Proceedings of the ASME 2013 Summer Bioengineering Conference
SBC2013
June 26-29, 2013, Sunriver, Oregon, USA

SBC2013-14718

SIMULTANEOUS PREDICTION OF MUSCLE, LIGAMENT AND TIBIOFEMORAL CONTACT FORCES VIA OPTIMALLY TRACKING SUBJECT-SPECIFIC GAIT DYNAMICS

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Introduction: Computational models are needed to estimate soft tissue loads during movement. It would be ideal to perform such estimates on a subject-specific basis, where the information could be used clinically for assessing injury risk, planning treatment and monitoring rehabilitation outcomes. Musculoskeletal simulation software has evolved to the point that it is now relatively straight forward to estimate muscle forces needed to emulate subject-specific joint kinematics and kinetics [1]. These muscle forces can subsequently be used as boundary conditions in a knee mechanics model to estimate the associated ligament and cartilage loads [2]. However, this serial simulation approach may ignore inherent interactions between musculoskeletal dynamics and internal joint mechanics. That is, cartilage contact forces and ligament tension can potentially contribute to joint moment equilibrium [3]. Further, ligament stretch may allow joint kinematics to vary in a way that affects muscle moment arms and lines of action about the joint. Thus, it would be preferable if muscle, ligament and cartilage contact loads were estimated simultaneously so that these interactions are accounted for. The objective of this study was to incorporate a six degree of freedom tibiofemoral model into an existing subject-specific gait simulation framework [4]. In this study, we introduce the computational model, and then use it to track measured gait dynamics of a subject with an instrumented knee joint replacement. A comparison of the model-predicted and measured tibiofemoral contact forces provides a basis for assessing the validity of this novel co-simulation framework.

Methods: We constructed a knee model that included a six degree of freedom (dof) tibiofemoral joint and a one dof patellofemoral joint. The patellofemoral joint allowed for the patella to translate within a constrained path relative to the femur, subject to quadriceps and patellar tendon loading. Seventeen knee ligament bundles were represented as nonlinear springs with origins, insertions, reference strains and stiffness values adapted from the literature [2, 5]. The knee modal was incorporated into an existing lower limb musculoskeletal model that included 44 musculotendons acting about the hip, knee and ankle joints [6]. We first generated gait simulations using an intact knee model, in which the femoral geometry was segmented from MRI images of an average size adult male. The medial and lateral tibia plateaus were modeled as planes with posterior slopes of 4 and 6 deg, respectively [7]. These geometries were later exchanged for the subject's (JW) joint replacement geometry provided through the Grand Challenge website [8]. For each case, tibiofemoral contact forces were computed via an elastic foundation model [9].

Upright, overground and treadmill walking motion analysis data were downloaded from the Grand Challenge website (simtk.org/home/kneeloads). The lower extremity model was incorporated into a 32 degree of freedom whole body model that included lower back, shoulder, and elbow joints. The whole body model was scaled to anatomical marker positions recorded with the subject standing in an upright trial. We then used global optimization inverse kinematics to compute the pelvis position, pelvis orientation and joint angles that produced the best agreement between measured and model marker positions at each time frame of the treadmill and overground gait cycles. We then created dynamic simulations by using a computed muscle control algorithm to determine muscle excitations that drove the right lower extremity to track the measured knee flexion trajectory, with the other five tibiofemoral dof and patella translation evolving from the mechanics. Pelvis, hip, ankle and upper extremity joint kinematics were prescribed to follow measured values, such that whole body dynamic effects were accounted for. Measured ground reaction forces were directly applied to the feet. Note that a whole body gait simulation was not feasible using the treadmill data set, since the subject was allowed to hold onto an un-instrumented handrail. Muscle redundancy was resolved by minimizing the sum of muscle volume-weighted squared activations. Gait simulations were first performed using an intact knee model for five overground walking gait cycles for which published lateral and medial tibiofemoral contact force information was available [8]. We then performed simulations of two speed-varying treadmill gait cycles, one of which was when the subject was accelerating from 0.8 to 1.4 m/s. The second gait cycle was when the subject began decelerating after reaching a peak speed of 1.4 m/s.

Results: The computed muscle control algorithm was able to closely track the measured knee flexion trajectory. During overground walking, the simulated tibiofemoral contact forces with the intact knee model were of comparable magnitude and patterns to those measured directly (Fig. 1). The estimated first peak medial contact force was 1.99 (±0.34) * body weight (BW), which was slightly greater than the average measured value of 1.84 BW for this subject [8]. However, the second peak medial contact force was substantially over-predicted, 2.46 (±0.09) vs. 1.45 BW. Estimated contact forces in the lateral compartment were substantially lower than on the medial side, which is similar to that seen empirically. During swing phase, the estimated contact loads were greatly reduced, but slightly higher than measured values in the lateral compartment.

