Concurrent Prediction of Muscle and Tibiofemoral Contact Forces During Treadmill Gait

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Detailed knowledge of knee kinematics and dynamic loading is essential for improving the design and outcomes of surgical procedures, tissue engineering applications, prosthetics design, and rehabilitation. This study used publicly available data provided by the “Grand Challenge Competition to Predict in-vivo Knee Loads” for the 2013 American Society of Mechanical Engineers Summer Bioengineering Conference (Fregly et al., 2012, “Grand Challenge Competition to Predict in vivo Knee Loads,” J. Orthop. Res., 30, pp. 503–513) to develop a full body, musculoskeletal model with subject specific right leg geometries that can concurrently predict muscle forces, ligament forces, and knee and ground contact forces. The model includes representation of foot/floor interactions and predicted tibiofemoral joint loads were compared to measured tibial loads for two different cycles of treadmill gait. The model used anthropometric data (height and weight) to scale the joint center locations and mass properties of a generic model and then used subject bone geometries to more accurately position the hip and ankle. The musculoskeletal model included 44 muscles on the right leg, and subject specific geometries were used to create a 12 degrees-of-freedom anatomical right knee that included both patellofemoral and tibiofemoral articulations. Tibiofemoral motion was constrained by deformable contacts defined between the tibial insert and femoral component geometries and by ligaments. Patellofemoral motion was constrained by contact between the patellar button and femoral component geometries and the patellar tendon. Shoe geometries were added to the feet, and shoe motion was constrained by contact between three shoe segments per foot and the treadmill surface. Six-axis springs constrained motion between the feet and shoe segments. Experimental motion capture data provided input to an inverse kinematics stage, and the final forward dynamics simulations tracked joint angle errors for the left leg and upper body and tracked muscle length errors for the right leg. The one cycle RMS errors between the predicted and measured tibia contact were 178 N and 168 N for the medial and lateral sides for the first gait cycle and 209 N and 228 N for the medial and lateral sides for the faster second gait cycle. One cycle RMS errors between predicted and measured ground reaction forces were 12 N, 13 N, and 65 N in the anterior-posterior, medial-lateral, and vertical directions for the first gait cycle and 43 N, 15 N, and 96 N in the anterior-posterior, medial-lateral, and vertical directions for the second gait cycle. [DOI: 10.1115/1.4026359]

1 Introduction

The knee joint is at the center of a kinetic chain that includes the foot, ankle joint, lower and upper leg, hip joint, and pelvis. The complexity of the knee, the fact that it bears most of the body's weight, and its centrality to human movement make the knee vulnerable to acute and chronic injuries. Contact loading is a contributing factor in the onset and progression of osteoarthritis [1], and it is equally important in artificial knee replacement wear [2,3], which leads to failure of total knee prostheses. Knowledge of in vivo joint loads and motions during activities of daily living is also crucial in tissue engineering, injury prevention, and in developing, evaluating, and optimizing therapeutic strategies. Instrumented knee prostheses capable of measuring joint loading during dynamic activities have been implanted in patients [4,5–7]. But implementing these devices is expensive, invasive, and the results may not necessarily transfer to other patients and healthy adults. Thus, in vivo joint loads must be predicted using computational models.

Estimation of the in vivo knee loads under dynamic conditions is highly dependent on correct prediction of muscle forces. This

requires resolution of the muscle redundancy problem, which occurs because the muscle force solutions to create a certain motion are not unique. A number of studies addressed the muscle redundancy problem by using optimization methods to converge to a set of muscle forces that recreate the kinematics [8–14]. This approach involves two stages. In the first stage the muscle and net joint torques are estimated by inverse dynamics (static optimization) or by forward dynamic optimization. Static optimization uses the ground reaction forces as an input and solves the dynamics equations at each time frame. Muscle forces are predicted at each time step such that the kinematics are consistent with the experimentally measured motions. In dynamic optimization, the dynamics equations are numerically integrated across all time frames; thus, the muscle forces and kinematics are predicted concurrently. Typically, articular contact models are not included in the first stage, and the computed forces are applied to an articular contact model of the knee to estimate the joint contact forces. There are several limitations in this approach. Because articular contacts are not included in the first stage and the knee is often modeled as a one degree-of-freedom idealized joint, the contact forces do not affect the predicted muscle forces. Implicitly, the assumption is that muscle forces do not contribute to motions outside of knee flexion-extension. In a study by Lin et al. [14], it was demonstrated that muscle forces are affected by the number of constraints included in the knee model. In the case of static optimization, the accuracy of the available motion and ground

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reaction force data influences the muscle force predictions [9]. Moreover, since the ground reaction forces are used as an input, they cannot be used to evaluate the performance of the model. Finally, optimization methods assume that human movement is bound by minimizing some cost function such as sum of squares of muscle activation or metabolic energy, but it is unknown if the nervous system produces motion in such a way.

