# OpenSim: Open-Source Software to Create and Analyze Dynamic Simulations of Movement

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***Abstract*—Dynamic simulations of movement allow one to study neuromuscular coordination, analyze athletic performance, and estimate internal loading of the musculoskeletal system. Simulations can also be used to identify the sources of pathological movement and establish a scientific basis for treatment planning. We have developed a freely available, open-source software system (OpenSim) that lets users develop models of musculoskeletal structures and create dynamic simulations of a wide variety of movements. We are using this system to simulate the dynamics of individuals with pathological gait and to explore the biomechanical effects of treatments. OpenSim provides a platform on which the biomechanics community can build a library of simulations that can be exchanged, tested, analyzed, and improved through a multi-institutional collaboration. Developing software that enables a concerted effort from many investigators poses technical and sociological challenges. Meeting those challenges will accelerate the discovery of principles that govern movement control and improve treatments for individuals with movement pathologies.**

***Index Terms*—Computed muscle control, forward dynamic simulation, musculoskeletal modeling, open-source software.**

## I. Introduction

MANY elements of the neuromusculoskeletal system interact to enable coordinated movement. Scientists fascinated by human movement have performed an extensive range of studies to describe these elements. As a result, there is a wealth of data that characterize the mechanics of muscle, the geometric relationships between muscles and bones, and the motions of joints. Clinicians who treat movement abnormalities in individuals with cerebral palsy, stroke, osteoarthritis and Parkinson's disease have examined the neuromuscular excitation patterns and movement kinematics of literally thousands of patients, both before and after treatment interventions. However, synthesizing detailed descriptions of the elements of the neuromusculoskeletal system with measurements of movement to create an integrated understanding of normal movement and to establish a scientific basis for correcting abnormal movement remains a major challenge.

Using experiments alone to understand movement dynamics has two fundamental limitations. First, important variables, including the forces generated by muscles, are not generally measurable in experiments. Second, it is difficult to establish cause-effect relationships in complex dynamic systems from experimental data alone. As a result, elucidating the functions of muscles from experiments is not straightforward. For example, electromyographic (EMG) recordings can indicate when a muscle is active, but examination of EMG recordings does not allow one to determine which motions of the body arise from a muscle's activity. Determining how individual muscles contribute to observed motions is difficult because a muscle can accelerate joints that it does not span and body segments to which it does not attach [1].

A theoretical framework is needed, in combination with experiments, to uncover the principles that govern the coordination of muscles during normal movement, to determine how neuromuscular impairments contribute to abnormal movement, and to predict the functional consequences of treatments. To achieve these goals, the theoretical framework must reveal the cause-effect relationships between neuromuscular excitation patterns, muscle forces, and motions of the body.

A dynamic simulation of movement that integrates models describing the anatomy and physiology of the elements of the neuromusculoskeletal system and the mechanics of multijoint movement provides such a framework. Muscle-driven dynamic simulations complement experimental approaches by providing estimates of important variables, such as muscle and joint forces, which are difficult to measure experimentally. Simulations also enable cause-effect relationships to be identified and allow "what if?" studies to be performed in which, for example, the excitation pattern of a muscle can be changed and the resulting motion can be observed.

Although the value of dynamic simulations of movement is broadly recognized [2]–[8], the field is fragmented. Many laboratories develop their own simulation software, and do not provide this software to others; thus, it is difficult for a simulation to be used or evaluated outside the laboratory where it is developed. The inability to reproduce results is a major limitation to advancing the science of biomedical simulation. Individual investigators have made elegant contributions to simulation technology, including the development of novel methods to model muscle [9]–[11], simulate contact [12], [13], and repre-

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sent musculoskeletal geometry [14]–[16], but it is difficult for others to make use of these new techniques because the software that implements them is generally unavailable. Since software tools are not freely accessible to assist in the development, analysis, and control of musculoskeletal dynamic simulations, researchers typically must spend a great deal of time implementing each new simulation and creating tools to analyze it. Developing dynamic simulations of movement is technically challenging, and many movement science laboratories lack the resources or technical expertise to generate their own simulations. These conditions create a major barrier to advancing simulation technology and achieving the scientific potential of musculoskeletal simulations.

In the early 1990s, Delp and Loan introduced a musculoskeletal modeling environment, called SIMM [17]–[19], that lets users create, alter, and evaluate models of many different musculoskeletal structures [20]–[22]. This software is now used by hundreds of biomechanics researchers to create computer models of musculoskeletal structures and to simulate movements such as walking [23]–[25], cycling [26]–[28], running [29], [30], and stair climbing [31]. Using SIMM, models of the lower and upper extremities were developed to examine the biomechanical consequences of surgical procedures including tendon surgeries [32]–[38], osteotomies [39]–[41] and total joint replacements [42]–[44]. A lower-extremity model was used to estimate muscle-tendon lengths, velocities, moment arms, and induced accelerations during normal and pathologic gait [45]–[52]. Studies have been conducted to investigate the treatment of individuals with spinal cord injury [53]–[56], to analyze joint mechanics in subjects with patellofemoral pain [57], [58], to calculate forces at the knee during running [59] and cutting [60], to examine the influence of foot positioning and joint compliance on the occurrence of ankle sprains [61], [62], and to investigate causes of abnormal gait [63]–[65]. These studies have demonstrated the utility of musculoskeletal models and dynamic simulations for analyzing the causes of gait abnormalities and the effects of various treatments. SIMM has helped bring simulation to biologists who have created computational models of the frog [66], [67], Tyrannosaur [22], cockroach [68], and other animals.

Although SIMM helps users formulate models of the musculoskeletal system and dynamic simulations of movement, it provides no assistance with the computation of muscle excitations that produce coordinated movement and has limited tools for analyzing the results of dynamic simulations. Furthermore, SIMM and other commercial packages, such as Visual 3-D (C-Motion Inc.), Anybody (Anybody Technology) or Adams (MSC Software Corp.), do not provide full access to source code, which makes it difficult for biomechanics researchers to extend their capabilities. Over the past decade, new software engineering methods have emerged that enable the development of software systems that are more extensible. We view this as an opportunity to develop a simulation platform that engages a broader spectrum of the biomechanics community.

We have established an open-source simulation environment, called OpenSim, to accelerate the development and sharing of simulation technology and to better integrate dynamic simulations into the field of movement science (Fig. 1). Open-source software development has become a successful strategy for both commercial and academic efforts (e.g., the Linux operating system). Making source code available enables researchers to reproduce results produced by other laboratories and to make improvements and adapt code to meet their needs. Modern plug-in technology, which we have adopted, lets users extend software functionality and allows new tools to be shared more easily. We believe that the biomechanics community will benefit from a greater degree of collaboration as a result of an open-source effort.

