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Decreased Knee Adduction Moment Does Not Guarantee Decreased Medial Contact Force during Gait

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Abstract

Excessive contact force is believed to contribute to the development of medial compartment knee osteoarthritis. The external knee adduction moment has been identified as a surrogate measure for medial contact force during gait, with an abnormally large peak value being linked to increased pain and rate of disease progression. This study uses *in vivo* gait data collected from a subject with a force-measuring knee implant to assess whether knee adduction moment decreases accurately predict corresponding decreases in medial contact force. Changes in both quantities generated via gait modification were analyzed statistically relative to the subject's normal gait. The two gait modifications were a "medial thrust" gait involving knee medialization during stance phase and a "walking pole" gait involving use of bilateral walking poles. Reductions in the first (largest) peak of the knee adduction moment (32 to 33%) did not correspond to reductions in the first peak of medial contact force. In contrast, reductions in the second peak and angular impulse of the knee adduction moment (15 to 47%) corresponded to reductions in the second peak and impulse of medial contact force (12 to 42%). Calculated reductions in both knee adduction moment peaks were highly sensitive to rotation of the shank reference frame about the superior-inferior axis of the shank. Both peaks of medial contact force were best predicted by a combination of peak values of the external knee adduction moment and peak absolute values of the external knee flexion moment ($R^2 = 0.93$). Future studies that evaluate the effectiveness of gait modifications for offloading the medial compartment of the knee should consider the combined effect of these two knee moments.

Keywords

Knee osteoarthritis; instrumented knee implant; joint loading; gait modification; biomechanics

INTRODUCTION

Medial compartment knee osteoarthritis affects a large portion of the population[1]. The disease process is often initiated by changes in knee joint motions and loads, such as those following anterior cruciate ligament or meniscal injury[2,3]. Since overloading of the medial compartment in particular has been hypothesized to play a critical role in the disease process[4], many treatment

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approaches (e.g., high tibial osteotomy, gait retraining) have made reduction of medial compartment contact force (MCF) during gait their primary target.

Unfortunately, no accurate method currently exists for determining *in vivo* MCF in individual patients during gait. Direct measurement of MCF is possible only under special circumstances using instrumented implants[5,6]. External measurements and inverse dynamic analyses can determine the net forces and moments experienced by a joint, but knowledge of muscle and ligament forces is needed to calculate contact forces accurately. Since contact force predictions from musculoskeletal computer models have yet to be thoroughly validated, researchers have sought an external measure as a surrogate for MCF. To date, the peak knee adduction moment (KAM) is the best candidate[7]. This moment is primarily caused by the moment of the ground reaction force vector about the knee center projected along a specified anterior-posterior axis. The KAM normally exhibits two peaks during stance phase, with the magnitude of the first peak (typically the largest) being highly correlated with increased disease severity[8], pain[9], and rate of disease progression[10]. Though the shape of the KAM curve within the gait cycle has been shown to be highly correlated with the shape of the MCF curve within the gait cycle[11], no study has demonstrated that changes in the peak values of the KAM are accurate indicators of changes in the peak values of MCF during stance phase.

This study addresses this important issue using gait data collected from a subject with a force-measuring knee implant. The primary goal was to evaluate whether changes in KAM measures (first peak, second peak, and angular impulse) are accurate indicators of corresponding changes in MCF measures (first peak, second peak, and impulse). Because the calculated KAM depends on the reference frame in which it is expressed[12,13], the secondary goal was to investigate how rotating the shank reference frame about the long axis of the shank affects the correlation between KAM and MCF measures. Since rotating the shank-fixed axis used for the KAM calculation mixes the original KAM with the external knee flexion moment (KFM), we also investigated whether a combination of these two moments is a better predictor of MCF than is the KAM alone. Having KAM and MCF data collected simultaneously from the same patient offered a unique opportunity to perform the investigation.

METHODS

One patient with a force-measuring knee implant (male, right knee, age 83, mass 68 kg, height 1.7 m, body mass index 23.5, neutral leg alignment, contralateral knee implanted as well) performed overground gait with simultaneous collection of internal knee contact force, external video motion (Motion Analysis Corporation, Santa Rosa, CA), and external ground reaction data (AMTI Corporation, Watertown, MA). The patient was tested three and a half years after implantation for primary knee osteoarthritis. Institutional review board approval and subject informed consent were obtained. The implant utilized a custom tibial prosthesis instrumented with four uniaxial force transducers, a micro-transmitter, and an antenna, where the transducers measured compressive force at the four corners of the tibial tray[5]. For the video motion data, a modified Cleveland Clinic marker set was used with additional markers placed on the feet[14].

The subject performed five trials each of three different gait patterns: 1) the subject's normal gait pattern, 2) a medial thrust gait pattern involving knee medialization during stance phase, and 3) a walking pole gait pattern involving use of bilateral walking poles commonly used for hiking. The two modified gait patterns have been previously shown to alter medial contact force relative to the subject's normal gait pattern[15]. For medial thrust gait, the subject was instructed to bring his stance leg knee toward the midline of his body by increasing his knee flexion[16,17] and internally rotating his hip[18]. For walking pole gait, he used two Leki Makalu Tour walking poles with rubber tips (LEKI Lenhart GmbH, Kirchheim, Germany), and pole height was adjusted based on the manufacturer's recommendations. The subject performed all three gait patterns at a self-selected walking speed of 1.23 m/s. In addition, the subject performed a static standing trial with the toes pointing forward (i.e., no toeing in or toeing out) and joint range of motion trials for each ankle, knee, and hip.

