Correlation between the Knee Adduction Torque and Medial Contact Force for a Variety of Gait Patterns

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ABSTRACT: The external knee adduction torque has been proposed as a surrogate measure for medial compartment load during gait. However, a direct link between these two quantities has not been demonstrated using in vivo measurement of medial compartment load. This study uses in vivo data collected from a single subject with an instrumented knee implant to evaluate this link. The subject performed five different overground gait motions (normal, fast, slow, wide, and toe-out) with simultaneous collection of instrumented implant, video motion, and ground reaction data. For each trial, the knee adduction torque was measured externally while the total axial force applied to the tibial insert was measured internally. Based on data collected from the same subject performing treadmill gait under fluoroscopic motion analysis, a regression equation was developed to calculate medial contact force from the implant load cell measurements. Correlation analyses were performed for the stance phase and entire gait cycle to quantify the relationship between the knee adduction torque and both the medial contact force and the medial to total contact force ratio. When the entire gait cycle was analyzed, R^2 for medial contact force was 0.77 when all gait trials were analyzed together and between 0.69 and 0.93 when each gait trial was analyzed separately ($p < 0.001$ in all cases). For medial to total force ratio, R^2 was 0.69 for all trials together and between 0.54 and 0.90 for each trial separately ($p < 0.001$ in all cases). When only the stance phase was analyzed, R^2 values were slightly lower. These results support the hypothesis that the knee adduction torque is highly correlated with medial compartment contact force and medial to total force ratio during gait. © 2007 Orthopaedic Research Society. Published by Wiley Periodicals, Inc. J Orthop Res 25:789–797, 2007

Keywords: knee joint; adduction moment; gait; in vivo knee loads; total knee arthroplasty; instrumented knee implant

INTRODUCTION

The human knee joint is critical for locomotion and is commonly affected by osteoarthritis (OA). Adverse mechanical loading, and in particular high medial contact force, is believed to contribute to the development of knee OA.[1] The ability to measure or predict high medial contact force in individual patients would be valuable for identifying those at highest risk for developing knee OA as well as for devising new treatment approaches. Unfortunately, noninvasive in vivo measurement of medial compartment contact force is not yet available.

For this reason, researchers have investigated the use of external measures available from gait analysis as surrogates for internal medial force. To date, the peak knee adduction torque has been identified as the best candidate, in part because of its ability to predict OA disease progression[2] and long-term outcome following high tibial osteotomy surgery.[3] Schipplein and Andriacchi[4] were the first to propose that the knee adduction torque is the primary determinant of medial compartment load during gait. Their conclusions were based on medial compartment load predictions made by a statically determinate muscle model. Using the same model, Noyes et al.[5] found a statistically

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significant correlation between the peak knee adduction torque and the predicted peak medial compartment load in a group of ACL-deficient patients. More recently, Hurwitz et al.[6] showed that the peak knee adduction torque was the best single predictor of the medial to lateral ratio of proximal tibial bone density. While these results support the hypothesis that the knee adduction torque during gait serves as a surrogate for medial compartment load, no study has been able to correlate these quantities based on internal medial load measurements.

This article investigates the relationship between the knee adduction torque and in vivo medial contact force during normal, fast, slow, wide, and toe-out gait. A single patient with an instrumented knee implant provided a unique opportunity to perform the investigation. The adduction torque curve for each gait pattern was obtained using standard external gait measurements. The corresponding internal axial contact force was measured by the instrumented knee implant.[7] A linear regression equation was used to determine medial contact force from the implant load cell measurements, where the regression coefficients were found by using a dynamic contact model to compute medial and lateral contact force from fluoroscopic and axial load data collected from the same patient performing treadmill gait. The hypothesis tested was that the knee adduction torque is highly correlated with medial compartment force and force ratio (i.e., ratio of medial to total contact force) during a variety of gait activities.