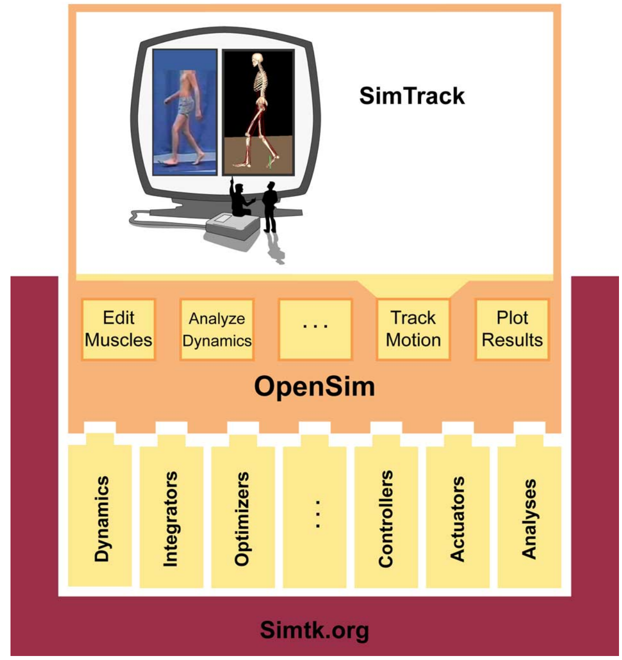


Fig. 1. Schematic of OpenSim, an open source software system for modeling, simulating, and analyzing the neuromusculoskeletal system. OpenSim is built on top of core computational components that allow one to derive equations of motion for dynamical systems, perform numerical integration, and solve constrained non-linear optimization problems. In addition, OpenSim offers access to control algorithms (e.g., computed muscle control), actuators (e.g., muscle and contact models), and analyses (e.g., muscle-induced accelerations). OpenSim integrates these components into a modeling and simulation platform. Users can extend OpenSim by writing their own plug-ins for analysis or control, or to represent neuromusculoskeletal elements (e.g., muscle models). In a graphical user interface, the user is able to access a suite of high-level tools for viewing models, editing muscles, plotting results, and other functions. SimTrack, one of the OpenSim tools, enables accurate muscle-driven simulations to be generated that represent the dynamics of individual subjects. OpenSim is being developed and maintained on Simtk.org; all of the software is freely available.

Enticing researchers to help develop and test open-source software requires the initial developers to provide tools that others can use and extend. OpenSim provides two. The first comprises a set of modeling and analysis tools that complement those included in SIMM [17], [19]. The second, SimTrack, enables researchers to generate dynamic simulations of movement from motion capture data.

This article first provides a brief overview of OpenSim. We then focus on SimTrack and how simulations that characterize the dynamics of individual subjects can assist in treatment planning. We describe a method to generate subject-specific simulations and present a case study, in which we used a dynamic simulation of a subject with stiff-knee gait to understand the causes of his abnormal movement and the effects of possible treatments. We close with a review of the challenges for the field.

## II. What Is OpenSim?

OpenSim is an open-source platform for modeling, simulating, and analyzing the neuromusculoskeletal system. It includes low-level computational tools that are invoked by an application (Fig. 1). A graphical user interface provides access to key functionality. OpenSim is being developed and maintained on Simtk.org by a growing group of participants. Simtk.org serves as a public repository for data, models, and computational tools related to physics-based simulation of biological structures.

The software is written in ANSI C++, and the graphical user interface is written in Java, allowing OpenSim to compile and run on common operating systems. Open-source, third-party tools are used for some basic functionality, including the Xerces Parser from the Apache Foundation for reading and writing XML files (xml.apache.org/xerces-c) and the Visualization Toolkit from Kitware for visualization (www.vtk.org). Use of plug-in technology allows low-level computational components such as dynamics engines, integrators, and optimizers to be updated as appropriate without extensive restructuring. For example, OpenSim initially used SDFast (Parametric Technology Corp.) as its dynamics engine; however, current releases will allow Simbody$^{\mathrm{TM}}$ to be used as well. Simbody™ is an open-source order-n dynamics engine under development at Simtk.org.

The plug-in architecture of OpenSim encourages users to extend functionality by developing their own muscle models, contact models, controllers, and analyses. For example, about a dozen analysis plug-ins, authored by different users, are available in OpenSim. These analysis tools calculate joint forces, muscle-induced accelerations, muscle powers, and other variables. Although these analyses were developed for different musculoskeletal models, they have general applicability and can be used with any OpenSim model. The plug-in architecture of OpenSim thus provides a means of rapidly disseminating new functionality to the biomechanics community.

To add a plug-in (e.g., an analysis), a user must write a new C++ class (e.g., InducedAcceleration) derived from the appropriate base class (e.g., Analysis), implement a number of required methods, and compile the class into a dynamically linked library. The new plug-in (e.g., the InducedAcceleration analysis) can then be used in simulations and shared with other users. Independently, plug-ins can also be developed to enhance the capabilities of the graphical user interface. The user interface gets nearly all its functionality from plug-ins. For example, the modules for motion viewing, plotting, and muscle editing are all plug-ins. A user interface plug-in example is provided with OpenSim that users can adapt to extend the functionality of the graphical interface. Like the low-level C++ plug-ins for analyses, muscle models, controllers, etc., user interface plug-ins can be shared with other users.

The OpenSim graphical user interface includes a suite of tools for analyzing musculoskeletal models, generating simulations, and visualizing results (Fig. 2). Some of the basic functionality of SIMM is available in OpenSim, including, for example, the ability to edit muscles and plot variables of interest. In addition, SIMM joint (*.jnt) and muscle (*.msl) files [18] can be imported. OpenSim provides simulation and control capabilities that complement SIMM. SimTrack, in particular, is a tool capable of generating muscle-actuated simulations of subject-specific motion quickly and accurately, as described below.

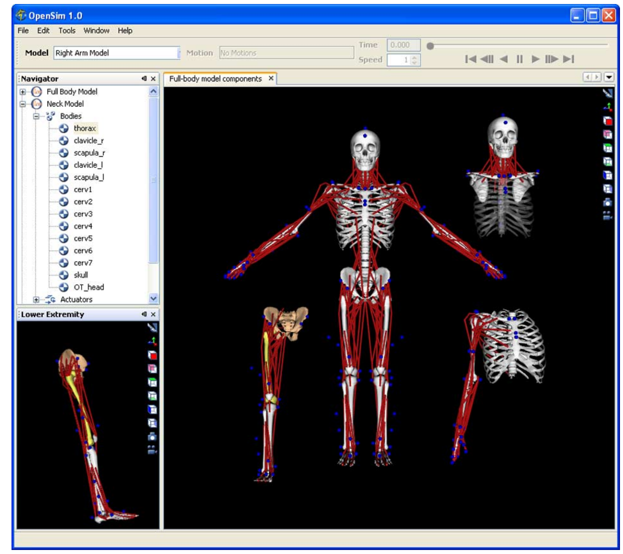


Fig. 2. Screenshot from OpenSim. Models of many different musculoskeletal structures, including the lower extremity, upper extremity, and neck, can be loaded, viewed and analyzed. Muscles are shown as red lines; virtual markers are shown as blue spheres.

