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Compressive tibiofemoral force during crouch gait

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ABSTRACT

Crouch gait, a common walking pattern in individuals with cerebral palsy, is characterized by excessive flexion of the hip and knee. Many subjects with crouch gait experience knee pain, perhaps because of elevated muscle forces and joint loading. The goal of this study was to examine how muscle forces and compressive tibiofemoral force change with the increasing knee flexion associated with crouch gait. Muscle forces and tibiofemoral force were estimated for three unimpaired children and nine children with cerebral palsy who walked with varying degrees of knee flexion. We scaled a generic musculoskeletal model to each subject and used the model to estimate muscle forces and compressive tibiofemoral forces during walking. Mild crouch gait (minimum knee flexion 20–35°) produced a peak compressive tibiofemoral force similar to unimpaired walking; however, severe crouch gait (minimum knee flexion > 50°) increased the peak force to greater than 6 times body-weight, more than double the load experienced during unimpaired gait. This increase in compressive tibiofemoral force was primarily due to increases in quadriceps force during crouch gait, which increased quadratically with average stance phase knee flexion (i.e., crouch severity). Increased quadriceps force contributes to larger tibiofemoral and patellofemoral loading which may contribute to knee pain in individuals with crouch gait.

1. Introduction

Crouch gait is a common pathological walking pattern adopted by individuals with cerebral palsy that is characterized by excessive hip and knee flexion. Walking in a crouched posture is inefficient [1,2] and can lead to joint pain and compromise an individual's walking ability [3]. Surgical and therapeutic treatments for crouch gait aim to produce a more upright posture to improve walking efficiency and prevent joint pain and deterioration.

Altered loads on the knee can have adverse effects on joint health. Cartilage and bone growth and maintenance depend on the loads experienced during daily life [4,5], and abnormal loading can lead to joint pain, cartilage degeneration [6], and the formation of bone deformities [7]. Joint pain can be a significant contributor to walking deterioration in adults with cerebral palsy. Jahnsen et al. [8] found that 41% of adults with diplegic cerebral palsy reported significant knee pain.

To develop successful treatment strategies for crouch gait, surgeons and therapists need to understand how joint loads change with increasing knee flexion during crouch gait. Treatments are aimed at reducing the excessive knee flexion associated with crouch gait, but it is unclear if changes in knee flexion will alter joint loads. Quantifying the relationship between knee flexion, muscle forces, and the compressive force on the tibia during gait could help clinicians determine if a more upright posture could reduce the risks caused by altered joint loading.

Perry et al. [9] examined knee forces in a static crouched posture using a cadaver model and reported increasing compressive tibiofemoral force with increasing knee flexion. In dynamic activities, such as walking, we expect larger joint forces than in a static posture due to the additional muscle forces required to support the body weight during movement and propel the body forward [10]. Compressive tibiofemoral forces during unimpaired walking have been reported in the range of 2–3 times body-weight [11–14]. During crouch gait, muscle forces in the stance-limb are higher than during unimpaired walking [15]. Since muscle forces are the primary contributors to joint loading [16,17], we expect that compressive tibiofemoral forces are higher during crouch gait, yet the relationship between crouch gait severity and the compressive tibiofemoral force remains unknown.

The purpose of this study was to estimate the magnitude of the compressive tibiofemoral force during crouch gait and examine how this force changes with crouch severity. To achieve this goal we estimated the muscle forces and the compressive force on the tibia in typically-developing children with unimpaired gait and children

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Table 1
Subject characteristics.