Another major challenge in estimating the in vivo joint loads is the correct resolution of the tibiofemoral contact force into the medial and lateral compartment contact forces. The two regions of contact between the femur and the tibia create a contact force redundancy issue, as well [4]. Most musculoskeletal modeling studies tend to overestimate the tibial contact forces during gait [2,15–18] when compared to in vivo measurements collected by instrumented implants [5–7,19]. The discrepancies between model predictions and measured joint loads are even greater when the medial to lateral compartment load distributions are considered. The distribution of the tibiofemoral contact forces depends on the magnitude of the moments acting about the knee joint and the individual contributions of the muscles and ligaments [15,18].

In this paper, we present a two stage modeling method to predict tibiofemoral contact loads during treadmill walking. Our work is distinguished from previous published studies in a number of ways. In the first stage of our method, kinematic data is the only input into the multibody model while the knee joint (including both the tibiofemoral and patellofemoral joints) is allowed 12 degrees-of-freedom. The tibiofemoral and patellofemoral joints are constrained by contact between articulating surfaces as well as ligament forces. In the second stage, muscle forces and joint torques are predicted by using a proportional integral derivative (PID) controller, and the contact forces are predicted concurrently. Moreover, the ground reaction forces, as well as electromyography (EMG) and segment kinematics are used to validate the performance of the forward simulation. Because the ground reaction forces are never used as an input to the model, a detailed model of the foot-shoe-floor is created.

Our specific aims are (a) to develop a full body, musculoskeletal model with subject specific lower extremity geometries that includes a detailed model of the foot/floor interactions that can concurrently predict muscle, ligament, and contact forces and (b) to evaluate the tibiofemoral joint load predictions with measured tibial loads for two different cycles of treadmill gait.

2 Methods

2.1 Experimental Data. The publicly available data provided by the “Grand Challenge Competition to Predict In-Vivo Knee Loads” for the 2013 American Society of Mechanical Engineers Summer Bioengineering Conference [4] were used to develop the model for this project. The experimental data were collected on one subject (166 cm height, 64.6 kg body weight) with an instrumented tibial tray (Generation I, eKnee) that calculated in vivo medial and lateral tibia contact forces from four load cells embedded in the tibial tray [19]. The provided data included motion capture, ground reaction forces, surface electromyography, and computed tomography scans. In addition, geometries of the subject’s pelvis and right femur, patella, tibia, fibula, talus, calcaneus, midfoot, metatarsals, and phalanges were provided along with the prosthetic implant geometries (femoral component, patellar button, tibial tray, and tibial insert).

Two gait cycles were simulated, accelerating and fast. Both cycles come from the treadmill gait trial “jw_ngait_tm_transition.” The accelerating cycle begins at 4.86 s into the trial. During the accelerating gait cycle the treadmill speed starts at 800 mm/s and then accelerates at 60 mm/s². The fast cycle begins 18.75 s into the trial, and the treadmill speed stays constant at 1400 mm/s throughout the gait cycle.

2.2 Musculoskeletal Model. A generic musculoskeletal model was created in the commercially available multibody

dynamic analysis program ADAMS (MSC Software Corporation, Santa Ana, CA) and was comprised of 21 rigid body segments, 44 muscles, and 53 revolute joints. The generic model was scaled based on height, weight, and gender. Subject specific geometries of the right leg replaced the geometries of the generic model, the right knee revolute joints were removed, and a 12 degrees-of-freedom anatomical right knee that included both patellofemoral and tibiofemoral articulations was created. Tibiofemoral motion was constrained by deformable contacts defined between the tibial insert and femoral component geometries and by ligaments. Patellofemoral motion was constrained by contact between the patellar button and femoral component geometries and the patellar tendon. The quadriceps muscle insertions were modified to insert on the patella. Shoe geometries were added to the feet, and shoe motion was constrained by contact between three shoe segments per foot and the treadmill surface. Six-axis springs constrained motion between the feet and shoe segments. Experimental motion capture data provided input to the first inverse kinematics stage of the model and the final forward dynamics simulations tracked joint angle errors for the left leg and upper body and tracked muscle length errors for the right leg.

