Contribution of Tibiofemoral Joint Contact to Net Loads at the Knee in Gait

Jonathan P. Walter,[1] Nuray Korkmaz,[2] Benjamin J. Fregly,[3] Marcus G. Pandy[1]

[1]Department of Mechanical Engineering, University of Melbourne, Melbourne, Australia, [2]Department of Mechanical Engineering, Istanbul University, Avcilar, Istanbul, Turkey, [3]Department of Mechanical and Aerospace Engineering, University of Florida, Gainesville, Florida

Received 18 September 2014; accepted 27 January 2015
Published online 2 March 2015 in Wiley Online Library (wileyonlinelibrary.com). DOI 10.1002/jor.22845

ABSTRACT: Inverse dynamics analysis is commonly used to estimate the net loads at a joint during human motion. Most lower-limb models of movement represent the knee as a simple hinge joint when calculating muscle forces. This approach is limited because it neglects the contributions from tibiofemoral joint contact forces and may therefore lead to errors in estimated muscle forces. The aim of this study was to quantify the contributions of tibiofemoral joint contact loads to the net knee loads calculated from inverse dynamics for multiple subjects and multiple gait patterns. Tibiofemoral joint contact loads were measured in four subjects with instrumented implants as each subject walked at their preferred speed (normal gait) and performed prescribed gait modifications designed to treat medial knee osteoarthritis. Tibiofemoral contact loads contributed substantially to the net knee extension and knee adduction moments in normal gait with mean values of 16% and 54%, respectively. These findings suggest that knee-contact kinematics and loads should be included in lower-limb models of movement for more accurate determination of muscle forces. The results of this study may be used to guide the development of more realistic lower-limb models that account for the effects of tibiofemoral joint contact at the knee. © 2015 Orthopaedic Research Society. Published by Wiley Periodicals, Inc. J Orthop Res 33:1054–1060, 2015.

Keywords: inverse dynamics; tibiofemoral contact; knee adduction moment; lower-limb musculoskeletal model

The net loads (forces and moments) acting at a joint are generated by a combination of muscle, ligament, and joint contact forces. Inverse dynamics analysis is commonly used to obtain estimates of these net loads when studying human motion. However, accurate knowledge of the internal loads applied individually by muscles, ligaments, and joint contact is needed for a more comprehensive understanding of musculoskeletal function. Because non-invasive measurement of the internal loads due to muscle/ligament tension and joint contact is not possible, computational models are often used in conjunction with gait analysis to determine musculoskeletal loading in vivo.[1–10]

Estimates of the forces and moments produced by muscles and joint contact (e.g., the moments about the knee joint center produced by the tibiofemoral joint contact forces) may be obtained by balancing these internal loads against the net loads calculated from inverse dynamics as follows (see Fig. 1):

$$\begin{gathered} F_{\mathrm{IVD},i} = F_{\mathrm{MUSC},i} + F_{\mathrm{CONT},i} + F_{\mathrm{LIG},i} \\ M_{\mathrm{IVD},i} = M_{\mathrm{MUSC},i} + M_{\mathrm{CONT},i} + M_{\mathrm{LIG},i} \\ i = x,y,z \end{gathered} \tag{1}$$

where F_{IVD} and M_{IVD} are the net forces and moments calculated from inverse dynamics, respectively; F_{MUSC} and M_{MUSC} are the forces and moments applied by muscles; F_{CONT} and M_{CONT} are the forces and moments arising from joint contact; F_{LIG} and M_{LIG} are the forces and moments applied by ligaments; and x, y, z represent the anterior–posterior, superior–inferior, and medial–lateral directions, respectively.

In most lower-limb models of movement, muscle forces are determined by representing the knee as a simple hinge joint, which neglects the contributions that ligament forces and tibiofemoral joint contact forces make to the net knee extension moment; that is, $M_{\mathrm{LIG,Z}}$ and $M_{\mathrm{CONT,Z}}$ are assumed to be zero in Equation (1).[8,9,11,12] This assumption may be valid if the moments produced by ligament forces and tibiofemoral joint contact forces are much smaller than those developed by the muscles; for example, if $M_{\mathrm{CONT,Z}}$ is less than 10% of $M_{\mathrm{IVD,Z}}$. However, if ligament and joint contact forces contribute more substantially to the net loads, then the above simplification may yield erroneous estimates of lower-limb muscle forces.

