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How tibiofemoral alignment and contact locations affect predictions of medial and lateral tibiofemoral contact forces

Zachary F. Lerner[a,*], Matthew S. DeMers[b], Scott L. Delp[b], Raymond C. Browning[a,c]

[a] School of Biomedical Engineering, Colorado State University, Fort Collins, CO, USA
[b] Departments of Bioengineering and Mechanical Engineering, Stanford University, Stanford, CA, USA
[c] Department of Health and Exercise Science, Colorado State University, Fort Collins, CO, USA

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ABSTRACT

Understanding degeneration of biological and prosthetic knee joints requires knowledge of the *in-vivo* loading environment during activities of daily living. Musculoskeletal models can estimate medial/lateral tibiofemoral compartment contact forces, yet anthropometric differences between individuals make accurate predictions challenging. We developed a full-body OpenSim musculoskeletal model with a knee joint that incorporates subject-specific tibiofemoral alignment (*i.e.* knee varus-valgus) and geometry (*i.e.* contact locations). We tested the accuracy of our model and determined the importance of these subject-specific parameters by comparing estimated to measured medial and lateral contact forces during walking in an individual with an instrumented knee replacement and post-operative genu valgum (6°). The errors in the predictions of the first peak medial and lateral contact force were 12.4% and 11.9%, respectively, for a model with subject-specific tibiofemoral alignment and contact locations determined through radiographic analysis, *vs.* 63.1% and 42.0%, respectively, for a model with generic parameters. We found that each degree of tibiofemoral alignment deviation altered the first peak medial compartment contact force by 51N (r^2=0.99), while each millimeter of medial-lateral translation of the compartment contact point locations altered the first peak medial compartment contact force by 41N (r^2=0.99). The model, available at www.simtk.org/home/med-lat-knee/, enables the specification of subject-specific joint alignment and compartment contact locations to more accurately estimate medial and lateral tibiofemoral contact forces in individuals with non-neutral alignment.

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1. Introduction

Abnormal knee loads are implicated in tibiofemoral osteoarthritis (Sharma et al., 1998), which affects more than 12% of US adults (Dillon et al., 2006). The distribution of tibiofemoral contact forces between the medial and lateral compartments can be influenced by frontal-plane tibiofemoral alignment and affect degeneration of biological (Sharma et al., 2001) and prosthetic (Ritter et al., 1994) knees. The treatment of orthopedic disorders of the knee is likely to benefit from an improved understanding of the *in-vivo* knee loading environment during activities of daily living.

Musculoskeletal models allow researchers to investigate medial/lateral tibiofemoral contact forces during activities such as walking (Fregly et al., 2012; Morrison, 1970). Some modeling approaches require complex, multi-step analyses, or the use of both full-body gait models and finite element or contact models (Bei and Fregly, 2004; Hast and Piazza, 2013; Lin et al., 2010; Thelen et al., 2014; Yang et al., 2010). Finite element and contact models rely on an accurate representation of the articulating joint surfaces and require imaging techniques that may be unavailable or prohibitively expensive. Resolving the magnitudes of medial/lateral forces by approximating medial/lateral compartment points of contact is a promising approach for estimating contact forces (Gerus et al., 2013; Kumar et al., 2012; Winby et al., 2009); however, no open-source, full-body gait model contains knee joint definitions that allow direct computation of medial/lateral contact forces.

Predictions of medial/lateral tibiofemoral contact forces in an individual using a musculoskeletal model with generic geometry may be inaccurate when the model does not accurately represent the individual. The specification of certain subject-specific model parameters may improve accuracy (Gerus et al., 2013). Two parameters, frontal-plane tibiofemoral alignment and medial/lateral compartment contact locations, are likely to influence model-predicted medial/lateral compartment contact forces by altering how muscle forces and external loads pass relative to each compartment. Frontal-plane tibiofemoral alignment affects loading of the knee (Halder et al., 2012; Hsu et al., 1990; Hurwitz et al., 2002; Yang et al., 2010), and can vary

* Correspondence to: 220 Moby B Complex Colorado State University Fort Collins, CO 80523-1582. Tel.: +814 571 4616; fax: +970 491 0445.
E-mail address: Zach.Lerner@ColoState.edu (Z.F. Lerner).