A dynamic patient-specific gait model was constructed from the video motion and ground reaction data using previously published optimization methods[14]. The full-body model possessed 27 degrees of freedom (DOFs) and the equations of motion were derived using symbolic manipulation software (Autolev™, OnLine Dynamics, Inc., Sunnyvale, CA). Six DOFs defined the position and orientation of the pelvis segment in the laboratory coordinate system. The remaining segments branched from the pelvis using ball-and-socket joints (hips, back), two intersecting pin joints (shoulders), two non-intersecting pin joints (ankles), and simple pin joints (knees, elbows). Model parameter values were calibrated to the subject's external kinematic and kinetic data from the static trial, isolated joint motion trials, and one normal gait trial using an established sequence of nonlinear least squares optimizations[14]. Model parameters included the positions and orientations of the joints in adjacent body segments and the masses, mass centers, and moments of inertia of the body segments. The knee center on each side was defined as the point on the optimized knee flexion axis that was equidistant from the static medial and lateral knee markers. The ankle center on each side was defined in a similar manner using the optimized ankle flexion axis and the static medial and lateral ankle markers. Marker locations on each body segment were calculated from the static trial and treated as fixed parameters.

The shank coordinate system in the model was defined such that the anterior axis of the shank was approximately aligned with the long axis of the foot in the static pose. To achieve this result, we used the knee and ankle centers and the heel and toe markers from the static trial in the coordinate system creation process. The knee center was selected as the origin of the shank coordinate system. The superior axis was defined by a line connecting the ankle center to the knee center. A temporary anterior axis was defined by a line connecting the heel marker to the toe marker. The lateral axis was then defined as the cross product of the temporary anterior axis and the superior axis, and the final anterior axis was defined as the cross product of the superior axis and the lateral axis.

For each of the 15 gait trials, joint kinematics for the patient-specific model were calculated by performing an inverse kinematics analysis. This analysis used nonlinear least squares optimization to adjust the joint angles in the model until markers on the model segments were aligned as closely as possible with the experimentally measured marker positions. During the alignment process, a higher weight was placed on matching the three shank markers to ensure that the knee center, which was fixed in the shank coordinate system, was in the correct position relative to the ground reaction force vector[14].

The net force and moment acting on the proximal shank were calculated about the knee center using the resulting joint kinematics in an inverse dynamics analysis. The net moment vector was expressed in terms of unit vectors aligned with the shank coordinate system. The external knee adduction moment (KAM) was defined as the negative of the internal three-dimensional knee moment projected along the anterior axis of the shank, while the external knee flexion moment (KFM) was defined as the negative of the internal three-dimensional knee moment projected along the lateral axis of the shank. Thus, the external KAM and external KFM were mutually perpendicular.

Changes in the KAM and MCF for the two modified gait patterns relative to the subject's normal gait pattern were evaluated statistically using a two-tailed Mann-Whitney U test ($p < 0.05$), which is the non-parametric equivalent to a two-tailed t-test. Each KAM and MCF curve during stance phase was characterized by the value of the first peak, the second peak, and the impulse (integral of moment or force with respect to time). To eliminate ambiguity in selection of peak values, we defined the first and second peaks as the local maximum (or minimum) closest to 25% and 75% of stance phase, respectively. In addition, peak values of the KFM, knee flexion angle, and hip internal rotation angle during stance phase were calculated by the patient-specific model and analyzed statistically, since medial thrust gait was expected to alter those quantities. For the hip internal rotation angle, local peaks did not exist, so window averages between 15–25% and 65–75% of stance phase were used to represent the first and second peak, respectively. The ability of KAM measures (first peak, second peak, and angular impulse) to predict corresponding changes in MCF measures (first peak, second peak, and impulse) was quantified using linear regression analysis.

Additional regression analyses were performed to investigate ways to maximize the accuracy of predicted MCF changes. Previous studies have shown that the knee adduction moment calculation is reference frame dependent[12,13]. Consequently, sensitivity of the statistical results to the selected anterior direction was evaluated by rotating the shank reference frame by ± 25 degrees in 1 degree increments about the shank superior axis and then recalculating the KAM. The optimal rotation for each MCF measure was defined as the rotation angle that produced the highest R^2 value from linear regression analysis. Since rotating the shank-fixed axis used for the KAM calculation mixes the original KAM with the KFM, we also performed multi-variable linear regression to assess whether a combination of the KAM and KFM predicts MCF better than does the KAM alone. Peak values of all quantities were used in the multi-variable regression analyses, and absolute values of KFM peaks were taken to account for the possibility of increased joint compression due to an increased flexion or extension moment.

RESULTS

Significant reductions in the KAM produced by both modified gait patterns (Fig. 1a) did not necessarily reflect corresponding reductions in medial contact force (Fig. 1b). During early stance phase, medial thrust and walking pole gait produced statistically significant reductions in the first peak of the KAM (Fig 2a, Table 1), but neither gait pattern produced a statistically significant decrease in the first peak of MCF (Fig. 2b). During late stance phase, walking pole gait produced a statistically significant reduction in the second peak of the KAM (with medial thrust gait showing a non-significant reduction) while both gait patterns yielded statistically significant reductions in the second peak of MCF. Over all of stance phase, medial thrust and walking pole gait produced statistically significant reductions in KAM angular impulse and similar significant reductions in MCF impulse.

