Knee Osteoarthritis Affects the Distribution of Joint Moments During Gait

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Abstract

Alterations in lower extremity kinetics have been shown to exist in persons with knee osteoarthritis (OA), however few investigations have examined how the intersegmental coordination of the lower extremity kinetic chain varies in the presence of knee joint pathology. The objective of this study was to evaluate the how knee OA and walking speed affect total support moment and individual joint contributions to the total support moment. Fifteen healthy subjects and 30 persons with knee OA participated in 3D walking analysis at constrained (1.0 m/s), self-selected and fastest tolerable walking speeds. Individual joint contributions to total support moment were analyzed using separate ANOVAs with one repeated measure (walking speed). Linear regression analysis was used to evaluate the relationship between walking speed and joint contribution. Persons with knee OA reduced the contribution of the knee joint when walking at constrained (p=0.04) and self-selected walking speeds (p=0.009). There was a significant increase in the ankle contribution and a significant decrease in the hip contribution when walking speed was increased (P<0.004), however individual walking speeds were not significantly related to joint contributions. This suggests that the relationship between walking speed and joint contribution is dependent on the individual's control strategy and we cannot estimate the joint contribution solely on walking speed. The slower gait speed observed in persons with knee OA is not responsible for the reduction in knee joint moments, rather this change is likely due to alterations in the neuromuscular strategy of the lower extremity kinetic chain in response to joint pain or muscle weakness.

Keywords

osteoarthritis; knee; gait; kinesiology; inverse dynamics

1. Introduction

Alterations in lower extremity kinetics and kinematics have been shown to exist in persons with knee osteoarthritis (OA) [1-7]. While many previous authors have focused on kinetic changes that occur in individual lower extremity joints, few have examined how the

2.2 Gait Analysis

Three-dimensional kinematics were recorded at 60 Hz using a six camera Motion Analysis reflective marker system (Santa Rosa, CA, USA). A modified Helen Hayes marker set was used. Ground reaction forces and center of pressure data were recorded at 1080 Hz during walking trials on an instrumented split-belt treadmill with dual forceplates (Bertec Corp., Columbus, OH, USA). This instrument has previously been validated [25]. Self-selected walking speed was determined by a timed 10 meter walk in the hallway. After allowing time for familiarization to the treadmill [26], subjects walked at the self-selected speed for 30 seconds, during which time the data were recorded. Treadmill speed was then increased to the subject's fastest tolerable speed at which a 30 second walking trial was recorded. Fastest tolerable speed was defined as the maximum speed at which the subjects were able to safely walk on the treadmill without running or holding onto the handrails. Data were also collected at a control speed of 1.0 m/s to control for the effect of walking speed on the gait variables. All subjects were attached to an overhead harness as a safety precaution and were instructed to walk without use of the handrails.

2.3 Data reduction

Coordinate positions of the reflective markers were smoothed using a phase-corrected Butterworth filter with a cutoff frequency of 6 Hz (EvaRT 5.0.4, Motion Analysis Corp) and joint angles were reconstructed using Euler angles. Inverse dynamic calculations of joint moments were performed using OrthoTrak 6.3.5 (Motion Analysis Corp.) and were normalized to the individuals' body mass. All of the moments are reported as external joint moments. Gait cycles were divided from heel strike to ipsilateral heel strike as determined from kinematic data of the foot and sacral markers [27]. These cycles were then time normalized to 101 points and averaged for each variable. Summation of the hip, knee and ankle extensor moments provided a total support moment at each time step [10]. In order to assess differences in total support moment, mean peak total support moment was determined. Relative hip, knee and ankle joint contributions were determined as a percent of the total support and reported at the time of peak support moment.

For the self-selected and fast walking speeds, differences in mean peak total support moment were assessed using an ANOVA with one repeated measure (walking speed). Similarly, differences in individual joint contributions were evaluated using separate ANOVAs with one repeated measure (walking speed). Differences in walking speed between groups and conditions were assessed using a repeated measure ANOVA. At 1.0 m/s, differences were evaluated using separate one-way ANOVAs for the hip, knee and ankle contributions. Relational comparisons between the individual's walking speed and joint contribution were separately evaluated for the OA and control groups at the self-selected and fast walking speeds using a linear regression analysis. All statistical tests were performed using SPSS software version 15 (Chicago, IL, USA).