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up to $\pm 3.75°$ in individuals without obvious genu valgum-varum (Moreland et al., 1987). Existing modeling approaches have limitations that hinder the accurate representation of a subject's frontal-plane alignment; for example, generic models typically lack or constrain the frontal-plane motion of the knee (Gerus et al., 2013; Hast and Piazza, 2013; Kumar et al., 2012; Winby et al., 2009) and subject-specific models based on geometry determined from MRI or CT images are of non-weight-bearing limbs (Bei and Fregly, 2004; Gerus et al., 2013). In addition, when medial/lateral compartment contact is approximated through single points, the locations of these points influence how the tibiofemoral loads are distributed. It has been assumed that the medial/lateral compartment contact locations are centered at the midline of the femoral condyles (Winby et al., 2009) in biological knees or located at set distances from the joint center in prosthetic knees (Gerus et al., 2013), but variability in alignment and joint degeneration may alter these locations.

To address the need to calculate tibiofemoral loads accurately this study had three goals. The first was to develop a musculoskeletal model that accounts for differences in tibiofemoral alignment and contact locations and computes medial/lateral contact forces during walking. The second goal was to quantify the accuracy of knee contact force estimates made using generic geometry and subject-specific geometry by comparing these estimates to *in-vivo* measurements from an individual with an instrumented knee replacement and genu valgum. The third goal was to evaluate the effects of model-specified frontal-plane knee alignment and contact point locations on medial/lateral contact force predictions. The model, experimental data, and contact force predictions are freely available at www.simtk.org.

2. Methods

2.1. Model development

To compute medial and lateral tibiofemoral contact forces during walking we developed a model of the tibiofemoral joint in OpenSim (Delp et al., 2007) and incorporated it within a published full body musculoskeletal model (DeMers et al., 2014). The published model, designed for studying gait, was comprised of 18 body segments and 92 muscle-tendon actuators. Model degrees of freedom (DOF) included a ball-and-socket joint between the third and fourth lumbar vertebra, three translations and three rotations of the pelvis, a ball-and-socket joint at each hip, and revolute ankle and subtalar joints. In our model, the sagittal plane rotation and translations of the tibia and patella relative to the femur were identical to those specified by (Delp et al., 1990); however, we augmented the mechanism defining the tibiofemoral kinematics.

The tibiofemoral model introduced components for configuring frontal-plane alignment of the knee and for resolving distinct medial and lateral tibiofemoral forces. We introduced a distal femoral component body and a tibial plateau body (represented by CAD geometry of the instrumented implant, Fig. 1, pink) with orientation parameters for configuring frontal-plane alignment in the femur (θ_1) and tibia (θ_2). Between the femoral component and the tibial plateau, we defined a series of joints to characterize the tibiofemoral kinematics and medial/lateral load distribution. Firstly, the knee joint from Delp et al. (1990) defined the sagittal-plane rotations and translations of the knee between the femoral component and the sagittal articulation frame of reference (Fig. 1A, hidden, Fig. 1B, translucent). Secondly, two revolute joints connected the sagittal articulation frame to medial and lateral tibiofemoral compartments (Fig. 1, purple). The axes for these two revolute joints were perpendicular to the frontal-plane. Lastly, the medial and lateral compartments were welded at the anterioposterior mid-point of the tibial plateaus such that they remained fixed to the tibia while articulating with the surface of the femoral component during flexion-extension. The patella segment articulated with the femoral-condyle segment according to (DeMers et al., 2014). The quadriceps muscles wrapped around the patella before attaching to the tibial tuberosity to redirect the quadriceps forces along the line of action of the patellar ligament and allow the resultant tibiofemoral contact forces to be computed (DeMers et al., 2014).

In this knee mechanism, the medial and lateral revolute joints cannot resist frontal-plane moments individually. However, by acting in parallel, the two joints share all loads transmitted between the femur and tibia and resolve them as the medial and lateral contact forces required to balance the net reaction forces and frontal-plane moments across the tibiofemoral joint. Correspondingly, the knee remained a single DOF joint with motion only in the sagittal plane. The medial and lateral contact forces were computed and reported using the Joint Reaction Analysis in OpenSim (Steele et al., 2012).