	N	Age (years)	Height (cm)	Weight (kg)	Speed/height (s^{-1})	Minimum KFA[a] (°)
Unimpaired	3	10.3 ± 3.4	145 ± 16	36.3 ± 8.8	0.79 ± 0.1	1.7 ± 5.5
Mild crouch	3	8.8 ± 0.8	123 ± 7	24.2 ± 3.6	0.67 ± 0.1	19.1 ± 3.8
Moderate crouch	3	9.2 ± 2.9	123 ± 15	43.1 ± 37	0.63 ± 0.1	36.1 ± 4.0
Severe crouch	3	14.0 ± 2.3	158 ± 12	40.1 ± 6.8	0.61 ± 0.1	58.6 ± 5.6

[a] KFA, knee flexion angle during walking.

with cerebral palsy who walked in varying degrees of crouch severity. We used a freely available biomechanics software package, OpenSim [18], to scale a musculoskeletal model to each individual and estimate joint loads based upon each individual's gait dynamics.

2. Methods

2.1. Subjects

The subjects for this study were selected from a database of patients treated at Gillette Children's Specialty Healthcare (St. Paul, MN; Table 1). Nine subjects with spastic diplegic cerebral palsy were selected to cover a broad range of crouch severity and were divided evenly into three groups: mild crouch gait (minimum knee flexion angle of 20–35°), moderate crouch gait (minimum knee flexion angle of 35–50°), and severe crouch gait (minimum knee flexion angle greater than 50°). All subjects walked with excess knee and hip flexion and had at least 5° of ankle dorsiflexion during stance. We excluded subjects that had greater than 30° of femoral or tibial torsion, which can affect muscle moment arms and the ability of muscles to generate accelerations [19]. Three unimpaired subjects were chosen who were representative of the age and stature of the subjects with cerebral palsy. Additionally, a subject with an instrumented total knee replacement (TKR, age: 80 years, weight: 64 kg, walking speed/height: 0.74 s^{-1}) was included to provide experimental measurements of the compressive tibiofemoral force for comparison with forces estimated from the musculoskeletal model. This subject was not included in subsequent comparisons between unimpaired gait and crouch gait due to differences in age and stature in relation to the other subjects.

2.2. Motion analysis

Motion analysis data was collected at Gillette Children's Specialty Healthcare (St. Paul, MN) using a 12-camera system (Vicon Motion Systems, Lake Forest, CA), four force plates (AMTI, Watertown, MA), and a standard marker protocol [20]. Ground reaction forces and moments were sampled at 1080 Hz and low-pass filtered at 20 Hz. Electromyography (EMG) was collected for six of the crouch gait subjects from the quadriceps, hamstrings, and gastrocnemius (Motion Laboratory Systems, Baton Rouge, LA). The EMG data was sampled at 1080 Hz, band-pass filtered between 20 and 400 Hz, rectified, and low-pass filtered at 10 Hz. All subjects walked at their self-selected speed and achieved two consecutive force plate strikes during which only one foot contacted each force plate. The motion analysis data for the subject with the instrumented TKR was obtained from www.simtk.org where it is freely available for researchers [21].

2.3. Musculoskeletal modeling

A generic musculoskeletal model based upon adult cadaver data [22] with 19 degrees of freedom and 92 musculotendon actuators was scaled to each subject according to anthropometric measurements. This musculoskeletal model has been used for studies involving unimpaired children and children with cerebral palsy [10,19,23]. The degrees of freedom in the musculoskeletal model included six degrees of freedom at the pelvis, a ball-and-socket joint at the third lumbar vertebra between the pelvis and torso, a ball-and-socket joint at each hip, a planar joint with coupled translations at each knee [24], and a revolute joint at each ankle. Joint angles during walking were calculated by minimizing the error between experimental marker trajectories and markers placed on the model at locations corresponding to the experimental markers.

Static optimization was used to calculate the muscle forces required to reproduce the joint moments of each subject throughout the gait cycle. To distribute muscle forces, static optimization was used to minimize the objective function:

$$\min \sum_{i=1}^{N} c_i a_i^2 \quad (1)$$

where N is the number of muscles in the model, a is the activation level (between zero and one) of each muscle, and c is an integer weighting constant for each muscle with a default value of one. The weighting constants were determined by comparing the calculated compressive tibiofemoral force to the experimentally measured force for the subject with the instrumented TKR, as described below.

