PREDICTION OF KNEE LOADING IN NATIVE KNEE AND TOTAL KNEE
ARTHROPLASTY: A FORWARD DYNAMICS COMPUTATIONAL STUDY

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ARTHROPLASTY: A FORWARD DYNAMICS COMPUTATIONAL STUDY

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CHAPTER 1

INTRODUCTION

Muscles and tendons actuate movement by developing force and generating moments about joints. When human movement is impaired by disease or trauma, function can sometimes be restored with surgical reconstruction of musculoskeletal structures. For example, patients with muscular spasticity often undergo tendon transfer and tendon lengthening operations aimed at correcting gait abnormalities. In osteoarthritic patients, bones are reconstructed to alter the joint geometry or to accommodate a prosthesis. Surgical reconstructions and joint pathologies often compromise the capacity of muscles to generate force and moment about a joint. For instance, when a joint becomes deformed from disease or is replaced by a prosthesis, the moment arms of the muscles that cross the joint may be reduced, thus decreasing the moment that can be generated at the joint. Patients that cannot generate sufficient muscle force or joint moment are associated with impairments such as muscle weakness.

At present, there are few quantitative tools to evaluate the effects of surgical procedures on muscle function. Surgeons rely on their experience and qualitative information from clinical exams to make surgical decisions. However, the combined effects of the active and passive force generating elements are complex and are being explored. The absence of a quantitative understanding has limited the success of various musculoskeletal reconstructions and hampers the development of new procedures.

Computer models can assist in understanding the biomechanical consequences of musculoskeletal reconstructions. Since musculoskeletal mechanics are complex, computer models are needed to understand the effects of surgery on the interactions of

bone, muscle, ligaments, and cartilage. Alternatively, developing effective surgical procedures without computer models may take significantly longer and accounting for variations in sex, geometry, ligament location, muscle forces, etc. is extremely challenging. Computer models can be especially useful to surgeons for investigating the effects of surgical decisions on a model rather than a patient. Computer models can be useful to biomechanical engineers and surgeons for analyzing problematic surgeries, and designing and evaluating various implant designs.

1.1 Focus of the Dissertation

The goal of this dissertation is to develop a musculoskeletal model and corroborate model predictions to experimentally measured in vivo knee contact forces, in order to study the biomechanical consequences of two different total knee arthroplasty designs.

The two main contributions of this dissertation are:

- (1) **Corroboration to experimental data:** The development of an EMG-driven, full-body, musculoskeletal model with subject-specific leg geometries including deformable contacts, ligaments, 6DOF knee joint, and a shoe-floor model that can concurrently predict muscle forces, ligament forces, and joint contact forces. Model predictions of tibiofemoral joint contact forces were evaluated against the subject-specific in vivo measurements from the instrumented TKR for three distinctly different styles of over ground gait.
- (2) **Virtual surgery in TKA:** The musculoskeletal modeling methodology was then used to develop a model for one healthy participant with a native knee and then virtually replacing the native knee with fixed-bearing and mobile-bearing total knee arthroplasty designs performing gait and step-up tasks. This approach

minimized the biomechanical impact of variations in sex, geometry, implant size, design and positioning, ligament location and tension, and muscle forces found across patients. The differences in biomechanics were compared for the two designs.

1.2 Significance of this Research

The world health organization ranks musculoskeletal disorders as the second largest contributor to disability worldwide. Conservative estimates put the national cost of direct care for musculoskeletal disease at \$212.7 billion a year [1].

Many people who suffer from neuromuscular or musculoskeletal diseases may benefit from the insights gained from surgery simulations, since musculoskeletal reconstructions are commonly performed on these individuals. Improved surgical outcomes will benefit these individuals not only in the short-term, but also in the long-term, since their future rehabilitation needs may be reduced. For example, although total knee arthroplasty is a common surgical procedure for the treatment of osteoarthritis with over 700,000 procedures performed each year [2], many patients are unhappy with the ultimate results [3]. Ten to 30% of patients report [4] pain, dissatisfaction with function, and the need for further surgery such as revision after the initial surgery resulting in costs exceeding \$11 billion [5]. Potentially, simulation studies that quantify the important biomechanical variables will reduce the need for revision surgeries in patients.

REFERENCES

1. Asfaw A, Pana-Cryan R, Bushnell T, Sauter S. Musculoskeletal disorders and associated healthcare costs among family members of injured workers. *American journal of industrial medicine* 58(11): 1205, 2015
2. Kurtz S, Ong K, Lau E, Mowat F, Halpern M. Projections of primary and revision hip and knee arthroplasty in the United States from 2005 to 2030. *The Journal of bone and joint surgery American volume* 89(4): 780, 2007
3. Bourne RB, Chesworth BM, Davis AM, Mahomed NN, Charron KDJ. Patient satisfaction after total knee arthroplasty: who is satisfied and who is not? *Clinical orthopaedics and related research* 468(1): 57, 2010
4. Beswick AD, Wylde V, Gooberman-Hill R, Blom A, Dieppe P. What proportion of patients report long-term pain after total hip or knee replacement for osteoarthritis? A systematic review of prospective studies in unselected patients. *BMJ open* 2(1): e000435, 2012
5. Losina E, Walensky RP, Kessler CL, Emrani PS, Reichmann WM, Wright EA, Holt HL, Solomon DH, Yelin E, Paltiel AD, Katz JN. Cost-effectiveness of total knee arthroplasty in the United States: patient risk and hospital volume. *Archives of internal medicine* 169(12): 1113, 2009

CHAPTER 2

EMG-DRIVEN FORWARD DYNAMICS SIMULATION TO ESTIMATE IN VIVO JOINT CONTACT FORCES DURING NORMAL, SMOOTH, AND BOUNCY GAITS

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ABSTRACT

Computational models that predict in vivo joint loading and muscle forces can potentially enhance and augment our knowledge of both typical and pathological gait. To adopt such models into clinical applications, studies validating modeling predictions are essential. This study created a full-body musculoskeletal model using data from the “Sixth Grand Challenge Competition to Predict in Vivo Knee Loads”. This model incorporates subject-specific geometries of the right leg in order to concurrently predict knee contact forces, ligament forces, muscle forces, and ground contact forces. The objectives of this paper are twofold: 1.) to describe an electromyography (EMG)-driven modeling methodology to predict knee contact forces, and 2.) to validate model predictions by evaluating the model predictions against known values for a patient with an instrumented total knee replacement (TKR) for three distinctly different gait styles (normal, smooth, and bouncy gait). The model integrates a subject-specific knee model onto a previously validated generic full-body musculoskeletal model. The combined model included six degrees-of-freedom (DOF) patellofemoral and tibiofemoral joints, ligament forces, and deformable contact forces with viscous damping. The foot/shoe/floor interactions were modeled by