MATERIALS AND METHODS

Data were collected from a single patient with an instrumented knee implant (male, right knee, age 80, mass 68 kg, height 1.7 m) 8 months after surgery.[8] Institutional review board approval and patient informed consent were obtained. In vivo tibial force data were recorded simultaneously with video motion (Motion Analysis Corporation, Santa Rosa, CA) and ground reaction (AMTI Corporation, Watertown, MA) data for five patterns of overground gait: normal (1.24 ± 0.03 m/s), fast (1.52 ± 0.04 m/s), slow (0.80 ± 0.05 m/s), wide stance (1.03 ± 0.05 m/s), and toe-out (1.10 ± 0.05 m/s). These gait patterns were chosen because walking speed and foot path have been shown to influence the knee adduction torque.[3,9–11] Four single-axis loads cells in the implant provided a measure of total axial load but not the load distribution between the medial and lateral compartments. The Cleveland Clinic marker set with additional markers placed on the foot segment was used to create segment coordinate systems and provide three-dimensional movement data. Raw marker data were filtered using a fourth-order, zero phase-shift, low pass Butterworth filter with a cutoff frequency of 6 Hz.[12–14] The subject performed three trials of each gait pattern using a self-selected walking speed and foot path. For each trial, one complete motion cycle, starting and ending with right heel strike, was chosen for analysis.

The knee adduction torque was calculated using traditional bottom-up inverse dynamics. The dynamical equations were derived for the foot-shank system with Autolev[15] using Kane's method.[16] The foot possessed six degrees of freedom (DOF) relative to the laboratory-fixed coordinate system. The ankle joint was modeled as two nonintersecting pin joints[13] whose axes were found via optimization of additional motion data collected from the patient performing an ankle circumduction movement.[17] Foot and shank masses, mass centers, and moments of inertia were estimated using regression relationships.[18] Optimal alignment of the shank segment with the shank markers was performed using an approach based on the singular value decomposition.[19] Subsequent alignment of the foot segment with the foot markers was performed using optimization.[17] The inverse dynamics reaction torque in the knee was calculated about the knee joint center, defined as the midpoint between the medial and lateral femoral epicondyles. The external knee adduction torque, which is due primarily to the moment of the ground reaction force vector about the knee center, was taken as the negative of the internal knee abduction torque calculated from inverse dynamics.

A two-step process was followed to convert the four implant load cell measurements into medial contact force measurements. This process was used because there was no guarantee a priori that the sum of the two medial load cell measurements would equal the medial contact force. First, we calculated medial contact force during treadmill (rather than overground) gait using a deformable contact model. The same subject performed treadmill gait with simultaneous collection of instrumented implant and single-plane fluoroscopic data. These data were used to drive an elastic foundation contact model[20–23] possessing nonlinear material properties.[24,25] The contact model was implemented within the Pro/MECHANICA MOTION simulation environment (PTC, Waltham, MA) as described in previous studies[20,22,25,26] (Fig. 1). In brief, a six DOF joint between a fixed femoral component and moving tibial insert was used to measure relative (i.e., joint) kinematics for contact calculations. The motion of three DOFs (i.e., flexion-extension, internal-external rotation, and anterior-posterior translation) was prescribed to match the fluoroscopic measurements, because medial and lateral contact forces are not sensitive to small errors in these DOFs.[27] The motion of the remaining three DOFs was predicted via forward dynamic simulation. Thus, each forward dynamic simulation accounted for the motion of all six DOFs simultaneously using numerical integration of a stiff system of ordinary differential equations. During a dynamic simulation, the total axial force measured by the implant was applied to the back surface of the insert at the