Representing the knee as a simple hinge joint also means that only the net knee flexion–extension moment equation is considered in the calculation of lower-limb muscle forces. More accurate estimates of muscle forces may be obtained by also balancing the net loads acting in the frontal and transverse planes, and in particular, accounting for the contributions of tibiofemoral joint contact forces to the net knee adduction moment, Mx, and the net knee superior force, Fy. The knee adduction moment is related to the distribution of forces between the medial and lateral compartments of the knee[13–15] and has been a focal point in studies aimed at modifying the gait patterns of patients with medial compartment osteoarthritis (OA).[16,17] In addition, the net knee superior force may be an important indicator of the total knee joint contact force.[18] Because muscle/ligament tension and joint contact forces cannot be measured directly, validation of models that have included these quantities has been difficult and limited to single subjects.[19] Recent studies have examined the

Grant sponsor: Australian Council Discovery Projects Grant; Grant number: DP120101973; Grant sponsor: Victorian Endowment for Science and Knowledge Innovation; Grant sponsor: NIH; Grant number: R01EB009351.
Correspondence to: Jonathan P. Walter (T: +61 3 8344 0405; F: +61 3 9347 8784; E-mail: walterj@unimelb.edu.au)

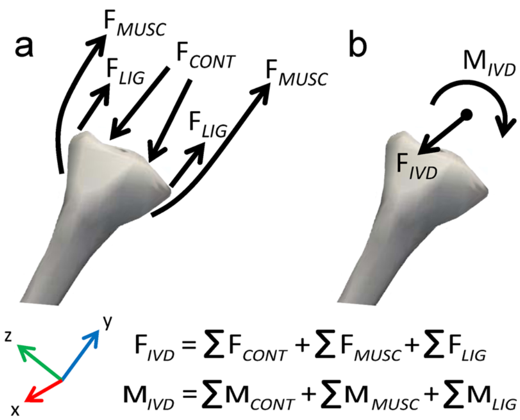

Figure 1. Diagram illustrating the forces and moments (loads) acting at the proximal tibia. (a) Internal loads acting at the proximal tibia; and (b) reaction loads calculated from inverse dynamics required to balance the internal loads acting at the proximal tibia.

relationships between net knee loads calculated from inverse dynamics and measurements of the medial, lateral, and total tibiofemoral joint contact forces obtained from instrumented knee implants,[13,14,20–22] but the contributions of measured tibiofemoral joint contact forces to multiple components of the net knee loads have yet to be identified.

The aim of this study was to quantify the contributions of tibiofemoral joint contact loads to the net knee loads calculated from inverse dynamics for multiple subjects and multiple gait patterns. Tibiofemoral contact loads were measured in four subjects with instrumented tibial prostheses as each subject walked at their preferred speed (normal gait) and performed prescribed gait modifications designed to treat medial knee OA.[23] We tested an assumption implicit in most musculoskeletal models used to simulate lower-limb movement: that the magnitude of the knee extension moment contributed by the tibiofemoral joint contact force is relatively small (i.e., less than 10%) compared to the magnitude of the net knee extension moment calculated from inverse dynamics. We also tested the proposition that the adduction moment contributed by the tibiofemoral joint contact force correlates with the net knee adduction moment.[13,14,20,21] Accurate knowledge of the relationships between tibiofemoral contact loads and the net loads at the knee would improve the accuracy of muscle force calculations in the lower limb.

METHODS

Experimental data obtained from the past five "Grand Challenge Competitions to Predict In Vivo Knee Loads"[23,24] were used in this study. Four subjects (one tested twice) implanted with load-measuring knee replacements performed over-ground walking with simultaneous collection of tibiofemoral contact loads, marker positions, and ground reaction forces (GRFs). Institutional review board approval and informed consent were obtained. Two types of tibial implants were used: (i) the eKnee, a custom tibial prosthesis equipped with four uniaxial force transducers that measured compressive forces in four quadrants of the tibial tray[25] (one subject; first and fourth Grand Challenge [GC] competition datasets, GC1 and GC4); and (ii) the eTibia, another custom tibial prosthesis comprised of strain gauges calibrated to measure three orthogonal forces and moments at the center of the tibial post and level with the top surface of the tibial tray[26] (three subjects; one subject each for GC2, GC3, and GC5). Each subject performed five trials of level walking using their natural gait pattern, followed by up to five trials for each of the following modified gait patterns: medial-thrust gait, trunk-sway gait, and walking-pole gait. These three gait modifications were chosen because they are purported to reduce the net knee adduction moment[16] and thereby offload the medial compartment of the knee.[17] Each subject walked at his or her preferred speed during all gait trials.