incorporating shoe geometries to the feet. Contact between shoe segments and the floor surface were used to constrain the shoe segments. A novel EMG-driven feedforward with feedback trim motor control strategy was used to concurrently estimate muscle forces and knee contact forces from standard motion capture data collected on the individual subject. The predicted medial, lateral, and total tibiofemoral forces represented the overall measured magnitude and temporal patterns with good root mean squared errors (RMSEs) and Pearson's correlation (p^2). The model accuracy was high: medial, lateral, and total tibiofemoral contact force RMSEs = 0.15, 0.14, 0.21 body weight (BW), and ($0.92 < p^2 < 0.96$) for normal gait; RMSEs = 0.18 BW, 0.21 BW, 0.29 BW, and ($0.81 < p^2 < 0.93$) for smooth gait; and RMSEs = 0.21 BW, 0.22 BW, 0.33 BW, and ($0.86 < p^2 < 0.95$) for bouncy gait, respectively. Overall, the model captured the general shape, magnitude and temporal patterns of the contact force profiles accurately. Potential applications of this proposed model include predictive biomechanics simulations, design of TKR components, soft tissue balancing, and surgical simulation.

INTRODUCTION

Estimating the forces applied to our joints by muscles, ligaments, and articulating surfaces as well as their contribution to joint loading is fundamental in understanding joint damage, function, and disease. Knee joint loading is an important parameter in the design and implantation of total knee replacement (TKR) components, as excessive joint loading can cause component wear and eventually lead to failure [1-3]. Excessive joint contact forces are also an important contributor to the development of osteoarthritis [4]. Knowledge of joint loading during daily activities is essential for understanding mechanisms of injury, in the development of tissue engineered bio-materials [5], as well

as developing, evaluating, and optimizing injury prevention strategies. Joint loads can be measured *in vivo* using instrumented TKRs [6, 7] during dynamic activities. However, implementation of such devices is not common because it is invasive and expensive. Furthermore, the results are subject-specific and may not transfer to a healthy population. Thus, computational models for dynamic simulation [8-11] are more commonly used to estimate joint loads and muscle forces. Accurate prediction of joint loads under dynamic conditions requires accurate estimates of muscle forces, component and skeletal alignment, and ligament stiffness [12]. The resulting system is overdetermined with more muscles crossing the joint than degrees of freedom. The most common approach to solving this redundancy problem is static optimization [1, 13]. In this approach, inverse dynamics is used to minimize a cost function such as muscle activation, contact energy, or muscle stress one frame at a time to find muscle forces that reproduce computed joint moments [14]. While this approach is computationally efficient, the physiological form of the cost function is unknown and this method may not account for variability in an individual's muscle activation patterns. Activation of a muscle can be task dependent and can vary for the same joint kinematics and kinetics. Another approach uses electromyography (EMG) signals in conjunction with a muscle model to estimate task specific muscle forces [15-17]. The reliance of EMG-driven models on measured muscle activity takes into account an individual's activation patterns and muscle co-contraction. With the EMG-driven approach, no assumptions are required about the form of the cost function being minimized. However, "flexibility" still remains in the solution process as the absolute amplitude of each muscle excitation is difficult to determine, the number of EMG measurements is often limited, and EMG data are typically unavailable from deep

muscles. Hybrid approaches [18, 19] that combine EMG-driven models with either static optimization or computed muscle control take advantage of both optimization and EMG-driven approaches.

During the development of musculoskeletal models various assumptions are made, a few of these are phenomenological rather than established in biological or physiological evidence. To increase the accuracy of these models, additional studies measuring *in vivo* joint loading, joint kinematics, and musculoskeletal geometry are required. Such studies will provide a better understanding of the impact of size, age, morphology, or surgical history on model predictions and help define model limitations. Current musculoskeletal models often simplify or ignore the effects of soft tissue. However, coupling soft tissue deformation and muscle loading in concurrent simulation is essential for realistic prediction of osteokinematic, and more importantly, arthrokinematic motion and joint loading. The strength and ultimate goal of musculoskeletal modeling is to predict outcomes of an intervention or surgery. Despite these potential advantages, predictive musculoskeletal simulation has not been widely explored in clinical applications [20]. These simulations are not widely used due to high demand for computational efficiency [21], difficulty in defining relevant objective functions [22], and the complex computer programming required to build these simulations.

This paper presents a dual stage modeling method to predict tibiofemoral contact forces during three styles of over ground gait. This study differs from earlier research in several respects. During the first stage, kinematic data is the sole input into the model, while the knee joint (tibiofemoral and patellofemoral) is allowed six degrees of freedom

(DOF). The knee joint is constrained by contact between articulating surfaces and ligament forces. In the second forward dynamics stage, muscle and joint loads are predicted in a single computational step using an EMG driven feedforward with feedback trim motor control strategy. Traditional inverse dynamics optimization calculates moments about a knee modeled as a simple hinge, or as a hinge with secondary motion prescribed based on flexion angle. The method proposed in this paper is conducive to models with six DOF knee joints. Because ground reaction forces (GRFs) are not used as an input to the model, this method provides a foundation for predictive biomechanical simulation, testing different motor control strategies and incorporating proprioceptive feedback during movement.

The objective of this study was development of an EMG-driven, full-body, musculoskeletal model with subject-specific leg geometries including deformable contacts, ligaments, six DOF knee joint, and a shoe-floor model that can concurrently predict muscle forces, ligament forces, and joint contact forces. Model predictions of tibiofemoral joint contact forces were evaluated against the subject-specific *in vivo* measurements from the instrumented TKR for three distinctly different styles of over ground gait.