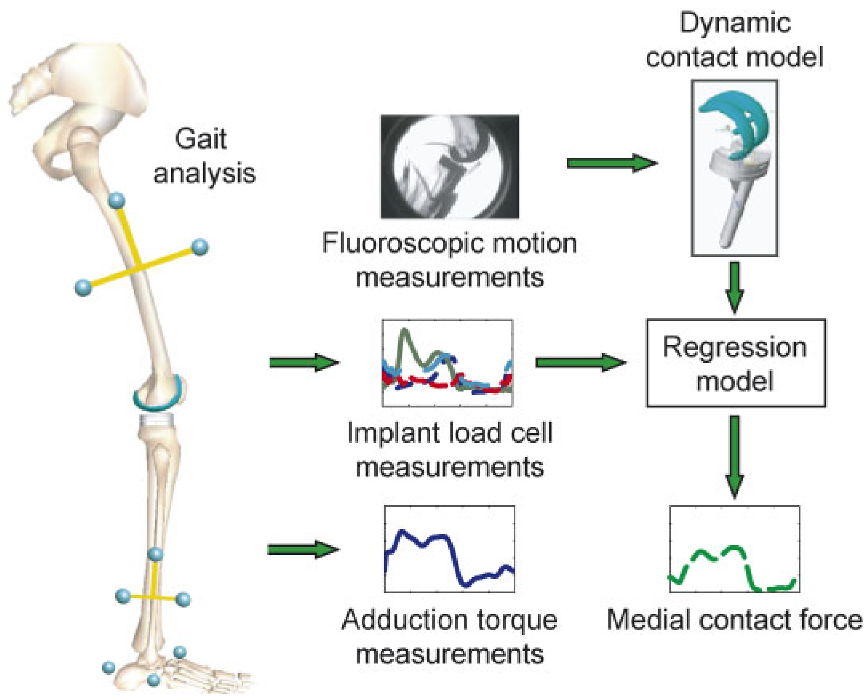

Figure 1. Overview of the experimental and computational methods used to quantify the correlation between the external knee adduction torque and internal medial contact force during gait. Gait analysis was used to make adduction torque measurements. Instrumented implant load cell measurements and fluoroscopic motion measurements were used in a dynamic contact model to develop a linear regression model for calculating medial contact force directly from the four load cell measurements. [Color scheme can be viewed in the online issue, which is available at http://www.interscience.wiley.com]

measured center of pressure location and the contact model used to calculate medial and lateral contact forces.

Second, we used the calculated medial contact forces in a multivariable linear regression analysis. The medial contact forces (F_M) from treadmill gait were fitted as a function of the four implant load cell measurements (F_{AM}, F_{PM}, F_{AL}, and F_{PL}) using linear least squares [Equation 1, $R^2 = 0.99$, root-mean-square (RMS) error = 0.01 body weight (BW)]. The regression equation was

$$F_M = C_1F_{AM} + C_2F_{PM} + C_3F_{AL} + C_4F_{PL} \quad (1)$$

where C_1, C_2, C_3, and C_4 are the regression coefficients and subscripts A, P, M, and L indicate anterior, posterior, medial, and lateral locations, respectively, of the instrumented implant force measurements F. This equation allowed us to calculate medial contact force from the four load cell measurements during overground gait, eliminating the need for fluoroscopic data and the deformable contact model.

Once medial contact force could be calculated for overground gait, the relationship between the knee adduction torque and the medial contact force or medial to total force ratio was analyzed using correlation analyses. Each analysis was performed on within-cycle data points from each of the 15 gait trials separately as well as all 15 gait trials together. R^2 values were calculated for the entire gait cycle and only the stance phase.