The compressive tibiofemoral force was calculated using the Joint Reaction analysis in OpenSim. A detailed description of this analysis is provided in Supplementary Material. Briefly, the tibiofemoral force was calculated as a point load acting on the tibial plateau using the Newton–Euler equation:

$$\vec{R}_{\text{knee}} = [M]_{\text{tibia}}\vec{\alpha}_{\text{tibia}} - (\vec{R}_{\text{ankle}} + \sum \vec{F}_{\text{muscles}} + \vec{F}_{\text{gravity}}) \quad (2)$$

Where, $\vec{R}_{\text{knee}}$ is the force from the femur on the tibia, $[M]_{\text{tibia}}$ is the matrix of inertial properties of the tibia, $\vec{\alpha}_{\text{tibia}}$ is the six dimensional angular and linear acceleration of the tibia, $\vec{R}_{\text{ankle}}$ is the force from the foot on the tibia, and $\vec{F}_{\text{muscles}}$ and $\vec{F}_{\text{gravity}}$ are the muscle forces and gravitational forces acting on the tibia. The compressive tibiofemoral force was calculated as the component of $\vec{R}_{\text{knee}}$ parallel to the longitudinal axis of the tibia and used for all subsequent analyses.

For the subject with the instrumented TKR, we varied the static optimization weighting constants for the major muscle groups that cross the knee: the hamstrings, gastrocnemius, and quadriceps. The hamstrings were modeled as four individual muscles: semimembranosus, semitendinousus, biceps femoris long head, and biceps femoris short head. The quadriceps included individual muscles for the rectus femoris, the vastus medialis, the vastus intermedius, and the vastus lateralis. The same weighting constant was applied to all muscles in each group. The results for the muscles within each group were compared and, if found to be similar, were combined to facilitate analysis. The weighting constants were given integer values between one and ten. We performed static optimization for all combinations of weighting constants and calculated the resulting compressive tibiofemoral force. We selected the combination of weighting constants that had the minimum average value and resulted in a difference between the estimated and experimental peak compressive force of less than 20% body-weight. The set of weighting constants that met this criterion was a weight of three for the hamstrings, seven for the gastrocnemius, and one for the quadriceps. This combination of weighting constants resulted in a root mean square error of 0.28 times body-weight and an average error of 0.02 times body-weight over the gait cycle between the estimated force and the experimental measurements (Fig. 1). These weighting constants were then used to perform static optimization for all other subjects. OpenSim's Joint Reaction analysis algorithm was used to calculate the compressive tibiofemoral force for one representative gait cycle for each subject.

To evaluate whether muscle activations calculated from static optimization reflected the subjects' muscle activity, we qualitatively compared the estimated muscle activations to EMG recordings during stance for the six subjects for whom EMG data was available (Fig. 2). EMG and estimated muscle activations indicated that the quadriceps were active during stance. Hamstring activity decreased during stance in both the EMG and estimated muscle activations; however, estimated muscle activations decreased earlier in stance than indicated by EMG for some of the subjects. For these subjects, increased hamstring activity during stance would have increased estimates of the compressive tibiofemoral contact force. The gastrocnemius muscle was active during the majority of stance in both the EMG and estimated muscle activations.

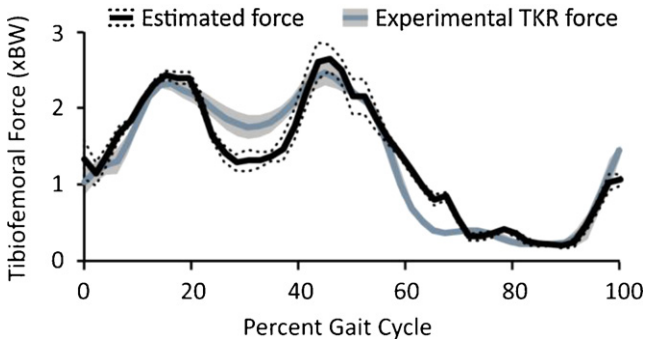

Fig. 1. Tibiofemoral contact force expressed in multiples of body-weight (×BW) from experimental data measured using an instrumented total knee replacement (TKR, gray) and estimated with the musculoskeletal model (black). The average ± 1 standard deviation is shown from four trials.