METHODS

Experimental Data

The data for this study were collected from an 83-year-old male subject (mass=70 kg and height=172 cm) with an instrumented TKR that measures the six loading components acting on the tibial tray. The experimental data were sourced from the sixth edition of the “*Grand Challenge Competition to Predict in Vivo Knee Loads*” [6, 23] and

includes motion capture marker trajectories (modified Cleveland Clinic marker set), GRFs and moments, EMG signals, dynamometer measurements, knee joint forces, geometries of the right leg bones and prosthetic, fluoroscopic, computed tomography (CT), and magnetic resonance (MR) images. Motion capture marker trajectories, EMG, GRFs and knee joint loads were collected simultaneously in a motion capture laboratory while the subject completed three different styles of over ground gait: normal, smooth and bouncy. One gait cycle for each gait style was chosen for modeling. The descriptions of the gait styles [24] were, smooth: reduced superior-inferior translation of the pelvis during the gait cycle, bouncy: increased superior-inferior translation of the pelvis during the gait cycle. GRF data were sampled at 1000 Hz and passed through a low-pass filter with a 50 Hz cut-off frequency. Motion capture data were sampled at 120 Hz and passed through a low pass filter with 6 Hz cut-off frequency. Measured total knee contact forces were distributed into medial and lateral components using an experimental regression equation [25].

Knee Model

The subject specific knee model was created using the implanted component geometries and bone geometries segmented from CT images of the subject (Fig. 1) in the multibody dynamic analysis program ADAMS (MSC Software Corporation, Santa Ana, CA). The right knee joint allowed six DOF for the tibiofemoral and patellofemoral joints. The motion for both joints was constrained by a compliant contact force model with viscous damping [26, 27], ligament forces, and the patellar tendon. The tibial insert was divided into medial and lateral geometries, with contacts created between each geometry

and the femoral component. The contact force in the medial and lateral tibial insert by means of a compliant contact was defined as:

$$F_c = k\delta^n + B(\delta)\dot{\delta} \quad (1)$$

Where F_c is the contact force, δ is the interpenetration distance between the geometries in contact, $\dot{\delta}$ is the velocity of interpenetration, k is the contact stiffness, n is the nonlinear power exponent, and $B(\delta)$ is a damping coefficient. The values implemented in this model were derived from a previous study [26] where $k = 30,000 \text{ N/mm}^n$, $n = 3/2$, and $B = 40 \text{ Ns/mm}$.

Figure 1: The knee model used subject-specific bone and knee prosthetic component geometries including ligaments and the patellar tendon. The knee model was integrated into a generic full-body model which included 44 muscle-tendon actuators acting about the hip, knee, and ankle. The standing radiograph was used to confirm limb alignment in the coronal plane.

The ligaments were modeled as multiple bundles with origin and insertion site footprints based on anatomical studies [28-37]. The posterior cruciate ligament was separated into two bundles [38]. Three bundles were used for the lateral collateral ligament [29]. Five bundles were used for the medial collateral ligament (MCL), two deep and three superficial [30, 39, 40]. Three bundles each were used for the lateral and medial patellofemoral ligaments. The anterior cruciate ligament was not included in the model as the TKR surgery involved resection of this ligament. A generic piecewise function [41] defining the force-length relationship of ligaments was used to model each bundle. Ligament bundle stiffness and zero-load length values were used to scale the generic force-length relationship for each bundle. Ligament bundle stiffness values were obtained from the literature [41-45] and have been verified in a previous knee joint model [46]. The zero-load lengths for each ligament bundle were determined from open-chain knee flexion-extension joint range-of-motion trials where ligament force was assumed to be small (under 20 N). To prevent the superficial MCL bundles penetrating into the bone and component geometries, wrapping was incorporated into the ligament. The quadriceps muscles inserted on the patella and the patellar tendon was modeled using three bundles with the same piecewise function used for the ligaments.

Shoe-floor Contact Model

The shoe-ground interface was modeled using deformable contacts between the shoe and force plate geometries. The shoe geometries were obtained by 3D scanning shoes of the same size, model, and make used by the subject during the motion capture measurements. The shoe geometries were divided into three rigid bodies: (1) the region containing the heel and mid-foot, (2) the region containing the metatarsals, and (3) the

region containing the phalanges (Fig. 2). The regions were defined by visually inspecting the compliance and the geometry of the shoe during gait. Regions were attached to each other using six-axis springs. The shoe was attached to the body of the foot segment via a six-axis spring. A hinge joint was applied where the mid-foot region joins the toe region to model the metatarsophalangeal joints. The six-axis springs serve as representative models for shoe compliance as well as relative movement of the foot within the shoe. Contacts were defined between the three rigid bodies and the force plates. The same shoe model was applied to both feet. Simple box representations were used to model the force plates.

Figure 2: Shoe and foot model. The shoe geometry is divided into three rigid bodies: the shoe, shoe toes, and shoe tip. The foot geometry is divided into two rigid bodies: the foot body and foot toes. Deformable contacts are defined between the three shoe parts and the force plate.

Generic Musculoskeletal Model

The generic full-body musculoskeletal model consisted of 21 rigid body segments, 53 revolute joints, and 44 right leg muscles. The lower limb extremity model is based on the model by Arnold et al. [47]. Regression equations from the US Air Force's Generator of Body Data (GEBOD) was used to determine generic joint center locations, mass, and inertial properties based on subject height, weight, and gender [48]. The generic bone geometries for the right leg were scaled to match the subject-specific bone

geometries. These scaling factors were also used to scale muscle origins, muscle insertions, and muscle via points for the right leg. Each joint center was represented by three orthogonal revolute joints with the exception of the ankle and the metatarsophalangeal joint.

Figure 3: Three-dimensional model of the lower limb. (a) Bony geometries including the wrapping surfaces for the pelvis, femur and tibia. (b) Wrapping surfaces for the medial and lateral gastrocnemius. (c) Via-points for ankle plantar flexors.

The ankle was modeled using two hinge joints that defined the talocrural and the talocalcaneal axes [49]. The metatarsophalangeal joint was modeled using a single revolute joint. The hip joint center and knee joint center were obtained using the symmetrical center of rotation estimation (SCoRE) method [50] and symmetrical axis of rotation approach (SARA) [51] applied to their respective joint range of motion trials.

The scaled hip and knee joint center from the generic model were replaced by the SCoRE and SARA method joint centers. The relative motion between the head and the neck joint were fixed for this study. For each marker a three-axis spring was defined between the body segment and the corresponding motion marker to allow relative movement between them. Markers defining the modified Cleveland clinic marker set locations were manually adjusted relative to their attached segment by minimizing the forces in the springs. The subject-specific femur and tibia along with the femoral and tibial components were manually aligned to the scaled generic model. Ellipsoidal wrapping surfaces and via points [47] (Fig. 3) were used to define muscle-tendon paths inhibited by bones and deeper muscles. Using wrapping surfaces allows the model to reflect more accurately operating lengths, muscle moment arms, and force production ability for muscles in the lower limb.

Figure 4: Feedforward with feedback control scheme for calculating muscle forces and joint contact forces.