RESULTS

Peak total and medial axial load as well as peak adduction torque varied between gait patterns (Table 1). Among the 15 trials, the largest peak

Table 1. Average ± Standard Deviation of Peak Values for Internal Total Force, Internal Medial Force, and External Adduction Torque Calculated Using Three Trials of Each Type of Gait Motion

	Normal	Fast	Slow	Wide	Toe-Out
Total force (BW)	2.49 ± 0.14	2.59 ± 0.13	2.35 ± 0.27	2.47 ± 0.31	2.44 ± 0.25
Medial force (BW)	1.58 ± 0.10	1.63 ± 0.09	1.51 ± 0.21	1.66 ± 0.13	1.57 ± 0.14
Torque (%BW × HT)	2.56 ± 0.18	2.53 ± 0.40	2.71 ± 0.29	2.44 ± 0.20	3.13 ± 0.41

axial force was 2.74 BW, which occurred during one fast gait trial, with a corresponding peak medial force of 1.73 BW. The smallest peak axial force was 2.06 BW, which occurred during one slow gait trial, with a corresponding peak medial force of 1.28 BW. For all trials, the shape of the medial contact force curve closely followed that of the total contact force curve (Fig. 2), and for 10 of the 15 trials, both force curves exhibited two distinct peaks. The corresponding adduction torque curves exhibited two distinct peaks for 9 of the 15 trials. However, the shape of each adduction torque curve did not necessarily follow that of the corresponding medial force curve, nor did the adduction torque peaks necessarily match those of the medial force (Fig. 2). Among the 15 trials, the largest adduction torque peak was 3.37% BW $\times$ HT, which occurred during a toe-out gait trial, while the smallest peak was 2.13% BW $\times$ HT during a fast gait trial.

Statistically significant correlations ($p < 0.001$ for all trials) were found between the knee adduction torque and both the medial contact force and the medial to total contact force ratio (Table 2, Fig. 3). When the entire gait cycle was analyzed, R^2 values for medial contact force were between 0.69 and 0.93 for the 15 gait trials analyzed separately and 0.77 for all 15 gait trials analyzed together. For the medial to total force ratio, R^2 values were between 0.54 and 0.90 for the individual trials and 0.69 for all trials together. When the regression analyses were repeated using only the stance phase, correlations were slightly weaker. For medial force, R^2 values were between 0.38 and 0.85 for each trial separately with a value of 0.60 for all trials together. For medial to total force ratio,