Joint center locations and generic bones were linearly scaled, based on subject height, from a generic skeletal model and placed within ADAMS. Height, weight, and gender scaled mass properties were added to the model as well as height scaled muscle origin and insertion locations and via points for the right leg [20]. Subject specific bone geometries of the right lower extremity were then placed in the model, and the joint locations of the right hip and ankle were further refined. Reference bone geometries of the left leg were created by mirroring right leg geometries, and joint center locations of the left leg hip, knee, and ankle were refined based on motion capture marker locations as well as the mirrored geometries. Three orthogonal revolute joints were defined at each joint center location (Table 1). Exceptions

Table 1 Joints used in the musculoskeletal model. Each joint location has three orthogonal revolute joints with the following exceptions. The metatarsophalangeal (toe) joints between the toes and foot body segments were comprised of a single revolute joint. The joint between the head and neck (head joint) was a rigid joint, and the right knee was comprised of tibiofemoral and patellofemoral joints that were constrained by contact between articulating surfaces and ligaments. Active joints were actuated by a torque generator during forward dynamics simulations. Passive joints had no active torque generators and joint motion resulted from muscle forces (with the exception of the rigid head joint).

Joint location	DOF	Body segments connected	Active (A) or passive (P)
Right Toe	1	Right: Toes, Foot	P
Right Ankle	3	Right: Foot, Shank	P
Right Knee	12	Right: Shank, Thigh, Patella	P
Right Hip	3	Right Thigh, Lower Torso	P
Left Toe	1	Left: Toes, Foot	A
Left Ankle	3	Left: Foot, Shank	A
Left Knee	3	Left: Shank, Thigh	A
Left Hip	3	Left Thigh, Lower Torso	A
Lumbar	3	Lower Torso, Central Torso	A
Thoracic	3	Central Torso, Upper Torso	A
Neck	3	Upper Torso, Neck	A
Head	0	Neck, Head	P
Right Scapula	3	Right: Upper Torso, Scapula	A
Right Shoulder	3	Right: Scapula, Upper Arm	A
Right Elbow	3	Right: Upper Arm, Lower Arm	A
Right Wrist	3	Right: Lower Arm, Hand	A
Left Scapula	3	Left: Upper Torso, Scapula	A
Left Shoulder	3	Left: Scapula, Upper Arm	A
Left Elbow	3	Left: Upper Arm, Lower Arm	A
Left Wrist	3	Left: Lower Arm, Hand	A

included the metatarsophalangeal joints between the foot and toes where only a single revolute joint was defined, and the joint between the head and neck was made rigid because head motion was not measured. To create three orthogonal revolute joints between two adjoining body segments, two massless dummy parts were created at each joint center. A torsional spring with damper was defined about the axis of each revolute joint. The torsional stiffness for each joint spring was 100 N-mm/deg, and the damping coefficient was 20 N-mm/(deg/s) with the exception of the passive torsional spring about the adduction/abduction axis of the hip. Based on hip kinematic errors, the stiffness of this torsional spring was increased to 800 N-mm/deg. The joint spring-dampers prevented high-frequency rotations about the joint during the motion capture driven inverse kinematics simulation. In addition, the torsional springs allowed higher gains to be set for the joint torque feedback controllers during the forward dynamics simulations. For joints that were muscle driven (passive) and not torque driven (active), the joint spring damper can also simulate soft tissue constraints. Motion capture marker locations for the first frame of the static trial with feet pointed forward (jw_staticfor) were imported to the model. The body was posed through rotation of joint axes to match the marker locations, thus, placing the model in the static trial position.

The model included 44 muscles on the right leg. Length measures were added to record the total length from origin to insertion through all via points of each muscle. Measures were also created to measure the angle about each hinge joint. Point-to-point force elements were added between origin/insertions and via points for all muscles and a torque generator was created about the axis of each joint for the upper body and left leg. During inverse kinematics simulations, the magnitude of all muscle force and joint torque generators was zero. For the forward dynamics simulations, the muscle force and joint torque magnitudes came from a feedback control scheme that tracked inverse kinematics measured muscle length or joint angle.

