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Effects of bilateral medial knee osteoarthritis on intra- and inter-limb contributions to body support during gait

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ABSTRACT

Patients with knee OA show altered gait patterns, affecting their quality of living. The current study aimed to quantify the effects of bilateral knee OA on the intra-limb and inter-limb sharing of the support of the body during gait. Fifteen patients with mild, 15 with severe bilateral knee OA, and 15 healthy controls walked along a walkway while the kinematic and kinetic data were measured. Compared with the controls, the patients significantly reduced their knee extensor moments and the corresponding contributions to the total support moment in the sagittal plane ($p < 0.05$). For compensation, the mild OA group significantly increased the hip extensor moments ($p < 0.05$) to maintain close-to-normal support and a more symmetrical inter-limb load-sharing during double-limb support. The severe OA group involved compensatory actions of both the ankle and hip, but did not succeed in maintaining a normal sagittal total support moment during late stance, nor a symmetrical inter-limb load-sharing during double-limb support. In the frontal plane, the knee abductor moments and the corresponding contributions to the total support moment were not affected by the changes in the other joints, regardless of the severity of the disease. The observed compensatory changes suggest that strengthening of weak hip muscles is essential for body support during gait in patients with knee OA, but that training of weak ankle muscles may also be needed for patients with severe knee OA.

1. Introduction

Knee osteoarthritis (OA) is the most common degenerative joint disease in the elderly, affecting the quality of life of over 27 million people in the USA (Lawrence et al., 2008). It is characterized by the loss of articular cartilage – more often in the medial compartment – with a varus alignment (Mündermann et al., 2005). Other symptoms include pain, joint swelling and stiffness, muscle weakness, limited range of motion (ROM) and joint deformity (Maurer et al., 1999; Semble et al., 1990), levels of which depend on the severity of the disease. Knee OA has been shown to affect gait patterns and increases the risk of falls in the elderly (Levinger et al., 2012) as a result of altered joint kinematics and loading patterns, as well as balance control (Leveille et al., 2002, 2009).

With malalignment primarily in the frontal plane, previous studies on knee OA have focused mainly on the increased adducting moment during gait and its relationship with disease severity and progression (Goh et al., 1993; Gok et al., 2002; Huang et al., 2008). Nonetheless, primary and compensatory changes were also found in the sagittal plane, including reduced knee flexion (Gok et al., 2002), decreased knee extensor moments (Gok et al., 2002; Huang et al., 2008), increased hip extensor moments and reduced flexor moments during single-limb support (SLS) (Chang et al., 2005). Most of the studies have based their clinical recommendations on the observed planar changes at individual joints even though walking is a three-dimensional (3D), multi-joint movement. Knowledge of how the joints work together to provide necessary support for the whole body and to compensate for the changes at the diseased knee may be helpful for the management of these patients, and for evaluating treatment outcomes.

For a specific kinematic pattern of the body segments, a combination of moments of the lower limb joints is necessary to support the body weight (BW) and to prevent collapse of the lower limbs while balancing and supporting the body (Winter, 1980). This combination of joint moments may vary with the severity of knee OA as the

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moments at the diseased knee are changed. In the literature, the changes of the moment combinations have been quantified in the sagittal plane, here referred to as the sagittal total support moment (sagittal Ms), which was defined as the numerical sum of the extensor moments at the hip and knee, and the plantarflexor moment at the ankle during stance (Winter, 1980). The sagittal total support moment and the individual joint moments as a percentage of the sagittal total support moment (CMs) together have been used to assess the synergistic strategy in healthy subjects (Hong et al., 2013) and patients with stroke (Nadeau et al., 1998), musculoskeletal pathology (Nadeau et al., 1997), as well as for total knee replacement (Mandeville et al., 2007). Patients with knee OA were also assessed during treadmill walking at different speeds but without differentiating their severity levels or sides of involvement (i.e., unilateral or bilateral) (Zeni and Higginson, 2011). Apart from in the sagittal plane, a combination of joint moments in the frontal plane is also necessary for the lower limbs while balancing and supporting the body. This can be accomplished by a collaboration of abductor muscles and non-contractile tissue at the joints of the lower limb. Since the ranges of motion of these joints are limited in the frontal plane, the distribution between the individual joint moments as a result of kinematic pattern changes, and thus the total support moment, may not be the same as that in the sagittal plane. Therefore, quantification of the changes of this total support moment, here referred to as frontal total support moment (frontal Ms), is especially critical in knee OA because the associated malalignment occurs primarily in the frontal plane.

Previous studies on the sagittal total support moments during walking have focused mainly on SLS. However, the weight transfer between limbs during double-limb support (DLS) has been shown to be a critical period of balance control (Hsu et al., 2010; Winter, 1995). During this period, the loading to the leading hip and knee joints will reach the first peak, the rate of which depends on the rate of release of the weight from the trailing limb. For patients with knee OA, the weight transfer between the limbs may be affected by the altered joint structures and the associated pain, and thus is an important factor for fall-prevention irrespective whether the OA severity in the limbs is similar or not. However, it remains unclear how the lower limbs in patients with knee OA are controlled to produce the necessary support for the stability and advancement of the whole body during weight transfer. The roles of the limbs in supporting the body during DLS could be studied by the changes in the total support moments of the individual limbs, both in the sagittal and frontal planes, as percentages of the corresponding whole body support moment (WMs), defined as the sum of the bilateral total support moments (Hong et al., 2013). To our best knowledge, no study has characterized quantitatively the intra- and inter-limb support synergies of the lower limb joints in terms of the Ms and CMs, both in the sagittal and frontal planes, in response to bi-lateral medial knee OA of different severity levels during level walking.

