Sensitivity of Knee Replacement Contact Calculations to Kinematic Measurement Errors

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ABSTRACT: The ability to measure in vivo knee kinematics accurately makes it tempting to calculate in vivo contact forces, pressures, and areas directly from kinematic data. However, the sensitivity of contact calculations to kinematic measurement errors has not been adequately investigated. To address this issue, we developed a series of sensitivity analyses derived from a validated in vivo computational simulation of gait. The simulation used an elastic foundation contact model to reproduce in vivo contact force, center of pressure, and fluoroscopic motion data collected from an instrumented knee replacement. Treating each degree of freedom (DOF) in the simulation as motion controlled, we first quantified how errors in measured relative pose of the implant components affected contact calculations. Pose variations of ± 0.1 mm or degree over the entire gait cycle changed maximum contact force, pressure, and area by 204, 100, and 117%, respectively. Larger variations of ± 0.5 mm or degree changed these same quantities by 1157, 108, and 578%, respectively. In both cases, the largest sensitivities were to errors in superior-inferior translation and varus-valgus rotation, with loss of contact occurring on one or both sides. We then quantified how switching the sensitive DOFs from motion to load control affected the sensitivity results. Pose variations of ± 0.5 mm or degree in the remaining DOFs changed maximum contact quantities by at most 3%. These results suggest that accuracy on the order of microns and milliradians is needed to estimate contact forces, pressures, and areas directly from in vivo kinematic measurements, and that use of load rather than motion control for the sensitive DOFs may improve the accuracy of in vivo contact calculations. © 2008 Orthopaedic Research Society. Published by Wiley Periodicals, Inc. J Orthop Res 26:1173–1179, 2008

Keywords: in vivo knee kinematics; elastic contact model; computational simulation; total knee arthroplasty

Technological advances allow measurement of in vivo knee kinematics with submillimeter accuracy. Such measurements are useful for investigating how changes in knee kinematics (e.g., following anterior cruciate ligament (ACL) injury) may contribute to the development of osteoarthritis and how differences in knee replacement design may affect wear performance. Various methods exist for performing these in vivo measurements, including single-plane fluoroscopy with model-based shape matching or implanted markers,[1–7] biplane X-ray with model-based shape-matching or implanted markers,[8–11] cortical bone pins with external marker triads,[12] and point cluster methods with redundant surface markers.[13,14] For X-ray-based methods, the accuracy of the translations and rotations representing the pose of the femur with respect to the tibia are on the order of ± 0.5 mm/degreee for single-plane methods of these measurements and ± 0.1 mm/degree for biplane methods.[8–11]

The high accuracy of these measurements makes it tempting to use them directly to estimate contact forces, pressures, and areas on the medial and lateral condyles and to determine contact status (i.e., in or out of contact). Based on kinematic data alone, studies have reported loss of contact on the medial or lateral sides of many total knee replacements,[15–17] while others have suggested that lift-off is difficult to assess given the uncertainty in current kinematic measurements.[18] It would be highly advantageous if in vivo kinematic measurements could be used directly to assess medial and lateral contact forces, pressures, and areas, since these quantities may play an important role in the development of osteoarthritis in natural knees and wear in artificial knees. However, the sensitivity of contact calculations to uncertainties in the kinematic inputs has not been adequately investigated.

This study uses a validated in vivo computational simulation of gait to quantify how in vivo kinematic measurement errors affect contact calculations for a total knee replacement. The simulation utilized an elastic foundation contact model to reproduce in vivo contact force, center of pressure, and fluoroscopic motion data collected from an instrumented knee replacement. The sensitivity of the model's contact calculations to errors in relative pose was investigated using simulations where either all degrees of freedom (DOFs) were motion controlled or selected DOFs were load controlled. The results provide insight into the kinematic accuracy required to estimate in vivo contact loads and lift-off accurately and into alternate modeling methods that may improve the estimation process.

MATERIALS AND METHODS

Nominal Simulation

The starting point for the sensitivity analyses was a validated in vivo contact simulation of gait. We constructed the simulation using data collected from a patient with an instrumented knee implant.[19] In brief, one patient (male, right knee, age 80, mass 68 kg) performed treadmill gait under fluoroscopic motion analysis. Institutional review board approval and informed consent were obtained. The patient walked at a self-selected speed (1.24 ± 0.03 m/s) with his hands resting on the treadmill handlebars for balance. He had a custom tibial

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prosthesis (DePuy PFC Sigma CR PLI design possessing a posterior-lipped insert) instrumented with four uniaxial force transducers, a microtransmitter, and an antenna. The instrumented implant provided in vivo contact force and center of pressure data, while fluoroscopy provided in vivo kinematic data. After synchronizing the kinematic and kinetic data, we defined one gait cycle for subsequent analysis.