## III. SimTrack: An OpenSim Tool for Generating Dynamic Simulations

To create a muscle-driven simulation of a movement, one must first formulate a dynamic model of the musculoskeletal system and its interactions with the environment. The elements of the musculoskeletal system are modeled by sets of differential equations that describe muscle contraction dynamics, musculoskeletal geometry, and body segmental dynamics. These equations characterize the time-dependent behavior of the musculoskeletal system in response to neuromuscular excitation. Once a dynamic model of the musculoskeletal system is formulated, the next step is to find a pattern of muscle excitations that produce a coordinated movement. Excitations may be found by solving an optimization problem in which the objective of a motor task is defined (e.g., jumping as high as possible) or in which the objective is to drive a dynamic model to "track" experimental motion data [69]. Simulations are generally evaluated by how well they agree with experimentally measured kinematics, kinetics, and EMG patterns. Once a simulation is created and its accuracy is tested, it can be analyzed to examine the contributions a muscle makes to the motions of the body and the consequences of a simulated treatment.

Determining a set of muscle excitations that produce a coordinated movement is one of the major challenges in creating a dynamic simulation. Historically, the computational cost of generating coordinated muscle-actuated simulations of movement has been high, requiring days, weeks, or months of computer time [23], [65], [70]. Recent breakthroughs in the application

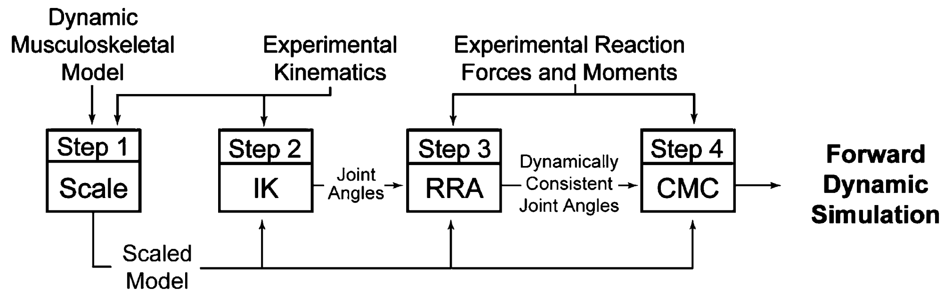


Fig. 3. Steps for generating a muscle-driven simulation of a subject's motion with SimTrack. The inputs are a dynamic musculoskeletal model, experimental kinematics (i.e., x-y-z trajectories of marker data, joint centers, and joint angles), and experimental reaction forces and moments obtained from a subject. In Step 1, the experimental kinematics are used to scale the musculoskeletal model to match the dimensions of the subject. In Step 2, an inverse kinematics (IK) problem is solved to find the model joint angles that best reproduce the experimental kinematics. In Step 3, a residual reduction algorithm (RRA) is used to refine the model kinematics so that they are more dynamically consistent with the experimental reaction forces and moments. In Step 4, a computed muscle control (CMC) algorithm is used to find a set of muscle excitations that will generate a forward dynamic simulation that closely tracks the motion of the subject.

of robotic control techniques to biomechanical simulation have dramatically reduced the time needed to generate such simulations [28], [71]. For example, the computed muscle control technique determines muscle excitations that reproduce measured pedaling dynamics in just ten minutes [28]; this is over two orders of magnitude faster than conventional dynamic optimization techniques. Thelen and Anderson extended this approach to compute muscle excitation patterns that drove a 21-degree-of-freedom, 92-muscle model to track experimental gait data of 10 healthy adults [25]. A simulation of a half-cycle of gait was generated in approximately 30 minutes. The speed of this technique makes it practical to generate subject-specific simulations of a wide variety of movements.

SimTrack guides users through four steps to create a dynamic simulation (Fig. 3). As input, SimTrack takes a dynamic model of the musculoskeletal system and experimentally-measured kinematics and reaction forces and moments. While this approach is general, we will describe it in the context of generating simulations of gait, since this is one of the most challenging applications.

In Step 1, a dynamic musculoskeletal model (e.g., a SIMM model [19]) is scaled to match the anthropometry of an individual subject. The dimensions of each body segment in the model are scaled based on relative distances between pairs of markers obtained from a motion-capture system and the corresponding virtual marker locations in the model (e.g., see blue spheres in Fig. 2). The mass properties of the body segments are scaled proportionally so that the total measured mass of the subject is reproduced. Muscle fiber lengths and tendon slack lengths of the muscle-tendon actuators are scaled so that they each remain the same percentage of total actuator length.

In Step 2, an inverse kinematics (IK) problem is solved to determine the model generalized coordinate values (joint angles and translations) that best reproduce the raw marker data obtained from motion capture. Step 2 is formulated as a least-squares problem that minimizes the differences between the measured marker locations and the model's virtual marker locations, subject to joint constraints [72]. If the experimental kinematics includes a set of joint centers or joint angles produced by motion-capture software, these may also be included in the formulation. Therefore, for each frame in the experimental kinematics, the inverse kinematics problem is to minimize the weighted squared error

$$\text{Squared Error} = \sum_{i=1}^{\text{markers}} w_i \left( \vec{x}_i^{\text{subject}} - \vec{x}_i^{\text{model}} \right)^2 + \sum_{j=1}^{\text{joint angles}} \omega_j \left( \theta_j^{\text{subject}} - \theta_j^{\text{model}} \right)^2 \tag{1}$$

where $\vec{x}_i^{\text{ subject}}$ and $\vec{x}_i^{\text{ model}}$ are the three-dimensional positions of the $i$th marker or joint center for the subject and model, $\theta_j^{\text{subject}}$ and $\theta_j^{\text{model}}$ are the values of the $j$th joint angle for the subject and model, and $w_i$ and $\omega_j$ are factors that allow markers and joint angles to be weighted differently.

Due to experimental error and modeling assumptions, measured ground reaction forces and moments are often dynamically inconsistent with the model kinematics. In Step 3, a residual reduction algorithm (RRA) is applied to make the model generalized coordinates (joint angles and translations) computed in Step 2 more dynamically consistent with the measured ground reaction forces and moments. From Newton's second law, the following equation relates the measured ground reaction force and gravitational acceleration to the accelerations of the body segments

$$\vec{F}_{\text{external}} = \sum_{i=1}^{\text{segments}} m_i \vec{a}_i - \vec{F}_{\text{residual}} \tag{2}$$

where $\vec{F}_{\text{external}}$ is the measured ground reaction force minus the body weight vector, $\vec{a_i}$ is the translational acceleration of the center of mass of the $i$th body segment, $m_i$ is the mass of the $i$th body segment, and $\vec{F}_{\text{residual}}$ is the residual force. An analogous equation relates the ground reaction moment to the model kinematics and the residual moment. In the absence of experimental and modeling error, the residual force should be zero (i.e., $\vec{F}_{\text{residual}} = \vec{0}$). In practice, this is never the case. Through a combination of slight, controlled perturbations to the motion trajectory, and small adjustments to the mass parameters of the model, it is possible to reduce the residual forces and moments required for dynamic consistency. To reduce the residual forces and moments, the residuals are computed and averaged over the duration of the movement. Based on these averages, the algorithm recommends changes in the model mass parameters, such as the location of the center of mass of the trunk, that reduce the average values of the residuals over the duration of the movement. Following any adjustments to the mass parameters, a control problem is solved in which all degrees of freedom of the model are actuated. In particular, the joints are actuated by idealized joint moments, and, in addition, three residual forces and three residual moments are applied to a chosen segment of the model to control the six degrees of freedom between the model and the ground (i.e., three translations and three rotations). If no limits are placed on the residuals, the kinematics can be tracked with little or no error. However, at the user's discretion, upper limits can be placed on the magnitudes of the residuals, in which case the motion of the model is altered yielding a new set of kinematics that are dynamically consistent with the limited residuals. A performance criterion is used to distribute tracking errors across the joint angles