The feedforward muscle scheme (Green) incorporates experimental EMG in conjunction with musculotendon (activation and contraction) dynamics to produce feedforward muscle forces. The feedback muscle scheme (Blue) uses the error between the current muscle length and the desired muscle length to produce feed-back trim muscle forces. The predicted muscle forces are the sum of the feedforward muscle forces and calculated feed-back trim muscle forces.

Subject-Specific EMG-Driven Muscle Model

Surface EMG data were collected for 16 lower extremity muscles (Table 1) on the right side. These muscles were adductor magnus (AddM), biceps femoris long head (BFLH), gluteus maximus (GMAX), gluteus medius (GMED), gracilis (GRA), lateral gastrocnemius (LG), medial gastrocnemius (MG), peroneus longus (PL), rectus femoris (RF), sartorius (SAR), semimembranosus (SM), soleus (SO), tensor fasciae lata (TFL), tibialis anterior (TA), vastus medialis (VM) and vastus lateralis (VL). After evaluating EMG patterns of the BFLH, GMED and VM for all measured trials, it was determined the EMG signals were in error and EMG signals for these muscles were replaced by average activations from an age-matched population during normal gait. Muscle activation for vastus intermedius (VI) was estimated as the average of VM and VL activation values. Semitendinosus (ST) was assumed to have the same activation as SM. Biceps femoris short head (BFSH) was presumed to have the same activation as BFLH [16]. To generate subject-specific and trial-specific muscle forces, recorded EMG for normal, smooth, and bouncy gait trials were high-pass filtered using a fourth order Butterworth filter with a cutoff frequency of 30 Hz, rectified. They were then low-pass filtered with a cutoff frequency of 4 Hz to obtain linear envelopes for each muscle similar to the procedure described by Lloyd et al. [16]. The linear envelopes were normalized to the peak values obtained from a series of available isometric and quasi-static force tasks. Two types of model parameter values were implemented: (1) Activation dynamics parameter values, and (2) Contraction dynamics parameter values. The EMG-to-activation model is a first-order dynamic model based on the work of Thelen [52] and Winters [53]. The activation and deactivation time constants are assumed to be 10 ms and

40 ms, respectively [54]. For the Hill-type muscle-tendon model, initial parameter values including optimal muscle fiber length, tendon slack length, and peak isometric muscle force, were taken from Arnold et al. [47] and scaled according to subject height and bone geometries. A custom Hill-type muscle model [47] with an inextensible tendon [55] was implemented in Simulink (The MathWorks, Inc., Natick, MA). Previously implemented EMG-driven models [16, 17, 56] tune or calibrate their models to identify the set of muscle parameters reproducing moments computed from inverse dynamics, no such calibration was performed on these models.

Table 1: Muscles with feedforward signal and feedback trim and their corresponding EMG signal inputs

Measured EMG	Muscle Model
Adductor magnus	Adductor magnus distal Adductor magnus ischial Adductor magnus middle Adductor magnus proximal
Biceps femoris long head*	Biceps femoris long head Biceps femoris short head
Lateral gastrocnemius	Lateral gastrocnemius
Medial gastrocnemius	Medial gastrocnemius
Gluteus maximus	Gluteus maximus superior Gluteus maximus middle Gluteus maximus inferior
Gluteus medius*	Gluteus medius anterior Gluteus medius middle Gluteus medius posterior
Gracilis	Gracilis
Peroneus longus	Peroneus longus
Rectus femoris	Rectus femoris
Sartorius	Sartorius
Semimembranosus	Semimembranosus Semitendinosus
Soleus	Soleus
Tensor fasciae latae	Tensor fasciae latae
Tibialis anterior	Tibialis anterior Vastus intermedius**
Vastus lateralis	Vastus lateralis
Vastus medialis*	Vastus medialis

* Replaced by average activations from a similar age matched population

** Equal to the average of the signals from the medial and lateral vasti

Gait Simulation

The experimental motion capture data for three gait styles (normal, smooth, bouncy) were used as inputs in inverse kinematics analyses. During these analyses, measured marker trajectories drove motion constraints connected to three-axis springs associated with the

corresponding body segments, during which, muscle-tendon lengths for the right leg and joint angles for the left leg and upper body were recorded. During inverse kinematics, the knee joint was constrained by ligament forces and contact forces, and the shoe ground interface contact constraints were included. For forward dynamics simulations (Fig. 4), the motion constraints were eliminated and muscle forces drove the right leg. Joint torques drove the upper body and left leg. Feedback controllers produced joint torques tracking inverse kinematics joint angles for the upper body and contralateral limb. A feedforward with feedback trim control scheme (Fig. 4) was used to produce muscle forces for the right limb. The feedforward muscle scheme incorporated experimental EMGs when available (Table 1) with a Hill-type muscle model to produce feedforward muscle forces. A feedback trim scheme was used to modulate the feedforward muscle force such that muscle-tendon lengths from the inverse kinematics step were maintained. The feedback trim controller adds to feedforward muscle forces if these forces are insufficient to maintain inverse kinematics musculotendon lengths during forward dynamics simulations to produce the total muscle force. In other words, muscle force is augmented if the forward dynamics length is greater than inverse kinematics length at a given time step. Similarly, the feedback trim controller will decrease muscle force if the current forward dynamics musculotendon length is too short. For muscles without measured EMG signals, the feedforward muscle forces come only from passive muscle properties. The feedback controller parameters were scaled based on physiological cross-sectional area in order to account for muscle size. As motion and EMGs were the inputs to the modeling scheme, tibiofemoral joint contact forces could be compared to measured forces.

Model Evaluation

The model predicted results were resampled to a time interval from 0% to 100% gait cycle using cubic spline interpolation. Differences between the measured and predicted forces were assessed by calculating the Pearson's correlation coefficient (ρ^2), root mean squared errors (RMSEs), and the coefficient of determination (R^2). The Pearson's correlation coefficient is used as a measure of shape differences. Values may range between -1 and 1, with values of 1 indicating a complete positive correlation, and 0 no correlation.

Figure 5: Medial, lateral, and total tibiofemoral contact forces compared with in vivo measurements obtained during three modifications of gait. Contact force is scaled to bodyweight with 1 BW equal to 686 N.

The root mean squared error (RMSE) is used as a measure of magnitude differences. The coefficient of determination is used as a measure of both magnitude and shape differences. The total muscle force, which is the output of the feedback trim controller, is used in the forward dynamics simulations. Comparison of normalized experimental EMG and predicted total and feedforward muscle forces are used to

evaluate the contribution of the feedforward muscle force to the total muscle force required for forward dynamics simulation.

