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Subject-specific knee joint geometry improves predictions of medial tibiofemoral contact forces

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ABSTRACT

Estimating tibiofemoral joint contact forces is important for understanding the initiation and progression of knee osteoarthritis. However, tibiofemoral contact force predictions are influenced by many factors including muscle forces and anatomical representations of the knee joint. This study aimed to investigate the influence of subject-specific geometry and knee joint kinematics on the prediction of tibiofemoral contact forces using a calibrated EMG-driven neuromusculoskeletal model of the knee. One participant fitted with an instrumented total knee replacement walked at a self-selected speed while medial and lateral tibiofemoral contact forces, ground reaction forces, whole-body kinematics, and lower-limb muscle activity were simultaneously measured. The combination of *generic* and *subject-specific* knee joint geometry and kinematics resulted in four different OpenSim models used to estimate muscle–tendon lengths and moment arms. The *subject-specific* geometric model was created from CT scans and the *subject-specific* knee joint kinematics representing the translation of the tibia relative to the femur was obtained from fluoroscopy. The EMG-driven model was calibrated using one walking trial, but with three different cost functions that tracked the knee flexion/extension moments with and without constraint over the estimated joint contact forces. The calibrated models then predicted the medial and lateral tibiofemoral contact forces for five other different walking trials. The use of subject-specific models with minimization of the peak tibiofemoral contact forces improved the accuracy of medial contact forces by 47% and lateral contact forces by 7%, respectively compared with the use of generic musculoskeletal model.

1. Introduction

Large joint contact forces are thought to be an important factor in the development and progression of osteoarthritis (Guilak, 2011; Hurwitz et al., 2001; Roemhildt et al., 2012; Solomon, 1976). The external knee adduction moment (KAM) has been used as a convenient surrogate for the medial–lateral load distribution at the knee and has been linked to the onset, progression, and severity of medial tibiofemoral osteoarthritis (Foroughi et al., 2009; Schipplein and Andriacchi, 1991). The KAM, estimated by inverse dynamics, does not account for the knee's other degrees of freedoms and the muscles' direct contribution to the knee contact forces, and does not always correlate strongly with medial contact force at the knee (Meyer et al., 2012). In this study, we hypothesized that computational neuromusculoskeletal models that include knee loading about multiple degrees of freedom and muscle forces may provide more accurate estimates of knee contact loads.

However, developing and validating these models is challenging because of the neuromusculoskeletal system complexity and inter-subject variability (Delp et al., 2007). The accuracy of computational models to predict tibiofemoral joint contact forces can be assessed using direct measures from instrumented total knee replacements (Fregly et al., 2012). Computational models that use generic anatomy tend to overestimate medial knee contact

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forces when compared to *in vivo* measurements (Fregly et al., 2012). Altered estimates of the muscle–tendon moment arms and muscle–tendon lengths from variations of musculoskeletal geometries have been reported for the knee (Ackland et al. 2012; Pal et al., 2007) and hip joints (Duda et al., 1996; Scheys et al., 2011). Tsai et al. (2012) found that the use of moment arms estimated from magnetic resonance imaging provides a more accurate prediction of the net joint moment compared to the measured net joint moment. In this context, it is possible that the aforementioned contact force overestimations are due to an underestimation of muscle moment arms, resulting in higher muscle forces to generate the same net joint moment. In addition, joint kinematics estimation errors may affect load computations.

Computational models to estimate muscle forces can be broadly classified as; (i) *optimization method,* which estimate a set of muscle activations based on an objective function (e.g. minimize muscle stress) (Crowninshield and Brand, 1981), or (ii) *electromyography* (EMG) *EMG-driven approach,* which determines muscle activations based on recorded EMG signals (Lloyd and Besier, 2003; Buchanan et al., 2004). In the case of musculoskeletal disorders, such as osteoarthritis, muscle activation strategies are highly variable and significantly different from normal healthy people (Zeni et al., 2010; Heiden et al., 2009). In this case, an EMG-driven approach appears warranted to account for an individual's unique muscle activation pattern (Kumar et al., 2012). The mapping from EMG to muscle force is not trivial and current EMG-driven methods use a calibration process to adjust EMG-to-activation and muscle–tendon parameters (Lloyd and Besier, 2003). Parameter calibration attempts to match experimental joint moments of the ankle, knee and/or hip measured from inverse dynamics. However, this calibration is a limitation of EMG-driven modeling because the solution space is large and the matching of the knee flexion/extension joint moment does not necessarily ensure accurate joint contact force estimations. Indeed, even though EMG-driven approaches were found to predict joint moments very well, they nevertheless overestimated the medial tibiofemoral knee joint contact forces (Fregly et al., 2012). The influence of adding further constraints beyond the magnitude of the contact forces during the calibration process has not been investigated.

The aim of this study was to investigate the influence of knee joint geometry, knee joint kinematics and calibration cost functions on the estimation of tibiofemoral contact forces using an EMG-driven neuromusculoskeletal approach. It was hypothesized that, subject-specific knee joint geometry and/or knee joint kinematics would improve the accuracy of medial and lateral contact force predictions, compared to a generic model. We also hypothesized that a calibration cost function including a minimization of the peaks of medial and lateral contact forces would improve joint contact forces predictions.