$$\text{Squared Error} = \sum_{j=1}^{\text{joints}} \Omega_j \cdot \left(\ddot{q}_j^{\text{desired}} - \ddot{q}_j^{\text{model}}\right)^2 \qquad (3)$$

where $\Omega_j$ is a factor weighting the relative importance of the $j$th joint, and $\ddot{q}_j^{\text{desired}}$ is the desired acceleration of the $j$th degree of freedom given by a proportional-derivative control law [28]. The values for the model degrees of freedom and mass properties output by the residual reduction algorithm are used as input to Step 4.

In Step 4, computed muscle control (CMC) is used to generate a set of muscle excitations that produce a coordinated muscle-driven simulation of the subject's movement. Computed muscle control uses a static optimization criterion to distribute forces across synergistic muscles and proportional-derivative control to generate a forward dynamic simulation that closely tracks the kinematics derived in Step 3 [25]. Although a static performance criterion is used, the full state equations representing the activation and contraction dynamics of the muscles are incorporated into the forward dynamic simulation. Activation dynamics is modeled by relating the time rate of change of muscle activation $(\dot{a})$ to muscle activation $(a)$ and excitation $(u)$

$$\dot{a} = \begin{cases} (u-a)\cdot[u/\tau_{\text{act}} + (u+1)/\tau_{\text{deact}}] & u > a \\ (u-a)/\tau_{\text{deact}} & u < a \end{cases} \qquad (4)$$

where $\tau_{\text{act}}$ and $\tau_{\text{deact}}$ are the time constants for activation and deactivation. Musculotendon contraction dynamics is described by a lumped-parameter model that accounts for the force-length-velocity properties of muscle and the elastic properties of tendon. In particular, the time rate of change of muscle length $(\dot{l}_m)$ is related to muscle length $(l_m)$, musculotendon actuator length $(l_{\text{mt}})$, and muscle activation $(a)$

$$\dot{l}_m = f_v^{-1}(l_m, l_{\text{mt}}, a) \qquad (5)$$

where $f_v$ is the force velocity relation for muscle. In our current implementation, the force between the foot and the ground is not modeled; rather, the measured ground reaction forces and moments are applied directly to the foot. When analyzing a simulation, as described in the case study below, spring-damper elements are introduced between the foot and the ground to allow the reaction forces and moments to respond to perturbations (e.g., altered muscle forces).

## IV. Case Study

We have generated dynamic simulations of individual subjects with abnormal gait using computed muscle control [25] to examine the causes of their abnormal walking pattern and to simulate treatment options. This case study demonstrates how simulations can provide insight into the causes of stiff-knee gait, a condition in which swing-phase knee flexion is substantially diminished. Reduced knee flexion is often attributed to excessive excitation of the rectus femoris during the swing phase [73]. However, factors that limit knee flexion velocity just before swing, such as excessive force in vasti or rectus femoris, or diminished force in iliopsoas or gastrocnemius, may also reduce knee flexion during swing [74]. Determining which, if any, of these factors limit an individual's knee flexion is challenging because current diagnostic methods cannot evaluate how forces produced by the rectus femoris or other muscles influence swing-phase knee motions.

There are several options for treatment of stiff-knee gait. One option, botulinum toxin injection, theoretically decreases the hip and knee moments generated by the rectus femoris. A second option, rectus femoris transfer, theoretically decreases the muscle's knee extension moment while leaving its hip flexion moment intact. At present, the mechanisms responsible for patients' improvements in swing-phase knee flexion following these treatments are not well understood. In this case study, we generated and analyzed a dynamic simulation of a subject with stiff-knee gait to determine the biomechanical cause of his diminished knee flexion and the potential consequences of different treatment options (Fig. 4).

The subject was a 12-year-old male diagnosed with spastic cerebral palsy. His left lower limb exhibited limited knee flexion during swing and abnormal activity of rectus femoris (preswing and swing) and vasti (preswing). We represented the subject's musculoskeletal system by a scaled, 21-degree-of-freedom linkage actuated by 92 muscles and generated a forward dynamic simulation of the subject's gait. The simulated joint angles reproduced the subject's measured knee flexion angle to within 2° (Fig. 5, "simulated").

We evaluated the contributions of rectus femoris, vasti, and other muscles to knee flexion by altering muscle excitations in the simulation and computing the resulting changes in peak knee flexion. Analysis of the subject's dynamic simulation suggested that excessive activity of the knee extensors in preswing was the major cause of his stiff-knee gait. Decreasing the excitation of rectus femoris or vasti during preswing increased peak knee flexion substantially (Fig. 5, curves A and B). Decreasing the excitation of rectus femoris in early swing had a negligible effect on peak knee flexion (Fig. 5, curve C).

We examined the potential biomechanical consequences of botulinum toxin injection and rectus femoris transfer. Botulinum toxin injection was simulated by decreasing the excessive excitation of rectus femoris while leaving its passive

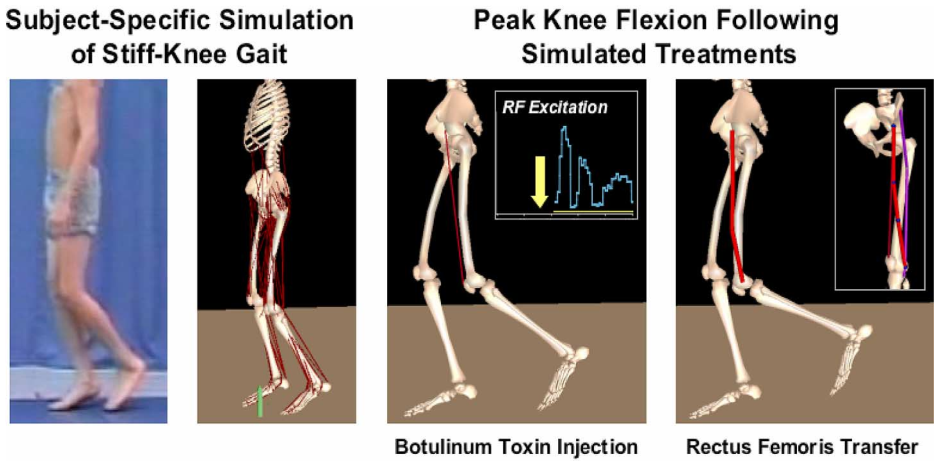


Fig. 4. Simulation-based treatment planning for stiff-knee gait. Stiff-knee gait is characterized by insufficient knee flexion during the swing phase. A muscle-driven simulation that reproduces an individual's movement dynamics (left) can provide a scientific basis for planning treatments; for example, by predicting whether an increase in knee flexion is likely to result following botulinum toxin injection to reduce rectus femoris excitation (right-center) or rectus femoris transfer surgery to change the muscle line of action (right).