A generic musculoskeletal model described by Arnold et al.[27] was modified and used to simulate each subject's gait pattern. The model included scalable segments representing the feet, shanks, thighs, pelvis, and torso/head. The model was scaled to each subject using individual segment scale factors calculated from anatomical marker locations and CT images. Femoral and tibial scale factors were calculated by comparing and aligning the nominal bone geometries to subject-specific bone geometries created from CT scans of the subject's lower limbs. The models of the subject-specific tibia and tibial implant were carefully aligned to the scaled tibial geometry to ensure that the measured contact loads were correctly positioned relative to the knee joint center. The models of the lumbosacral, hip, ankle and subtalar joints were identical with those represented in the generic model.[27] The knee was represented as a translating ball-and-socket joint by adding two degrees-of-freedom (DOF) to the existing translating hinge joint to include internal–external and adduction–abduction rotations of the tibia relative to the femur. An additional six-DOF joint with all coordinates locked to zero was added at the knee joint center in the tibial reference frame to allow three orthogonal forces and three orthogonal moments (including Fy, Mx, and Mz) to be calculated during an inverse dynamics analysis.[10]

Net joint loads were calculated using the scaled musculoskeletal models in OpenSim.[28] An inverse kinematics analysis was performed for each gait trial to calculate the rotations and translations at each joint. The inverse kinematics results and the GRFs were then used to perform an inverse dynamics analysis to determine the net loads (i.e., forces and moments) acting at the knee joint center. GRFs were low-pass filtered at 20 Hz using a fourth-order zero phase-lag Butterworth filter while joint displacements were filtered similarly at 6 Hz. The six net knee-joint loads were expressed in the scaled-generic tibial reference frame (Fig. 1). All moments were calculated about the knee joint center in the tibial reference frame as defined in the generic OpenSim knee model.

The four forces measured by the eKnee implant were summed to obtain the total compressive joint contact force, Fy. The corresponding joint contact moments, Mx and Mz, were calculated by multiplying the force measured by each eKnee transducer by the distance from the transducer to the center of the top surface of the tibial tray.[25] The eTibia implant provided measurements of all six load components applied to the tibial tray. The joint contact loads obtained from both instrumented implants were then re-expressed in the same tibial reference frame using the same point (i.e., center of the knee joint) as that used to calculate the net knee-joint loads from inverse dynamics.

Contact-to-net load ratios were used to quantify the contributions of tibiofemoral joint contact to the net knee loads acting at the knee joint center. These contact-to-net load ratios were calculated during the mid-stance and terminal-stance phases of the gait cycle, where peak joint contact loads were observed. Means and standard deviations (s.d.) of the ratios were calculated for each gait pattern across all time frames of the aforementioned phases as well as across all trials and all subjects. A statistical correlation measure was also computed to determine whether a general relationship could be assumed between a given contact load and the corresponding net load. These contact-to-net correlation coefficients were calculated for each gait pattern across all time frames during the entire stance phase of gait as well as across all trials and all subjects. Because these data were non-normally distributed and a linear relationship therefore could not be assumed, a Spearman rank correlation test was performed. Linear regression analysis between the contact loads and the net loads was also performed and the corresponding R^2 (coefficient of determination) errors and RMS errors calculated. Both correlation and R^2 measures were classified as poor (0.0–0.5), moderate (0.5–0.75), good (0.75–0.9), or strong (0.90–1.0).[21]

Estimates of the remaining loads (i.e., those arising from the muscles and ligaments) were found by subtracting the tibiofemoral joint contact loads from the net knee loads at each time frame. The result represented the combined contribution of muscles and ligaments to the net knee loads at the knee joint center. Moments were normalized to a percentage of the subject's body weight (BW) multiplied by the subject's height (HT) while forces were normalized to the subject's body weight.

RESULTS

The tibiofemoral joint contact force contributed substantially to the net knee adduction moment in all four gait patterns. For normal gait, the contact-to-net adduction moment ratio varied from -109% to 180% with a mean $\pm$ s.d. of $54 \pm 44\%$ across all subjects (Figs. 2 and 3, Table 1). The magnitudes of this ratio were slightly lower for the medial thrust and walking-pole gait patterns (51% and 47%, respectively) and markedly lower for the trunk sway gait pattern (28%) compared to that calculated for normal gait.

The contribution of the tibiofemoral joint contact force to the net knee extension moment was higher than expected for normal gait; the contact-to-net extension moment ratio varied from -47% to 66% with a mean $\pm$ s.d. of $16 \pm 13\%$ (Figs. 2 and 3). The corresponding mean values for the medial thrust, walking-pole, and trunk sway gait patterns were lower at 7%, 8%, and 6%, respectively (Table 1).

The superior–inferior tibiofemoral joint contact force was higher than the superior–inferior net joint force calculated from inverse dynamics in all subjects and for all gait patterns (Figs. 2 and 3). For normal gait, the contact-to-net superior force ratio ranged from 141% to 369% with a mean $\pm$ s.d. of $230 \pm 39\%$ across all subjects (Table 1). The magnitudes of this ratio were higher for the medial thrust and trunk sway gait patterns (267% and 261%, respectively) and practically the same for the walking-pole gait pattern (228%) compared to that calculated for normal gait (Table 1).