During the treadmill gait trial, the estimated first medial peak contact loads were 1.65 and 1.99 BW during the acceleration and deceleration phases, respectively (Fig. 2). The second medial peak contact loads were higher, with values of 2.35 and 2.75 BW. Corresponding peak contact loads on the lateral side were lower than on the medial side, ranging from 0.4 to slightly >1.0 BW. The net peak contact loads were 3.4 and 3.7 BW during the deceleration and acceleration phases, respectively.

Discussion: This study demonstrates that it is feasible to simultaneously predict muscle, ligament and cartilage contact loads within the context of a dynamic gait simulation. This represents the first time that the computed muscle control algorithm has been used for this purpose, as prior applications of CMC required that simple kinematic joint constraints be used in the whole body gait model [4]. By incorporating a knee with joint laxity, one is able to obtain predictions of joint contact forces and secondary kinematics (e.g. anterior tibia translation and internal tibia rotation). Hence, this co-simulation framework provides an ideal framework to look at how soft tissue properties, e.g. ligament tensioning in a joint replacement surgery, could alter the secondary knee motion and loading at the knee. Such information is relevant for assessing joint wear, stability and function. We are now in the process of replacing the intact knee model with the artificial joint geometry of the subject for which the contact load was measured. A challenge in computing contact forces within the context of a whole body gait simulation is the time involved in detecting and computing contact between complex surfaces. Our approach for this challenge is to use the parallel computing speed of the GPU to detect contact and compute forces. In pilot implementations, we have achieved approximately a 10 fold speed-up in implementing the contact algorithm on the GPU compared to the CPU. This approach makes it feasible to use the co-simulation framework to simulate joint contact loads on complex joint surfaces within the context of whole body movement.

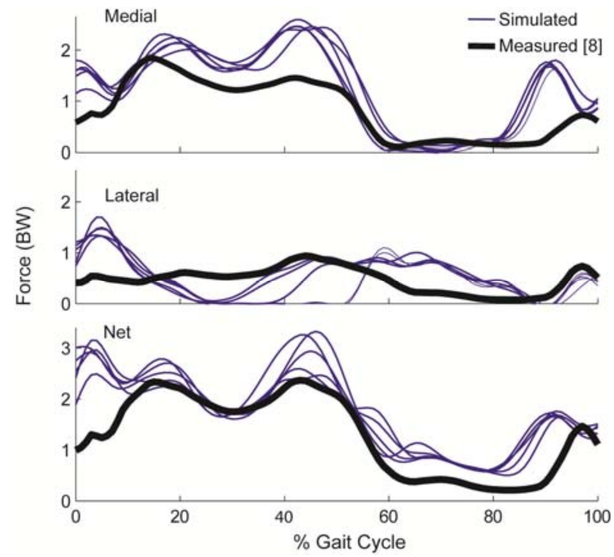

Fig. 1. Simulated (thin lines) and measured (thick lines, [8]) tibia contact forces during overground gait.

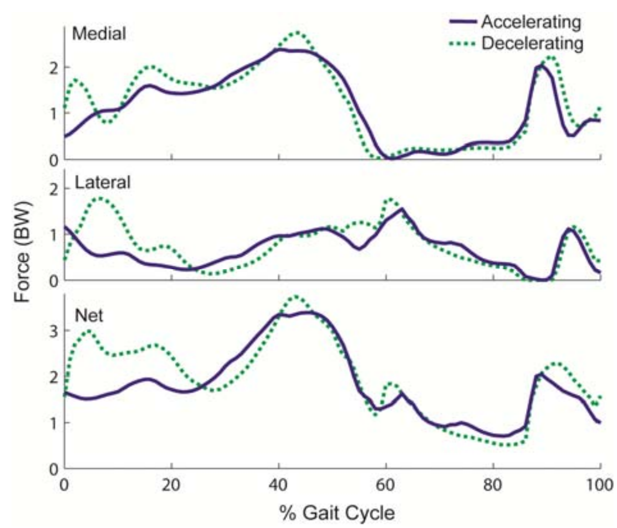

Fig. 2. Simulated tibia contact forces during accelerating (solid line) and decelerating (dashed line) phased of treadmill gait.

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Acknowledgements: NSF 0966535