3 Results

3.1 Walking Speed

In this study, subjects with OA chose a self-selected walking speed that was significantly slower than the freely chosen walking speed of the control group. The control group walked at 1.24 ± 0.17 m/s and the OA group at 1.08 ± 0.19 m/s ($p < 0.001$). The trend was the same at the fast walking speed and subjects in the control group had a mean fast speed of 1.75 ± 0.22 m/s and the OA group walked at 1.43 ± 0.22 m/s ($p < 0.001$). There was a significant speed by group interaction effect as the control group demonstrated a larger increase in speed between the self-selected and fast walking conditions ($p < 0.001$).

3.2 Total Support Moment

At 1.0 m/s, no significant differences existed between the groups for peak total support moment (p=0.15) (Figure 1). Peak total support moment for the control and OA group were 0.86 ± 0.16 Nm/kg and 0.96 ± 0.31 Nm/kg, respectively. There was a significant increase in total support moment peaks for both groups between the self-selected and fast walking speeds (p<0.001). At the self-selected walking speed, peak total support moment for the control and OA groups were 1.05 ± 0.25 Nm/kg and 0.96 ± 0.31 Nm/kg, respectively. This difference was not significantly different (p=0.36). At the fast walking speed, significant differences in peak total support moment were seen between the control (1.62 ± 0.33 Nm/ kg) and OA group (1.20 ± 0.40 Nm/kg) (p=0.001).

3.3 Individual Joint Contribution

At 1.0 m/s, persons with knee OA demonstrated reduced knee contribution to total support (p=0.04), but no significant differences were seen for the hip (p=0.96) or ankle (p=0.07) (Figure 2). Knee contribution for the control and OA groups were $47\% \pm 17\%$ and $32\% \pm 25\%$, respectively. A similar trend was seen at the self-selected walking speed and the OA group demonstrated a reduction in the contribution from the knee (p=0.009) and ankle (p=0.024) (Figure 2). Knee contribution to total support in the control group was $51\% \pm 15\%$, which was higher than $32\% \pm 26\%$ in the OA group. The control group demonstrated a lower contribution from the ankle ($25\% \pm 16\%$) when compared to the OA group ($45\% \pm 32\%$). In the fast walking condition, only the contribution from the ankle demonstrated significant differences between groups. Ankle contribution to total support at the fast speed from the control and OA groups were $14\% \pm 12\%$ and $30\% \pm 22\%$, respectively. Knee contribution from the control group was $49\% \pm 16\%$ and was $37\% \pm 31\%$ in the OA group. Although the mean knee contribution at the fast speed was 12.2% higher in the control group, there was large variability and the difference was not significant (p=0.09).

For both groups, significant differences were found between self-selected and fast walking speeds for the hip (p=0.004) and ankle (p<0.001) contributions. There was a significant increase in the ankle contribution and a significant decrease for the hip contribution. No differences were found between speeds for the knee contribution. No group by speed interaction effects were seen for the hip, knee or ankle contributions. Using a regression model, self-selected and fast walking speeds were not significantly related to the respective hip, knee or ankle contribution, with the exception of the hip contribution in the control group at the fast walking speed. The significant relationship between the hip contribution and fast walking speed in the control group had an R^2 value of 0.353 (p=0.019), with faster walking speeds resulting in higher hip contribution.

4 Discussion

The evaluation of knee contribution to total support adds to the understanding of compensatory strategies utilized by persons with knee OA. The magnitude of the total support moment remains unchanged between groups at the self-selected and controlled walking speeds, but individuals with knee OA alter the distribution of the load shared by the lower extremity kinetic chain. While previous investigations have revealed a reduction in the magnitude of the knee flexion moment in persons with knee OA [28,29], we have found that subjects with OA adopt a reorganized control strategy to achieve a similar total support moment by reducing the knee contribution and increasing the ankle joint contribution.

Using the regression model, there was no consistent relationship between walking speed and joint contributions; however, there was an effect of speed on the joint contributions in both groups. Between self-selected and fast walking speeds, there was a consistent increase in the

hip joint contribution and reduction in the ankle contribution. This suggests that the relationship between walking speed and joint contribution is dependent on the individual's muscle coordination patterns (or control strategy) and we cannot estimate the joint contribution solely on walking speed. However, for each person, we can assume that an increase in walking speed above the self-selected speed will reduce the ankle contribution and increase the hip contribution, regardless of whether or not the individual has knee pathology. This supports previous research that has implicated the hip moment as playing a larger role during faster walking speeds [21,22].