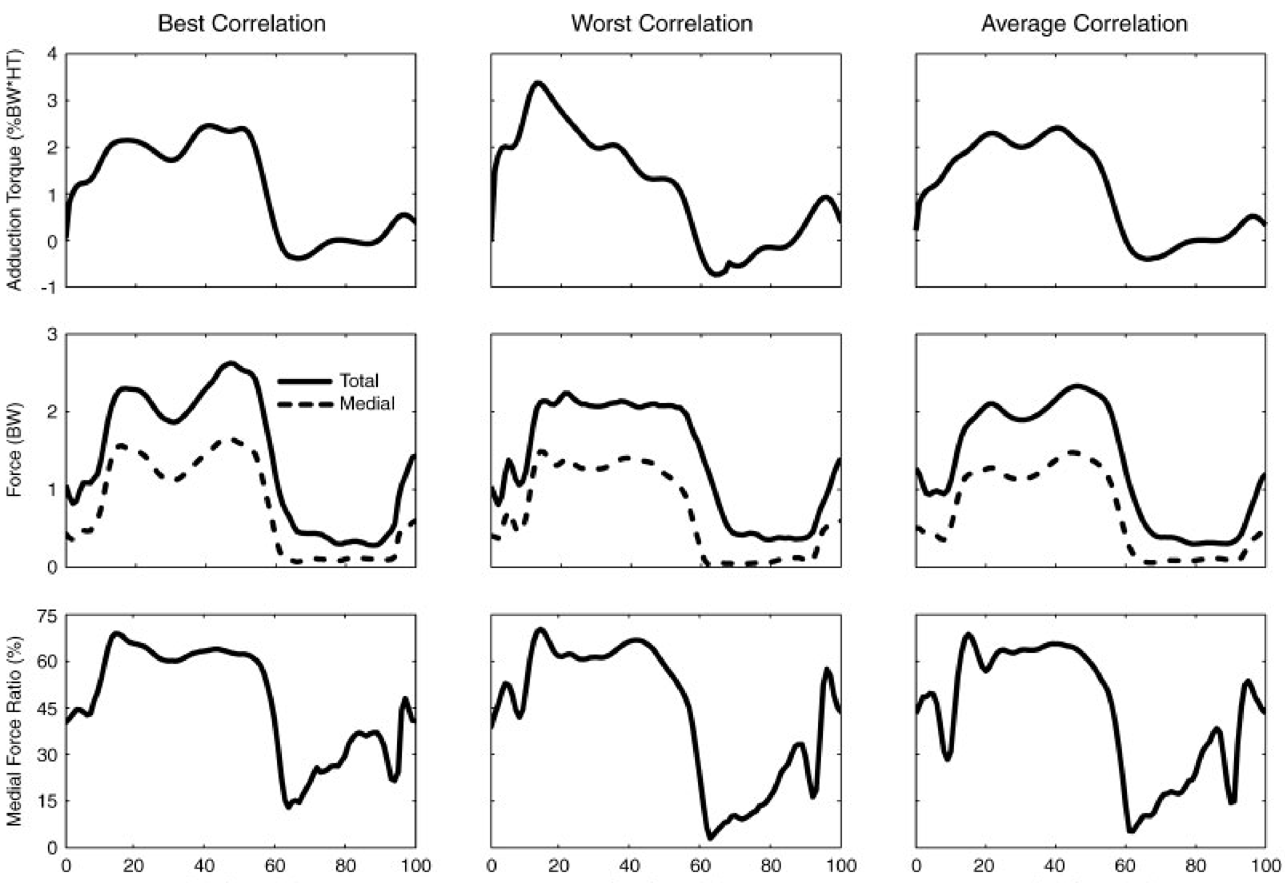

Figure 2. Visualization of best (first column), worst (second column), and average (third column) correlation results between the external knee adduction torque (first row) and the internal medial contact force (second row). Medial to total contact force ratio (third row) is also shown for comparison. The curves in the best column were from a normal gait trial, the curves in the worst column were from a toe-out gait trial, and the curves in the average column were from the mean of all 15 gait trials.

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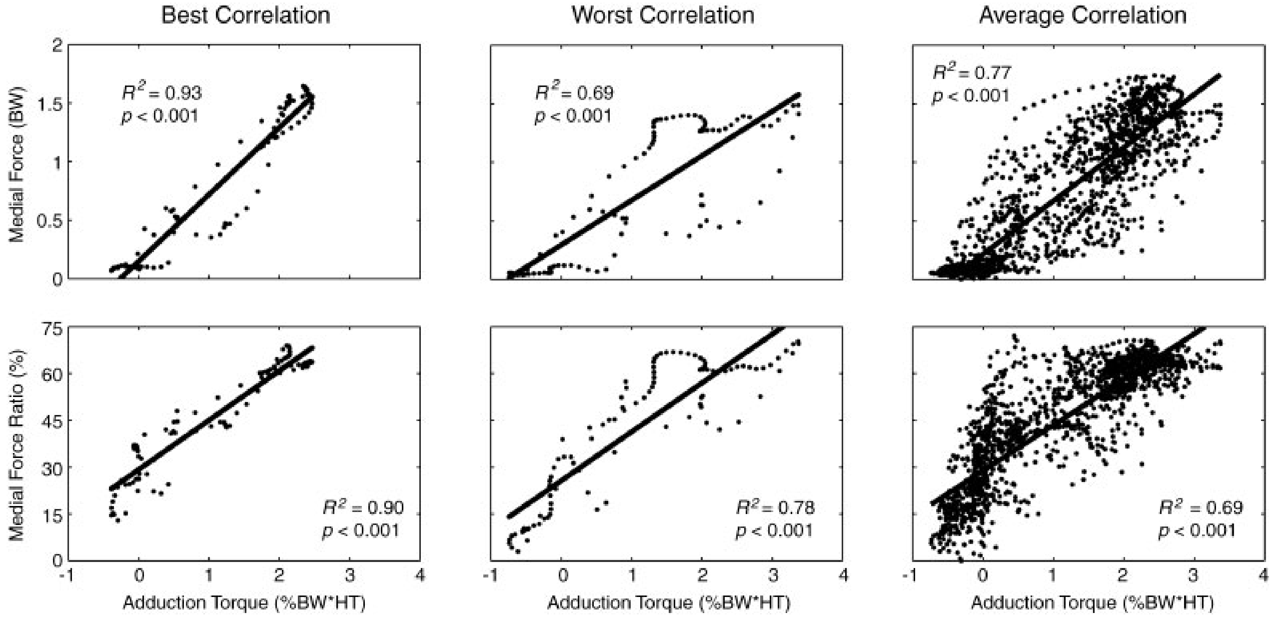