The contact-to-net load correlations varied from poor to good. The contact-to-net adduction moment correlations were poor to moderate with coefficients ranging from 0.43 to 0.50 across the four gait patterns (Table 1). The contact-to-net extension moment correlations were also poor to moderate with coefficients ranging from 0.02 to 0.64. The contact-to-net superior force correlations were the highest with coefficients ranging from 0.73 to 0.76. Linear regression R^2 values were poor for the knee adduction and extension moments and moderate for the knee superior force. Corresponding RMS errors ranged from 0.34%BW*HT to 0.74%BW*HT for the adduction and extension moments and from 0.32 BW to 0.54 BW for the superior force.

DISCUSSION

The aim of this study was to quantify the contributions of tibiofemoral joint contact loads to the net loads acting at the knee for multiple subjects and multiple gait patterns. Spearman correlation coefficients and ratios were used to describe the relationships between the joint contact loads and net loads. We found that tibiofemoral joint contact loads contributed substantially to the net knee extension moment calculated from inverse dynamics (Fig. 2); for normal gait, the mean value of the contact-to-net extension moment ratio was 16% (Table 1). The mean value of the contact-to-net adduction moment ratio was higher at 54%, but this value varied widely between time frames as well as across subjects and gait patterns, and resulted in poor correlation measures. The magnitudes of the ratio of contact-to-net adduction moment were lower for the trunk-sway and medial-thrust gaits compared to normal gait, but the corresponding values of the ratio of contact-to-net superior force were higher (Fig. 3 and Table 1).

Trepczynski et al.[21] reported a value of 0.65 for a linear regression between the contact and net knee adduction moments for normal gait, which is similar to the mean ratio of 0.54 (54%) calculated in the present study (Table 1). These authors also reported an R^2 value of 0.90 for a linear regression between the contact and net knee adduction moments across multiple tasks.[21] We performed a linear regression analysis across four different gait patterns and obtained a much lower R^2 value of 0.19 and a correspondingly low correlation coefficient of 0.48 (Table 1). Previous studies investigating the relationship between the net knee adduction moment and medial contact force reported poor (0.25) to good (0.77) R^2 values for linear regressions performed on multiple gait trials.[13,14,20,22] A correlation coefficient of 0.50 and a linear regression R^2 value of 0.21 for the relationship between the contact and net knee adduction moments for normal gait obtained in this study are at the lower end of the range found in the literature, which may be explained