RESULTS

The model predictions of medial, lateral, and total tibiofemoral contact force followed the measured temporal patterns with good Pearson's correlation (p^2) during normal ($0.92 < p^2 < 0.96$), smooth ($0.81 < p^2 < 0.93$) and bouncy ($0.86 < p^2 < 0.95$) gaits respectively (Fig. 5). The temporal patterns of the stance phase were well predicted in both compartments for all three trials with the predicted terminal stance phase having the least correlation. Much of this trend is attributed to the shoe-floor model and a premature heel strike on the contralateral limb. On average, the greatest agreement between the measured and predicted forces was in normal gait. For normal gait, the RMSEs between predicted and measured medial, lateral, and total contact force were 0.15 body weight (BW) , 0.14 BW and 0.21 BW and the R^2 were 0.9, 0.85, and 0.93 respectively. The greatest differences were observed in bouncy gait, with RSMEs between predicted and measured medial, lateral and total contact forces of 0.21 BW, 0.22 BW, and 0.33 BW respectively. The corresponding R^2 values were 0.85, 0.64, and 0.87.

The predicted pattern and timing of the feedforward and total muscle forces were compared along with the measured EMG signals for the primary muscles involved during gait. In general, the total muscle forces used to drive the forward dynamics simulations was similar to the feedforward muscle forces generated by the Hill-type muscle models for the major muscles of gait (Fig. 6). An exception was the medial gastrocnemius (MG) where feedforward muscle forces during terminal stance are decreased by the feedback trim controller, particularly for bouncy gait. Feedforward forces for the vastus lateralis

(VL) were also decreased during the loading response phase of gait. For the gluteus medius (GMED), the feedback trim controller added muscle force during mid-stance and terminal stance, indicating that the feedforward force for this muscle was not sufficient to maintain the muscle lengths derived during inverse kinematics. Measured EMG signals for the gluteus maximus (GMAX) and vastus lateralis (VL) muscles increased from normal to smooth gait and again from smooth to bouncy gait. There was a corresponding match in both feedforward and total muscle force for these muscles. The ground reaction force RMSEs were less than 0.08 BW for the anterior-posterior and medial-lateral shear directions and less than 0.22 BW in the vertical direction for the three different styles of gait.

Figure 6: Comparison of normalized experimental EMG and predicted total and feedforward muscle forces for the muscles of biceps femoris long head (BFLH), medial gastrocnemius (MG), gluteus maximus (GMAX), semimembranosus (SM), gluteus medius (GMED), tibialis anterior (TA), vastus lateralis (VL), and soleus (SO) for the three versions of gait. Note: scale for muscle forces of VL and SO (bottom row) is greater than scale for other muscles.

DISCUSSION

Various studies have demonstrated the potential of EMG-driven musculoskeletal models [15-17] in estimating muscle forces, providing understanding of normal and pathological movements, and complementing interpretations obtained via standard motion capture studies. Our primary objective for this paper was to describe our EMG-driven, subject-specific musculoskeletal modeling methodology. This methodology has the capability of concurrently simulating joint contact mechanics, shoe-ground interactions, and muscle forces. The second objective for this paper was to provide a comparison between our predictions of joint contact forces to known values recorded for a subject for three distinctly different styles of gait with the goal of validating our model. The Sixth Grand Challenge Data set provides a unique opportunity to evaluate our modeling method by providing data for a subject with an instrumented TKR which continuously measured six loading components on the tibial tray. Experimental data was collected for three gait styles: normal, smooth, and bouncy. A subject-specific knee model was created from the experimental data and incorporated into a generic full-body musculoskeletal model. The muscle force estimates were based on subject specific muscle activation patterns derived from recorded EMG and knee contact forces were used as a means for indirect validation of muscle force estimates.

Thanks in large part to the publicly available data set provided by the “*Grand Challenge Competition to Predict In-Vivo Knee Loads*”, musculoskeletal modeling and simulation techniques have seen notable advancements in recent years. An array of musculoskeletal modeling methods have been used to predict muscle forces and the resulting contact forces during movement including EMG-driven models [17], static and

dynamic optimization techniques [8-11], hybrid methods [18, 19], and parametric methods [57] to solve the muscle redundancy problem. The 2012 “Grand Challenge” winner [17] introduced an EMG-driven model which estimated muscle forces by solving an inverse dynamics based optimization problem. However, the modeling approach prescribed motion to the pelvis, assumed two DOF at the knee, and used GRFs as model inputs. We have extended this approach by developing a full-body musculoskeletal model that is torque driven for the upper body and left leg, includes a six DOF knee joint and includes a shoe-floor contact model in lieu of GRFs as an input. Our knee contact force predictions (RSMEs: medial = 0.15 BW and lateral = 0.14 BW in normal gait) are slightly better than obtained using the previous EMG-driven model (RSMEs: medial = 0.16 BW and lateral = 0.22 BW in normal gait). The 2016 “Grand Challenge” winner [9] introduced a full-body model with a simple knee contact model that used an inverse dynamics based optimization method to estimate muscle forces. However, the modeling approach assumed three DOF at the knee, and used GRFs as model inputs. Our joint contact load predictions errors (RSMEs: medial = 0.18 BW and lateral = 0.21 BW in smooth gait; medial = 0.21 BW and lateral = 0.22 BW in bouncy gait) are comparable and even slightly better to that obtained using the proposed approach (RSMEs: medial = 0.22 BW and lateral = 0.27 BW in smooth gait; medial = 0.20 BW and lateral = 0.25 BW in bouncy gait); the results obtained were also more accurate than those that have been acquired using forward dynamic simulations or traditional optimization [8-11] in the 2016 competition. Only one [10] of the 2016 competitors modeled normal gait using a one DOF knee model, our joint load predictions errors (RSMEs: medial = 0.15 BW and lateral = 0.14 BW in normal gait) are better to that obtained (RSMEs: medial = 0.26 BW

and lateral = 0.63 BW in normal gait) using an inverse dynamics based optimization model.