The KFM and knee flexion angle for the two modified gait patterns generally showed opposite changes to the KAM (Fig. 3a, b). In early stance, the first peak in KFM exhibited statistically significant increases of 79 and 82% for medial thrust and walking pole gait, respectively (Fig. 4a), while the first peak in knee flexion angle showed significant increases of 52 and 41%, respectively (Fig. 4b). In late stance, the second peak in KFM (a minimum) exhibited statistically significant increases of 117 and 107%, respectively, while only medial thrust gait produced a statistically significant increase (142%) in the second peak of the knee flexion angle (also a minimum). For the hip internal rotation angle, only the first “peak” for medial thrust gait showed a statistically significant change (40% increase) (Fig. 3c, 4c).

Rotating the shank coordinate system by ± 25 degrees about the superior axis of the shank created large variations in the correlations between KAM and MCF measures (Fig. 5; Table 2). For zero rotation (i.e., nominal anterior axis direction), regression analysis revealed that the first peak of the KAM was a poor predictor of the first peak of the MCF. Correlation between corresponding second peaks and impulses was higher. The highest correlation between corresponding first peaks (R^2 = 0.57) was obtained when the shank coordinate system was internally rotated by 18 degrees. In contrast, for corresponding second peaks, the highest correlation (R^2 = 0.71) was obtained for an external rotation of 4 degrees. For corresponding impulses, the highest correlation (R^2 = 0.74) was found for an internal rotation of 15 degrees.

Adding the KFM to the regression analyses generally improved the predictions (Table 2). For all MCF measures except the first peak, including the KFM as a second independent variable increased R^2 and decreased root-mean-square (RMS) fitting error compared to the optimal results for the KAM alone. Though fitting results for the first peak had the lowest R^2 values, they also had the lowest RMS errors for regression analyses involving peak values. Using absolute rather than actual values of the KFM improved fitting of the second peak and both peaks together, with the latter case achieving the highest R^2 value (= 0.93). The RMS error for this case was less than that of the optimal fit for the second peak using only the KAM.

DISCUSSION

This study evaluated whether reductions in the external knee adduction moment guarantee corresponding reductions in medial contact force. The evaluation was performed using unique gait data collected from a subject with a force-measuring knee implant, where the subject performed three gait patterns known to produce different medial contact force results[15]. Our findings suggest that reducing the peak KAM does not necessarily guarantee a corresponding decrease in peak MCF and that a corresponding increase in peak KFM is the most likely explanation.

Rotation of the shank reference frame used for calculating the KAM had a significant influence on the KAM peaks and angular impulse. Rotating the anterior axis used for the KAM calculation is equivalent to linearly mixing the original KAM and original KFM. Though an optimal amount of rotation could be identified to achieve the largest possible R^2 values between corresponding KAM and MCF measures (first peak, second peak, and impulse), the amount was different for each measure. Furthermore, it is not clear whether similar anterior axis directions would be optimal for other subjects. For zero rotation, the low R^2 value for the first peak may be due to the fact that little change in MCF occurred for the three gait patterns tested. Thus, the constant term in the linear regression analysis accounted for the most variability in the first peak. In contrast, the second peak and impulse of MCF showed larger changes for the two modified gait patterns, providing better data for investigating the correlation between the KAM and MCF. Adding the KFM to the regression analysis produced comparable or better R^2 values without the need to identify an optimal amount of rotation for the shank reference frame. This finding is consistent with a previous study where a reduced peak KFM in patients with knee OA was identified as a compensatory mechanism for reducing contact force[19].

Reductions in MCF due to a decrease in the KAM appear to be attenuated by increases in the absolute value of the KFM. The regression analyses showed that peak values of MCF were best fitted by a combination of peak values of the KAM and peak absolute values of the KFM. This finding suggests that increases in the peak knee flexion or extension moment may tend to increase joint compression. A more in-depth analysis of this phenomenon will likely require EMG data and a validated musculoskeletal model to analyze how individual muscles act to counterbalance the external KAM and KFM and contribute to contact forces. At a minimum, these observations indicate that prediction of changes in MCF due to gait modification may be more complex than indicated by changes in the KAM alone.

Though seemingly contradictory at first glance, our results for the first peak may not be inconsistent with results from previous studies reporting a high correlation between the peak KAM and clinical outcome measures (e.g. pain[9]; disease severity[8,20]; disease progression[10]).

In the absence of any change in peak absolute KFM, our results remain consistent with the idea that an increased peak KAM is associated with a corresponding increase in peak MCF and hence with knee OA[21]. In studies where the peak KAM was found to be a good predictor of clinical outcome (e.g., for high tibial osteotomy surgery[22]), a reduced peak KAM may have been accompanied by an unchanged or reduced peak KFM[23], possibly magnifying the apparent effect of the peak KAM on clinical outcome. In studies where the peak KAM was found to be a poor predictor of clinical outcome (e.g., pain[24]), lack of analysis of the peak KFM may have contributed to an incomplete picture of medial compartment loading. Future studies should report changes in both the peak KAM and the peak absolute KFM, as well as the reference frame used to perform the calculations, to provide as much information as possible for assessing how these moments may influence MCF and clinical outcome.

Our findings also have important implications for gait retraining studies that seek to reduce MCF via a reduction in the peak KAM. The medial thrust gait pattern used in the present study is currently being investigated by several labs[12,16,18]. As shown here and in a previous study[16], medial thrust gait has the tendency to increase the peak KFM, which may attenuate the benefits of reducing the peak KAM. Thus, efforts to train patients with medial compartment knee OA to perform medial thrust gait should emphasize a minimal increase in knee flexion angle. As suggested by another study[18], increased hip internal rotation without a corresponding increase in knee flexion may prove to be effective in this regard. Acute gait modifications that do not alter the peak KFM do not appear to have this issue[25].