2.3 Knee Model. After posing the body in the static trial position, the three hinge joints for the right knee were removed and an anatomical right knee was created. The anatomical right knee included deformable contacts defined between the tibial insert and femoral component geometries and between the patellar button and femoral component (Fig. 1). The tibial insert was divided into medial and lateral geometries with separate contacts created for each. The deformable contact with viscous damping [21] was defined as

$$F_c = k_c \delta^n + B_c(\delta) \dot{\delta} \quad (1)$$

Fig. 1 Sagittal plane (a) and (c) and frontal plane (b) view of the right knee. Also shown are the contact patches on the medial and lateral side of the tibial insert during a forward dynamics simulation (d). The small spheres denote the location of muscle via points. The large spheres denote the location of motion capture markers associated with the body segments.

where F_c is the contact force, δ is the interpenetration of geometries, $\dot{\delta}$ is the velocity of interpenetration, k_c is the contact stiffness, n is the nonlinear power exponent, and $B_c(\delta)$ is a damping coefficient. The damping coefficient $B_c(\delta)$ is defined as

$$B_c(\delta) = \begin{cases} 0 & \delta \leq 0 \\ B_{\max} \left(\frac{\delta}{d_{\max} - \delta} \right)^2 \left(3 - \frac{2\delta}{d_{\max} - \delta} \right) & 0 < \delta < d_{\max} \\ B_{\max} & \delta > d_{\max} \end{cases} \quad (2)$$

where $d_{\max}$ is the penetration at which the maximum damping $B_{\max}$ value is applied. All geometries used in contacts were converted to the Parasolid geometric modeling kernel, creating a three-dimensional solid with continuous representation. For the tibiofemoral and patellofemoral contact, k_c was 30,000 N/mm^{3/2} and n was 3/2 [22]. In addition, $B_{\max}$ was set to 40 N/(mm/s) and $d_{\max}$ was 0.1 mm.

The quadriceps muscle insertions on the right leg were modified to attach to the patella bone. The collateral ligaments as well as the posterior bundle of the posterior cruciate ligament were modeled in the right knee. Ligaments were represented using a nonlinear force-displacement relationship to capture the toe-region and to account for the fact that ligaments are only loaded in tension. The model included three bundles for the lateral collateral ligaments (LCL), three bundles for the medial collateral ligaments (MCL), and one bundle for the posterior cruciate ligament (PCL). The anterior cruciate ligament was not included, as this ligament was removed during knee implant surgery. Ligament bundles were represented in the model as a single force element with a nonlinear force-strain curve

$$f = \begin{cases} \frac{1}{4} k \varepsilon^2 / \varepsilon_l & 0 \leq \varepsilon \leq 2\varepsilon_l \\ k(\varepsilon - \varepsilon_l) & \varepsilon > 2\varepsilon_l \\ 0 & \varepsilon < 0 \end{cases} \quad (3)$$

$$\varepsilon = \left(\frac{l - l_0}{l_0} \right) \quad (4)$$

where k is a stiffness parameter and ε is defined as the engineering strain. The spring parameter ε_l is a constant value, and it was assumed to be 0.03 [23,24]. The ligament stiffness parameter k was 2000 N for all bundles of the LCL, 2750 N for all bundles of the MCL, and 2000 N for the single bundle of the PCL. The zero-load lengths l_0 and ligament insertions and origins were estimated based on previous cadaveric studies performed by Guess et al. [25–27]. Each force element also included a parallel damper with a small damping coefficient value of 0.5 Ns/mm to remove the possibility of high frequency vibrations during simulation [26]. A custom ADAMS subroutine was written to implement Eqs. (3) and (4) in the model. Inputs to the subroutine included the ligament stiffness and strain, damping coefficient, the length of each ligament in the position it was constructed, and the zero-load length.

2.4 Foot Ground Contact Model. Deformable contacts between treadmill force plate geometries and shoe geometries were used to model the shoe-ground interface. Geometries for the shoe were estimated from the provided shoe size as well as images of the shoe type available on the Internet. The shoe geometry was then separated into three rigid bodies: (1) a shoe rigid body consisting of the heel and midfoot regions, (2) a shoe toes rigid body consisting of the region distal to the metatarsophalangeal joint and proximal to the shoe tip, and (3) a shoe tip (Fig. 2). The shoe tip region was determined from visual inspection of shoe bending during gait. The shoe was connected to the foot body segment

Fig. 2 Shoe Model. The shoe geometry is separated into three rigid bodies: the shoe, shoe toes, and shoe tip. The shoe is connected to the foot via a six-axis spring, the shoe toes are connected to the toes via a six-axis spring, and the shoe tip is connected to the shoe toes through a six-axis spring. The foot and toe body segments are connected through a revolute joint (metatarsophalangeal joint). Deformable contacts are defined between the three shoe parts and the treadmill force plate.