Our knee model, shoe-floor contact model, and simulation technique include several unique features. Typically, during inverse kinematics or inverse dynamics the knee joint is constrained by a single DOF (flexion-extension) joint and secondary DOFs are constrained to be functions of knee flexion-extension. This assumption removes any interdependence of contact forces on the muscles as well as muscle force influence to motion in the frontal and transverse 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 six DOF. Accurately modeling the shoe-ground interface is important for accurate prediction of joint contact forces. Our shoe-floor contact model included three rigid shoe segments per shoe, translational motion between the foot and shoe, and deformable contact between shoe and ground geometries. In the absence of experimental knee contact forces, evaluations of the predicted ground reaction forces (GRF) against measured GRF may offer indirect validation of joint load predictions. Also, the proposed framework is more conducive for predictive musculoskeletal simulations where, in the absence of experimental data or in efforts to predict new motions, contact with the environment (e.g. ground) must be modeled.

The feedforward with feedback trim control scheme allows simultaneous estimates of muscle forces, ligament forces, and medial and lateral knee contact forces. Hill-type muscle models rely on accurate estimation of subject-specific model parameters, such as optimal muscle fiber length, tendon slack length, and peak isometric

muscle force. Given the errors associated with EMG measurement and Hill-type muscle models, a true EMG driven forward dynamics model is not reasonably possible. To reduce sensitivity to measurement and model parameter uncertainty, a feedback trim controller is employed to ensure the model tracks measured kinematics by reducing the error between the current forward dynamics musculotendon length and the inverse kinematics recorded musculotendon length. Muscle activation can be task dependent, even for gait. This is evidenced by the different activation profiles, for similar joint angles, of the gluteus maximus (GMAX), vastus lateralis (VL), and tibialis anterior (TA) during the different styles of gait (Fig. 6). Hence, we believe that in order to produce better physiological estimates of muscle forces, EMG data as model inputs should be used. The computational performance is relatively quick for both inverse kinematics and forward dynamics taking under 25 minutes on a desktop PC (Windows 7, 3.4GHz Intel Core i7-4770 CPU, 16 GB RAM), when compared to dynamic optimization based approaches. The faster simulation time facilitates sensitivity analyses and predictive simulations, promoting understanding of causes for movement deviation, and ultimately assisting assessment of treatment options in response to clinical questions. However, the major challenge of applying this methodology in the clinical setting remains subject-specific geometry generation.

In order to accurately interpret the results of this study, multiple limitations in methodology must be noted. First, a single gait trial from each gait style from a single subject was analyzed. Data from additional trials and subjects are necessary in order to assess the extent to which these results can be generalized. In the multibody framework, joint loading predictions were estimated using a compliant contact force model with

viscous damping rather than a finite-element model which could better represent component deformation. Co-simulating multibody dynamics and linear or non-linear finite-element methods to predict refined estimates of deformation and stress is an area we are pursuing. The muscle modeling parameters and architecture parameters were based on a published generic musculoskeletal model [47] and assumed linear scaling. Inclusion of MRI-based subject-specific muscle parameter prediction [58] and more complex muscle-tendon dynamics models such as equilibrium or damped-equilibrium models [55] would improve feedforward muscle force predictions. Our models used a feedforward with feedback trim control scheme to predict forward dynamics muscle forces. The maximum EMG signal from various trials was used to normalize EMG data, but the accuracy of the EMG normalization is unknown. The feedforward control scheme relies on experimental EMG traces, when available, coupled with Hill-type models to estimate muscle force. The EMG signals from three muscles (VM, GMED, BFLH) were replaced with age matched average EMG signals from normal gait, contributing to errors in model force predictions, especially for smooth and bouncy gait. The feedback trim controller will add to feedforward muscle forces if this force is insufficient to maintain inverse kinematics musculotendon lengths during forward dynamics simulations. The feedback trim controller is necessary for forward dynamics simulations, but it has no physiological basis and may reduce co-contraction of antagonistic muscles. MR images were not available for this subject and ligament attachment footprints in the model were obtained in relation to anatomical landmarks. Hence, incorrect attachment sites and zero-load length determination could affect model predictions. Precisely modeling the shoe-floor interface is critical in accurately predicting joint loading. Errors in prediction of

vertical GRF and a premature heel strike on the contralateral limb indicate that improvements in the foot/shoe/floor model are necessary. Experimental measurement of foot/shoe motion and subject specific shoe compliance is recommended for future studies.

In summary, this study presented an EMG-driven musculoskeletal model with subject-specific joint anatomy to predict in-vivo TKR mechanics during walking. Joint loading predictions agreed well with in-vivo measurements obtained via an instrumented TKR. Thus the proposed framework is conducive for application in predictive biomechanics simulations, design of TKR components, soft tissue balancing, and surgical simulation.

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REFERENCES

1. Kellett CF, Short A, Price A, Gill HS, Murray DW. In vivo measurement of total knee replacement wear. *The Knee* 11(3): 183, 2004
2. Sathasivam S, Walker PS. Computer model to predict subsurface damage in tibial inserts of total knees. *J Orthop Res* 16(5): 564, 1998
3. Wimmer MA, Andriacchi TP. Tractive forces during rolling motion of the knee: implications for wear in total knee replacement. *J Biomech* 30(2): 131, 1997
4. Andriacchi TP, Mundermann A, Smith RL, Alexander EJ, Dyrby CO, Koo S. A framework for the in vivo pathomechanics of osteoarthritis at the knee. *Ann Biomed Eng* 32(3): 447, 2004
5. Zhang L, Hu J, Athanasiou KA. The Role of Tissue Engineering in Articular Cartilage Repair and Regeneration. *Critical reviews in biomedical engineering* 37(1-2): 1, 2009
6. D'Lima DD, Townsend CP, Arms SW, Morris BA, Colwell Jr CW. An implantable telemetry device to measure intra-articular tibial forces. *Journal of Biomechanics* 38(2): 299, 2005
7. Kutzner I, Heinlein B, Graichen F, Bender A, Rohlmann A, Halder A, Beier A, Bergmann G. Loading of the knee joint during activities of daily living measured in vivo in five subjects. *Journal of Biomechanics* 43(11): 2164, 2010
8. Ding Z, Nolte D, Kit Tsang C, Cleather DJ, Kedgley AE, Bull AM. In Vivo Knee Contact Force Prediction Using Patient-Specific Musculoskeletal Geometry in a Segment-Based Computational Model. *J Biomech Eng* 138(2): 021018, 2016