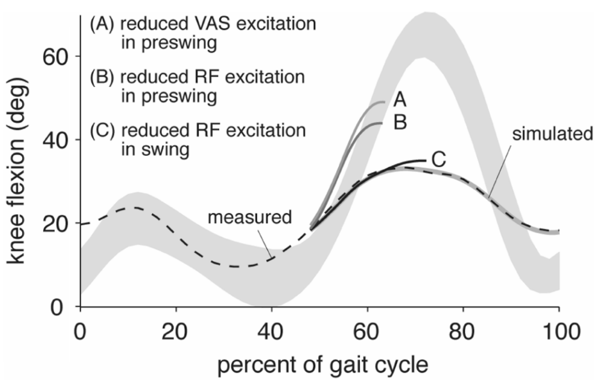


Fig. 5. Knee flexion trajectories for different quadriceps excitation levels. The subject's pre-operative measured knee angle is shown for comparison. Shaded area is the average knee angle for unimpaired subjects ±1 SD. Note that reducing the excitations of the vasti (VAS) and rectus femoris (RF) in preswing had a greater affect on knee motion than reducing excitation during swing.

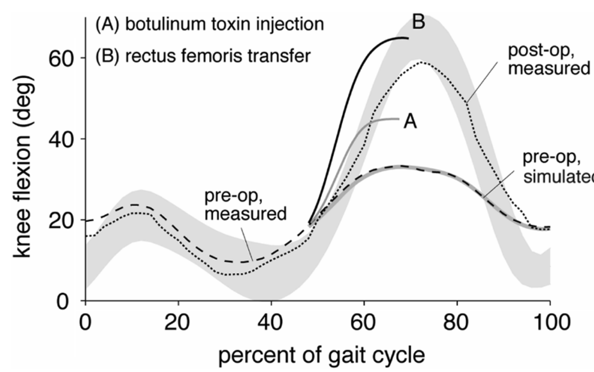


Fig. 6. Knee flexion trajectories for different simulated treatments. The subject's pre- and post-operative measured knee angles are shownfor comparison. Shaded area is the average knee angle for unimpaired subjects ±1 SD.

force-length properties intact. Rectus femoris transfer was simulated by transferring the muscle's insertion in the model to the iliotibial band, a common transfer site [75]. We assumed that the pattern of rectus femoris excitation was not changed by the surgery.

Peak knee flexion increased following each of the simulated treatments. Decreasing the excessive excitation of rectus femoris in the model, simulating the effects of botulinum toxin injection, increased knee flexion by about 10° (Fig. 6, curve A). Eliminating the excessive knee extension moment of rectus femoris in preswing and swing while leaving the hip moment intact, simulating a rectus femoris transfer, increased the peak knee flexion by about 30° (Fig. 6, curve B). This result suggests that preserving the capacity of the rectus femoris to generate a hip flexion moment is important when attempting to correct stiff-knee gait caused by rectus femoris overactivity. This subject underwent a rectus femoris transfer as part of his surgical treatment and achieved significant improvement in both knee flexion velocity at toe-off and knee flexion in swing. The improvement in knee flexion following a simulated tendon transfer were similar to the subject's actual improvements following surgery (Fig. 6).

Simulations of normal walking (e.g., [23], [63], [70]) have enabled investigators to identify the actions of muscles with a level of specificity and certainty that surpasses insights gained with experimental methods alone. Simulations of abnormal walking offer similar potential, but are challenging to develop, in part, because they require determination of muscle excitations that generate the abnormal movement dynamics exhibited by persons with movement disorders. The computed muscle control method [28] provides a computationally efficient means to generate these simulations and is now available for use by researchers around the world.

## V. Opportunities and Challenges for Biomechanical Simulation

We believe simulations will advance movement science by facilitating interactions between modelers and experimentalists. Modelers need experimentalists to acquire parameters used in simulations and to test the accuracy of results derived from simulations. Experimentalists need modelers to provide a theoretical framework within which to interpret experimental observations, and to help gain perspective from the wealth of data derived from biomechanical experiments. With access to open-source software for developing and analyzing muscle-driven simulations, biomechanics researchers are now in a position to establish quantitative, cause-effect relationships between the neuromuscular excitation patterns, muscle forces, external reaction forces, and motions of the body that are observed in the laboratory. Coupled with high-quality experimental measurements, simulations will help elucidate how elements of the neuromusculoskeletal system interact to produce movement and, we hope, improve the outcomes of treatments for persons with movement disorders.

A variety of software packages have been used to create and analyze models of the lower limb [17], [76], upper limb [21], [55], cervical spine [20], lumbar spine [77], and other musculoskeletal structures. Although these models are implemented in different modeling packages, they include similar model parameters. One challenge for the field is to define modeling standards and promote interchange between modeling packages.

Another challenge for the field is to demonstrate that the use of simulations can improve treatment outcomes for individuals with movement disorders. The potential to use subject-specific simulations to better understand the causes of movement deviations and to assess treatment options is exciting, and the case study above provides specific and relevant insights into stiff-knee gait for one subject. Future studies, in which simulations of many subjects are conducted, are needed to determine if general principles for treatment planning can be elucidated from the insights gained from analyzing simulations. Studies that retrospectively compare predictions from subject-specific simulations to the subjects' actual outcomes are also needed to evaluate whether existing musculoskeletal models are sufficiently accurate, and to establish the conditions under which the results of simulations are applicable. The simulation environment we have created makes such large-scale studies possible, though more development is needed to streamline the process of creating and validating simulations of individuals with impairments. Ultimately, prospective clinical trials are needed to determine if simulations can improve treatment outcomes.

The ability to rapidly create coordinated muscle-driven simulations provides new research opportunities. Many previous simulation studies include results from a single simulation. With SimTrack it is possible to generate and analyze 3-D simulations of many subjects, and to establish norms describing the muscle functions for subjects with a range of sizes, strengths, and movement patterns. It is also practical to perform sensitivity studies to determine whether the conclusions drawn from a simulation are sensitive to variations in model parameters. This is especially valuable when a direct comparison with experimental data (e.g., muscle force trajectories) is not feasible. It is also possible, as shown in the case study, to investigate how impairments, such as abnormal muscle excitations, contribute to abnormal movements in individual subjects (Fig. 5), and to explore the functional consequences of treatments (Fig. 6).