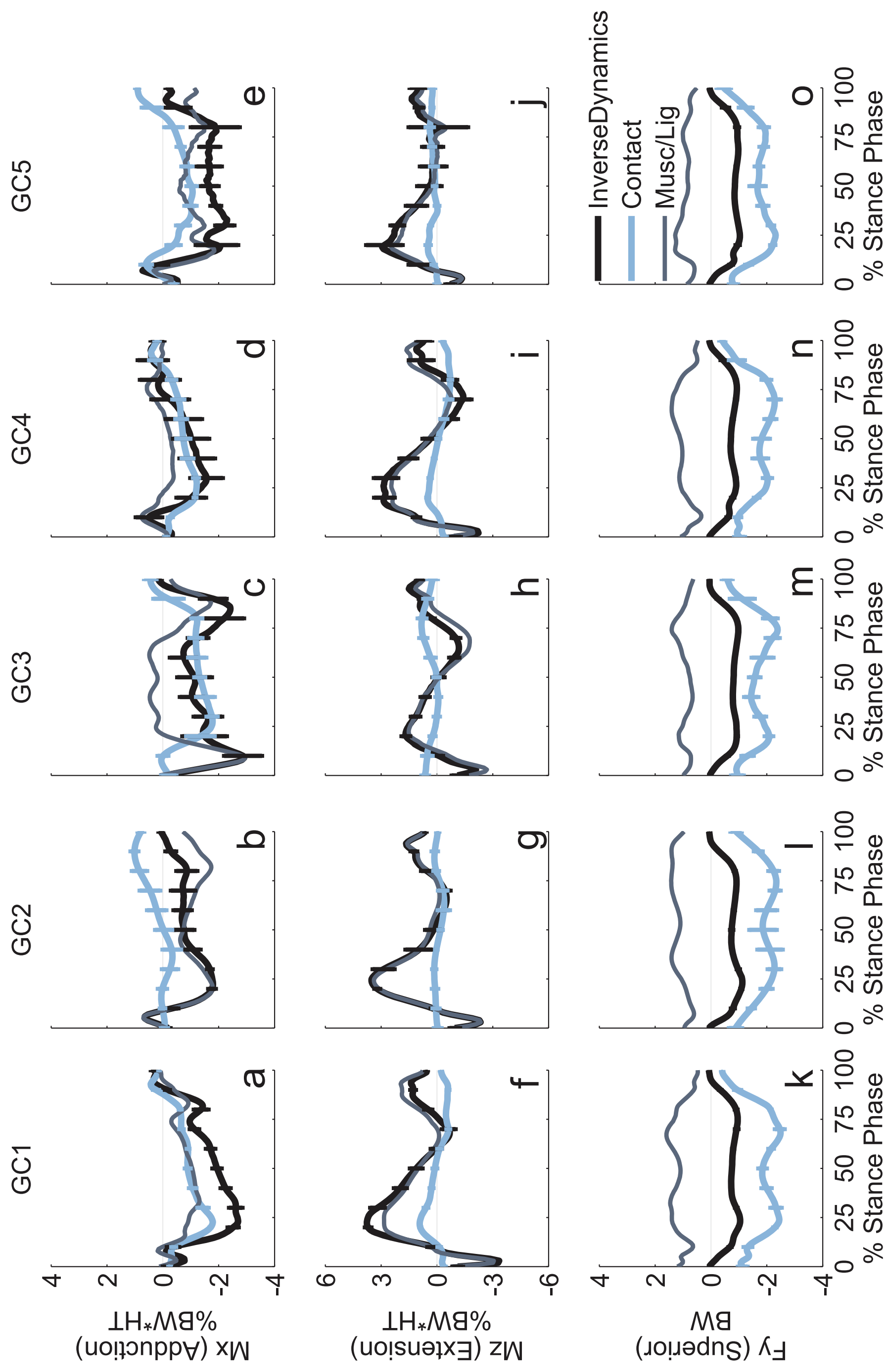

Figure 2. Mean joint contact, net, and muscle/ligament loads calculated for the knee joint during the stance phase of normal gait. Data are shown for the three load conditions: knee adduction moment (M*x*, a–e); knee extension moment (M*z*, f–j); and knee superior force (F*y*, k–o). GC1(a,f,k), GC2(b,g,l), GC3(c,h,m), GC4(d,i,n) and GC5(e,j,o) represent the "Grand Challenge datasets" used for the analysis. The vertical lines indicate ± 1 s.d. for the joint contact and net loads.

in part by the large inter-subject variability observed in our data (see Fig. 2 and Supplementary Material).

Although the contribution of the contact adduction moment to the net knee adduction moment was substantial throughout most of the stance phase, the contributions from muscles and ligaments were not inconsiderable. These results imply that neither tibiofemoral joint contact nor muscle action alone is sufficient to balance the net knee adduction moment and that both are required for dynamic equilibrium at the knee in the frontal plane.[29] In addition, the relationship between the contact and net knee adduction moments was poorly correlated and the ratio of these moments varied considerably across subjects and gait patterns. This result suggests that the contact adduction moment cannot be predicted from knowledge of the net knee

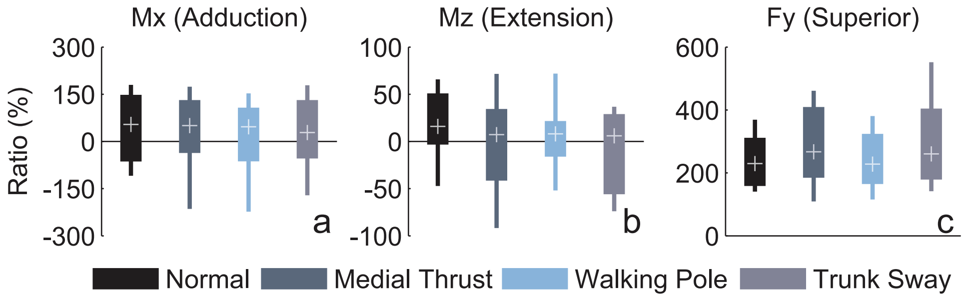

Figure 3. Contact-to-net load ratios (expressed as a percentage) for each gait pattern analyzed in this study: normal, medial thrust, walking-pole, and trunk sway. Data are shown for the three load conditions: knee adduction moment (Mx, a); knee extension moment (Mz, b); and knee superior force (Fy, c). The vertical lines indicate the entire range of data; the thick bars represent ± 3 s.d.s; the + symbol represents the mean. Positive values of the contact-to-net moment load ratio indicate that the contact moment acted in the same direction as the net adduction moment whereas negative values indicate that the contact moment acted in the opposite direction. Note that the joint contact force was higher than the knee superior force in all gait patterns.