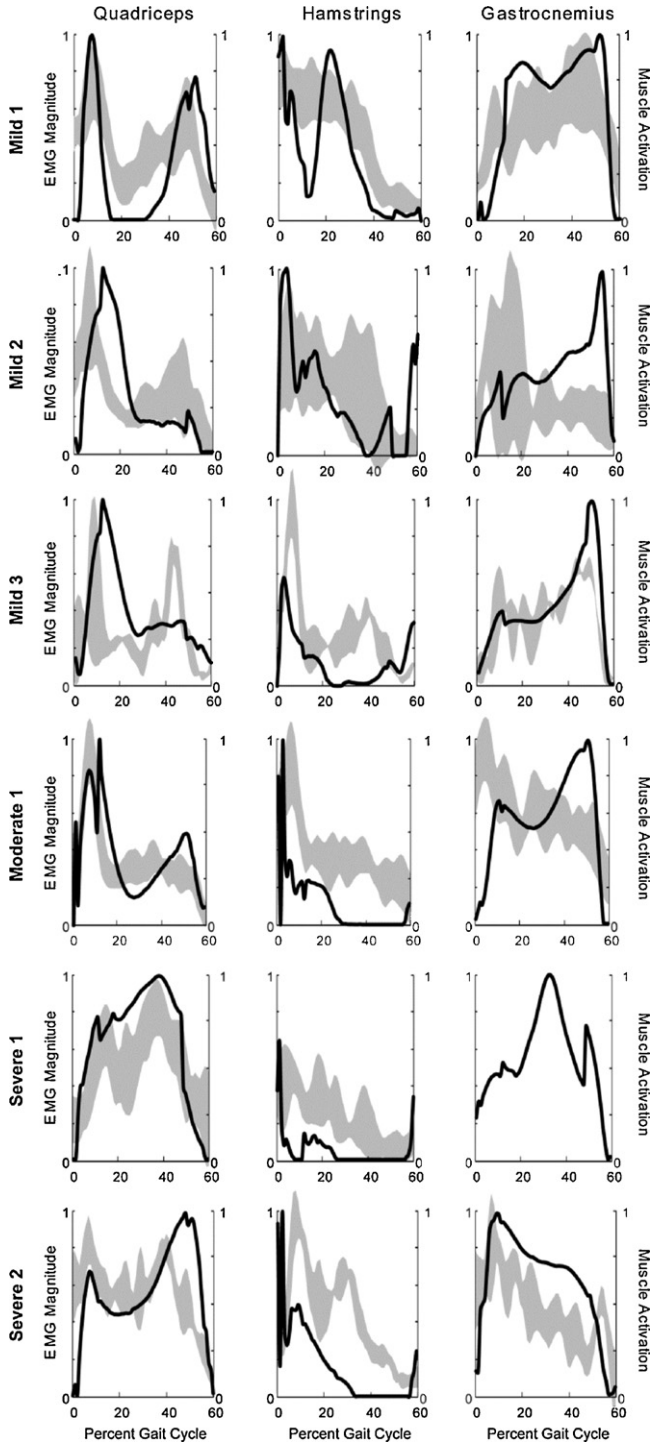

Fig. 2. Comparison of EMG (gray, average ± one standard deviation over all gait cycles) and muscle activations from static optimization (black line) for the six subjects with crouch gait for whom EMG data was available. EMG and activations were normalized from zero to one for each subject based upon the minimum and maximum values over the gait cycle. Note that subject "Severe 1" did not have EMG data from the gastrocnemius.