Walking pole gait may be more effective at reducing MCF than indicated in this study. Similar to medial thrust gait, walking pole gait exhibited increased knee flexion (and hence increased external KFM) during stance phase, while in contrast to medial thrust gait, it did not exhibit increased hip internal rotation. Since walking pole gait was performed after medial thrust gait, the increase in knee flexion relative to normal gait may have been due to a carry-over effect. If so, then walking pole gait could potentially produce significant reductions in both peaks of medial contact force. Further investigation of this possibility would be worthwhile, as would investigation of whether an optimal pole length and tip placement on the ground exist that maximize the reduction in the first peak of medial contact force.

The results of our study are limited by several important factors. First and foremost, only a single subject was studied with an implanted knee, so it is unclear the extent to which these results are generalizable to subjects with medial compartment knee OA. The subject also had neutral leg alignment, whereas most patients with medial compartment knee OA have a varus malalignment[26]. Because our results were generated using acute gait modifications, they may not reflect the role of the KAM in the process of natural disease progression over time, especially without corresponding changes in the KFM. Furthermore, because the first peak in MCF was not changed significantly for either gait modification, it is difficult to determine the correlation between this quantity and the first peak in KAM and KFM. Finally, we did not seek to train the subject to alter his KAM over the long term, so it is unknown whether the acute changes observed in this study would persist over time.

In conclusion, this study has shown that gait modifications are able to reduce significantly both peaks of the KAM curve without producing corresponding significant reductions in both peaks of MCF. Though this outcome was neither anticipated nor desired, gait modifications that seek to reduce the largest peak of the KAM curve may still offload the medial compartment of the knee. Minimizing corresponding increases in the absolute value of the peak KFM may be essential to achieving this goal.

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References

1. Lawrence RC, Felson DT, Helmick CG, et al. Estimates of the prevalence of arthritis and other rheumatic conditions in the United States. Part II. Arthritis and Rheumatism 2008;58:26–35. [PubMed: 18163497]
2. Roos H, Adalberth T, Dahlberg L, Lohmander LS. Osteoarthritis of the knee after injury to the anterior cruciate ligament or meniscus: the influence of time and age. Osteoarthritis and Cartilage 1995;3:261–267. [PubMed: 8689461]
3. Setton LA, Mow VC, Howell DS. Mechanical behavior of articular cartilage in shear is altered by transection of the anterior cruciate ligament. Journal of Orthopaedic Research 1995;13:473–482. [PubMed: 7674064]
4. Frost HM. Perspectives: A biomechanical model of the pathogenesis of arthroses. The Anatomical Record 1994;240:19–31. [PubMed: 7810912]
5. D'Lima DD, Townsend CP, Arms SW, et al. An implantable telemetry device to measure intra-articular tibial forces. Journal of Biomechanics 2005;38:299–304. [PubMed: 15598457]
6. Kaufman KR, Kovacevic N, Irby SE, Colwell CW. Instrumented implant for measuring tibiofemoral forces. Journal of Biomechanics 1996;29:667–671. [PubMed: 8707796]
7. Andriacchi TP. Dynamics of knee malalignment. Orthopedic Clinics of North America 1994;25:395–403. [PubMed: 8028883]
8. Sharma L, Hurwitz DE, Thonar EJMA, et al. Knee adduction moment, serum hyaluronan level, and disease severity in medial tibiofemoral osteoarthritis. Arthritis & Rheumatism 1998;41:1233–1240. [PubMed: 9663481]
9. Thorp LE, Sumner DR, Wimmer MA, Block JA. Relationship between pain and medial knee joint loading in mild radiographic knee osteoarthritis. Arthritis & Rheumatism 2007;57:1254–1260. [PubMed: 17907211]
10. Miyazaki T, Wada M, Kawahara H, et al. Dynamic load at baseline can predict radiographic disease progression in medial compartment knee osteoarthritis. Annals of the Rheumatic Diseases 2002;61:617–622. [PubMed: 12079903]
11. Zhao D, Banks SA, Mitchell KH, et al. Correlation between the knee adduction torque and medial contact force for a variety of gait patterns. Journal of Orthopaedic Research: Official Publication of the Orthopaedic Research Society 2007;25:789–797. [PubMed: 17343285]
12. Schache AG, Fregly BJ, Crossley KM, et al. The effect of gait modification on the external knee adduction moment is reference frame dependent. Clinical Biomechanics 2008;23:601–608. [PubMed: 18280623]
13. Manal K, McClay I, Richards J, et al. Knee moment profiles during walking: errors due to soft tissue movement of the shank and the influence of the reference coordinate system. Gait & Posture 2002;15:10–17. [PubMed: 11809576]
14. Reinbolt JA, Haftka RT, Chmielewski TL, Fregly BJ. A computational framework to predict post-treatment outcome for gait-related disorders. Medical Engineering & Physics 2008;30:434–443. [PubMed: 17616425]
15. Fregly BJ, D'Lima DD, Colwell CW Jr. Effective gait patterns for offloading the medial compartment of the knee. Journal of Orthopaedic Research 2009;27:1016–1021. [PubMed: 19148939]
16. Fregly BJ, Reinbolt JA, Rooney KL, et al. Design of patient-specific gait modifications for knee osteoarthritis rehabilitation. IEEE Transactions on Biomedical Engineering 2007;54:1687–1695. [PubMed: 17867361]
17. Fregly, BJ.; Rooney, KL.; Reinbolt, JA. Predicted gait modifications to reduce the peak knee adduction torque. Proceedings of the XXth Congress of the International Society of Biomechanics; Cleveland, OH. 2005.