We used the one-cycle gait data to drive a dynamic contact model of the patient's implant.[19,20] The model was implemented within the Pro/MECHANICA MOTION simulation environment (PTC, Waltham, MA) (Fig. 1). A 6 DOF joint between the femoral component and tibial insert was used to measure relative (i.e., joint) kinematics for contact calculations. The experimentally measured tibial force was applied to the back side of the tibial tray at the experimentally measured center of pressure location. Accordingly, the femoral component was fixed to ground and the tibial component was allowed to move relative to it. Anterior–posterior (AP) translation, internal–external (IE) rotation, and flexion-extension were prescribed to match the fluoroscopically measured kinematics while the other three DOFs were predicted via forward dynamic simulation.

Contact was modeled using a custom elastic foundation contact model with nonlinear material properties.[21–25] The model utilized computer-aided design surface geometry provided by the manufacturer and was incorporated into the dynamic model using the Pro/MECHANICA MOTION custom load interface. To prevent excessive interpenetration, the contact model utilized springs distributed uniformly over the bearing surfaces of the tibial insert, where each spring was treated as independent from its neighbors and was associated with a single tibial surface element of known area. The contact pressure p for each element was calculated from

$$p = \frac{(1-\nu)E(p)}{(1+\nu)(1-2\nu)}\frac{d}{h} \qquad (1)$$

where $E(p)$ is the modulus of the elastic layer (a nonlinear function of p in the model), as in Eq. (1) is Poisson's ratio of the elastic layer, h is the layer thickness at the element location, and d is the element's spring deflection, defined as the interpenetration of the undeformed surfaces in the direction of the local surface normal. The distance d was computed at each time instant from the relative position and orientation of the insert with respect to the femoral component as obtained from the 6 DOF joint in the dynamic model. The 3D contact force vector acting on each side of the insert was computed by multiplying the element pressures by their respective areas and performing a vector summation over all elements. Medial and lateral forces were calculated by taking the axial components of the two contact force vectors, consistent with how contact force was measured by the instrumented implant. Medial and lateral contact pressures were calculated by finding the element on each side with the largest pressure. Finally, medial and lateral contact areas were calculated by summing areas from all elements with non-zero pressure.

The contact pressure calculations utilized a published nonlinear material model with a Poisson's ratio of 0.46.[21,22,26] Elastic modulus was set to a different value for each element depending on its current level of contact pressure. The relationship between modulus and pressure was derived from a modified nonlinear power law material model:[22,23]

$$\varepsilon = \frac{1}{2}\varepsilon_0\frac{p}{p_o} + \frac{1}{2}\varepsilon_o\left(\frac{p}{p_o}\right)^n \qquad (2)$$