2. Method

2.1. Gait experiments

This study used data previously collected from an adult male fitted with an instrumented total knee replacement (right knee, age 83, mass 68 kg, height 1.7 m) (Fregly et al., 2012). Institutional review board approval and the participant's informed, written consent were obtained prior to data collection.

We used data recorded from two gait tasks. The first was walking on an instrumented treadmill (Bertec, Columbus, USA) where a C-arm fluoroscope (GE Medical Systems, Salt Lake City, USA) was used to record rotations and translations of the tibia relative to the femur. The second task involved walking overground at a naturally selected walking speed (n=six trials). The whole body segmental motion was recorded at 120 Hz using a VICON motion analysis system (Vicon, Oxford, UK). Ground reaction forces (GRF) were recorded at 1200 Hz from three force plates (Bertec, Columbus, USA), and surface EMG recorded at 1200 Hz using a 16-channel Bagnoli system (Delsys, Boston, USA) with custom double differential preamplified electrode leads. The motion capture markers were attached according to a full-body marker set reported by Besier et al. (2003) and EMG activity on the involved side was recorded from 8 muscles: *biceps femoris long-head* (BicFemlh), *gastrocnemius lateralis* (GasLat), *gastrocnemius medialis* (GasMed), *rectus femoris* (RectFem), *semi-membranous* (SemiMem), *tensor fascia lata* (TFL), *vastus lateralis* (VastLat), and *vastus medialis* (VastMed). Medial and lateral tibiofemoral contact forces were recorded at 120 Hz, synchronously with motion capture, GRFs, and EMG.

2.2. Description of the OpenSim models

Various models were created with generic and subject-specific elements.

(1) The *generic geometry anatomical model* was based on a full-body OpenSim model, which consisted of 14 rigid-linked skeletal segments with 37 degrees of freedom (DOF) (Hamner et al., 2010; Donnelly et al., 2012). This model was scaled in three-dimensions to match each subject's anthropometry based on marker trajectories measured from motion capture and calculated hip, knee and ankle joint centers. The positions of the lower limb joint centers and axes were estimated from functional tasks (Besier et al., 2003; Donnelly et al., 2012). Of importance, the lower limbs had a 3 DOF ball joint for the hip and 1 DOF hinge joint for the ankle (Hamner et al., 2010; Donnelly et al., 2012). The knee joint kinematics are described in more detail below (see Section 2.2 iii and iv).

(2) The *subject-specific geometry anatomical model* was an adaptation of the generic full-body model. The upper body was the generic scaled model. However, the lower limb model was created using a subject-specific knee from the implant's geometry, bone geometry from CT scans (i.e., femur, tibia, fibula, and patella), and generic bone geometry for the other bodies. The position of the knee joint center was located at the midpoint of the femoral condyles when the knee was in the fully extended posture, and the hip-joint center was located in the center of the femoral head (Arnold et al., 2010). The ankle-joint center was calculated as for the scaled generic anatomical model. The vertical length of the femur, tibia, and fibula were adjusted to match the position of the calculated hip, knee, and ankle joint centers. The scale factor was 1.05 and 1.03 for the femur and tibia-fibula, respectively. Each muscle–tendon path was adjusted manually to fit with the new bone geometry using the bony landmarks from the generic model as a reference. The moving path definition of some muscles was adjusted to avoid penetration into bone. The translations of the patella as a function of knee flexion were redefined to fit the shape of the implant. The moving path of the quadriceps muscle group was modified to follow the new motion of the patella and to avoid penetration into the femur.

(3) The *generic knee joint kinematic model* had 3 rotational and 2 translational DOFs (Donnelly et al., 2012). The knee comprised of a sagittal planar joint with a flexion/extension axis going through the knee joint center and perpendicular to the plane. A spline defined the anterior–posterior and superior–inferior translations of the tibia in this plane as a function of knee flexion angle (Fig. 1), which was the translation of the knee joint center relative to the origin of the femur (femoral head) (Delp et al., 1990). The knees also had an internal/external rotation hinge joint with its axis going through the ankle joint and knee joint centers, and two hinge joints for adduction/abduction, the axes perpendicular to the tibial frontal plane with one going through the medial condyle contact point and the other through the lateral condyle contact point. The position of the medial and lateral condyle contact points were the same as used for the subject-specific knee (see below).

(4) For the *subject-specific knee kinematic model* the generic spline functions were adjusted to represent the experimental translations recorded using fluoroscopy without penetration between the femur and tibia (Fig. 1). The position of the knee joint center was not modified. Additionally, as in the generic knee, the medial and lateral condyle contact points were based on the inter-condyle distance and contact positions relative to the knee joint extracted from instrumented knee data and were 40 mm and 20 mm respectively (Zhao et al., 2007).

The combination of generic and subject-specific knee joint geometry and kinematics resulted in four different OpenSim models:

1. *Generic* geometry and *generic* kinematics (G-Geom & G-Kin).
2. *Generic* geometry and *subject-specific* kinematics (G-Geom & SS-Kin).
3. *Subject-specific* geometry and *generic* kinematics (SS-Geom & G-Kin).
4. *Subject-specific* geometry and *subject-specific* kinematics (SS-Geom & SS-Kin).

2.3. Estimation of joint angles and joint moments in gait

The OpenSim inverse kinematics and inverse dynamics analysis tools were used to estimate the joint kinematics and moments from the gait data (Delp et al., 2007). In the inverse kinematics solution all DOFs were free to move except at the knee where internal/external rotation and adduction/abduction were fixed and