

Development of an Open-Source Cosimulation Method of the Knee

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Abstract—Rigid body dynamics and soft tissue loads are solved simultaneously in a cosimulation framework to couple musculoskeletal dynamics and tissue mechanics. The goal of this work was to implement a validated, open-source cosimulation framework of the knee to determine how this coupling affects computed cartilage loads. The kinematic knee joint of a generic whole body model in the open-source software OpenSim was replaced by a previously developed discrete element knee model that consisted of a six degree of freedom (dof) tibiofemoral joint and one dof patellofemoral joint. A serial approach was initially used to estimate muscle forces and cartilage contact loads for a simple flexion movement. Then, a cosimulation framework was implemented for a simple knee flexion movement in which neuromusculoskeletal dynamics and knee mechanics were simultaneously solved using a computed muscle control (CMC) algorithm. This work highlights that the choice of computation method and the precise acquisition of all the dofs of the knee are important factors to consider when estimating soft tissue loads.

I. INTRODUCTION

Osteoarthritis occurs more in the knee than any other joint in the human body [1]. Altered joint arthrokinematics resulting from injury are thought to contribute to the progression of osteoarthritis. For example, anterior cruciate ligament (ACL) deficient knees exhibit increased joint laxity that shifts cartilage contact loads from thicker areas of cartilage to thinner areas that are not adapted to loading [2]. Hence, the progression of cartilage thinning that accompanies osteoarthritis [2]. Knee models are valuable tools that can be used to estimate these hard to measure internal soft tissue loads.

Soft tissue loads can be computed using two methods: a serial and cosimulation approach. In a serial methodology [3], muscle forces are calculated using an optimization strategy and a kinematically constrained knee joint. These muscle forces are then used as boundary conditions to drive a more detailed joint model, e.g. finite or discrete element models. Hence, the serial approach is a two step method. This approach has been used to investigate the use of quadriceps and hamstring muscles to compensate for increased joint laxity in ACL deficient knees [4]. However, the serial approach does not include the effects of joint laxity in the calculation of muscle forces. This has led to the development of the cosimulation method where soft tissue mechanics and rigid body dynamics are computed simultaneously, making this a single step algorithm. This method has been used to investigate neuromuscular coordination patterns to optimize jumping while simultaneously calculating soft tissue loading in the foot [5]. Notably, cosimulation has also been used to simultaneously predict cartilage contact loads and muscle

forces during functional movement like gait [6]. Cosimulation is advantageous in that it provides more realistic estimates of muscle forces than the serial approach [7]. What remains unclear is the influence of simulation method on the cartilage contact loads, which is an important variable in the study of osteoarthritis initiation and progression [2].

Open-source models provide a way to make cosimulation a more accessible method to study these interactions. Recently, a discrete element knee model has been developed that is completely open-source [8]. As a next step, the goal of this work was to develop an open-source cosimulation framework of the knee to determine the influence of simulation method on cartilage loading.

II. METHODS

A. Serial Simulation

A serial approach was initially used to calculate cartilage contact loads during a simple knee flexion motion. First, a generic whole body model (i.e. gait2392 model [9]) in the open-source software OpenSim [10] was scaled for a 77.5 kg female to be consistent with the size of the discrete element knee model [8]. This generic model consists of a knee joint that is kinematically constrained as a modified hinge joint.

Next, inverse dynamics was used to calculate the torque needed to move the model (i.e. kinematic knee joint) through a right knee flexion angle ranging from 0 to 50 deg. Then, computed muscle control (CMC, [11]) was used to calculate the muscle forces needed to track this knee flexion motion. This knee angle was input to the computed muscle control algorithm as the prescribed motion to track. The CMC algorithm uses a PD controller to calculate the acceleration needed to drive the model towards the prescribed motion (1).

$$\ddot{\theta}^*(t+T) - \ddot{\theta}_p(t+T) = k_v(\dot{\theta}_p(t) - \dot{\theta}(t)) + k_u(\theta_p(t) - \theta(t)) \quad (1)$$