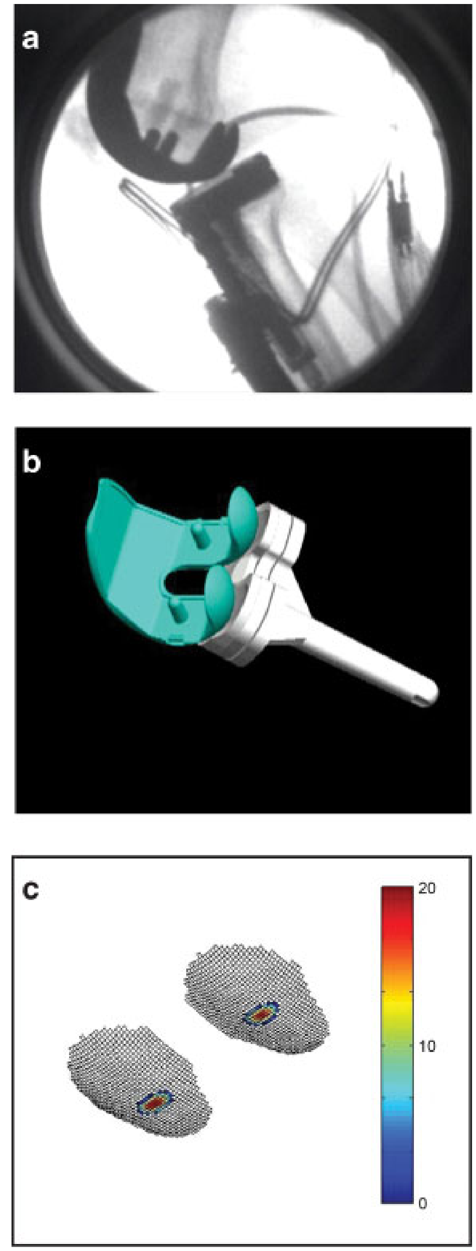

Figure 1. Steps used to create the validated nominal in vivo gait simulation from which four sensitivity analyses were derived. In vivo fluoroscopic motion data and instrumented implant load data collected simultaneously (a) were used to drive a dynamic contact model of the implant (b) that calculated medial and lateral contact forces (c) consistent with the experimental data. [Color figure can be seen in the online version of this article, available on the website, www.interscience.wiley.com.]

where ε is element strain and ε_o, p_o and n are material parameters. These parameters were set to $\varepsilon_o = 0.0257$, $p_o = 15.9$ MPa, and $n = 3$ based on experimental stress–strain data for polyethylene.[27] For any specified value of element pressure, $E = dp/d\varepsilon$ was calculated from Equation (2) as:

$$E(p) = 1/\left\{\frac{1}{2}\frac{\varepsilon_o}{p_o}\left[1 + n\left(\frac{p}{p_o}\right)^{n-1}\right]\right\} \qquad (3)$$

Equation (3) was then substituted into Equation (1) to produce a single nonlinear equation with p as the only unknown. During a simulation, this equation was solved for each element using a nonlinear root-finding algorithm. Further details can be found in Reference 21.

We validated the simulation results by comparing predicted and measured total contact force and center of pressure (CoP) over the gait cycle.[19] Root-mean-square (RMS) error in predicted total contact force was less than 1 N. During stance phase, RMS error in predicted CoP was 0.5 mm in the AP direction and 0.6 mm in the medial–lateral (ML) direction. Over the entire gait cycle, RMS errors in predicted CoP were 3.5 mm and 0.6 mm, respectively.

Sensitivity Analyses

We treated the validated nominal gait simulation as defining perfect in vivo kinematic and kinetic measurements for four subsequent sensitivity analyses. For each analysis, we performed multiple simulations that were modified versions of the nominal simulation. The first two investigated performing contact calculations directly from imperfect kinematic measurements. For these analyses, all 6 DOFs in the model were motion controlled and prescribed to follow their nominal motions but with constant offsets representing measurement errors. These offsets were applied to one motion curve at a time. The first analysis used offsets of ± 0.1 mm or degree while the second used offsets of ± 0.5 mm or degree, representing typical measurements errors reported for biplane and single-plane fluoroscopy, respectively. With six motion curves and two offsets (positive and negative) per curve, these two analyses required 12 simulations.

The second two sensitivity analyses investigated performing contact calculations from a combination of imperfect kinematic measurements and assumed loads. For these analyses, the sensitive DOFs in the model determined by the first two analyses were switched to load control, while the remaining DOFs remained motion controlled ML translation was also switched to load control since it is not measured accurately by single-plane fluoroscopy. The third analysis applied the tibial load as in the nominal simulation but added constant offsets representing measurement error to each prescribed motion. These offsets were applied to one motion curve at a time using values of ±0.5 mm or degree. With three motion curves and two offsets (positive and negative) per curve, this analysis required six simulations. The fourth analysis used the nominal prescribed motions but altered the ML location at which the tibial load was applied. Rather than using a time-varying location as in the nominal simulation, this analysis used a constant 50–50% or 70–30% ML load split, thereby requiring two simulations.

For each analysis, we calculated changes in medial and lateral contact force, pressure, and area with respect to the nominal simulation. To simplify reporting, we tabulated the maximum percent changes in the maximum values from the medial and lateral sides. Maximum values for the nominal simulation occurred near 50% of the gait cycle during late stance phase.