adduction moment alone and vice versa. We conclude, therefore, that more accurate estimates of the internal loads at the knee may be obtained by including both muscle and joint contact forces in lower-limb musculoskeletal models of movement.[1,30]

The largest values of the ratio of contact-to-net knee extension moment (i.e., greater than 10%) coincided with a posterior shift of the contact center-of-pressure on the tibial plateau during the mid-stance phase of gait. A posterior shift of the contact center-of-pressure would increase the extension moment applied to the tibia by the tibiofemoral contact force, and consequently reduce the extension moment needed from the quadriceps to balance the flexion moment applied by the ground reaction force. Posterior displacement of the contact center-of-pressure may have been caused by excessive anterior translation of the tibia relative to the femur, which has been found in anterior-cruciate-ligament-deficient[31] and post-total-knee-arthroplasty[32] patients at 0 and 15° of loaded knee flexion. If this interpretation is correct, then the magnitude of the contact extension moment calculated during mid-stance may not be as high in people with healthy knees, as anterior tibial translation will presumably be less due to tension in the anterior cruciate ligament (ACL).[12] Future studies should be directed at investigating the cause of a larger-than-expected contribution of the tibiofemoral joint contact force to the net extension moment by including the effects of ligament and joint contact forces in musculoskeletal simulations of knee-joint movement.

Many studies have related the net knee adduction moment to the medial contact force by assuming that an increase in the net knee adduction moment would correspond to an increase in the ratio of the medial-to-lateral contact force.[33–41] While the contact adduction moment is directly proportional to the ratio of the medial-to-lateral contact force (assuming the total contact force remains unchanged), this may not be the case for the net knee adduction moment because muscles and ligaments also contribute to this quantity.[8,9,11] We found only a moderate correlation between the contact and net adduction moments, which suggests that a simple linear relationship

Table 1. Linear Regression Values (R^2 and RMSE), Spearman Correlation Coefficients (ρ), and Ratios of Contact-to-Net Loads Expressed as Percentages

Load	Measure	Normal	Medial Thrust	Walking Pole	Trunk Sway	All Gait
Mx, adduction	ρ	0.50	0.46	0.43	0.44	0.48
	R^2	0.21	0.15	0.17	0.20	0.19
	RMSE(%BW*HT)	0.62	0.65	0.49	0.74	0.60
	Ratio mean(STD) %	54(44)	51(43)	47(40)	28(42)	48(43)
Mz, extension	ρ	0.33	0.41	0.64	0.02	0.40
	R^2	0.11	0.21	0.42	0.00	0.19
	RMSE(%BW*HT)	0.40	0.51	0.34	0.50	0.45
	Ratio mean(STD) %	16(13)	7(17)	8(9)	6(16)	9(14)
Fy, superior	ρ	0.73	0.73	0.76	0.73	0.75
	R^2	0.66	0.54	0.57	0.61	0.55
	RMSE(BW)	0.32	0.54	0.32	0.40	0.42
	Ratio mean(STD) %	230(39)	267(62)	228(37)	261(57)	241(50)

Values in parentheses represent ± 1 s.d.

between the net knee adduction moment and ratio of medial-to-lateral contact force (or medial contact force) may not exist.

The magnitudes of the contact adduction moment and the contact-to-net adduction moment ratio were lower for each of the gait modifications than for normal walking, implying that the medial compartment of the knee may have been off-loaded when the subjects adopted a modified gait pattern. Unfortunately, however, the tibiofemoral joint contact force was higher for the trunk-sway and medial-thrust gaits as evidenced by an increase in the mean value of the contact-to-net superior force ratio. A higher joint contact force, which may have been caused by an increase in knee muscle co-contraction, would negate any reduction in the medial contact force that was caused by a lower knee adduction moment. In contrast, the contact-to-net superior force ratio was slightly lower for walking-pole gait than normal gait, implying that the medial compartment force may have been reduced. It should be noted that this result reflects the average obtained across all trials and all subjects; the effect of walking with poles on load distribution at the knee in any given individual was variable (see Supplementary Material).[42]

Perhaps the most significant limitation of this study was the cost associated with surgically implanting load-measuring prostheses in patients with end-stage knee OA, which limited the current dataset to four subjects. The age of the subjects tested together with the removal of the ACL at the time of surgery also limit the applicability of our results to the general population. The analysis undertaken was also constrained by the fact that only three joint contact loads (i.e., knee adduction moment, knee extension moment, and knee superior force) were measured by both implant systems. Finally, it is acknowledged that the results obtained in this study depend on the location of the knee joint center assumed in the model. A post-hoc analysis showed that a 1 cm posterior displacement of the knee joint center (relative to the location assumed in the nominal model) decreased the mean contact-to-net extension moment ratio calculated for normal gait from 16% to 10% and increased the standard deviation from 13% to 74% (see Supplementary Material). In contrast, a 1 cm anterior displacement of the knee joint center increased both the mean and standard deviation of the contact-to-net extension moment ratio for normal gait from 16% to 55% and from 13% to 22%, respectively. These results demonstrate the sensitivity of the calculated values of knee-joint moments to changes in the location of the knee joint center in the model.