In this equation $\ddot{\theta}^*$ is the desired acceleration calculated after the time interval T needed to drive the model towards the prescribed kinematics; $\dot{\theta}$ and θ are the angular velocity and displacement reached by the model under the effect of the muscle forces; $\dot{\theta}_p$, θ_p , and θ_p are the kinematic variables of the prescribed motion; k_v and k_u are the velocity and position feedback constants of the PD controller. Once the acceleration is calculated, optimization is used to determine the muscle forces needed to achieve this model acceleration. In OpenSim, the CMC algorithm calculates the acceleration induced by each muscle at each time step through a forward dynamics simulation. This formulation is important to note as

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this differs between packages. For example, other formulations calculate the muscle induced accelerations by assuming temporary kinematic joint constraints [6]. The optimization algorithm (2) minimizes J , which is the sum of squared controls, x_i (i.e. muscle activations), across all muscles n_x . Along with the cost function, equality constraints $C_j=0$ are included to require the desired accelerations to be within the tolerance set for the optimizer.

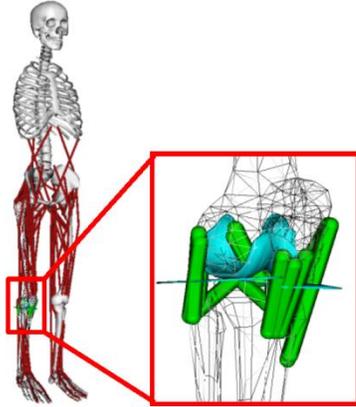


Figure 1: The kinematic knee joint of a scaled, generic musculoskeletal model was replaced with a detailed knee model.

$$J = \sum_{i=1}^{n_x} x_i^2 \quad (2)$$

$$C_j = \ddot{\theta}^* - \ddot{\theta}$$

In addition to the musculoskeletal components, a single coordinate actuator was included for the knee angle. This residual reserve actuator accounts for forces the model could not resolve with muscle actuators alone.

Following the calculation of muscle forces, the kinematic knee joint was replaced by a previously developed open-source, discrete element knee model that consisted of a six degree of freedom (dof) tibiofemoral joint and one dof patellofemoral joint (Fig. 1) [8]. The knee model contains eighteen ligament bundles, elastic foundation contact between the tibia and femur, and a kinematically constrained patella. This detailed knee model is available from <https://simtk.org/home/kneemodel/> and can be freely downloaded and recreated using OpenSim. Then, the muscle forces calculated from the kinematic joint were used in a forward dynamics simulation to actuate the detailed joint model.

B. Cosimulation

A cosimulation framework free to download from <https://simtk.org/home/uwcosim> was implemented in OpenSim. Cosimulation was performed for the same simple knee flexion movement as the serial simulation but neuromusculoskeletal dynamics and knee mechanics were simultaneously solved by using the detailed joint model directly. The CMC algorithm was used to track the knee flexion angle from 0-50 deg.

For the cosimulation approach, CMC was used to track the same prescribed knee flexion movement as the serial approach but directly using the discrete element knee model.

To track the knee flexion angle accurately, k_u was set to 8000, k_v was set to 179, and the CMC look ahead window T (i.e. integration step) was set to 0.001 sec. At each point in time, the CMC algorithm performs a forward dynamic simulation to calculate how each muscle accelerates the model and compare these accelerations to the prescribed kinematics to determine which muscle to actuate. Therefore, these high tracking weights and small look ahead window were needed so the algorithm would not integrate forward in time too much and miss the subtle interaction between the soft tissue loading and the accelerations induced by the muscle. Most notably, this occurs with the quadriceps muscles since the quadriceps pull on the patellar tendon to induce motion.

A single coordinate actuator was included for the knee angle in the cosimulation method. The CMC algorithm currently implemented in OpenSim does not account for passive forces (Appendix) [12]. Therefore, adding a reserve actuator to the model quantifies the amount of torque due to these soft tissue restraints.