RESULTS

When calculated directly from imperfect kinematic measurements, contact forces, pressures, and areas were highly sensitive to small errors in some DOFs but not others (Figs. 2 and 3). Specifically, errors in superior-inferior (SI) translation and varus-valgus (VV) rotation produced much larger changes in maximum contact quantities than did errors in the remaining four DOFs. For kinematic measurement errors of ±0.1 mm or degree, errors in superior-inferior (SI) translation and varus-valgus (VV) rotation changed maximum contact quantities by as much as 204 and 77%, respectively, while errors in the remaining DOFs changed them by at most 13% (Table 1). Furthermore, SI translation and VV rotation errors caused loss of contact on the medial or lateral side, or both, during stance phase (Fig. 2, gray shaded areas touching the x-axis). For measurement errors of ±0.5 mm or deg, the same trends were observed except that changes in contact

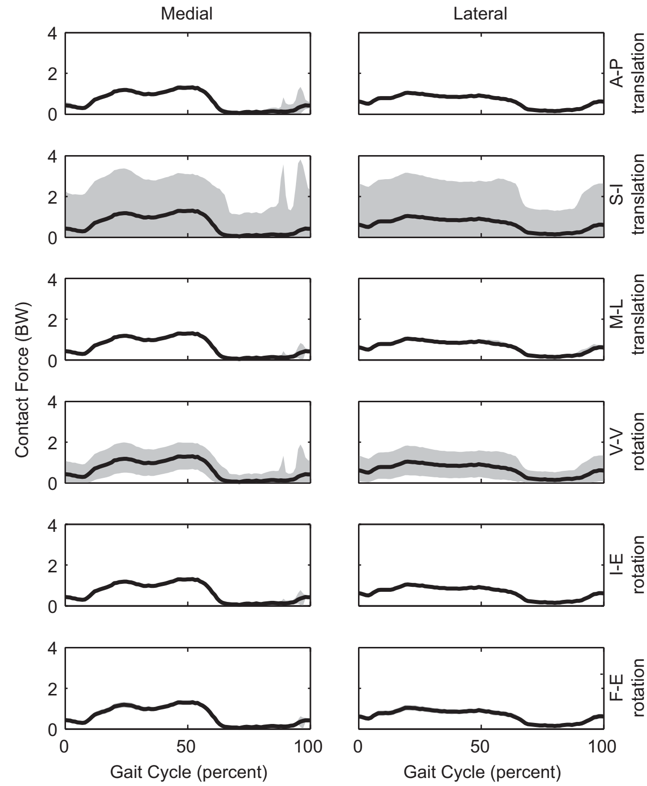

Figure 2. Sensitivity of medial (first column) and lateral (second column) contact force calculations to kinematic measurement errors of ±0.1 mm or deg. First row is sensitivity to errors in AP translation, second to errors in SI translation, third to errors in ML translation, fourth to errors in VV rotation, fifth to errors in IE rotation, and sixth to errors in FE rotation. Solid lines are calculated forces from the nominal in vivo gait simulation. Shaded regions indicate range of calculated contact forces when each DOF was motion controlled with errors of ±0.1 mm or degree. Condylar lift-off occurred, even though none occurred in vivo, wherever a shaded region touches the x-axis. Changes in calculated contact pressures and areas exhibited similar trends.

Table 1. Maximum % Change in Maximum Contact Force, Pressure, and Area Corresponding to the Results in Figure 2

Error	Force	Pressure	Area
AP Trans	8.6	2.5	6.8
SI Trans	204.2	100.0	117.4
ML Trans	7.1	3.7	6.6
VV Rot	76.7	24.7	49.5
IE Rot	3.6	0.9	3.3
FE Rot	13.3	3.7	10.9

All six DOFs were motion controlled with errors of ±1 mm or deg.

Table 2. Maximum % Change in Maximum Contact Force, Pressure, and Area Corresponding to the Results in Figure 3

Error	Force	Pressure	Area
AP Trans	64.3	11.9	111.8
SI Trans	1156.6	107.8	577.8
ML Trans	102.8	29.5	66.7
VV Rot	411.0	100.0	244.5
IE Rot	17.6	5.0	13.9
FE Rot	68.5	21.9	45.8

All six DOFs were motion controlled with errors of ±0.5 mm or deg.

quantities were larger. Errors in SI translation and VV rotation changed maximum contact quantities by as much as 1157 and 411%, respectively, while errors in the remaining DOFs changed them by at most 112% (Table 2). For these larger errors, loss of contact occurred during stance phase for all DOFs except ML translation (Fig. 3, gray shaded areas touching the x-axis).