Figure 3. Plots of best (first column), worst (second column), and average (third column) correlation results between the external knee adduction torque (first row) and the internal medial contact force (first row). Medial to total contact force ratio (second row) is also plotted for comparison. The data points in the best and worst columns were from the corresponding curves shown in Figure 2, while the data points in the average column were from all 15 gait trials.

they were between 0.50 and 0.90 for each trial separately and 0.63 for all trials together.

DISCUSSION

This study used a combination of an instrumented knee implant, video-based motion analysis, fluoroscopic motion analysis, and a dynamic contact model to evaluate the hypothesis that the knee adduction torque measured externally can be used as an indicator of internal medial contact force and medial to total contact force ratio during gait. Based on data from this patient, the large R^2 values found in our study strongly support this hypothesis. No assumptions about muscle origin and insertion sites, moment arms, strength, or indeterminacy were required to predict medial contact force,[4,5] which instead was calculated directly from the load cell measurements provided by the instrumented implant. Our findings may be useful for studies that seek to monitor or alter medial compartment load in response to various treatment paradigms for medial compartment OA.

The strong correlation between medial compartment load and the knee adduction torque has at least two clinical implications. First, clinicians

Table 2. Average ± Standard Deviation of R^2 Values between the Internal Knee Abduction Torque and Medial Contact Force or Medial to Total Force Ratio for Three Trials of Each Type of Gait Motion and for All 15 Trials Analyzed Together

	Medial Force		Medial Force Ratio	
Gait	Stance Phase	Full Cycle	Stance Phase	Full Cycle
Normal	0.75±0.09	0.88±0.04	0.81±0.11	0.86±0.05
Fast	0.53±0.13	0.78±0.06	0.68±0.16	0.78±0.07
Slow	0.71±0.09	0.85±0.05	0.77±0.03	0.86±0.02
Wide	0.66±0.07	0.75±0.03	0.60±0.04	0.60±0.08
Toe-Out	0.59±0.09	0.75±0.06	0.76±0.05	0.81±0.04
All trials	0.60	0.77	0.63	0.69

The internal knee abduction torque is the negative of the external knee adduction torque.

should consider screening individuals with previous medial compartment injury, such partial or total medial menisectomy, for high knee adduction torque peaks. Andriacchi and colleagues have reported that following high tibial osteotomy (HTO) surgery, patients with a peak adduction torque below approximately 2.5% BW $\times$ HT have the best long-term outcome.[3,28] This level may be an appropriate target for deciding whether or not intervention is warranted to slow or even avoid the development of medial compartment OA. Second, clinicians should consider conservative interventions such as strengthening of the quadriceps[29,30] and iliotibial band[31,32] or gait retraining[33,34] to lower the adduction torque to below the target level. For example, a medial thrust gait motion designed in a recent computational study has been shown to reduce the peak adduction torque by 35–50%,[35] an amount comparable to HTO surgery, in the one patient studied thus far. To implement these suggestions, clinical gait labs would need to play a central role in initial screening, treatment design, and evaluation of treatment efficacy.

