (mild OA,  $p < 0.0001$ ; severe OA,  $p = 0.002$ ). Similar results were also found at toe-off, while the differences were smaller than those during the early stance phase (mild OA,  $p = 0.003$ ; severe OA,  $p = 0.015$ ). Following heel-strike, peak ankle dorsiflexor moments decreased in all patients with knee OA (mild OA,  $0.007 \pm 0.004 \text{ BWs}^{-1}$ ; severe OA,  $0.008 \pm 0.005 \text{ BWs}^{-1}$ ; control,  $0.014 \pm 0.007 \text{ BWs}^{-1}$ ) compared with those in the healthy subjects (mild OA,  $p = 0.003$ ; severe OA,  $p = 0.023$ ).

During the mid-stance and terminal stance phases, the severe group had significantly greater peak abduc-

tor moments at the hip than those found in the healthy subjects ( $p = 0.009$ ), while those in the mild group were similar to those of the healthy controls (Fig. 2). Similarly, peak knee abductor moments during the early and late stance in the patients with severe OA were also greater as compared to the healthy subjects (early stance,  $p = 0.005$ ; late stance,  $p = 0.047$ ), while those between patients with mild knee OA and healthy subjects were similar. At heel-strike, all patients with knee OA had smaller ankle internal rotation moments (mild OA,  $0.002 \pm 0.001 \text{ BWs}^{-1}$ ; severe OA,  $0.003 \pm 0.002 \text{ BWs}^{-1}$ ; normal,  $0.004 \pm 0.002 \text{ BWs}^{-1}$ ) as compared with the healthy subjects ( $p < 0.024$ ).

#### 4. Discussion

The current study supported the hypothesis that patients with mild and severe medial knee OA adopt different compensatory gait patterns in the pelvis-leg apparatus to unload the diseased knee, not only in the frontal plane but also the sagittal plane.

Gait velocity was reported to affect lower limb biomechanics in both healthy subjects and patients with knee OA [17,22]. Increased walking speed increases stride length and stride rate, as well as peak knee flexion, during the loading response phase and swing phase. In the sagittal plane, joint moments also increased in magnitude as walking speed increased [25], while the results in the frontal plane remain controversial. A significant negative correlation was found

Table 3  
Means and standard deviations (S.D.) of the peak angles at five critical times (T1: loading response; T2: beginning of single leg stance; T3: end of single leg stance; T4: stance phase; T5: swing phase)

Angle (°)	Group					
	Control		Mild OA		Severe OA	
	Mean	S.D.	Mean	S.D.	Mean	S.D.
<b>Sagittal plane</b>						
<b>Hip</b>						
T1	27.57	4.66	24.84	3.32	22.77	7.47
T2	19.04	6.39	15.44	5.17	12.42	9.57
T3	-10.81	3.64	-9.08	3.78	-10.19	6.94
T5	30.16	3.89	27.67	4.08	26.03	7.97
<b>Knee</b>						
T1 <sup>b</sup>	17.26	9.01	15.05	7.63	9.78	6.44
T2 <sup>b</sup>	16.25	9.04	13.46	7.14	8.58	6.32
T3	10.75	6.93	9.38	6.14	6.46	4.45
T5	62.45	15.43	62.08	7.24	57.29	7.74
<b>Ankle</b>						
T1 <sup>b</sup>	-7.25	3.92	-7.31	2.92	-7.64	4.14
T2	0.98	3.22	0.21	3.82	-0.04	4.58
T3	6.82	5.60	5.71	3.22	6.99	4.62
T5	5.26	3.63	4.70	3.19	3.52	3.50
<b>Frontal Plane</b>						
<b>Hip</b>						
T2 <sup>b</sup>	10.65	4.80	7.92	3.83	6.83	3.96
T3	5.01	3.82	5.99	3.95	6.38	3.46
T4	11.13	4.67	9.58	4.86	8.10	3.81
T5	-3.23	4.93	-3.12	2.61	-1.61	3.27
<b>Knee</b>						
T2 <sup>b</sup>	-5.50	4.49	-1.07	2.73	0.14	3.22
T3 <sup>a,b</sup>	-5.00	3.90	-1.24	2.30	-0.18	3.15
T4 <sup>a,b,c</sup>	-1.80	1.52	0.27	2.46	1.78	2.11
T5 <sup>a,b</sup>	-20.74	14.28	-10.79	4.22	-11.20	7.02
<b>Ankle</b>						
T2	-3.14	1.90	-2.64	2.03	-2.58	4.74
T3	-1.12	3.18	0.56	3.72	-2.50	4.40
T4	7.78	5.57	7.79	4.14	6.00	5.21
T5	-5.51	2.15	-4.14	2.22	-5.45	3.98
<b>Pelvic list</b>						
T2	2.75	1.67	2.93	2.06	1.60	2.87
T3 <sup>b</sup>	-0.76	1.53	0.68	2.25	1.26	2.32
T5 <sup>a,b</sup>	-0.48	1.26	1.54	1.74	1.03	1.40
<b>Pelvis tilt</b>						
T2 <sup>b</sup>	7.49	3.41	12.09	3.47	15.62	6.11
T3 <sup>a,b</sup>	7.16	3.22	12.69	2.39	15.88	7.05