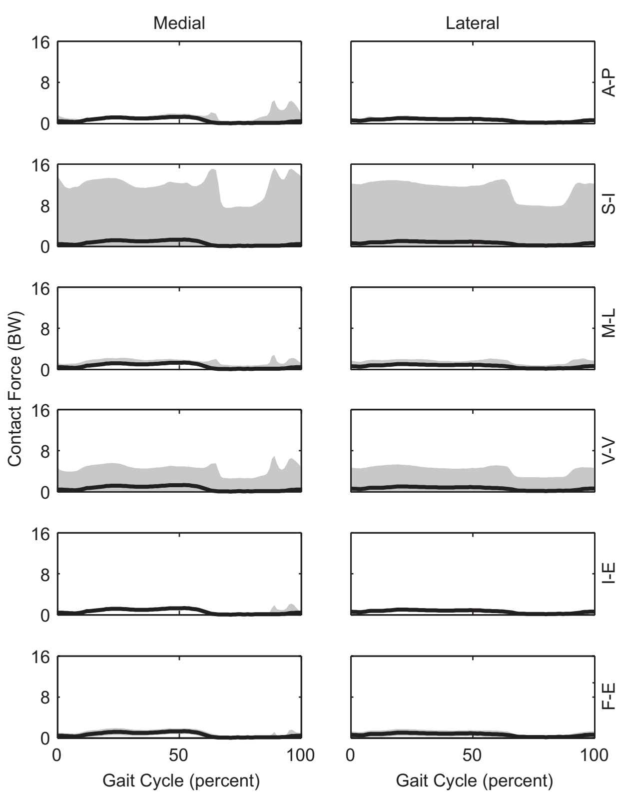

Figure 3. Sensitivity of medial (first column) and lateral (second column) contact force calculations to kinematic measurement errors of ±0.5 mm or deg. First row is sensitivity to errors in AP translation, second to errors in SI translation, third to errors in ML translation, fourth to errors in VV rotation, fifth to errors in IE rotation, and sixth to errors in FE rotation. Solid lines are calculated forces from the nominal in vivo gait simulation. Shaded regions, which require a y-axis scale four times larger than in Figure 2, indicate range of calculated contact forces when each DOF was motion controlled with errors of ±0.5 mm or deg. Condylar lift-off occurred, even though none occurred in vivo, wherever a shaded region touches the x-axis. Changes in calculated contact pressures and areas exhibited similar trends.

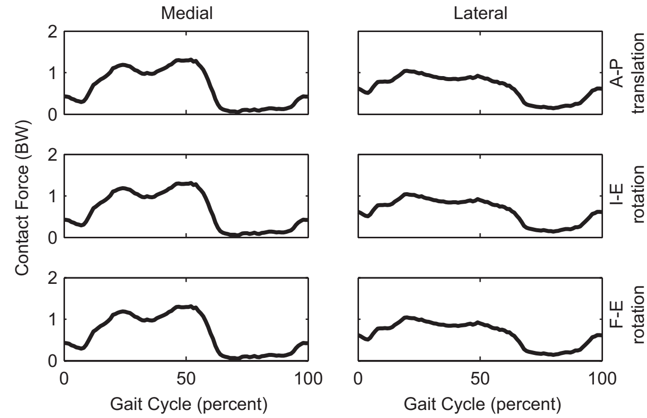

Figure 4. Sensitivity of medial (first column) and lateral (second column) contact force calculations to kinematic measurement errors of ±0.5 mm or deg when sensitive DOFs are switched to load rather than motion control. First row is sensitivity to errors AP translation, second to errors in IE rotation, and third to errors in FE rotation. Solid lines are calculated forces from the nominal in vivo gait simulation. Shaded regions, indicating range of calculated forces when AP translation, IE rotation, and FE rotation were motion controlled with errors of ±0.5 mm or deg and the remaining DOFs were load controlled with no errors, are too small to be visible. Condylar lift-off never occurred, consistent with the in vivo situation, despite the presence of kinematic measurement errors. Changes in calculated contact pressures and areas exhibited similar trends.