The present study quantified the contribution of joint contact to the net loads acting at the knee for multiple subjects and multiple gait patterns. We found that the tibiofemoral joint contact force contributed substantially to the net knee adduction moment and the net knee extension moment, implying that knee-contact kinematics and loads should be included in lower-limb models of movement for more accurate determination of muscle forces. The results of this study may be used to guide the development of more realistic lower-limb models that account for the effects of tibiofemoral joint contact at the knee.

ACKNOWLEDGMENTS

This work was funded by an Australian Council Discovery Projects Grant (DP120101973) and an Innovation Fellowship from the Victorian Endowment for Science and Knowledge Innovation to M.G.P. Partial funding was also provided by NIH grant R01EB009351. We thank Dr. Yi-Chung Lin for his comments on a previous version of this paper.

REFERENCES

1. Lin Y-C, Walter JP, Banks SA, et al. 2010. Simultaneous prediction of muscle and contact forces in the knee during gait. J Biomech 43:945–952.
2. Taylor WR, Heller MO, Bergmann G, et al. 2004. Tibiofemoral loading during human gait and stair climbing. J Orthop Res 22:625–632.
3. Gerus P, Sartori M, Besier TF, et al. 2013. Subject-specific knee joint geometry improves predictions of medial tibiofemoral contact forces. J Biomech 46:2778–2786.
4. Sasaki K, Neptune RR. 2010. Individual muscle contributions to the axial knee joint contact force during normal walking. J Biomech 43:2780–2784.
5. Thelen DG, Won Choi K, Schmitz AM. 2014. Co-simulation of neuromuscular dynamics and knee mechanics during human walking. J Biomech Eng 136:021033.
6. Demers MS, Pal S, Delp SL. 2014. Changes in tibiofemoral forces due to variations in muscle activity during walking. J Orthop Res 32:769–776.
7. Hast MW, Piazza SJ. 2013. Dual-joint modeling for estimation of total knee replacement contact forces during locomotion. J Biomech Eng 135:021013.
8. Pandy MG, Andriacchi TP. 2010. Muscle and joint function in human locomotion. Annu Rev Biomed Eng 12:401–433.
9. Sritharan P, Lin Y-C, Pandy MG. 2012. Muscles that do not cross the knee contribute to the knee adduction moment and tibiofemoral compartment loading during gait. J Orthop Res 30:1586–1595.
10. Walter JP, Kinney AL, Banks SA, et al. 2014. Muscle synergies may improve optimization prediction of knee contact forces during walking. J Biomech Eng 136:021031.
11. Shelburne KB, Torry MR, Pandy MG. 2006. Contributions of muscles, ligaments, and the ground-reaction force to tibiofemoral joint loading during normal gait. J Orthop Res 24:1983–1990.
12. Shelburne KB, Torry MR, Pandy MG. 2005. Muscle, ligament, and joint-contact forces at the knee during walking. Med Sci Sports Exerc 37:1948–1956.
13. Zhao D, Banks SA, Mitchell KH, et al. 2007. Correlation between the knee adduction torque and medial contact force for a variety of gait patterns. J Orthop Res 25:789–797.
14. Kutzner I, Trepczynski A, Heller MO, et al. 2013. Knee adduction moment and medial contact force-facts about their correlation during gait. PloS ONE 8:e81036.
15. Schipplein OD, Andriacchi TP. 1991. Interaction between active and passive knee stabilizers during level walking. J Orthop Res 9:113–119.
16. Van den Noort JC, Schaffers I, Snijders J, et al. 2013. The effectiveness of voluntary modifications of gait pattern to reduce the knee adduction moment. Hum Mov Sci 32:412–424.
17. Fregly BJ, D'Lima DD, Colwell CW Jr. 2009. Effective gait patterns for offloading the medial compartment of the knee. J Orthop Res 27:1016–1021.