The purpose of the current study was to quantify the effects of bilateral knee OA on the intra-limb and inter-limb contributions to the support of the body during level walking in terms of total support moments in the sagittal and frontal planes, contributions of individual joints to these total support moments, and the whole body support moment. It was hypothesized that the severity of knee OA would have different effects on the total support moments with altered contributions of individual joints, and on the contribution of the limbs to the whole body support.

2. Material and methods

2.1. Subjects

Thirty patients with bilateral medial knee OA and 15 normal controls participated in the current study with written informed consent as approved by the Institutional Research Board. The patients were divided into two groups, each with 15 subjects, according to their Kellgren–Lawrence (K/L) grades for both knees: mild group (grade 1 or 2; 9 female, 6 male, age: 63.1 ± 11.9 years, height: 161.9 ± 6.5 cm, mass: 68.4 ± 10.3 kg) and severe group (grade 3 or 4; 13 female, 2 male, age: 63.1 ± 8.2 years, height: 156.0 ± 8.9 cm, mass: 64.0 ± 8.5 kg). Patients with knee OA were excluded from the study if they had other neuromusculoskeletal diseases which might affect gait and/or cognitive function, or if they had received an intra-articular corticosteroid injection in the previous two months, or were planning for a total knee replacement in the next year. The healthy controls were matched to the OA groups by age, height and mass (9 female, 6 male, age: 63.2 ± 9.9 years, height: 159.3 ± 8.1 cm, mass: 60.5 ± 8.4 kg). They were free from any neuromusculoskeletal, cardiovascular or neurological pathology, and any other disorders that might have affected their gait and/or cognitive function. For the subjects with knee OA, their knee pain was assessed using a 100-mm visual analog scale (VAS: 0 = no pain, 100 = worst imaginable pain). The varus/valgus alignment of the knee joint was measured on full limb radiographs as the angle between the line connecting the femoral head center and the femoral intercondylar notch center, and the line connecting the mid-point of the tibial spine tips and the talus center, with an angle of 0° indicating a neutral alignment (Chao et al., 1994; Sharma et al., 2001; Tetsworth and Paley, 1994).

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 (Chen and Lu, 2006). Three-dimensional marker trajectories were measured using a 7-camera motion analysis system (Vicon 512, Oxford Metrics Group, UK) at a sampling rate of 120 Hz. The ground reaction forces (GRF) were collected at a frequency of 1080 Hz using two forceplates (OR-6-7-1000, AMTI, USA). Three successful gait cycles for each limb were recorded for each subject.

With the measured GRF and kinematic data, moments at the lower limb joints were calculated using inverse dynamics analysis. Inertial properties for each body segment were obtained using an optimization-based method (Chen et al., 2011). A global optimization method was used to reduce the effects of soft tissue artefacts associated with the skin markers (Lu and O'Connor, 1999). All the calculated joint moments were normalized to body weight (BW) and leg length (LL). The curves of each joint moment component from all the subjects of each group were ensemble-averaged to obtain mean curves for plotting the joint kinetics data.

The sagittal total support moment (sagittal Ms) of a limb was calculated as the sum of the extensor moments at the hip and knee, and the plantarflexor moments at the ankle during stance, which showed a characteristic double-peak pattern similar to that of the GRF curve during gait (Winter, 1980). The contributions of the hip, knee and ankle moments to the first and second peaks of the sagittal total support moment (i.e., Ms1 and Ms2) were calculated as percentages of the corresponding peak values of sagittal Ms. The whole body support moment (WMs) was also calculated as the summation of the sagittal Ms of both limbs. Note that joint moments during the swing phase were not included in the calculations of Ms and WMs. For studying the coordination between the limbs in providing the WMs for the overall support of the body, the time integral of the sagittal Ms for each limb during DLS was calculated as a percentage of the time integral of the WMs. For SLS, inter-limb sharing was not necessary as the WMs was provided completely by the stance limb. In the frontal plane, the frontal total support moment (frontal Ms) of a limb was calculated as the sum of the abductor moments at the lower limb joints during stance, and the variables for intra- and inter-limb sharing of frontal Ms were calculated following the same procedure used for those of the sagittal plane. During DLS, a limb was designated as either leading or trailing limb according to its position relative to the other limb. For each trial of each subject, apart from Ms1, Ms2, the time integrals over DLS, individual joint contributions, and the moment components for each joint at which Ms1 and Ms2 occurred (P1 and P2) were also extracted for subsequent statistical analysis. The spatial–temporal variables, including walking speed, cadence and the durations of SLS and DLS, were also calculated.

2.3. Statistical analysis

For all the calculated variables, one-way analysis of variance (ANOVA) was performed to compare between the mild group, severe group and control group. Once a significant main effect was found, a *post-hoc* analysis using a paired *t*-test with Bonferroni correction was performed. 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

The gait speed, cadence, step widths and stride lengths for the OA groups were not significantly different from those of the normal controls (Table 1). There were no significant differences