When calculated from a combination of imperfect kinematic measurements and assumed loads, contact quantities became much less sensitive to small errors in kinematic measurements (Figs. 4 and 5). After SI translation, VV rotation, and ML translation were switched to load rather than motion control, errors of ±0.5 mm or deg in the remaining three DOFs had almost no influence on calculated contact quantities (Fig. 4, gray shaded areas no longer visible). Changes in maximum contact quantities were at most 3% (Table 3). When the model DOFs were controlled in the same manner but with an assumed constant ML load split of 50–50 or 70–30 and with no errors present in the prescribed DOFs, changes in maximum contact quantities ranged between 5 and 36% (Fig. 5, Table 4). For both sensitivity analyses, no loss of contact occurred at any point in any of the simulations.

DISCUSSION

When imperfect kinematic measurements were applied to all 6 DOFs between the femoral component and tibial

Table 3. Maximum % Change in Maximum Contact Force, Pressure, and Area Corresponding to the Results in Figure 4

Error	Force	Pressure	Area
AP Trans	2.9	0.5	1.9
IE Rot	2.5	1.3	1.4
FE Rot	0.5	0.0	1.6

AP translation, IE rotation, and FE rotation were motion controlled with errors of ±0.5 mm or deg and the remaining DOFs were load controlled with no errors.

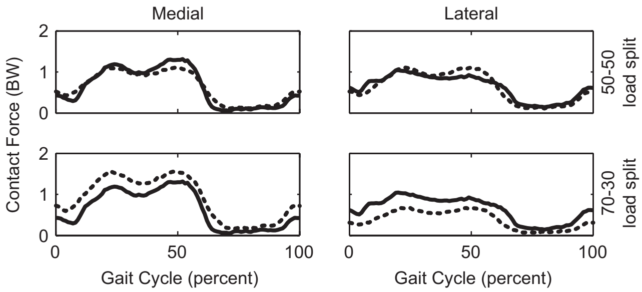

Figure 5. Sensitivity of medial (first column) and lateral (second column) contact force calculations to assumed ML load split. First row presents simulation results from a 50–50 ML load split; the second from a 70–30 load split. Solid lines are calculated forces from the nominal in vivo gait simulation. Dotted lines indicate contact forces calculated for each specified constant load split when AP translation, IE rotation, and FE rotation were motion controlled with no errors and the remaining DOFs were load controlled with the assumed constant load split. Changes in contact pressures and areas exhibited similar trends.

insert, calculated contact quantities were highly sensitive to small errors in SI translation and VV rotation. Errors as small as ± 0.1 mm or deg were enough to cause loss of contact or excessively deep penetration on the medial or lateral side, or both, suggesting that it is difficult to determine reliably whether lift-off is occurring based on kinematic measurements alone. When imperfect kinematic measurements were used for AP translation, IE rotation, and FE, along with assumed loads for the remaining DOFs, the calculated quantities were much less sensitive to kinematic measurement errors. For these simulations, lift-off never occurred, and errors were due primarily to the assumed ML load split.

Our sensitivity analyses involved several assumptions. First, we assumed that the nominal simulation represented perfect in vivo kinematic and kinetic measurements. This assumption provided a reasonable starting point for the analyses because the nominal simulation closely matched in vivo kinematic and kinetic measurements. The availability of in vivo contact force, CoP, and fluoroscopic data provided a unique opportunity to create a nominal simulation representative of in vivo conditions. Second, we assumed that an elastic foundation contact model provided a good estimation of contact forces, pressures, and areas. We previously reported that our contact model can reproduce experimentally measured contact pressures,[22] in vivo wear contours,[20] in vivo contact forces and CoPs,[19] and in vitro wear contours and volumes.[28] Thus, we believe that the sensitivity results produced by our contact model provide a close approximation of the in vivo situation. Furthermore, the magnitude of the predicted changes makes it unlikely that small errors in calculated contact quantities would alter the qualitative nature of the results. Third, we implicitly assumed that the results reported for the implant used in our study can be generalized to other designs. Our results should apply to many unconstrained posterior cruciate-retaining implants. Designs with greater AP conformity or posterior cruciate substitution may possess additional DOFs to which contact quantities are highly sensitive (i.e., IE rotation and AP translation, respectively). Finally, we added a constant offset to only one DOF at a time in our analyses. Kinematic measurement errors occur simultaneously in multiple DOFs. Consequently, true errors in predicted contact quantities will likely be larger than those presented here (e.g., due to coupled errors in SI translation and VV rotation).