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18. Barrios, JA.; Davis, IS. A gait modification to reduce the external knee adduction moment at the knee: a case study. Proceedings of the 31st Annual Meeting of the American Society of Biomechanics; Palo Alto, CA. 2007.
19. Kaufman KR, Hughes C, Morrey BF, et al. Gait characteristics of patients with knee osteoarthritis. Journal of Biomechanics 2001;34:907–915. [PubMed: 11410174]
20. Thorp LE, Sumner DR, Block JA, et al. Knee joint loading differs in individuals with mild compared with moderate medial knee osteoarthritis. Arthritis and Rheumatism 2006;54:3842–3849. [PubMed: 17133592]
21. Baliunas AJ, Hurwitz DE, Ryals AB, et al. Increased knee joint loads during walking are present in subjects with knee osteoarthritis. Osteoarthritis and Cartilage 2002;10:573–579. [PubMed: 12127838]
22. Prodromos CC, Andriacchi TP, Galante JO. A relationship between gait and clinical changes following high tibial osteotomy. Journal of Bone and Joint Surgery 1985;67A:1188–1194. [PubMed: 4055843]
23. Kean CO, Birmingham TB, Garland JS, et al. Moments and muscle activity after high tibial osteotomy and anterior cruciate ligament reconstruction. Medicine and Science in Sports and Exercise 2009;41:612–619. [PubMed: 19204589]
24. Maly MR, Costigan PA, Olney SJ. Mechanical factors relate to pain in knee osteoarthritis. Clinical Biomechanics 2008;23:796–805. [PubMed: 18346827]
25. Erhart, JC.; Dyrby, C.; D'Lima, DD., et al. Reduction in the Medical Compartment Force Correlates with the Reduction in Knee Adduction Moment Using a Variable-Stiffness Shoe with an Instrumented Knee. Proceedings of the ASME 2009 Summer Bioengineering Conference; Lake Tahoe, CA. 2009.
26. Sharma L, Song J, Felson DT, et al. The role of knee alignment in disease progression and functional decline in knee osteoarthritis. The Journal of the American Medical Association 2001;286:188–195.