C. Comparison of Serial Simulation and Cosimulation

To assess the results of both simulation methods, the following variables were investigated: a comparison of the model predicted knee flexion angle with that of the prescribed knee flexion motion and a comparison of the knee angle torque obtained by inverse dynamics with the torque of the reserve actuator and the muscles. Additionally, the two methods were compared by investigating the total knee flexor muscle force, total knee extensor muscle force, medial cartilage load, lateral cartilage load, and total (medial + lateral) cartilage force.

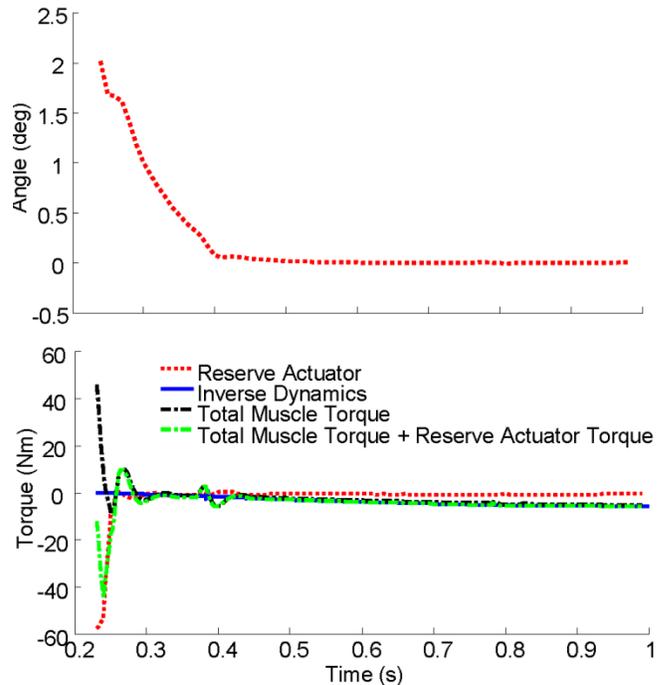


Figure 2: (top) Difference between prescribed and CMC tracked knee flexion angle. CMC tracks knee flexion within 2 degrees. (bottom) Up to -5.9 Nm at the most flexed knee angle ($t = 1$ sec) were needed to actuate the model: -0.5 Nm from the reserve actuator and -5.4 Nm from the muscles.

III. RESULTS

When the muscle forces were calculated in the serial simulation (i.e. assuming a hinge knee joint), CMC tracks knee flexion within 2 degrees (Fig. 2). As the knee flexed, up to -5.9 Nm were needed to actuate the model: -0.5 Nm from the reserve actuator and -5.4 Nm from the muscles. When the muscle forces from the hinge joint were used to actuate the detailed joint model, the joint kinematics differed from those assumed in the hinge joint with differences of 29 deg of flexion, 5 deg of adduction, 5 deg of external rotation, 16 mm of anterior translation, 5 mm of superior translation, and 2 mm of medial translation (Fig. 3).

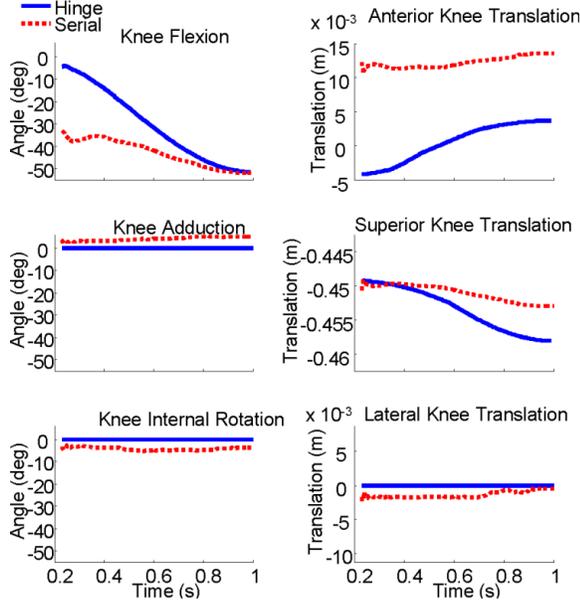


Figure 3: The kinematics predicted by the detailed model in the simulation approach vary from those assumed in the hinge model used to calculate the muscle forces.