2.2. Experimental data

We used experimental data from a subject with an instrumented knee replacement (right knee, male, age 83, mass 67 kg, height 1.72 m) to generate dynamic simulations of walking. These data have been made available by the Knee Load Grand Challenge (Fregly et al., 2012). Researchers collected kinematic, kinetic, and instrumented implant data simultaneously during over-ground walking. Validated regression equations were used to calculate separate medial and lateral tibiofemoral compartment contact forces from the instrumented knee joint (Meyer et al., 2001).

Established methods (Moreland et al., 1987) were used to quantify the frontal-plane alignment of the subject's right lower-extremity from a standing anterioposterior radiograph (Fig. 2). The angle formed between the intersection of the mechanical axes of the femur and tibia was used to specify subject-specific model alignment. To model lower-extremity alignment, θ_1 and θ_2 from Fig. 1 are each specified as one half of the varus-valgus alignment angle ($180°–\theta$ from Fig. 2). To estimate subject-specific medial/lateral compartment contact locations, we measured the distance between the centerline of the femoral implant component and the centerline of the tibial implant component using a higher resolution anterioposterior radiograph of the knee (Fig. 3). A measurement scale was established from the known width of the implant. Contact model predictions using *in-vivo* measurements of a similar implant have indicated an intercondylar distance of 40 mm (Zhao et al., 2007), and this distance has been used previously to inform model contact points (Gerus et al., 2013). Therefore, we maintained this intercondylar distance while shifting the medial/lateral contact locations medially by the distance (d) measured from the radiograph.

2.3. Varying tibiofemoral specificity in the musculoskeletal model

To isolate the effects of specifying each subject-specific parameter we conducted simulations with the following four conditions of our musculoskeletal model.

2.3.1. Fully-informed model

This model had subject-specific tibiofemoral alignment ($\theta = 174°$) and contact locations informed through radiographic analysis. Medial compartment contact was located 23 mm medial of the knee joint center and lateral compartment contact was located 17 mm lateral of the knee joint center.

2.3.2. Uninformed model

Based on data from an instrumented implant contact model for a neutrally aligned lower-extremity (Zhao et al., 2007), and matching assumptions for an artificial knee implant made previously (Gerus et al., 2013), we specified the generic frontal-plane locations of the medial/lateral compartment structures 20 mm medial and lateral of the knee joint center. The tibiofemoral alignment for this model ($\theta = 180°$) was maintained from skeletal geometry originally defined by (Delp et al., 1990).

2.3.3. Alignment-informed model

This model had subject-specific alignment ($\theta = 174°$) but uninformed contact locations (20 mm medial and lateral of the joint center).

2.3.4. Contact-point-informed model

This model had subject-specific contact locations (medial compartment: 23 mm medial of the joint center, lateral compartment: 17 mm lateral of the joint center) but uninformed alignment ($\theta = 180°$).

To investigate the effects of model-specified tibiofemoral alignment on model-predictions, we created contact-point-informed models with variable tibiofemoral alignment ranging from 0°–8° valgus, at 2° increments. To investigate the effects of model-specified medial/lateral compartment contact locations on model-predictions, we created alignment-informed models with variable medial/lateral contact point locations spanning reported translations (± 4 mm) at 2 mm increments with 40 mm inter-condylar distances.

2.4. Musculoskeletal simulation of walking

We used marker location data from anatomical landmarks collected during a standing calibration trial to scale our models in OpenSim. For each scaled model, we used OpenSim's inverse kinematics analysis, which minimized the errors between markers fixed to the model and experimentally measured marker trajectories (Delp et al., 2007), to determine the joint angles during four over-ground walking trials. Model kinematics were recalculated for every model condition while the ground reaction forces remained the same. Because muscle forces are the main determinant of compressive tibiofemoral contact forces (Herzog et al., 2003), variations in muscle activity greatly influence the magnitude and accuracy of knee joint contact force predictions (DeMers et al., 2014). We resolved individual muscle forces using a weighted static optimization approach that was calibrated to the

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