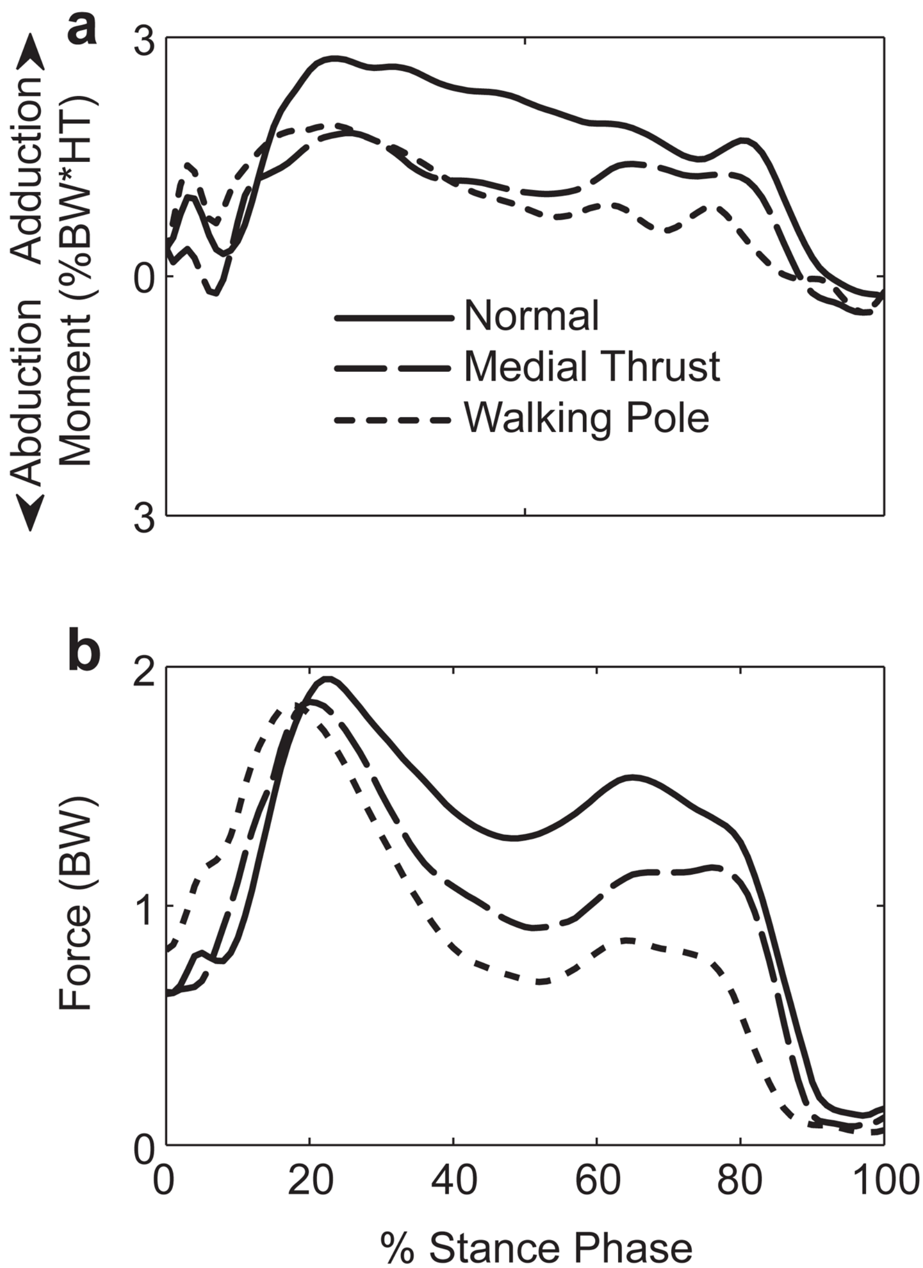

Figure 1.
Mean curves for (a) external knee adduction moment and (b) internal medial contact force over stance phase calculated from five trials of normal, medial thrust, and walking pole gait.

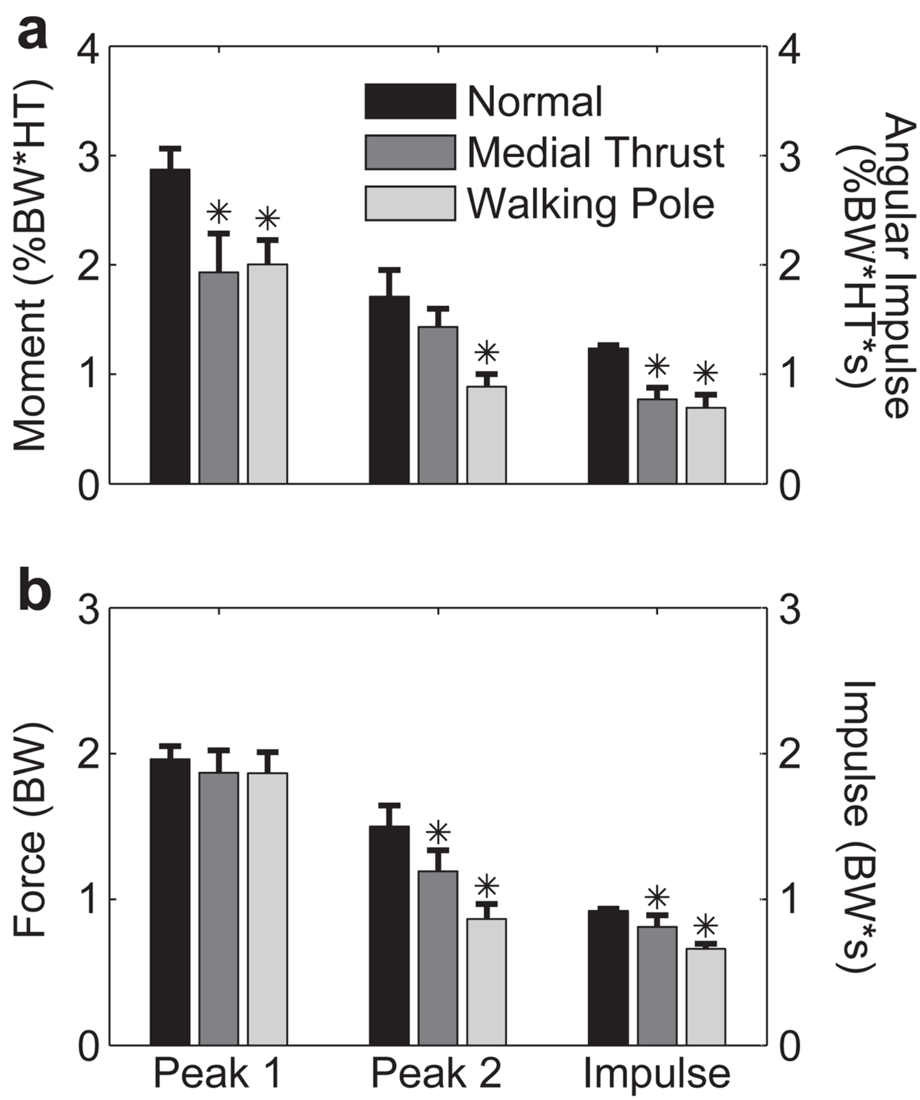

Figure 2.
Peak and impulse values of (a) external knee adduction moment and (b) internal medial contact force for the curves shown in Figure 1. Error bars indicate ± 1 standard deviation. Star (*) denotes statistically significant difference from normal trials.