9. Jung Y, Phan CB, Koo S. Intra-Articular Knee Contact Force Estimation During Walking Using Force-Reaction Elements and Subject-Specific Joint Model. *J Biomech Eng* 138(2): 021016, 2016
10. Moissenet F, Cheze L, Dumas R. Influence of the Level of Muscular Redundancy on the Validity of a Musculoskeletal Model. *J Biomech Eng* 138(2): 021019, 2016
11. Smith CR, Vignos MF, Lenhart RL, Kaiser J, Thelen DG. The Influence of Component Alignment and Ligament Properties on Tibiofemoral Contact Forces in Total Knee Replacement. *J Biomech Eng* 138(2): 021017, 2016
12. Erdemir A, McLean S, Herzog W, van den Bogert AJ. Model-based estimation of muscle forces exerted during movements. *Clinical Biomechanics* 22(2): 131, 2007
13. Marra MA, Vanheule V, Fluit R, Koopman BH, Rasmussen J, Verdonschot N, Andersen MS. A subject-specific musculoskeletal modeling framework to predict in vivo mechanics of total knee arthroplasty. *J Biomech Eng* 137(2): 020904, 2015
14. Kuo AD. A least-squares estimation approach to improving the precision of inverse dynamics computations. *J Biomech Eng* 120(1): 148, 1998
15. Lloyd DG, Buchanan TS. A model of load sharing between muscles and soft tissues at the human knee during static tasks. *J Biomech Eng* 118(3): 367, 1996
16. Lloyd DG, Besier TF. An EMG-driven musculoskeletal model to estimate muscle forces and knee joint moments in vivo. *J Biomech* 36(6): 765, 2003
17. Manal K, Buchanan TS. An Electromyogram-Driven Musculoskeletal Model of the Knee to Predict in Vivo Joint Contact Forces During Normal and Novel Gait Patterns. *Journal of Biomechanical Engineering* 135(2): 0210141, 2013

18. Higginson JS, Ramsay JW, Buchanan TS. Hybrid models of the neuromusculoskeletal system improve subject-specificity. *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine* 226(2): 113, 2012
19. Sartori M, Farina D, Lloyd DG. Hybrid neuromusculoskeletal modeling to best track joint moments using a balance between muscle excitations derived from electromyograms and optimization. *Journal of Biomechanics* 47(15): 3613, 2014
20. Mansouri M, Clark AE, Seth A, Reinbolt JA. Rectus femoris transfer surgery affects balance recovery in children with cerebral palsy: A computer simulation study. *Gait & Posture* 43: 24, 2016
21. Anderson FC, Pandy MG. Dynamic Optimization of Human Walking. *Journal of Biomechanical Engineering* 123(5): 381, 2001
22. Ackermann M, van den Bogert AJ. Optimality principles for model-based prediction of human gait. *Journal of Biomechanics* 43(6): 1055, 2010
23. Fregly BJ, Besier TF, Lloyd DG, Delp SL, Banks SA, Pandy MG, D'Lima DD. Grand challenge competition to predict in vivo knee loads. *J Orthop Res* 30(4): 503, 2012
24. Kinney AL, Besier TF, D'Lima DD, Fregly BJ. Update on grand challenge competition to predict in vivo knee loads. *J Biomech Eng* 135(2): 021012, 2013
25. Meyer AJ, D'Lima DD, Banks SA, Coburn J, Harman M, Mikashima Y, Fregly BJ. Evaluation of Regression Equations for Medial and Lateral Contact Force From Instrumented Knee Implant Data. (54587): 389, 2011
26. Guess TM, Stylianou AP, Kia M. Concurrent prediction of muscle and tibiofemoral contact forces during treadmill gait. *J Biomech Eng* 136(2): 021032, 2014

27. Kia M, Stylianou AP, Guess TM. Evaluation of a musculoskeletal model with prosthetic knee through six experimental gait trials. *Med Eng Phys* 36(3): 335, 2014
28. Meister BR, Michael SP, Moyer RA, Kelly JD, Schneck CD. Anatomy and kinematics of the lateral collateral ligament of the knee. *The American journal of sports medicine* 28(6): 869, 2000
29. Park SE, DeFrate LE, Suggs JF, Gill TJ, Rubash HE, Li G. The change in length of the medial and lateral collateral ligaments during in vivo knee flexion. *Knee* 12(5): 377, 2005
30. Hartshorn T, Otarodifard K, White EA, Hatch GF, 3rd. Radiographic landmarks for locating the femoral origin of the superficial medial collateral ligament. *Am J Sports Med* 41(11): 2527, 2013
31. LaPrade RF, Engebretsen AH, Ly TV, Johansen S, Wentorf FA, Engebretsen L. The anatomy of the medial part of the knee. *J Bone Joint Surg Am* 89(9): 2000, 2007
32. Amis AA, Firer P, Mountney J, Senavongse W, Thomas NP. Anatomy and biomechanics of the medial patellofemoral ligament. *The Knee* 10(3): 215, 2003
33. Stephen JM, Lumpaopong P, Deehan DJ, Kader D, Amis AA. The medial patellofemoral ligament: location of femoral attachment and length change patterns resulting from anatomic and nonanatomic attachments. *The American journal of sports medicine* 40(8): 1871, 2012
34. Viste A, Chatelet F, Desmarchelier R, Fessy MH. Anatomical study of the medial patello-femoral ligament: landmarks for its surgical reconstruction. *Surgical and radiologic anatomy : SRA*, 2014

35. Bowman Jr KF, Sekiya JK. Anatomy and Biomechanics of the Posterior Cruciate Ligament and Other Ligaments of the Knee. *Operative Techniques in Sports Medicine* 17(3): 126, 2009
36. Chandrasekaran S, Ma D, Scarvell JM, Woods KR, Smith PN. A review of the anatomical, biomechanical and kinematic findings of posterior cruciate ligament injury with respect to non-operative management. *The Knee* 19(6): 738, 2012
37. Race A, Amis AA. The mechanical properties of the two bundles of the human posterior cruciate ligament. *Journal of biomechanics* 27(1): 13, 1994
38. Osti M, Tschann P, Kunzel KH, Benedetto KP. Anatomic characteristics and radiographic references of the anterolateral and posteromedial bundles of the posterior cruciate ligament. *Am J Sports Med* 40(7): 1558, 2012
39. Liu F, Gadikota HR, Kozanek M, Hosseini A, Yue B, Gill TJ, Rubash HE, Li G. In vivo length patterns of the medial collateral ligament during the stance phase of gait. *Knee Surg Sports Traumatol Arthrosc* 19(5): 719, 2011
40. Liu F, Yue B, Gadikota HR, Kozanek M, Liu W, Gill TJ, Rubash HE, Li G. Morphology of the medial collateral ligament of the knee. *J Orthop Surg Res* 5: 69, 2010
41. Blankevoort L, Huiskes R. Ligament-bone interaction in a three-dimensional model of the knee. *J Biomech Eng* 113(3): 263, 1991
42. Hara K, Mochizuki T, Sekiya I, Yamaguchi K, Akita K, Muneta T. Anatomy of normal human anterior cruciate ligament attachments evaluated by divided small bundles. *Am J Sports Med* 37(12): 2386, 2009
43. Petersen W, Zantop T. Anatomy of the anterior cruciate ligament with regard to its two bundles. *Clin Orthop Relat Res* 454: 35, 2007