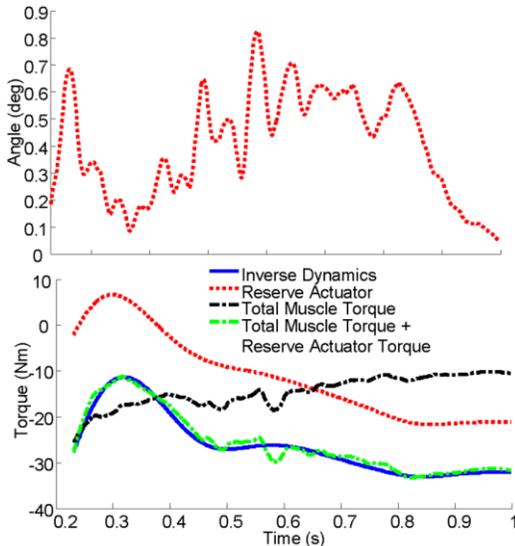


Figure 4: (top) In the cosimulation approach, CMC tracks knee flexion within 1 degree. (bottom) Up to -32 Nm were needed to actuate the model: -21 Nm from the reserve actuator and -11 Nm from the muscles.

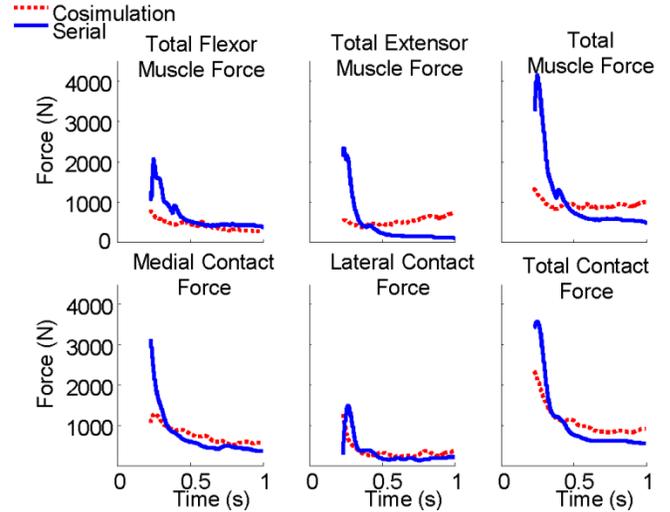


Figure 5: Compared to the cosimulation approach, the serial methodology predicted an overall lower muscle force and cartilage contact load.

When the joint laxity was included in the calculation of muscle forces during the cosimulation method, CMC tracks knee flexion within 1 degree (Fig. 4). As the knee flexed, up to -32 Nm were needed to actuate the model: -21 Nm from the reserve actuator and -11 Nm from the muscles.

At the most flexed knee angle, the serial simulation predicted a lower extensor and total muscle force by 606 N and 484 N, respectively (Fig. 5). Similarly, the serial approach also predicted lower cartilage contact loads in the medial compartment, lateral compartment, and overall by 218 N, 142 N, and 352 N, respectively. Conversely, the serial approach estimated a higher flexor muscle force by 116 N.

IV. DISCUSSION

OpenSim was used to estimate muscle forces while simultaneously accounting for soft tissue loads, e.g. contact between bones. The model, setup files, and results can be freely downloaded from <https://simtk.org/home/uwcosim>, making this an open-source cosimulation framework. The CMC algorithm in OpenSim was able to predict the knee flexion angle within one degree of the prescribed input knee angle (Fig. 2).