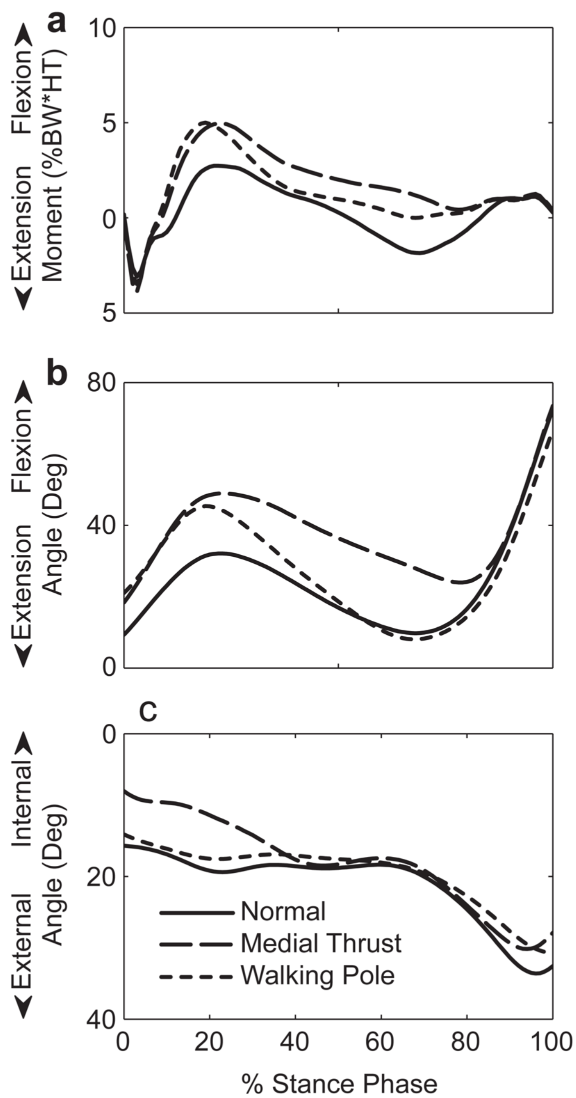

Figure 3.

Mean curves for (a) external knee flexion moment, (b) knee flexion angle, and (c) hip internal rotation angle over stance phase calculated from five trials of normal, medial thrust, and walking pole gait.

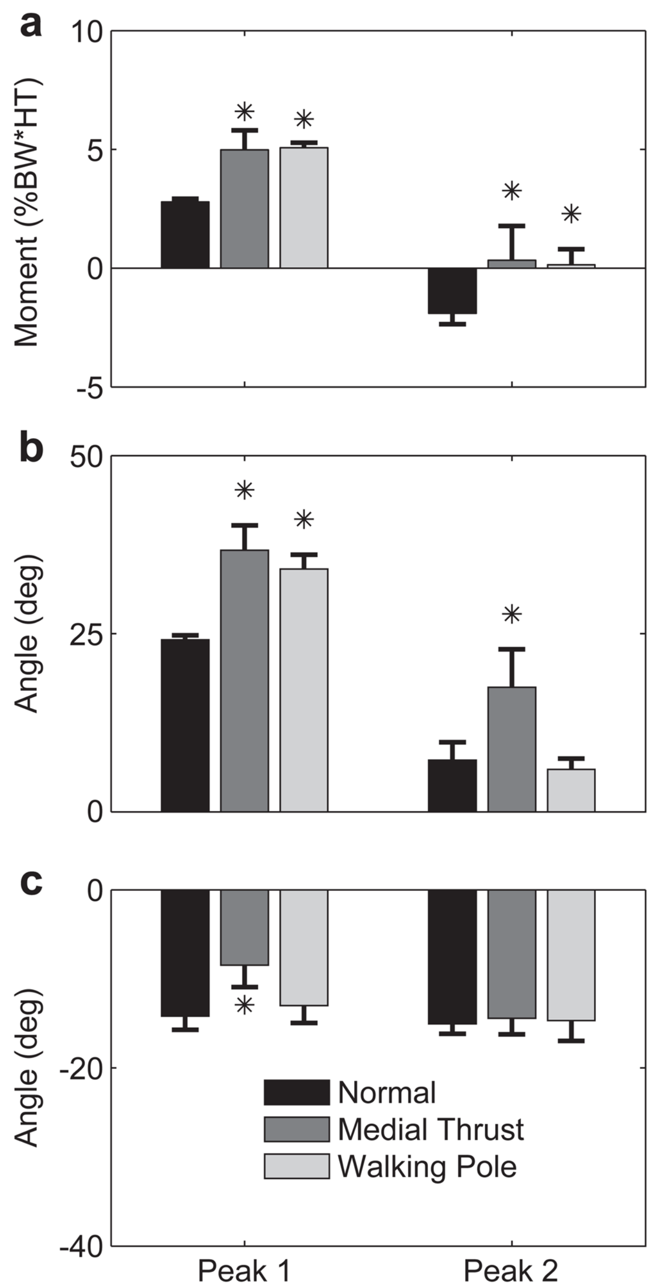

Figure 4.
Peak values of (a) external knee flexion moment, (b) knee flexion angle, and (c) hip internal rotation angle for the curves shown in Figure 3. Error bars indicate ± 1 standard deviation. Star (*) denotes statistically significant difference from normal trials.

