THE IMPORTANCE OF ACCOUNTING FOR KNEE LAXITY WHEN SIMULATING GAIT

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INTRODUCTION

Musculoskeletal simulations of gait are commonly used to estimate muscle and joint forces. Such models typically represent the knee as a kinematic constraint that is independent of load [1]. However, the tibiofemoral joint bas been shown to exhibit load-dependent behavior during functional tasks [2,3]. For example, at the same extended knee posture, internal tibia rotation varies markedly between early stance, late stance, and terminal swing [2]. These variations in kinematics may be important to consider in gait simulations, since they alter muscle moment arms and the instantaneous joint axes [4]. Therefore, the goal of this study was to determine if a musculoskeletal simulation model that included a knee with laxity could predict secondary knee kinematics seen in walking.

METHODS

We started with a lower limb musculoskeletal model that included 44 musculotendons acting about the hip, knee and ankle joints [1]. The one degree of freedom (dof) knee in the model was replaced by a six dof tibiofemoral joint [5,6] and a one dof patellofemoral joint. Nineteen ligament bundles were represented including the MCL (5 bundles), LCL, popliteofibular ligament, ACL (2 bundles), PCL (2 bundles), posterior capsule (4 bundles), iliotibial band (ITB), and patellar ligament (3 bundles). Each ligament was represented as a nonlinear spring with origins and insertions based on [5] and wrapping about the femoral condyles accounted for. The geometry of the distal femur and cartilage was segmented from high resolution MRI images of a young male knee with average femoral geometry. The medial and lateral tibia plateaus were modeled as planes with posterior slopes of 2 and 7 deg, respectively [5]. Tibiofemoral contact forces were computed via an elastic foundation model [5]. The one dof patellofemoral joint allowed for the patella to translate within a constrained path relative to the femur, subject to quadriceps and patellar ligament forces acting on either end. The reference strains and stiffness values of the ligaments were adapted from the literature [5,6], with a minimal amount of tuning to ensure the model replicated literature measures of passive motion, anterior-posterior stiffness, and axial rotational tibiofemoral stiffness.

The model was used to simulate knee motion and loading during gait. To do this, computed muscle control (CMC) [7] was used to determine muscle excitations that drive the model to track normal knee flexion throughout a gait cycle. Muscle redundancy was resolved by minimizing the sum of muscle volume-weighted squared activations. During the simulation, measured ground reaction forces were directly applied to the foot and 3D pelvis, hip, and ankle motion were prescribed to track measured trajectories. Note that CMC was only used to track knee flexion, such that the other five dof at the tibiofemoral joint and the patella translation were predicted (Fig. 1). These predictions were compared to a gait simulation conducted with a one dof kinematic knee [1], in which all secondary tibiofemoral kinematics and patellofemoral motion were constrained functions of knee flexion. The kinematic constraint functions were obtained by passively flexing the knee model.

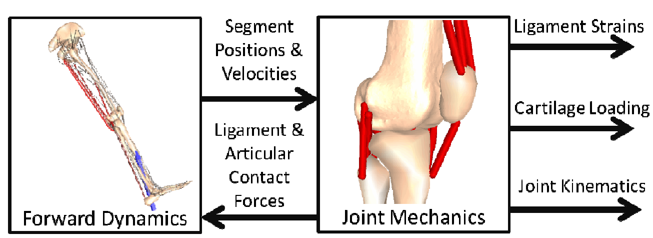

Figure 1. Forward dynamics and joint mechanics models were integrated simultaneously when simulating gait, providing predictions of muscle forces, ligament loads, cartilage contact, and secondary knee kinematics.

RESULTS

Predicted tibiofemoral kinematics differed from that assumed in a kinematic knee model but were generally consistent with direct bone pin measures during gait [2] (Fig. 2). Notably, internal tibia rotation occurs throughout much of stance and peaks near toe-off. Anterior tibia translation was greater during stance than assumed in the kinematic model, with the difference attributable to quadriceps activity during the load acceptance phase of gait. The patella also translated more superiorly in the co-simulation model due to compliance of the patellar ligament. A major advantage of the co-simulation model formulation is that it provides direct estimates of muscle, ligament, and joint contact forces. The shape of the simulated knee contact loading was consistent with measures obtained via instrumented implants [8]. ACL tension peaked during the load acceptance phase of gait, as suggested in [5].

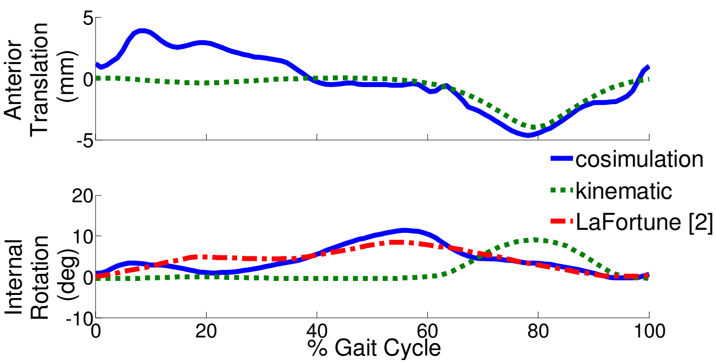

Figure 2: Predictions of secondary tibiofemoral kinematics during gait using cosimulation. Internal tibia rotation differs markedly from a kinematic model but is consistent with [2].

DISCUSSION

This study represents the first use of CMC to simulate movement using a joint that exhibits laxity. Previous formulations of CMC were only applicable for kinematics joints in which the multi-joint acceleration capacity of a muscle can be directly computed [7]. With a deformable joint, numerical integration is required to determine how individual muscles will induce motion as ligament and contact forces change over time. To overcome this challenge, we used short forward integrations (30ms) within CMC to assess muscle acceleration capacities, which were in turn used to determine excitations needed to track desired trajectories. We note that the full complement of musculoskeletal dynamics and joint mechanic equations are integrated together during the actual simulation.

It is noteworthy that the goal of tracking normal knee flexion directly predicts the secondary knee kinematics seen experimentally. Hence, secondary motions seem to arise naturally from the interaction of muscle, ligament, articular contact, and external forces acting on the systems. Such interactions likely become very important to consider when soft tissue restraints are compromised (e.g. ACL deficiency), resulting in the joint exhibiting even larger deviations from passive behavior. We note that this gait simulation only considered actuation at the knee. Future studies will include multi-joint muscle actuation to better understand the effect of inter-segmental dynamics and coordination on movement at the knee.

CONCLUSIONS

We conclude that a co-simulation model is important to account for load-dependent changes in knee kinematics during gait. This more rigorous formulation could be particularly important when investigating the effects of soft tissue injury, surgical reconstruction, and rehabilitation protocols on internal joint mechanics during movement.

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