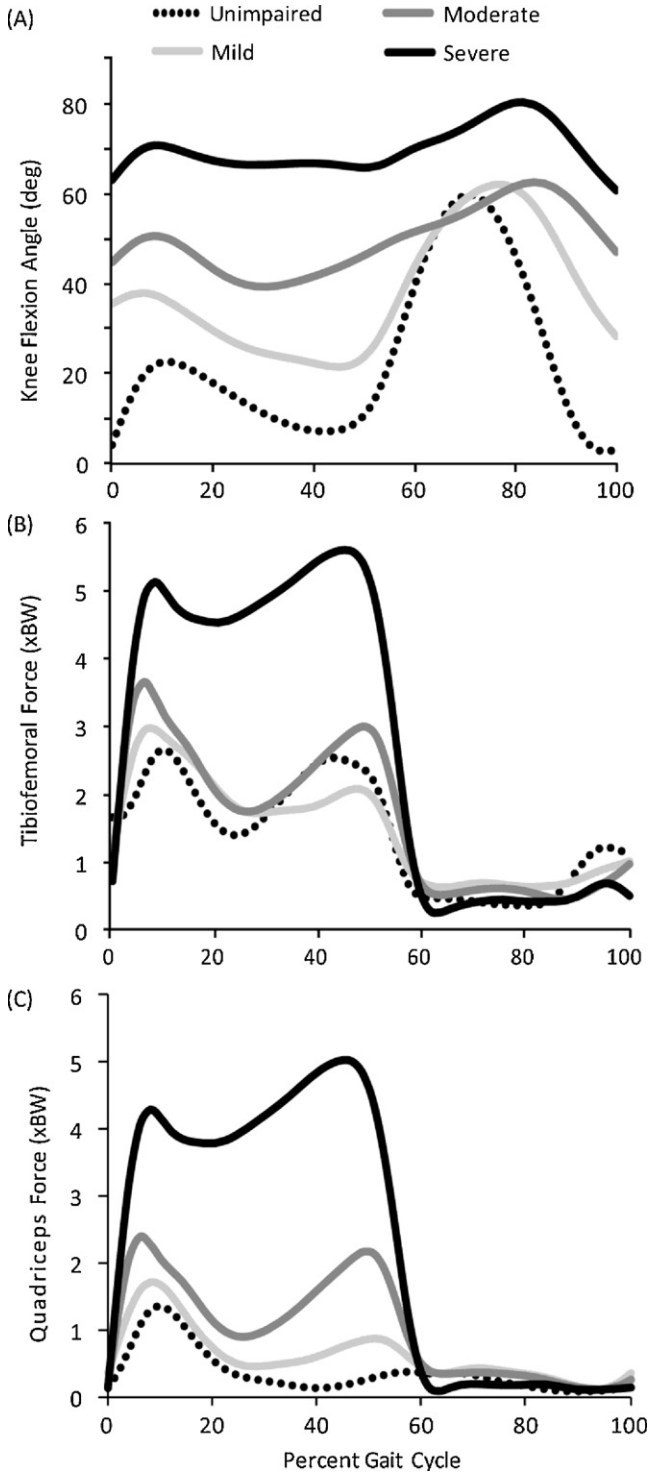

Fig. 3. (A) Average knee flexion angle, (B) average compressive tibiofemoral force, and (C) average quadriceps force expressed as multiples of body-weight (×BW) during one gait cycle for the subjects who walked with an unimpaired gait and mild, moderate, and severe crouch gait.

3. Results

Compressive tibiofemoral force was higher during moderate and severe crouch gait than during unimpaired gait (Fig. 3). Subjects with a mild crouch gait had similar compressive tibiofemoral forces to subjects with unimpaired gait. The maximum force during mild crouch gait was 3.2 ± 0.4 times body-weight compared to 3.0 ± 0.5 times body-weight during unimpaired gait. Maximum force during a moderate crouch gait was 4.2 ± 1.2 times body-weight. During a severe crouch gait maximum force was 6.5 ± 0.7 times body-weight.