Table 4. Maximum % Change in Maximum Contact Force, Pressure, and Area Corresponding to the Results in Figure 5

Split	Force	Pressure	Area
50-50	16.1	5.1	6.3
70-30	35.8	12.0	26.4

AP translation, IE rotation, and FE rotation were motion controlled with no errors and the remaining DOFs were load controlled assuming a 50–50 or 70–30 ML load split.

The presence of highly sensitive and relatively insensitive DOFs in the contact model led us to specify motion- and load-controlled DOFs in the nominal simulation. We chose to use load control for the DOFs to which contact quantities are most sensitive, namely SI translation and VV rotation. We also chose load control for ML translation because this DOF is measured the least accurately by single-plane fluoroscopy[1] and the contact model is able to produce good predictions of ML translation directly from the articular geometry. We chose motion control for the DOFs to which contact conditions are least sensitive, those being AP translation, IE rotation, and FE. All three of these DOFs can be measured accurately under in vivo conditions, to within ± 0.5 mm or deg by single-plane fluoroscopy[2–4,6] and even more accurately by biplane methods.[8–11] Our analyses indicate that these choices led to calculated contact quantities that were insensitive to kinematic measurement errors.

The sensitivity plots revealed significant changes in contact location during certain portions of the gait cycle. In Figures 2 and 3, gray shaded regions where contact force increased abruptly (e.g., toward the end of stance phase in the medial compartment) were indicative of a large change in contact location. Several studies used biplane fluoroscopic methods to estimate contact location or contact area on the medial and lateral tibial surfaces.[29–32] Although contact occurs over an area, published studies usually report contact location as a single point, calculated as either the closest point between the contacting surfaces or the centroid of a calculated contact area.[30] If the calculated area would simply grow or shrink due to errors in the estimated relative pose, then the calculated location would change little. However, in some situations, a small error in relative pose can cause the area to shift to a different location on the surface, making the calculated location inaccurate. Future studies should account for these issues when estimating in vivo contact locations or areas from fluoroscopic motion measurements.

Our sensitivity results also have important implications for identifying the occurrence of condylar lift-off in knee replacements. Previous studies reported lift-off based on calculating the minimum distance between opposing surfaces placed in fluoroscopically measured poses.[15–17] Our results indicate that when the implant surfaces are in contact, the accuracy of fluoroscopic kinematic measurements is insufficient to determine whether the surfaces are deeply penetrating or out of contact. To determine lift-off reliably, we suggest placing the components in the fluoroscopically measured relative pose and then applying offsets to SI translation and VV rotation based on their known uncertainties (e.g., 0.1 mm and 0.1 degree). If the surfaces remain out of contact for all possible poses within the envelope of uncertainty, then it is reasonable to report the occurrence of lift-off. Otherwise, no reliable conclusions can be drawn.

Although contact force, pressure, and area calculations (and hence sensitivities) are interrelated, all three were investigated here because they are important for developing computational predictions of insert wear or cartilage damage under in vivo conditions.[20] Since experimentally measured kinematics cannot be used directly to estimate contact forces, pressures, and areas, a simulation approach is needed that eliminates sensitivity to measurement errors. If SI translation, VV rotation, and ML translation are left free to equilibrate under the applied axial load, then the sensitivity of contact calculations to errors in the remaining DOFs is essentially eliminated. Thus, given fluoroscopic measurements of implant or bone motion, the problem reduces to one of estimating the in vivo axial load and ML load split for any particular subject.

Because in vivo contact force and load split data are not typically available, a future challenge is to develop ways to estimate internal loads from external measurements. Given in vivo data from additional patients with instrumented implants, regression relationships could be developed to predict total axial load and ML load split as a function of forces and moments at the knee calculated via inverse dynamics. Factors such as gait speed, stride length, and foot placement (e.g., stance width or toe out angle) will likely be important in these relationships.

In summary, we have shown that calculated in vivo contact forces, pressures, and areas in a total knee replacement are highly sensitive to errors in experimentally measured SI translation and VV rotation. Consequently, it is difficult to assess lift-off in knee replacements without access to kinematic measurements with an accuracy on the order of microns and milliradians. To estimate in vivo contact quantities, we suggest that computational simulations use estimated axial load and load split inputs in place of prescribed SI translations and VV rotations. We also suggest caution when attempting to assess condylar lift-off from kinematic measurements alone.

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