Table 2 Spring constants for the shoe model

Six-axis spring connection	Translation (N/mm)			Rotation (N-mm/deg)		
	Med-Lat	Ant-Post	Vertical	Med-Lat	Ant-Post	Vertical
Shoe to Foot body segment	100	10	1000	2000	2000	2000
Shoe toes to Toes body segment	200	100	1000	2000	2000	2000
Shoe toes to Shoe tip	100	100	100	250	1000	1000

through a six-axis spring. The shoe toes rigid body was connected to the toes body segment through a six-axis spring, and the shoe tip was connected to the shoe toes through a six-axis spring. Stiffness values for the springs used in the shoe model are given in Table 2. The stiffness values were chosen based on comparisons of predicted ground reaction forces with measured ground reaction forces. The six-axis springs between the shoe geometries and the body segments model the compliance of the shoe as well as relative motion between bone and shoe sole. Contacts were defined using Eq. (1), between the shoe, shoes toes, and shoe tip and treadmill force plates. For these contacts, k_c was $50 \text{ N/mm}^{3/2}$ and n was $3/2$; $B_{\max}$ was set to 0.5 N/(mm/s) and $d_{\max}$ was 0.1 mm . Hertzian contact theory was used to determine contact parameters

using rubber as the contacting material. A coulomb friction was also defined at shoe ground contact with a static friction coefficient of 0.8 and a dynamic friction coefficient of 0.9 [28]. The shoe model was the same for both the right and left feet.

The left and right treadmill force plate surfaces were modeled as simple boxes ($500 \text{ cm} \times 2 \text{ cm} \times 50 \text{ cm}$). A velocity based motion constraint was defined between each force plate and ground. The treadmill speed during each gait trial was determined by taking the average velocity of each marker on the foot in the direction of treadmill motion (anterior/posterior direction). The average marker velocity for each foot was minimum when the entire shoe surface was in contact with the treadmill.

2.5 Inverse Kinematics Simulation. At each motion capture marker position, two spheres were created. One sphere is associated with the body segments while the second colocated sphere is a separate part (motion sphere). Each motion sphere and body segment sphere is connected via a three-axis spring. A motion constraint is applied to the motion sphere using experimentally measured motion data. The model simulations included two stages, an inverse kinematics simulation and a forward dynamics simulation. In the inverse kinematics simulation, the model is moved from the initial build position to the body position just prior to the start of either the accelerating or fast gait cycle and then run through 3 s of gait. To allow the model time to reach the forward dynamics starting position and for the dynamic model to reach equilibrium, the experimental motion data is modified such that the initial position is held constant for 3 s. After three seconds, the gait trial begins. The total length of the inverse kinematics simulations were 6 s. The forces generated in the three-axis springs connecting the motion sphere to the rigid body segment sphere moved the model from the initial position to the starting position as well as through the gait cycle. During the inverse kinematics simulation, the model is constrained by the joints, contacts, and ligaments at the knee, and by contacts between the shoe parts and treadmill. The angle of each joint axis as well as the total length of each muscle is recorded during the inverse kinematics simulation. In addition, the rigid body motion of the pelvis is recorded.

2.6 Forward Dynamics Simulations. The model is then saved in the starting position of either the accelerating or fast gait trial and a 3 s forward dynamics simulation run (Fig. 3). In the forward dynamics simulation, the motion constraints of the motion spheres are removed and the model is driven by muscle forces and by joint torques. The force and torque magnitudes are calculated via feedback controllers implemented in Simulink (The

Fig. 3 Musculoskeletal simulation of treadmill gait with a right total knee replacement. The right leg is driven by muscle forces and the upper body and left leg are driven by joint torques. The arrows represent forces (contact, muscle, ligament, spring) calculated during the forward dynamics simulation.

Fig. 4 Feedback control scheme for calculating muscle force. The Simulink “saturation” limits muscle forces such that a muscle can only pull and not push.