In contrast to our study, previous studies investigated the correlation between *peak* knee adduction torque and *peak* medial compartment load during only the stance phase of gait. Some of these studies used larger numbers of subjects to facilitate statistical analysis of the data.[4,5] A recent study performed with the same patient reported good correlation between the first adduction torque peak and the maximum medial compartment load ($R^2 = 0.44$).[36] That study used different types of shoes to produce peak adduction torque changes on the order of 1% BW $\times$ HT. When we repeated our linear regression analyses using only the peak values of knee adduction torque and medial compartment load, we did not find any statistically significant correlations. However, our different gait motions were less effective at modulating the peak adduction torque.

A previous study also reported a strong correlation between peak external knee extension torque and peak internal medial compartment load during gait.[5] To assess whether a similar strong correlation exists for our within-cycle data, we calculated the knee flexion-extension torque from inverse dynamics for the 15 trials. For the entire cycle, R^2 values correlating the flexion-extension torque with medial contact force ranged from 0.02 ($p = 0.12$) to 0.33 ($p < 0.001$) for the separate gait trials with a value of 0.03 ($p < 0.001$) for all gait trials together. When only the stance phase was analyzed, R^2 values ranged from 0.00 ($p = 0.90$) to 0.16 ($p < 0.001$) with a value of 0.00 ($p = 0.41$) for all trials together. Analysis of either the magnitude of the flexion-extension torque or the medial force ratio produced weaker correlations. Thus, given the weakness and wide variability of these correlation results, we did not find the knee flexion-extension torque to be a consistent indicator of medial compartment load for our subject.

We also assessed the extent to which the external knee adduction torque is balanced by internal medial and lateral contact forces. To perform the assessment, we calculated the moment of the four axial forces, as measured by the implant load cells, about the geometric center of the insert and took the component in the anterior-posterior direction. For the complete gait cycle, R^2 values correlating the external knee adduction torque with the internal moment due to axial contact forces were 0.49 to 0.85 ($p < 0.001$) for the individual gait trials and 0.66 ($p < 0.001$) for all gait trials together. However, the magnitude of the internal moment was roughly a factor of 10 smaller than that of the external adduction torque. This finding suggests that muscles and ligaments play a critical role in balancing knee loads in the coronal plane. Future data collected with an implant capable of measuring six components of internal knee load should be used to investigate this issue further.

The hypothesis that muscles and ligaments balance the majority of the external knee adduction torque can be evaluated using a simple back-of-the-envelope analysis. For this subject, the peak external knee adduction torque during normal gait was approximately 2.5% BW $\times$ HT. From the instrumented implant load cell measurements, one can estimate how much of the corresponding internal abduction torque is due to joint contact forces. The peak internal axial load from the instrumented implant was approximately 2.5 BW. On each condyle, the contact model predicted that the center of pressure was always approximately 20 mm, or roughly 0.01 HT, from the knee center in the medial-lateral direction. Thus, if the entire internal axial load passed through the medial compartment, the contribution of joint contact forces to the internal abduction torque would be approximately 2.5% BW $\times$ HT, resisting 100% of the external adduction torque. In contrast, if only half of the axial load passed through the medial compartment, then the moment about the knee center due to lateral contact force would cancel the moment about the knee center due to medial contact force. In this case, the contribution of joint contact forces to the internal abduction torque would be 0%, and all of the external adduction torque would be resisted by muscles and ligaments. The in vivo situation for this subject was well approximated by a 60–40 medial-lateral load split over much of stance phase, suggesting that roughly 20% of the external adduction torque was resisted by contact forces. When we explicitly calculated the percentage of the external adduction torque supported by contact forces during midstance phase, the maximum value was 12% on average (largest maximum was 16%) and the mean value was 7% on average, confirming the reasonableness of our back-of-the-envelope analysis. Thus, for this subject, the majority of the external adduction torque during midstance was likely counterbalanced by muscle and ligament forces.