Compressive tibiofemoral force during stance exhibited two peaks in unimpaired and crouch gait (Fig. 3B). These two peaks in the tibiofemoral force coincided with the two characteristic peaks

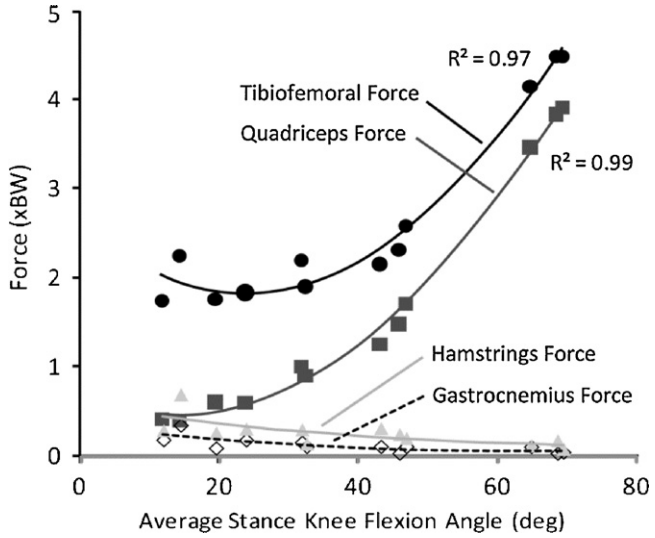

Fig. 4. Correlation of average knee flexion angle during stance with average compressive tibiofemoral force during stance (black circles), average quadriceps force during stance (dark gray squares), average hamstrings force during stance (light gray triangles), and average gastrocnemius force during stance (black outlined diamonds). Tibiofemoral force and average quadriceps force are expressed as multiples of body-weight (×BW). A quadratic relationship described the change in both tibiofemoral force and quadriceps force with increasing crouch.

of the ground reaction force. The largest tibiofemoral forces occurred during early and late stance with smaller forces in mid-stance and swing. During unimpaired gait, the primary contributors to compressive tibiofemoral force were the quadriceps in early stance and the gastrocnemius during late stance. During crouch gait, the quadriceps were the primary contributors to tibiofemoral force throughout stance (Fig. 3C).

There was a quadratic relationship between the average knee flexion angle during stance and the average compressive tibiofemoral force during stance (r^2 = 0.97, Fig. 4). The relationship is described by:

$$F_{knee} = 0.0013\theta^2 - 0.06\theta + 2.54 \quad (3)$$

where F_{knee} is the average compressive tibiofemoral force during stance, and θ is the average knee flexion angle during stance with values from 15° to 70° of flexion.

The increase in average compressive tibiofemoral force during stance with increasing crouch severity was primarily due to an increase in quadriceps force. The average quadriceps force during stance also increased quadratically with knee flexion angle (r^2 = 0.99, Fig. 4) with the relationship:

$$F_{quad} = 0.0011\theta^2 - 0.03\theta + 0.7 \quad (4)$$

The average force produced by the hamstrings during stance did not change with knee flexion; however, the average force of gastrocnemius decreased with crouch severity. Individuals with crouch gait had smaller ankle plantarflexor moments during terminal stance.

4. Discussion

Individuals who walk in a moderate or severe crouch gait experience substantially greater compressive tibiofemoral forces than individuals with an unimpaired gait; however, individuals who walk in a mild crouch gait have similar compressive tibiofemoral forces to those experienced in unimpaired gait. The increase in tibiofemoral force was primarily due to the increase in quadriceps force required to support the body during crouch gait. There was a quadratic increase in quadriceps force with increasing knee flexion which is similar to a reported quadratic increase in EMG magnitude in static, crouch postures [25]. The increase in quadriceps force with crouch severity not only contributes to increased tibiofemoral load but would also increase patellofemoral load [26] and may give rise to knee pain in individuals with cerebral palsy and crouch gait. To reduce the average compressive tibiofemoral force and quadriceps force during stance to within one standard deviation of the average during unimpaired gait, individuals with crouch gait need to achieve an average knee flexion angle less than 25° during stance.