The accuracy of a simulation depends on the fidelity of the underlying mathematical model of the neuromusculoskeletal system. Many assumptions are made in the development of musculoskeletal models, and some of these assumptions are based on limited experimental evidence. To improve the accuracy of musculoskeletal models, more *in vivo* measurements of musculoskeletal geometry and joint kinematics are needed to understand how variations due to size, age, deformity, or surgery influence the predictions of a model, and to determine the conditions under which simulations based on a generic model are applicable to individual subjects [79]. Experiments that characterize the effects of pathology and surgery on muscle force generation are needed to test assumptions made in musculoskeletal models and to assess their impact on movement. Advances in the neurosciences are needed to allow development of simulations that incorporate representations of sensory-motor control. Given that simulations include assumptions and approximations, it is critically important that each simulation be tested to establish its limitations. As more investigators use simulations of musculoskeletal dynamics, it is essential that each scientist test the accuracy of their simulations in the context of their specific scientific study.

OpenSim provides new opportunities for collaboration and peer review. The code that comprises OpenSim is being tested, analyzed, and improved through a multi-institutional collaboration. Users are encouraged to modify the code to suit their applications and to share their contributions with others. As a result, simulation-based studies can now be reproduced and tested outside the laboratory where the simulation is first developed. Such rigorous tests are essential if biomechanical simulation is to become more of a science and less of an art.

The development of a digital human (a computational model of the human neuromusculoskeletal system with complexity comparable to a human) is a grand challenge. If a general and comprehensive model were available, then users could choose how to simplify the model to address a particular scientific question. The Physiome Project [78] outlines this challenge and some of the important benefits of its success. Future work in this area is likely to involve musculoskeletal models that represent different temporal and spatial scales. The development of software to unite such multiscale models poses additional challenges.

To simulate whole-body movements such as walking or running, motion-capture protocols that accurately describe patients' joint axes, trunk motions, and foot motions are needed, along with ground reaction force data from consecutive strides. Conventional protocols for clinical motion analysis were not designed with the intent of creating simulations, and they could be improved. Developing simulations of movement highlights the limitations of current motion capture data and demonstrates the need for improved experimental protocols.

Muscle-driven simulations generate a wealth of data. Using simulations to elucidate the principles that govern muscle coordination and to achieve improved clinical outcomes, therefore, requires tools that can help reveal insights from these data. Developing and disseminating analysis and visualization tools that provide new insights poses an important challenge for advancing biomechanical simulation. Our goal is to provide a platform on which the biomechanics community can build tools that help uncover the principles that govern human movement and design better treatments for individuals with physical disabilities.

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## References

[1] F. E. Zajac and M. E. Gordon, "Determining muscle's force and action in multi-articular movement," *Exercise Sports Sci. Rev.*, vol. 17, pp. 187–230, 1989.

[2] D. G. Thelen, E. S. Chumanov, M. A. Sherry, and B. C. Heiderscheit, "Neuromusculoskeletal models provide insights into the mechanisms and rehabilitation of hamstring strains," *Exercise Sports Sci. Rev.*, vol. 34, pp. 135–141, 2006.

[3] S. J. Piazza, "Muscle-driven forward dynamic simulations for the study of normal and pathological gait," *J. Neuroeng. Rehab.*, vol. 3, pp. 5–5, 2006.

[4] M. G. Pandy, "Computer modeling and simulation of human movement," *Ann. Rev. Biomed. Eng.*, vol. 3, pp. 245–273, 2001.

[5] F. E. Zajac, "Understanding muscle coordination of the human leg with dynamical simulations," *J. Biomech.*, vol. 35, pp. 1011–1018, 2002.

[6] F. E. Zajac, R. R. Neptune, and S. A. Kautz, "Biomechanics and muscle coordination of human walking—Part II: Lessons from dynamical simulations and clinical implications," *Gait Posture*, vol. 17, pp. 1–17, 2003.

[7] J. W. Fernandez and M. G. Pandy, "Integrating modelling and experiments to assess dynamic musculoskeletal function in humans," *Experimental Physiolog.*, vol. 91, pp. 371–382, 2006.

[8] T. S. Buchanan, D. G. Lloyd, K. Manal, and T. F. Besier, "Neuromusculoskeletal modeling: Estimation of muscle forces and joint moments and movements from measurements of neural command," *J. Appl. Biomech.*, vol. 20, pp. 367–395, 2004.

[9] E. J. Cheng, I. E. Brown, and G. E. Loeb, "Virtual muscle: A computational approach to understanding the effects of muscle properties on motor control," *J. Neurosci. Methods*, vol. 101, pp. 117–130, 2000.

[10] M. M. Forcinito, M. M. Epstein, and W. W. Herzog, "Can a rheological muscle model predict force depression/enhancement?," *J. Biomech.*, vol. 31, pp. 1093–1099, 1998.

[11] L. J. Bhargava, M. G. Pandy, and F. C. Anderson, "A phenomenological model for estimating metabolic energy consumption in muscle contraction," *J. Biomech.*, vol. 37, pp. 81–88, 2004.

[12] B. J. Fregly, Y. Bei, and M. E. Sylvester, "Experimental evaluation of an elastic foundation model to predict contact pressures in knee replacements," *J. Biomech.*, vol. 36, pp. 1659–1668, 2003.

[13] P. G. Kry and D. K. Pai, "Interaction capture and synthesis," *ACM Trans. Graphics*, vol. 25, pp. 872–880, 2006.

[14] B. A. Garner and M. G. Pandy, "The obstacle-set method for representing muscle paths in musculoskeletal models," *Comput. Methods Biomech. Biodmed. Eng.*, vol. 3, pp. 1–30, 2000.

[15] L. L. Menegaldo, D. T. Fleury, and H. I. Weber, "Moment arms and musculotendon lengths estimation for a three-dimensional lower-limb model," *J. Biomech.*, vol. 37, pp. 1447–1453, 2004.

[16] F. Gao, M. Damsgaard, J. Rasmussen, and S. T. Christensen, "Computational method for muscle-path representation in musculoskeletal models," *Biolog. Cybern.*, vol. 87, pp. 199–210, 2002.

[17] S. L. Delp, J. P. Loan, M. G. Hoy, F. E. Zajac, E. L. Topp, and J. M. Rosen, "An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures," *IEEE Trans. Biomed. Eng.*, vol. 37, no. 8, pp. 757–767, Aug. 1990.

[18] S. L. Delp and J. P. Loan, "A graphics-based software system to develop and analyze models of musculoskeletal structures," *Comput. Biol. Med.*, vol. 25, pp. 21–34, 1995.

[19] S. L. Delp and J. P. Loan, "A computational framework for simulating and analyzing human and animal movement," *IEEE Comput. Sci. Eng.*, vol. 2, no. 5, pp. 46–55, Sep./Oct. 2000.

[20] A. N. Vasavada, S. Li, and S. L. Delp, "Influence of muscle morphometry and moment arms on the moment-generating capacity of human neck muscles," *Spine*, vol. 23, pp. 412–422, 1998.

[21] K. R. Holzbaur, W. M. Murray, and S. L. Delp, "A model of the upper extremity for simulating musculoskeletal surgery and analyzing neuromuscular control," *Annals of Biomedical Engineering*, vol. 33, pp. 829–840, 2005.

[22] J. R. Hutchinson, F. C. Anderson, S. S. Blemker, and S. L. Delp, "Analysis of hindlimb muscle moment arms in Tyrannosaurus rex using a three-dimensional musculoskeletal computer model: Implications for stance, gait, and speed," *Paleobiology*, vol. 31, pp. 676–701, 2005.