MathWorks, Inc., Natick, MA). During the forward dynamics simulations, ADAMS and Simulink are linked in cosimulation where ADAMS sends the current muscle lengths and joint angles to Simulink and then Simulink sends muscle forces and torques to ADAMS for the next calculation step. The error signal between the current forward dynamics joint angle and the joint angle measured during the inverse kinematics simulation is fed into a proportional integral derivative feedback control scheme. The output of the PID controller is the torque required to track the inverse kinematics joint angle during the forward dynamics simulation.

Similarly, a PID controller was used to generate muscles forces that track the inverse kinematics muscle length (Fig. 4). The muscle force is limited such that the muscles can only pull and cannot push. In addition, the PID parameters for each individual muscle are scaled based on the following:

$$P^i = \frac{PCSA^i}{\text{Reference } PCSA} * \text{Global } P \quad i = 1, 2, \dots, \# \text{ of muscles} \quad (5)$$

where P^i is the proportional gain for muscle i , $PCSA^i$ is the physiological cross-sectional area of muscle i , the Reference $PCSA = 1500 \text{ mm}^2$, and Global P is a global proportional gain that is scaled by Eq. (5) for each muscle. Similar equations exist for the integral and derivative gains. Muscles with a $PCSA$ less than the reference $PCSA$ will have smaller PID gains while larger muscles will have larger PID gains. The $PCSA$ values for each muscle originate from the work by Schumacher and Wolff [29–31] and the Reference $PCSA$ is the average of all muscles.

Three combinations of Global PID values were evaluated. The three Global PID values for the muscle controllers were: (1) $P = 300$, $I = 30$, and $D = 0.003$, (2) $P = 100$, $I = 10$, and $D = 0.001$, and (3) $P = 50$, $I = 5$, and $D = 0.0005$. Tibiafemoral contact predictions were used to evaluate the three Global PID combinations.

2.7 Stabilizer. During the inverse kinematics simulations, rigid body motion of the pelvis was recorded. During forward dynamics simulations, a motion constraint using this pelvis motion was applied to a dummy part. A constraint was not applied in the vertical direction and the dummy part was free to move in this direction. A six-axis spring (100 N/mm axial stiffness, torsional stiffness 1000 N-mm/deg) with damper connected the dummy part to the pelvis. This stabilizer prevented the model from tipping over due to accumulated position errors during the walk cycle. During the forward dynamics simulations, the maximum spring force for either the medial-lateral or anterior-posterior directions did not exceed 175 N during the gait cycle. The average force was 12 N in the anterior-posterior direction and 5 N in the med-lateral direction.

3 Results

Increasing the global gains of the muscle force PID controller increased the normalized force prediction of the hamstrings and gastrocnemius muscles during the forward dynamics simulation (Fig. 5). Both the semimembranosus and biceps femoris hamstring muscles had increased force magnitude as the PID gains increased and the biceps femoris had increased activity levels for the highest PID gains ($P = 300$, $I = 30$, and $D = 0.003$, PID = 300). The

Fig. 5 Normalized muscle forces during the forward dynamics simulation of the accelerating gait cycle for the global PID values: PID 50 ($P = 50$, $I = 5$, and $D = 0.0005$), PID 100 ($P = 100$, $I = 10$, and $D = 0.001$), and PID 300 ($P = 300$, $I = 30$, and $D = 0.003$). Muscle forces are normalized to the maximum force that can be produced by each muscle as determined by its $PCSA$.

medial gastrocnemius muscles also had increased force magnitude as PID gains increased. The global gains of the muscle force PID controller changed the contact force predictions with the higher gains creating greater contact force (Fig. 6). The peaks in contact force correspond to peaks in the semimembranosus muscle. The lowest PID parameters of $P = 50$, $I = 5$, and $D = 0.0005$ provided the best contact predictions, and these values were used for all subsequent simulation results.

The predicted medial and lateral tibia contact forces for both the accelerating and fast trials track measured eKnee forces well with the exception of peaks in the predicted forces at heel strike (Fig. 7). The one cycle RMS errors between the predicted and eKnee tibia contact values are less than 180 N for the accelerating cycle and less than 230 N for the fast cycle (Table 3). The vertical and shear right foot ground reaction forces follow measured forces well for the accelerating gait cycle with some deviation in predicted vertical and anterior-posterior shear reaction forces for the fast gait cycle (Fig. 8). The one cycle ground reaction force RMS errors were less than 14 N for the anterior-posterior and medial-lateral shear directions and less than 66 N in the vertical direction for the accelerating gait cycle (Table 4). Right leg hip and ankle kinematics for both the inverse kinematics and forward dynamics simulations are shown in Fig. 9. The hip has a greater forward dynamics external rotation in the stance phase for both the accelerating and fast trials, with the difference being more pronounced in the fast cycle. At heel strike, the forward dynamics simulation has greater ankle plantar flexion than the motion driven inverse kinematics simulation, and the foot has more internal rotation during stance with the difference being greater for fast. The timing of