The serial approach computed higher flexor and lower extensor muscle force (Fig. 5). This is likely due to the large difference in kinematics between the two methods (Fig. 3). When the muscle forces were computed in the serial approach with a hinge joint, the model assumed the tibia was more posterior than the detailed joint model (Fig. 3). A more posteriorly positioned tibia would decrease the flexion moment arm of the flexors and increase the moment arm of the extensors. To properly actuate the flexion motion, the flexors would need to increase in force to produce the same amount of torque with the decrease moment arm. Conversely, the extensors would decrease in force to produce the same torque with an increased moment arm. Hence, the differences in muscle forces seen between the two simulation methods (Fig. 5). Muscle forces are the main contributor to cartilage loads, especially in the absence of foot-ground contact [13].

In this study, ligaments, muscle, and cartilage contact loading were simultaneously estimated with rigid body mechanics. Although a simple knee flexion motion was used, this is a pivotal first step for the use of the cosimulation method in later dynamic simulations of movement. Before this model can be used to simulate gait, an accurate in vivo measurement of the kinematics of the additional dofs of the knee is required. Knee translational movements are present even under light load [14, 15]. As observed in the results, forcing the model to follow a single rotational dof gives rise to elevated forces of the reserve actuators which compensate when the combination of muscle forces are not able to achieve the required accuracy [16]. We can exclude that these additional forces are due to the passive components of muscle forces since the simulations done using a hinge returns negligible reserve actuators forces.

With proper kinematic measurements, this cosimulation method can be used to investigate the influence of ligament properties and geometry on knee mechanics. For example, how does ACL placement affect quadriceps loading during functional movement? This might have implications for ACL reconstructive surgery. This work demonstrates that a modified hinge model is not “good enough” when representing the movement of the knee, especially when the estimation of the knee loads needs to be accurate for optimal ligament reconstruction.

APPENDIX

Consider a sphere (e.g. femoral epicondyles) rolling on a plane (e.g. tibial plateau) constrained by a hinge (Fig. A1 top). The hinge joint has a clearly defined center of rotation in space. The torque needed to actuate the rolling motion is due to active force actuators, i.e. muscles. When passive soft tissue restraints, such as contact and ligaments, are included in the model (Fig. A1 bottom), the center of rotation is no longer fixed in space. Hence, soft tissue forces can contribute to the rotation motion of the joint, typically to oppose and hinder the motion. CMC minimizes the sum of muscle activations squared. Then, these activations are multiplied by their respective maximum isometric strength and muscle moment arm to calculate the torque driving the model. This inherently ignores the effect of soft tissue forces on the CMC tracked motion.

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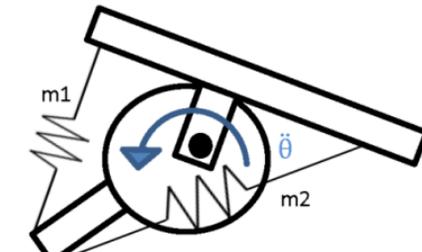
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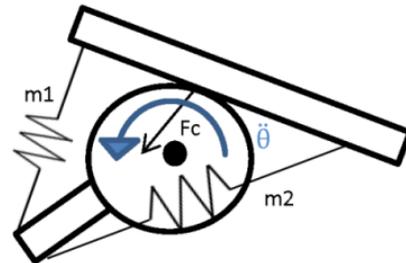
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$$I\ddot{\theta} = T_{m1} + T_{m2}$$



$$I\ddot{\theta} = T_{m1} + T_{m2} + T_{F_c}$$

$$\phi = \min(a_{m1}^2 + a_{m2}^2)$$

$$a_{m1} * F_{m1,strength} * r_{m1} + a_{m2} * F_{m2,strength} * r_{m2} = I\ddot{\theta}$$

Figure A1: The CMC algorithm does not include passive forces, F_c , when calculating muscle forces needed to actuate a model to track a motion. Here, the variables r_{m1} and r_{m2} refer to the moment arms of the muscles about the knee joint center.