It appears that the strategy adopted by persons with knee OA is aimed at reducing the knee flexion moment at all walking speeds. It can be assumed that higher knee joint moments will result in higher muscle forces crossing the joint [30,31]. An increase in muscle force may result in higher knee joint compression forces [32], which in turn may accelerate cartilage degeneration [33]. The reduction in contribution from the knee joint to total support may represent a compensatory strategy to reduce the need for quadriceps force during walking in order to reduce joint loading and joint pain. It is also possible that this strategy is a consequence of persistent quadriceps weakness in persons with knee OA [34-36]. If there is substantial weakness of the knee extensors, persons with knee OA may rely on the antigravity hip and ankle muscles during locomotion. As seen in the OA group, the reduction in knee contribution was augmented by an increase in the ankle moment. This tradeoff is in accordance with previous work that found increased ankle contribution may propagate a reduction in knee joint moment through segmental interaction [37].

As cartilage degrades in the medial compartment of the knee, there is an inherent loss of stability as the passive structures become lax. It is suggested that persons with knee OA ambulate with increased co-contraction of the quadriceps and hamstrings to overcome the passive laxity of the knee joint [38]. Although we did not analyze muscle activity, increased co-contraction of the quadriceps and hamstring muscle groups may lead to a knee stiffening strategy and subsequent reduction of the knee flexion moment during gait. Despite reducing the external knee flexion moment, this method of stabilization may increase the compression forces at the knee joint.

In the absence of antagonistic muscle activity, increasing the external knee flexion moment has been suggested to be a compensatory method to overcome the condylar lift-off induced by the frontal plane knee adduction moment [39]. However, in this study, we found a relative decrease in the contribution of the knee flexion moment in persons with knee OA and an increase in the contribution from the ankle. This supports previous studies that found increased activity of the plantarflexor muscles in addition to the quadriceps, may help control the frontal plane knee laxity [40,41]. It should be pointed out that the gastrocnemius is a biarticular muscle that does cross the knee joint, and therefore increased moments at the ankle may coincide with higher gastrocnemius muscle activity and, subsequently, higher compressive force at the knee [42].

The use of the support moment in gait analysis provides us with information about how the individual joints are coordinated to maintain support of the body during walking. Comparison of the individual joint contributions to this moment allows us to formulate potential control strategies used to coordinate human movement and support. The results from our study show that persons with knee OA adopt a kinetic coordination strategy that reduces the contribution from the knee and increases contributions from ipsilateral joints, in particular, the ankle joint. This strategy was seen irrespective of freely chosen or constrained walking speed, suggesting that the reduction in knee contribution in persons with knee OA is not dependent on the lower freely chosen walking speed. Furthermore, increasing walking speed in all of the subject populations resulted in greater contribution from the hip joint. The

slower gait speed observed in persons with knee OA is not responsible for the reduction in contribution from the knee joint. Instead, this change is likely due to alterations in the neuromuscular strategy of the lower extremity kinetic chain in response to joint pain, muscle weakness or a generalized stiffening strategy of antagonistic knee muscles.

Acknowledgments

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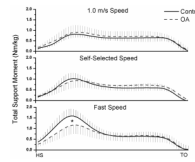
Figure 1.
Peak total support moment was significantly greater in the control group at the fast walking speed (p<0.001). There was a significant increase in peak total support between the self-selected and fast walking condition in both groups (p<0.001).
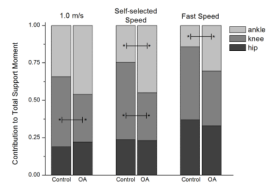
Figure 2.
Individual joint contributions to total support moment (*p<0.05). There was a significant increase in the hip moment (p=0.004) and reduction in the ankle moment (p<0.001).

Table 1

Subject demographics

	Control (n=15)	OA (n=30)	p-value
Age	58 (9)	63 (7)	0.067
Height	1.66 (0.05)	1.71 (0.10)	0.076
Weight	70.13 (8.5)	86.83 (13.3)	**<0.001** [*]
Body mass index	25.58 (3.5)	29.75 (4.2)	**0.002** [*]
Quadriceps Strength (Nm/BW)	1.76 (0.68)	1.26 (0.45)	**0.010** [*]
Hamstring Strength (Nm/BW)	1.06 (0.45)	0.82 (0.33)	0.077
KOOS	481 (23)	270 (86)	**<0.001** [*]
Self-selected Speed	1.24 (0.17)	1.08 (0.20)	**0.014** [*]
Fast Speed	1.75 (0.22)	1.42 (0.22)	**<0.001** [*]

Nm/BW = Newton*meters/body weight; KOOS = Knee Osteoarthritis and Outcome Score [max score = 500]

[*] $p < 0.05$