Fig. 6 Measured and predicted lateral (a) and medial (b) contact forces over the accelerating gait cycle for global PID values of 50, 100, and 300

Fig. 7 Measured eKnee and model predictions for the accelerating gait cycle (lateral (a), medial (b)) and fast gait cycle (lateral (c), medial (d)). Global PID values equal 50.

predicted muscle force generally follows the measured muscle excitations (EMG) for both the accelerating and fast gait cycles (Fig. 10). Experimentally, there is muscle activation of the vastus lateralis prior to and during heel strike. In the forward dynamics simulation, the vastus lateralis does not produce force until after initial heel contact, and these forces are small compared to the maximum force that can be generated by the muscle. Experimentally, muscle activations are greater for the muscles crossing the knee for the fast cycle and the rectus femoris has greater activation in the first 20% of the cycle. In the simulation, normalized muscle forces are greater for the fast gait cycle, including the vastus lateralis and the rectus femoris. Also, the rectus femoris produces force just after heel strike in the fast simulation but not in the accelerating cycle.

4 Discussions

The developed model includes simultaneous prediction of muscle forces, medial and lateral knee contact forces, and ground reaction forces during treadmill gait. The model is also capable of predicting contact locations on the tibial, femoral, and patellar components during gait as well as knee ligament forces. Identical models were used for both the accelerating and fast simulations and the only inputs to the simulations were motion capture data, leaving other experimental measures for indirect or direct model validation. The computation time for the inverse kinematics and forward dynamics simulations was under 20 min on a desktop PC (Windows 7, 2.8 GHz Intel Xeon CPU, 24 GB RAM), and predicted contact forces (tibiofemoral and ground reaction) compared favorably to experimental measurements.

Most previous computational musculoskeletal model studies have overestimated the total tibiofemoral contact forces during gait. The predicted contact forces have ranged from 1.8 to 8.1 times body weight (BW) while in vivo measurements made by instrumented implants have ranged from 1.8 to 3.0 BW [4]. The

Table 3 RMS error (N) for lateral and medial contact force predictions

	Lateral	Medial
Accelerating	178.2	168.2
Fast	208.5	227.6

distribution of the contact forces between the medial and lateral compartments has also been overestimated on the medial side. Typically the knee joint is constrained by a single one degree-of-freedom (flexion-extension) revolute joint when estimating muscle forces. This assumption removes any influence of contact forces on the muscles as well as muscle contributions to motion beyond the sagittal plane. In this study, during both the inverse kinematics and forward dynamics simulations, the knee was constrained by ligament forces and tibiofemoral and patellofemoral contact forces allowing 12 degrees-of-freedom.

The foot/shoe/floor contact model used during body level musculoskeletal simulation is more elaborate than what has been previously published allowing more accurate estimation of the ground reaction forces as well as a more accurate distribution of those forces on the bottom of the shoe during gait. The resulting foot/shoe/floor contact model is similar to the bed of springs model of Neptune et al. [32], but it includes multiple rigid shoe segments, motion between the foot and shoe, and deformable contact between shoe and ground geometries. In the absence of experimental knee contact forces, comparisons of the ground reaction forces may provide indirect validation of joint force predictions. In this study, the accelerating gait cycle had smaller ground reaction force RMS errors and smaller tibia contact force errors compared to the fast gait cycle. Accurately modeling the foot-ground interface is important for predicting joint contact forces. Errors in the prediction of vertical ground reaction forces, especially for the

Table 4 RMS error (N) for ground reaction force predictions

	Anterior-Posterior	Medial-Lateral	Vertical
Accelerating	12.1	13.3	65.5
Fast	42.9	14.8	96.1

fast cycle and errors in ankle flexion at heel strike demonstrate that the foot/shoe/floor contact model can be improved. Experimental measurement of foot/shoe motion and compliance is recommended for future studies to determine the degrees-of-freedom of the foot and shoe during gait. In addition, mechanical tests that can determine foot/ground contact and shoe stiffness parameters are also recommended.