Positive values indicate flexion, dorsiflexion, and adduction, while negative values indicate extension, plantarflexion, and abduction at hip, knee and ankle joints. Pelvic list was positive when the ASIS was higher than that of the contralateral side and the positive values of pelvis tilt indicate anterior tilt angles of the pelvis. Three symbols indicate a significant difference between two groups.

<sup>a</sup>  $p < 0.05$  as normal group vs. mild OA group.

<sup>b</sup>  $p < 0.05$  as normal group vs. severe OA group.

<sup>c</sup>  $p < 0.05$  as mild OA group vs. severe OA group.

between the peak knee abductor moment in terminal stance and gait speed [13], while Mundermann et al. [8] reported that a decreased abductor moment was related to slower speeds. In the current study, the OA and healthy control groups showed no significant differences in the self-selected gait speed, and their mean walking speeds were also similar to those reported in some previous studies whose subjects were also similar in height to the current ones [7,30]. Although there are other studies reporting higher walking speeds, their subjects were taller [11,19,20]. It seems that normalization of gait speeds to body height or leg length may be necessary when comparing gait velocities and their effects on lower limb mechanics between subject groups and between studies [30,31]. In the current study, the results of walking speed showed that the observed changes in the lower limb mechanics of the subjects with knee OA were not associated with gait speed. These observed changes were also largely similar to those available in the literature despite the differences in gait speeds [19,20]. From the current results, the subjects with knee OA appeared to adopt compensatory gait strategies to maintain normal gait speeds and stride lengths under biomechanical changes, such as varus alignment at the diseased knee.

Similar knee abductor moments were found between patients with mild knee OA and their control group. Varus alignment of the standing knee will cause the body center of mass (COM) to displace away from the knee joint center, and thus increase the abductor moments. In order to keep the standing knee abductor moments at a normal level, patients with mild knee OA appeared to list the swing-side of the pelvis so as to shift the body's COM toward the stance leg and thus to decrease forces across the medial tibiofemoral compartment cartilage of the stance limb. These kinematic compensations at the joints of the lower limb allow people with mild knee OA to unload the diseased knee, which may be helpful in reducing the pain and maintaining the spatiotemporal parameters at a normal level. In order to achieve these kinematic changes, sufficient hip abductor moments throughout the stance phase were needed in the OA group, in agreement with Chang et al. [18] and Mundermann et al. [19]. Sufficient hip abductor moments to protect the progression of medial compartment knee OA were reported [18,19].

Gait changes in the frontal plane were found to be associated with the severity of the disease in the medial compartment of the OA knee. Similar results were also found by Miyazaki et al. [5]. Greater knee abductor moments throughout the stance phase and greater hip abductor moment at the second peak during stance phase were found in the severe OA group, as compared to the healthy controls. Increased peak hip abductor moments seemed to reduce corresponding knee abductor moments. Compared to the first peak of the knee abductor moment, the difference of the second peak between the OA and healthy groups had decreased by 97%, although it remained greater than that of the controls. It seemed that patients with severe knee OA tried not only to list the swing-side of the pelvis, similar to those with mild knee OA, but also to increase standing hip abduction in order to reduce



