INTRODUCTION
It is widely accepted that alterations in the mechanical environment within the knee (i.e., altered stresses or strains generated within the cartilage) is a leading cause of knee osteoarthritis (OA). There are two methods of investigating musculoskeletal joint mechanics that have been used to date: 1) forward and inverse multibody dynamic simulations of human movement and 2) detailed quasi-static finite element (FE) modeling of individual joints. The overwhelming majority of work has been focused on musculoskeletal multibody dynamics modeling. However, multibody dynamics simulations do not allow for the detailed continuum level analysis of the mechanical environment of the cartilage and other knee joint structures (meniscus, ligaments, and underlying bone) within the knee during physical activities. This is a critical technology gap that is required to understand the relationship between functional or injurious loading of the knee and cartilage degradation. The objective of this investigation was to develop a detailed neuromuscularly activated dynamic finite element model of the human lower body in order to simultaneously determine the dynamic muscle forces, joint kinematics, contact forces, and detailed (e.g., continuum) stresses and strains within the knee (cartilage, meniscus, ligaments, and bone) during controlled and actuated lower limb movements.

METHODS
A first generation finite element model of the lower body has been developed using individual digitized anatomic surface models based on the Visible Human project from the Digimation human anatomy collection (www.digimation.com). Volumetric finite element meshes were created from the digitized anatomic surfaces using LS-Prepost (v. 3.0 LSTC, Livemore CA). Contact conditions were defined at the knee joint between the femoral cartilage, tibial cartilage, menisci, and patella cartilage to constrain articular joint motion based on anatomic geometry; all other joints are fixed in this version of the model for the leg extension simulation. All the ligaments of the knee (anterior cruciate, posterior cruciate, medial, and lateral ligaments) were modeled as non-linear one-dimensional springs; the femoral cartilage, tibial cartilage, menisci were modeled as linear elastic materials using published material properties [1]. Fifty-nine muscles of the lower limbs were implemented using one-dimensional active contraction Hill-type muscle springs (LS-DYNA v.971, LSTC, Livemore, CA) using published insertion points and Hill muscle model parameters [3]. The model was configured to simulate a seated leg extension exercise and vastus lateralis, vastus medialis, rectus femoris, and vastus intermedius (quadriceps) were activated while the activations for the remaining muscles were kept at zero or at a minimal activation level for stability.

We use a parameter optimization method to directly optimize muscle activation parameters within the dynamic FE model based on a theoretical cost function [8]. This approach has been shown to result in muscle activation patterns that correlate with empirical surface EMG data and that produce musculoskeletal motion that compares with experimentally measured motion with a high level of accuracy. The optimization procedure determined the optimal muscle control parameters for a leg extension using only the quadriceps muscles with 3 activation times using an objective function of the form:

\[ J = \sum_{m=1}^{m} (a_{m}(t))^2 \]

where \( J \) is the total energy, \( a_{m}(t) \) is the muscle activation for muscle \( m \) at discrete time step \( i \), \( n \) is the total number of muscles activated, and \( t \) is the total time for the desired motion. With 4 muscles and 3 activations times, this results in 13 finite element analyses required to perform a forward difference gradient calculation required for the optimizer. The initial conditions for all \( a_{m}(t) \) were set at the manually determined activations required to perform a leg extension without additional weight added at the ankle and the optimization was performed for a single leg extension with a 30 pound weight added at the ankle. All muscles used identical initial activation curves. The optimal values of \( a_{m}(t) \) were determined that resulted in a 90 degree extension of the knee joint while minimizing the objective function. For each model solution, the dynamic finite element model was solved using the explicit finite element code LS-DYNA (v. 971, LSTC, Livemore, CA).

RESULTS
A single solution of the dynamic finite element model required ~20 hours using 2 CPUs. The initial muscle activations (Figure 1) required to perform a leg extension result in a smooth controlled dynamic movement that results in the tibia articulating from approximately 90 degrees with respect to the femur to approximately 180 degrees with respect to the femur. The muscle activation parameter optimization required 13 iterations to converge to a solution and required approximately 260 CPU hours to complete. Gradient analyses were performed in parallel on a 172 CPU compute cluster.

DISCUSSION
In this project, we have utilized multibody continuum FE dynamics, active muscle modeling, and parameter optimization methods to generate a high fidelity dynamic model of the human lower body. We have explicitly modeled the lower limbs (pelvis down to the foot), including a detailed representation of the knee, using finite elements. Motion at the knee joint is controlled explicitly via deformable surface contact at each articular surface (rather than idealized as simple revolute or ball and socket joints). The major muscles activating the lower limb are explicitly modeled using anatomical muscle insertion points and geometric wrapping. The dynamic muscle forces, joint kinematics, contact forces, and detailed (e.g., continuum) stresses and strains within the knee (cartilage, meniscus, ligaments, and bone) were simultaneously determined for a neuromuscularly controlled seated leg extension with a weight of 30 lbs. added to the ankle. This methodology will be used to investigate how structural and material alterations in this complex environment due to ACL rupture, obesity, and sarcopenia (each of which are risk factors for knee OA) potentially affect the detailed dynamic mechanics of each component of the knee joint.

REFERENCES

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SIGNIFICANCE
It is widely accepted that alterations in the mechanical environment within the knee (i.e., altered stresses or strains generated within the cartilage) is a leading cause of knee osteoarthritis (OA). The methodology developed herein will be used to investigate how structural and material alterations in the complex environment of the knee due to ACL rupture, obesity, and sarcopenia (each of which are risk factors for knee OA) potentially affect the detailed dynamic mechanics of each component of the knee joint, particularly the articular cartilage.

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