[23] R. R. Neptune, S. A. Kautz, and F. E. Zajac, "Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking," *J. Biomech.*, vol. 34, pp. 1387–1398, 2001.

[24] I. Jonkers, C. Stewart, and A. Spaepen, "The study of muscle action during single support and swing phase of gait: Clinical relevance of forward simulation techniques," *Gait Posture*, vol. 17, pp. 97–105, 2003.

[25] D. G. Thelen and F. C. Anderson, "Using computed muscle control to generate forward dynamic simulations of human walking from experimental data," *J. Biomech.*, vol. 39, pp. 1107–1115, 2006.

[26] C. C. Raasch and F. E. Zajac, "Locomotor strategy for pedaling: Muscle groups and biomechanical functions," *J. Neurophysiolog.*, vol. 82, pp. 515–525, 1999.

[27] R. R. Neptune and M. L. Hull, "A theoretical analysis of preferred pedaling rate selection in endurance cycling," *J. Biomech.*, vol. 32, pp. 409–415, 1999.

[28] D. G. Thelen, F. C. Anderson, and S. L. Delp, "Generating forward dynamic simulations of movement using computed muscle control," *J. Biomech.*, vol. 36, pp. 321–328, 2003.

[29] D. G. Thelen, E. S. Chumanov, D. M. Hoerth, T. M. Best, S. C. Swanson, L. Li, M. Young, and B. C. Heiderscheit, "Hamstring muscle kinematics during treadmill sprinting," *Medicine Sci. Sports Exercise*, vol. 37, pp. 108–114, 2005.

[30] K. Sasaki and R. R. Neptune, "Muscle mechanical work and elastic energy utilization during walking and running near the preferred gait transition speed," *Gait Posture*, vol. 23, pp. 383–390, 2006.

[31] S. J. Piazza and S. L. Delp, "Three-dimensional dynamic simulation of total knee replacement motion during a step-up task," *J. Biomech. Eng.*, vol. 123, pp. 599–606, 2001.

[32] S. L. Delp and F. E. Zajac, "Force- and moment-generating capacity of lower-extremity muscles before and after tendon lengthening," *Clinical Orthopaed. Related Res.*, vol. 284, pp. 247–259, 1992.

[33] S. L. Delp, K. Statler, and N. C. Carroll, "Preserving plantar flexion strength after surgical treatment for contracture of the triceps surae: A computer simulation study," *J. Orthopaed. Res.*, vol. 13, pp. 96–104, 1995.

[34] A. Erdemir and S. J. Piazza, "Changes in foot loading following plantar fasciotomy: A computer modeling study," *J. Biomech. Eng.*, vol. 126, pp. 237–243, 2004.

[35] S. L. Delp, D. A. Ringwelski, and N. C. Carroll, "Transfer of the rectus femoris: Effects of transfer site on moment arms about the knee and hip," *J. Biomech.*, vol. 27, pp. 1201–1211, 1994.

[36] A. M. Herrmann and S. L. Delp, "Moment arm and force-generating capacity of the extensor carpi ulnaris after transfer to the extensor carpi radialis brevis," *Journal of Hand Surgery—American Volume*, vol. 24, pp. 1083–1090, 1999.

[37] K. R. Saul, W. M. Murray, V. R. Hentz, and S. L. Delp, "Biomechanics of the Steindler flexorplasty surgery: A computer simulation study," *J. Hand Surgery*, vol. 28A, pp. 979–986, 2003.

[38] W. M. Murray, A. M. Bryden, K. L. Kilgore, and M. W. Keith, "The influence of elbow position on the range of motion of the wrist following transfer of the brachioradialis to the extensor carpi radialis brevis tendon," *J. Bone Joint. Surgery*, vol. 84A, pp. 2203–2210, 2002.

[39] S. L. Delp, E. E. Bleck, F. E. Zajac, and G. Bollini, "Biomechanical analysis of the Chiari pelvic osteotomy: Preserving hip abductor strength," *Clinical Orthopaed. Related Res.*, vol. 254, pp. 189–198, 1990.

[40] S. A. Free and S. L. Delp, "Trochanteric transfer in total hip replacement: Effects on the moment arms and force-generating capacities of the hip abductors," *J. Orthopaed. Res.*, vol. 14, pp. 245–250, 1996.

[41] D. J. Schmidt, A. S. Arnold, N. C. Carroll, and S. L. Delp, "Length changes of the hamstrings and adductors resulting from derotational osteotomies of the femur," *J. Orthopaed. Res.*, vol. 17, pp. 279–285, 1999.

[42] S. L. Delp and W. Maloney, "Effects of hip center location on the moment-generating capacity of the muscles," *J. Biomech.*, vol. 26, pp. 485–499, 1993.