Decreasing the global PID values for the muscle length feedback controller improved contact force predictions (Fig. 6). Forward dynamic passive joint motion was similar for the three global PID values, but hamstring and gastrocnemius muscle forces were greater for higher PID values (Fig. 5). At the highest global PID gain (PID 300), the semimembranosus and biceps femoris long head create force in late stance (~40% to 50% of gait cycle). Gastrocnemius forces also increase with PID gains at this point in the gait cycle, creating increased tibiofemoral loading for the same motion. Measured EMG for the hamstrings do not show activation in the accelerating cycle during the late stance phase (Fig. 10).

Properties for the ligament models (zero-load lengths and insertion/origins) were estimated based on previous cadaveric studies and were not subject specific. Ligament forces were small (under 200 N peak loading for any ligament) for both treadmill gait cycles, consistent with ligament forces predicted in a simulation of normal gait for a natural knee [17]. Simulation of movements more complicated than a walk cycle (e.g., sidestep) would place greater load on knee ligaments and may require more accurate determination of ligament insertion/origins as well as zero-load lengths.

The peak transverse plane forces in the stabilizer are in the same range as the ground reaction force shear forces. The spring constant for the stabilizer is 100 N/mm, and a force of 175 N will be generated if the transverse plane forward dynamics pelvis motion deviates from the inverse kinematics motion by 1.75 mm. Damping is also modeled, and added forces will result from velocity differences. During forward dynamic simulations, the stabilizer force oscillates around the average force (12 N in the anterior-posterior direction and 5 N in the med-lateral direction) at a frequency much higher than the gait cycle. The purpose of the stabilizer is to minimize accumulated position errors over the forward dynamics cycle, particularly errors on the global position of the model. Future work is needed to evaluate the dynamic response of the spring and damper used for the stabilizer, and the sensitivity of the stiffness and damping values on muscle and contact force predictions. Another limitation for the treadmill gait simulations is the fact that the subject held onto a hand rail while walking. Forces imparted to the body by the handrail were not measured and were not included in the model.

The stiffness for the torsional spring about the adduction/abduction axis of the right hip was 800 N-mm/deg. The stiffness of this axis is eight times larger than the passive torsional spring about other axes at the hip and ankle. Increasing the stiffness of this axis decreased errors between inverse kinematics and forward dynamics hip joint rotations. Soft tissue constraints at the hip, such as the iliofemoral ligament, were not included. The need to increase the hip abduction/adduction torsional spring stiffness may indicate a need to explicitly model these ligament constraints, or it may indicate that the muscle models are inadequate to maintain proper hip adduction/abduction. In addition, the ankle is modeled as three-orthogonal revolute joints, functionally behaving as a spherical joint. This does not model the geometric constraints of the joint, which would influence muscle forces about the ankle.

Fig. 8 Measured and predicted ground reaction forces for the right leg during the accelerating (a) and fast (b) gait cycles. Global PID values equal 50.

Fig. 9 Right leg ankle and hip rotations for the accelerating and fast gait cycles. Shown are the inverse kinematics rotations and the forward dynamics rotations for the passive joints of the muscle force driven right leg. For the hip, the x-axis corresponds to flexion/extension, the y-axis corresponds to internal/external rotation, and z-axis rotations correspond to abduction/adduction. For the ankle, the x-axis corresponds to plantarflexion/dorsiflexion, the y-axis corresponds to internal/external rotation, and z-axis rotations correspond to inversion/eversion. Global PID values equal 50.

Fig. 10 Normalized experimental EMG and normalized predicted muscle forces for the accelerating and fast gait cycles. Muscle forces are normalized to the maximum force that can be produced by each muscle as determined by its PCSA. Global PID values equal 50.

A limitation of the muscle modeling scheme is the fact that it does not have a physiological basis and only uses feedback control of muscle length measured during inverse kinematics simulations. This control scheme does not account for the force-length and force-velocity relationships in muscle. In addition, only normalized muscle force can be compared to measured EMG, which is a measure of muscle activation and not the force produced by muscle. As demonstrated in this study, estimation of the in vivo knee loads under dynamic conditions is highly dependent on muscle force predictions. This requires resolution of the muscle redundancy problem, which occurs because the muscle force solutions to create a certain motion are not unique. Muscle cocontractions that occur during movement may not be captured by the feedback control scheme and additional optimization of muscle forces would be required.

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