If true, this finding would have important implications for musculoskeletal computer models that seek to predict muscle forces across the knee. Historically, such models have treated the knee as a pin joint and assumed that muscles crossing the knee only act to balance the external flexion-extension torque (e.g., ref. 37). For the subject in our study, a better assumption would be that muscles and ligaments crossing the knee act to balance both the external flexion-extension torque and about 90% of the external adduction torque. That is, muscles crossing the knee act to flex as well as stabilize the joint. Including a portion of the adduction torque in the muscle force prediction process will likely result in larger muscle force estimates and more co-contraction than when only the flexion-extension torque is included.[38]

While most studies use inverse dynamics to calculate the knee adduction torque, it is also possible to use the moment of the ground reaction force vector about the knee center as an estimate. We therefore investigated whether use of this alternate approach would significantly alter the results of our study. We treated the ground reaction force vector as a bound vector applied at the center of pressure under the foot. After calculating the moment of this vector about the knee center, we used the component in the anterior-posterior direction of the shank to approximate the external knee adduction torque. During stance phase where the alternate method produces nonzero results, the adduction torque trends for the two methods were similar with an RMS difference of $0.28 \pm 0.09\%$ BW $\times$ HT for the 15 trials. When we repeated the correlation analyses, R^2 values for medial contact force and medial to total force ratio changed little (-0.01 ± 0.04) for the 15 gait trials analyzed separately or together. Thus, omission of inertia forces acting on the shank and foot segments would not have altered our adduction torque or correlation results significantly.

The five different gait patterns analyzed in our study were selected because of their potential to alter the peak knee adduction torque. Increased toe-out angle[3,9,11] and reduced walking speed[10] have received the most attention as possible ways to reduce the adduction torque. Walking with an increased stance width may also have a minor positive effect.[39] Compared to the subject's normal gait pattern, these walking modifications did not produce clear adduction torque reductions. However, we only had access to a single subject, the patient had an artificial rather than natural knee, and the subject performed only three trials of each gait pattern. More extensive data would be required to draw any conclusions about the effectiveness of these gait modifications for reducing the knee adduction torque and medial contact force in the general population.

Three important modeling assumptions were involved in the generation of the medial contact force results. The first was that the dynamic contact model could predict accurate medial and lateral contact forces during treadmill gait. We evaluated this assumption by comparing the total contact force and center of pressure measured by the instrumented implant with the same quantities predicted by the contact model. Over the entire cycle, the RMS error in total contact force was 0.002 BW, while the RMS error in medial-lateral and anterior-posterior center of pressure was within 0.6 mm. A second but related assumption was that the contact force predictions were insensitive to choice of polyethylene material model. When we compared contact force predictions made using linear and nonlinear polyethylene material models,[25] RMS differences were within 0.003 BW. The third important modeling assumption was that a linear regression model could be used to calculate medial contact force from the four implant load cell measurements. To evaluate this assumption, we used additional fluoroscopic data collected from the same subject performing a stair rise/descent activity. When we compared medial contact force calculated by the dynamic contact model to medial contact force calculated by the regression equation from treadmill gait, the RMS difference between the two approaches was 0.04 BW. In addition, the RMS difference between the total contact force measured during overground gait and the sum of the medial and lateral contact forces predicted by separate regression equations was 0.002 BW. Given the small magnitude of these differences, we do not believe that our modeling assumptions had a significant influence on our results.

In summary, this study demonstrated that the external knee adduction torque is highly correlated with internal medial contact force as well as medial to total contact force ratio. The extent to which these results apply to the general population, to knee OA patients, or to activities other than gait is not known. Despite these limitations, our results strongly support the hypothesis that the knee adduction torque can be used as a surrogate measure for medial compartment load during gait.

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