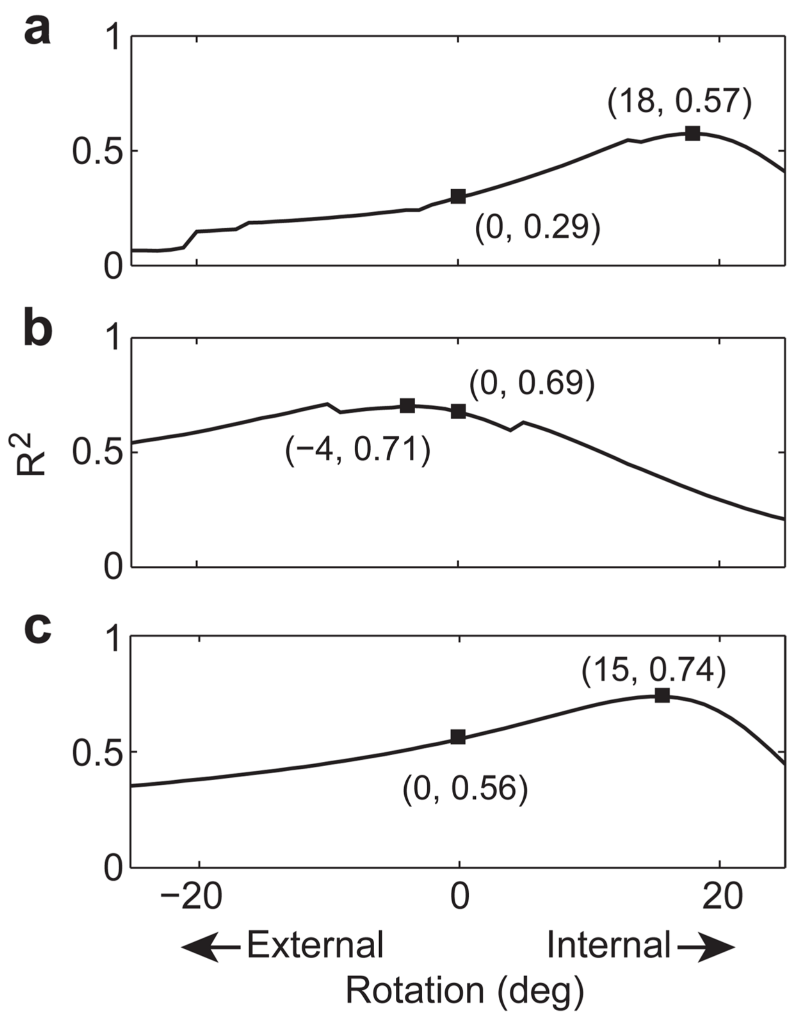

Figure 5.

Sensitivity of R^2 values between internal medial contact force and external knee adduction moment to rotation of the shank reference frame about the superior-inferior axis of the shank. Results are reported for corresponding a) first peak, b) second peak, and c) impulse values. Linear regression analyses were performed using data from 15 gait trials (five trials from each of the three gait patterns). The two data points on each curve indicate R^2 values for nominal (0 degrees) and optimal rotation values.

Table 1

Mean (standard deviation) percent reductions in external knee adduction moment and internal medial contact force peak and impulse relative to normal gait. Star (*) denotes statistically significant difference from normal trials.

		Gait Pattern	
Quantity	**Measure**	**Medial Thrust**	**Walking Pole**
Knee Adduction Moment	First Peak	32.4 (0.8)*	33.1 (0.9)*
	Second Peak	14.9 (1.4)	47.2 (1.0)*
	Angular Impulse	37.8 (0.2)*	43.9 (0.2)*
Medial Contact Force	First Peak	4.5 (0.3)	4.7 (0.3)
	Second Peak	19.9 (0.9)*	41.8 (0.7)*
	Impulse	11.9 (0.2)*	28.0 (0.1)*

Table 2

Linear regression equations and corresponding R^2 and root-mean-square (RMS) error values for medial contact force F_{med} fitted as a function of corresponding external knee adduction moment M_{add} and external knee flexion moment M_{flex} measures (first peak, second peak, both peaks, or impulse). Linear regression analyses were performed using data from 15 gait trials (five trials from each of the three gait patterns).

F_{med} Measure	Shank Ref. Frame	Regression Equation	c_1	c_2	c_3	R^2	RMS Error
Peak 1	Nominal	$F_{med} = c_1 M_{add} + c_2$	.14	1.58	–	**.29**	**.41**
	Optimal	$F_{med} = c_1 M_{add} + c_2$	.36	.68	–	**.57**	**.32**
	Nominal	$F_{med} = c_1 M_{add} + c_2 M_{flex} + c_3$	.31	.09	.82	**.54**	**.33**
	Nominal	$F_{med} = c_1 M_{add} + c_2 \lvert M_{flex}\rvert + c_3$	.31	.09	.82	**.54**	**.33**
Peak 2	Nominal	$F_{med} = c_1 M_{add} + c_2$	.60	.33	–	**.69**	**.63**
	Optimal	$F_{med} = c_1 M_{add} + c_2$	.54	.40	–	**.71**	**.60**
	Nominal	$F_{med} = c_1 M_{add} + c_2 M_{flex} + c_3$	.53	−.04	.40	**.70**	**.61**
	Nominal	$F_{med} = c_1 M_{add} + c_2 \lvert M_{flex}\rvert + c_3$	.33	.21	.49	**.85**	**.43**
Peaks 1 & 2	Nominal	$F_{med} = c_1 M_{add} + c_2$	.57	.49	–	**.68**	**1.29**
	Optimal	$F_{med} = c_1 M_{add} + c_2$	.46	.53	–	**.81**	**.98**
	Nominal	$F_{med} = c_1 M_{add} + c_2 M_{flex} + c_3$	.46	.06	.58	**.82**	**.96**
	Nominal	$F_{med} = c_1 M_{add} + c_2 \lvert M_{flex}\rvert + c_3$	.38	.13	.50	**.93**	**.59**
Impulse	Nominal	$F_{med} = c_1 M_{add} + c_2$	.34	.49	–	**.56**	**.30**
	Optimal	$F_{med} = c_1 M_{add} + c_2$	.66	.08	–	**.74**	**.23**
	Nominal	$F_{med} = c_1 M_{add} + c_2 M_{flex} + c_3$	.63	.16	.10	**.74**	**.23**
	Nominal	$F_{med} = c_1 M_{add} + c_2 \lvert M_{flex}\rvert + c_3$	.63	.16	.10	**.74**	**.23**