[43] S. L. Delp, A. V. Komattu, and R. L. Wixson, "Superior displacement of the hip in total joint replacement: Effects of prosthetic neck length, neck-stem angle, and anteversion angle on the moment-generating capacity of the muscles," *J. Orthopaed. Res.*, vol. 12, pp. 860–870, 1994.
[44] S. L. Delp, R. L. Wixson, A. V. Komattu, and J. H. Kocmond, "How superior placement of the joint center in hip arthroplasty affects the abductor muscles," *Clinical Orthopaed. Related Res.*, vol. 328, pp. 137–146, 1996.
[45] A. S. Arnold, D. J. Asakawa, and S. L. Delp, "Do the hamstrings and adductors contribute to excessive internal rotation of the hip in persons with cerebral palsy?," *Gait Posture*, vol. 11, pp. 181–190, 2000.
[46] A. S. Arnold and S. L. Delp, "Rotational moment arms of the medial hamstrings and adductors vary with femoral geometry and limb position: Implications for the treatment of internally-rotated gait," *J. Biomech.*, vol. 34, pp. 437–447, 2001.
[47] S. L. Delp, A. S. Arnold, R. A. Speers, and C. A. Moore, "Hamstrings and psoas lengths during normal and crouch gait: Implications for muscle-tendon surgery," *J. Orthopaed. Res.*, vol. 14, pp. 144–151, 1996.
[48] S. A. Kimmel and M. H. Schwartz, "A baseline of dynamic muscle function during gait," *Gait Posture*, vol. 23, pp. 211–221, 2006.
[49] L. M. Schutte, S. W. Hayden, and J. R. Gage, "Lengths of hamstrings and psoas muscles during crouch gait: Effects of femoral anteversion," *J. Orthopaed. Res.*, vol. 15, pp. 615–621, 1997.
[50] A. S. Arnold, M. Q. Liu, M. H. Schwartz, S. Ounpuu, and S. L. Delp, "The role of estimating muscle-tendon lengths and velocities of the hamstrings in the evaluation and treatment of crouch gait," *Gait Posture*, vol. 23, pp. 273–281, 2006.
[51] A. S. Arnold, M. Q. Liu, M. H. Schwartz, S. Ounpuu, L. S. Dias, and S. L. Delp, "Do the hamstrings operate at increased muscle-tendon lengths and velocities after surgical lengthening?," *J. Biomech.*, vol. 39, pp. 1498–1506, 2006.
[52] I. Jonkers, C. Stewart, K. Desloovere, G. Molenaers, and A. Spaepen, "Musculo-tendon length and lengthening velocity of rectus femoris in stiff knee gait," *Gait Posture*, vol. 23, pp. 222–229, 2006.
[53] A. J. Wilkenfeld, M. L. Audu, and R. J. Triolo, "Feasibility of functional electrical stimulation for control of seated posture after spinal cord injury: A simulation study," *J. Rehab. Res. Dev.*, vol. 43, pp. 139–152, 2006.
[54] R. F. Kirsch, A. M. Acosta, F. C. van der Helm, R. J. Rotteveel, and L. A. Cash, "Model-based development of neuroprostheses for restoring proximal arm function," *J. Rehab. Res. Dev.*, vol. 38, pp. 619–626, 2001.
[55] R. Davoodi, I. E. Brown, and G. E. Loeb, "Advanced modeling environment for developing and testing FES control systems," *Med. Eng. Phys.*, vol. 25, pp. 3–9, 2003.
[56] L. M. Schutte, M. M. Rodgers, and F. E. Zajac, "Improving the efficacy of electrical stimulation-induced leg cycle ergometry: An analysis based on a dynamic musculoskeletal model," *IEEE Trans. Rehab. Eng.*, vol. 1, no. 2, pp. 109–125, Jun. 1993.
[57] R. R. Neptune and S. A. Kautz, "Knee joint loading in forward versus backward pedaling: Implications for rehabilitation strategies," *Clinical Biomech.*, vol. 15, pp. 528–535, 2000.
[58] T. F. Besier, G. E. Gold, G. S. Beaupre, and S. L. Delp, "A modeling framework to estimate patellofemoral joint cartilage stress in vivo," *Medicine Sci. Sport Exercise*, vol. 37, pp. 1924–1930, 2005.
[59] D. G. Lloyd and T. F. Besier, "An EMG-driven musculoskeletal model to estimate muscle forces and knee joint moments in vivo," *J. Biomech.*, vol. 36, pp. 765–776, 2003.
[60] S. McLean, X. Huang, A. Su, and A. J. Van den Bogert, "Sagittal plane biomechanics cannot injure the ACL during sidestep cutting," *Clinical Biomech.*, vol. 19, pp. 828–838, 2004.
[61] I. C. Wright, R. R. Neptune, A. J. van den Bogert, and B. M. Nigg, "The effects of ankle compliance and flexibility on ankle sprains," *Medicine Sci. Sport Exercise*, vol. 32, pp. 260–265, 2000.
[62] I. C. Wright, R. R. Neptune, A. J. van den Bogert, and B. M. Nigg, "The influence of foot positioning on ankle sprains," *J. Biomech.*, vol. 33, pp. 513–519, 2000.
[63] S. J. Piazza and S. L. Delp, "The influence of muscles on knee flexion during the swing phase of gait," *J. Biomech.*, vol. 29, pp. 723–733, 1996.
[64] D. C. Kerrigan, R. S. Roth, and P. O. Riley, "The modelling of adult spastic paretic stiff-legged gait swing period based on actual kinematic data," *Gait Posture*, vol. 7, pp. 117–124, 1998.
[65] J. S. Higginson, F. E. Zajac, R. R. Neptune, S. A. Kautz, and S. L. Delp, "Muscle contributions to support during gait in an individual with post-stroke hemiparesis," *J. Biomech.*, vol. 39, pp. 1769–1777, 2006.
[66] W. J. Kargo, F. Nelson, and L. C. Rome, "Jumping in frogs: Assessing the design of the skeletal system by anatomically realistic modeling and forward dynamic simulation," *J. Experiment. Biol.*, vol. 205, pp. 1683–1702, 2002.
[67] W. J. Kargo and L. C. Rome, "Functional morphology of proximal hindlimb muscles in the frog Rana pipiens," *J. Experiment. Biol.*, vol. 205, pp. 1987–2004, 2002.
[68] R. J. Full and A. N. Ahn, "Static forces and moments generated in the insect leg: Comparison of a three-dimensional musculo-skeletal computer model with experimental measurements," *J. Experiment. Biol.*, vol. 198, pp. 1285–1298, 1995.
[69] R. R. Neptune and M. L. Hull, "Evaluation of performance criteria for simulation of submaximal steady-state cycling using a forward dynamic model," *J. Biomech. Eng.*, vol. 120, pp. 334–341, 1998.
[70] F. C. Anderson and M. G. Pandy, "Dynamic optimization of human walking," *J. Biomech. Eng.*, vol. 123, pp. 381–390, 2001.
[71] A. Seth and M. G. Pandy, "A neuromusculoskeletal tracking method for estimating individual muscle forces in human movement," *J. Biomech.*, vol. 40, pp. 356–366, 2007.
[72] T. W. Lu and J. J. O'Connor, "Bone position estimation from skin marker coordinates using global optimization with joint constraints," *J. Biomech.*, vol. 32, pp. 129–134, 1999.
[73] J. Perry, "Distal rectus femoris transfer," *Develop. Med. Child Neurolog.*, vol. 29, pp. 153–158, 1987.
[74] S. R. Goldberg, F. C. Anderson, M. G. Pandy, and S. L. Delp, "Muscles that influence knee flexion velocity in double support: Implications for stiff-knee gait," *J. Biomech.*, vol. 37, pp. 1189–1196, 2004.
[75] J. R. Gage, J. Perry, R. R. Hicks, S. Koop, and J. R. Werntz, "Rectus femoris transfer to improve knee function of children with cerebral palsy," *Develop. Med. Child Neurolog.*, vol. 29, pp. 159–166, 1987.
[76] T. M. Kepple, K. L. Siegel, and S. J. Stanhope, "Relative contributions of the lower extremity joint moments to forward progression and support during gait," *Gait Posture*, vol. 6, pp. 1–8, 1997.
[77] M. deZee, L. Hansen, C. Wong, J. Rasmussen, and E. B. Simonsen, "A generic detailed rigid-body lumbar spine model," *J. Biomech.*, vol. 40, pp. 1219–1227, 2006.
[78] P. J. Hunter and T. K. Borg, "Integration from proteins to organs: The Physiome Project," *Nature Rev. Molec. Cell Biol.*, vol. 4, pp. 237–243, 2003.
[79] S. Van Sint Jan, "Introducing anatomical and physiological accuracy in computerized anthropometry for increasing the clinical usefulness of modeling systems," *Critical Rev. Phys. Rehab. Med.*, vol. 17, pp. 249–274, 2005.

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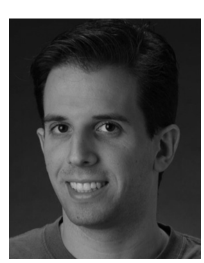

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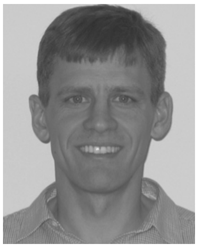

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