18. Mündermann A, Dyrby CO, Andriacchi TP. 2005. Secondary gait changes in patients with medial compartment knee osteoarthritis: increased load at the ankle, knee, and hip during walking. Arthritis Rheum 52:2835–2844.
19. Kim HJ, Fernandez JW, Akbarshahi M, et al. 2009. Evaluation of predicted knee-joint muscle forces during gait using an instrumented knee implant. J Orthop Res 27:1326–1331.
20. Meyer AJ, D'Lima DD, Besier TF, et al. 2013. Are external knee load and EMG measures accurate indicators of internal knee contact forces during gait? J Orthop Res 31:921–929.
21. Trepczynski A, Kutzner I, Bergmann G, et al. 2014. Modulation of the relationship between external knee adduction moments and medial joint contact forces across subjects and activities. Arthritis Rheumatol 66:1218–1227.
22. Walter JP, D'Lima DD, Colwell CW Jr, et al. 2010. Decreased knee adduction moment does not guarantee decreased medial contact force during gait. J Orthop Res 28:1348–1354.
23. Fregly BJ, Besier TF, Lloyd DG, et al. 2012. Grand challenge competition to predict in vivo knee loads. J Orthop Res 30:503–513.
24. Kinney AL, Besier TF, D'Lima DD, et al. 2013. Update on grand challenge competition to predict in vivo knee loads. J Biomech Eng 135:021012.
25. D'Lima DD, Townsend CP, Arms SW, et al. 2005. An implantable telemetry device to measure intra-articular tibial forces. J Biomech 38:299–304.
26. Kirking B, Krevolin J, Townsend C, et al. 2006. A multiaxial force-sensing implantable tibial prosthesis. J Biomech 39: 1744–1751.
27. Arnold EM, Ward SR, Lieber RL, et al. 2010. A model of the lower limb for analysis of human movement. Ann Biomed Eng 38:269–279.
28. Delp SL, Anderson FC, Arnold AS, et al. 2007. OpenSim: open-source software to create and analyze dynamic simulations of movement. IEEE Trans Biomed Eng 54:1940–1950.
29. Morrison JB. 1970. The mechanics of the knee joint in relation to normal walking. J Biomech 3:51–61.
30. Shelburne KB, Pandy MG. 2002. A dynamic model of the knee and lower limb for simulating rising movements. Comput Method Biomech 5:149–160.
31. Hosseini A, Van de Velde S, Gill TJ, et al. 2012. Tibiofemoral cartilage contact biomechanics in patients after reconstruction of a ruptured anterior cruciate ligament. J Orthop Res 30:1781–1788.
32. Yue B, Varadarajan KM, Moynihan AL, et al. 2011. Kinematics of medial osteoarthritic knees before and after posterior cruciate ligament retaining total knee arthroplasty. J Orthop Res 29:40–46.
33. Erhart JC, Dyrby CO, D'Lima DD, et al. 2010. Changes in in vivo knee loading with a variable-stiffness intervention shoe correlate with changes in the knee adduction moment. J Orthop Res 28:1548–1553.
34. Fregly BJ, Reinbolt JA, Rooney KL, et al. 2007. Design of patient-specific gait modifications for knee osteoarthritis rehabilitation. IEEE Trans Biomed Eng 54:1687–1695.
35. Barrios JA, Crossley KM, Davis IS. 2010. Gait retraining to reduce the knee adduction moment through real-time visual feedback of dynamic knee alignment. J Biomech 43:2208–2213.
36. Hunt MA, Birmingham TB, Bryant D, et al. 2008. Lateral trunk lean explains variation in dynamic knee joint load in patients with medial compartment knee osteoarthritis. Osteoarthritis Cartilage 16:591–599.
37. Fantini Pagani CH, Potthast W, Brüggemann G-P. 2010. The effect of valgus bracing on the knee adduction moment during gait and running in male subjects with varus alignment. Clin Biomech 25:70–76.
38. Kakihana W, Akai M, Nakazawa K, et al. 2005. Effects of laterally wedged insoles on knee and subtalar joint moments. Arch Phys Med Rehabil 86:1465–1471.
39. Kemp G, Crossley KM, Wrigley TV, et al. 2008. Reducing joint loading in medial knee osteoarthritis: shoes and canesshoes and canes. Arthritis Rheum 59:609–614.
40. Thorp LE, Sumner DR, Wimmer MA, et al. 2007. Relationship between pain and medial knee joint loading in mild radiographic knee osteoarthritis. Arthritis Rheum 57:1254–1260.
41. Guo M, Axe MJ, Manal K. 2007. The influence of foot progression angle on the knee adduction moment during walking and stair climbing in pain free individuals with knee osteoarthritis. Gait Posture 26:436–441.
42. Kinney AL, Besier TF, Silder A, et al. 2013. Changes in in vivo knee contact forces through gait modification. J. Orthop Res. 31:434–440.

SUPPORTING INFORMATION

Additional supporting information may be found in the online version of this article at the publisher's web-site.