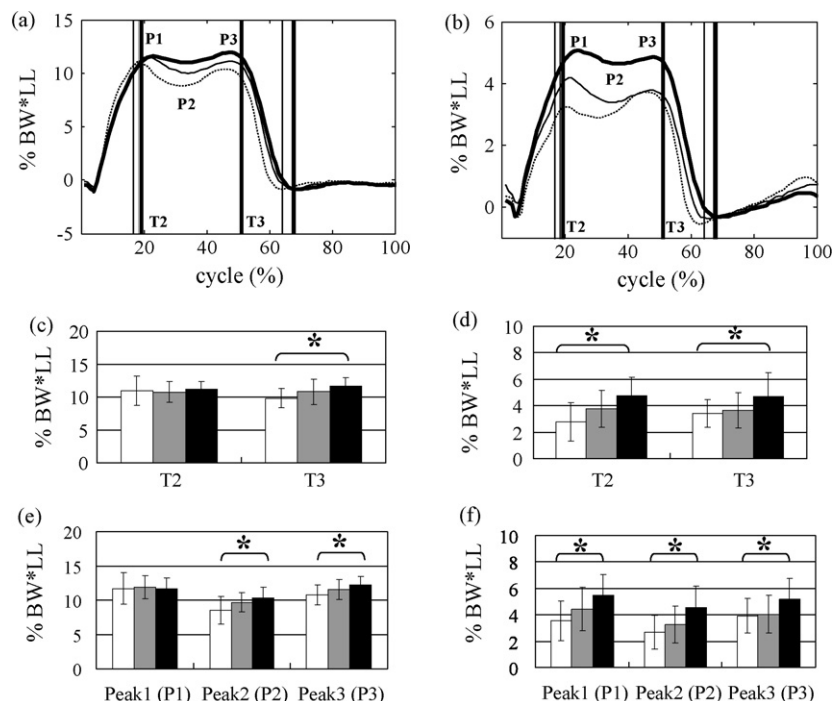


Fig. 2. Frontal patterns, peaks during the early stance and late stance, values at the beginning and the end of the single leg stance of hip and knee moments between the control (dashed lines or white bars), mild OA (thin lines or grey bars), and severe OA (thick lines or black bars) groups ( $*p < 0.05$ ). Positive values indicate abductor moments, while negative values indicate adductor moments. (a) Hip abductor/adductor moment, (b) knee abductor/adductor moment, (c) hip abductor moment, (d) knee abductor moment, (e) peak hip abductor moment and (f) peak knee abductor moment.

loads at the diseased knee. However, they were unable to achieve that goal, thus failing to have the potential protection against OA progression. Failure to reduce loads at the diseased knee in patients with severe knee OA was also reported by Mundermann et al. [19].

Both mild and severe OA groups exhibited smaller knee extensor moments throughout the stance phase when compared with the healthy subjects, in agreement with the literature [20]. It reported that the walking speed mostly affects the joint moments in the sagittal plane [25]. However, this is not the case in the current study, as there were no significant differences in the walking speeds between the OA and the control groups. The most likely reason was the increased anterior pelvic tilt in both OA groups, which would displace the COM forward to be closer to the knee joint center but away from the hip center, decreasing the knee extensor moments but increasing those of the hip. For the severe group, increased knee extension was also adopted, possibly related to the weakness of the quadriceps, which was reported to accompany knee OA [32]. Reduced knee extensor moments, approximately less than 60% of the peak moments during the stance phase, were reported to decrease the stability of the knee [33]. Prolonged maintenance of the hip extensor moments during SLS may be helpful for the OA group to stabilize the lower limb for weight transfer during stance. The static and dynamic stability provided by the musculoskeletal system through balancing external moments is one major mechanical function of muscles [34]. However, the muscles apply tensile forces along the bones, thus increasing the com-

pressive axial force levels in the bone shaft. It was reported that 70% of the value of the maximum axial force in the femur during walking was contributed by the action of the extensor or flexor muscle activity during stance phase [35]. Increased loading on the hip muscles may increase the articular compressive axial force and may thus increase the likelihood of hip OA. A similar suggestion was also made by Mundermann et al. [19] based on comparisons of the inter-segmental axial loading rate at the hip between normal and OA subjects. A recent study showed that 37.6% of patients with knee OA were found also to have hip OA [36]. While the strategy of gait compensation in the sagittal plane was helpful for reducing the knee loading, the increased hip moments may not be beneficial to the hip.

## 5. Conclusions

Patients with mild and severe medial knee OA adopt different compensatory gait patterns to unload the diseased knee, not only in the frontal plane but also the sagittal plane. The mild OA group successfully reduced the extensor moment and maintained normal abductor moment at the diseased knee mainly through listing and anterior tilting the pelvis. With extra compensatory changes at other joints and increased hip abductor moment, the severe group successfully reduced the knee extensor moment but failed to reduce the abductor moment. These results suggest that, apart from training of the knee muscles, training of the hip muscles and pelvic control

are essential in the rehabilitative intervention of patients with knee OA, especially for more severe patients.

### Conflict of interest

None.

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