Compressive tibiofemoral force during crouch gait reported here are slightly higher than those estimated by Perry et al., who used statically loaded cadavers in a crouch posture [9]. They determined the compressive tibiofemoral force at 30° and 45° of knee flexion to be 2.9 and 3.8 times body-weight, respectively, whereas we found the maximum force during a crouch gait with an average knee flexion angle of 30° and 45° to be 3.3 and 4.1 times body-weight. The static cadaver testing implemented by Perry et al. did not include contributions from the gastrocnemius or hamstring muscles to compressive tibiofemoral force. The small difference in compressive tibiofemoral force between standing and walking demonstrates that, although walking requires additional muscle force to propel the body forward [15], the increased quadriceps demand arising from a static crouched posture accounts for the majority of the increased tibiofemoral force. The tibiofemoral contact force of the unimpaired children included in this analysis was also similar to previously reported results [11–14].

Our calculation of compressive tibiofemoral force depends on the accuracy of estimated muscle activations. The estimated muscle activations showed patterns similar to EMG such as increased activity of the quadriceps; however, EMG activity was available for a limited number of muscles in six of the subjects. When muscle activations differed from the EMG signals, the optimization tended to underestimate muscle activity compared to EMG signals. This suggests that the optimization functions commonly used for unimpaired walking may not be appropriate for individuals with cerebral palsy who have altered motor control and muscle physiology. Muscle over-activity and excess co-contraction are common in individuals with cerebral palsy. Greater muscle forces due to co-contraction would increase the estimated tibiofemoral contact forces, suggesting that our calculations of compressive tibiofemoral force may be low estimates.

We compared our calculated tibiofemoral forces to experimental forces from an instrumented total knee replacement, but this did not provide a robust evaluation of knee forces during crouch gait. The total knee replacement data was used to select the static optimization weighting constants that reduced the error between the estimated and measured compressive tibiofemoral force. Different weighting constants may be appropriate for younger patients or patients with gait pathology. In this study, the weighting constants penalized recruitment of the hamstrings and gastrocnemius, which resulted in the recruitment of other muscles to actuate the hip and ankle without increasing the compressive load on the tibia. Although the quadriceps are the major contributors to compressive tibiofemoral force, increasing the quadriceps' weighting constant did not reduce the estimated tibiofemoral force since no other muscles could replace the quadriceps' function at the knee.

To test the sensitivity of our results to the objective function we evaluated how estimated tibiofemoral contact force changed when altering the weighting constants and the power of activation. The quadratic relationship between knee flexion angle and tibiofemoral contact force and quadriceps force was similar in all tested objective functions (Supp. Fig. 1). Using a linear objective function resulted in an average reduction in tibiofemoral contact force

during stance of 7% while an objective function that minimized activation cubed increased tibiofemoral contact force during stance by 11%. Using weighting constants of one for all muscles also increased the estimated tibiofemoral contact force during stance by an average of 15% primarily due to a 10% average increase in gastrocnemius force during stance. Future studies that measure compressive tibiofemoral force from individuals with instrumented total knee replacements walking in pathologic gait patterns, such as crouch gait, could provide further points of comparison for model-based estimates of compressive tibiofemoral force and help to determine the optimal objective functions.

This study has demonstrated that walking in a moderate or severe crouch gait increases the compressive tibiofemoral force, which could be contributing to joint pain and cartilage degeneration. Surgeries and therapies that produce a more upright walking posture will reduce forces at the knee and may help moderate the adverse effects of excessive joint loading.

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Conflict of interest

None of the authors had financial or personal conflict of interest with regard to this study.

Appendix A. Supplementary data

Supplementary data associated with this article can be found in the online version, at doi:10.1016/j.gaitpost.2011.11.023
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