44. Robinson JR, Bull AM, Amis AA. Structural properties of the medial collateral ligament complex of the human knee. *J Biomech* 38(5): 1067, 2005
45. Woo SL, Abramowitch SD, Kilger R, Liang R. Biomechanics of knee ligaments: injury, healing, and repair. *J Biomech* 39(1): 1, 2006
46. Guess TM, Razu S, Jahandar H. Evaluation of Knee Ligament Mechanics Using Computational Models. *J Knee Surg* 29(2): 126, 2016
47. Arnold EM, Ward SR, Lieber RL, Delp SL. A model of the lower limb for analysis of human movement. *Ann Biomed Eng* 38(2): 269, 2010
48. Cheng H, Obergefell L, Rizer A. Generator of Body Data (GEBOD), Manual. In.: DTIC Document. 1994
49. Isman RE, Inman VT, Poor P. Anthropometric studies of the human foot and ankle. *Bull Prosthet Res* 11: 97, 1969
50. Ehrig RM, Taylor WR, Duda GN, Heller MO. A survey of formal methods for determining the centre of rotation of ball joints. *Journal of biomechanics* 39(15): 2798, 2006
51. Ehrig RM, Taylor WR, Duda GN, Heller MO. A survey of formal methods for determining functional joint axes. *J Biomech* 40(10): 2150, 2007
52. Thelen DG. Adjustment of muscle mechanics model parameters to simulate dynamic contractions in older adults. *J Biomech Eng* 125(1): 70, 2003
53. Winters JM. An improved muscle-reflex actuator for use in large-scale neuromusculoskeletal models. *Ann Biomed Eng* 23(4): 359, 1995
54. Zajac FE. Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control. *Crit Rev Biomed Eng* 17(4): 359, 1989

55. Millard M, Uchida T, Seth A, Delp SL. Flexing computational muscle: modeling and simulation of musculotendon dynamics. *J Biomech Eng* 135(2): 021005, 2013
56. Buchanan TS, Lloyd DG, Manal K, Besier TF. Neuromusculoskeletal Modeling: Estimation of Muscle Forces and Joint Moments and Movements From Measurements of Neural Command. *Journal of applied biomechanics* 20(4): 367, 2004
57. Knowlton CB, Wimmer MA, Lundberg HJ. Grand Challenge Competition: A Parametric Numerical Model to Predict In Vivo Medial and Lateral Knee Forces in Walking Gaits. (44809): 199, 2012
58. Valente G, Pitto L, Testi D, Seth A, Delp SL, Stagni R, Viceconti M, Taddei F. Are subject-specific musculoskeletal models robust to the uncertainties in parameter identification? *PLoS One* 9(11): e112625, 2014

CHAPTER 3

**CHANGES IN TIBIOFEMORAL KNEE BIOMECHANICS
AFTER FIXED-BEARING OR MOBILE-BEARING TOTAL
KNEE ARTHROPLASTY: A FORWARD DYNAMICS
COMPUTATIONAL STUDY**

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ABSTRACT

Total knee arthroplasty (TKA) is widely used treatment for advanced knee osteoarthritis, however, many patients are dissatisfied with the final results. One proposed reason behind this dissatisfaction is the apparent inability of prosthetic knees to satisfactorily restore the complex biomechanics of the natural knee. The aim of this investigation was to evaluate how well two conceptually different TKA designs, the fixed-bearing (FB) design and mobile-bearing (MB) design, replicate the biomechanics of the normal knee during daily activities. For this investigation, a three-dimensional (3D) computational patient-specific musculoskeletal model was created which incorporated: 3D imaging studies, gait laboratory data, force-plate measurements and 13-muscle electromyographic data. Two virtual surgeries were performed to “replace” the knee using reverse engineering software and scanned prosthetic components with an accuracy of +/- 0.125

mm. The model was then used to predict biomechanical differences in performing gait and step-up tasks. Outcomes measured included: femorotibial contact stresses, bone-prosthesis interface forces, rotational motion, and quadriceps forces. Agreement of the model-derived predictions with published fluoroscopic studies was found to be good to excellent. Our results indicate that neither the MB nor FB TKA design fully restored the normal knee biomechanics in gait and step-up activities. Lower femorotibial contact stresses were observed for the FB design, while axial rotation was found to be superior with the MB design. Similar interface forces and quadriceps forces were seen in the simulations for both prosthetic designs. Further investigation into the biomechanical consequences of TKA design differences is warranted to improve patient satisfaction and outcomes.

INTRODUCTION

Total knee arthroplasty (TKA) is a common surgical procedure for the treatment of osteoarthritis with over 700,000 procedures performed each year[1], yet many patients remain dissatisfied with the results[2]. Ten to 30% of patients report[3] pain, dissatisfaction with function, and the need for subsequent revision surgery, resulting in costs exceeding \$11 billion[4] in the year 2005. Appropriate restoration of knee biomechanics[5] is key to the success of TKA. Factors related to biomechanical restoration include implant design, component placement, soft tissue balance, active muscle forces, and individual patient movement strategies[6-9]. Our understanding of the impact of TKA on knee biomechanics across a spectrum of deformity and ligamentous laxity may be enhanced by studying the composite effects of TKA on a normal, healthy subject. Few studies of patients with normal anatomy have assessed the impact of TKA

and its effects on knee kinematics, kinetics, and resultant muscle forces. Although several *ex vivo* investigations[10, 11] have explored these effects, they lacked the capacity to incorporate the influence of muscles, gravity, and weight-bearing during functional activities. Similarly, few *in vivo* studies have reported the kinematics of normal[12], osteoarthritic[13], ACL-deficient[12, 14-16], or post-TKA knees[13]. Such investigations do not include muscle, ligament, or joint contact forces because their direct measurement is not feasible. In addition, patients with instrumented prosthetics[17, 18] are rare and its implementation is expensive.

TKA implants include several design features that collectively affect knee biomechanics: femoral sagittal plane shape, trochlear contour and depth, tibial baseplate design, polyethylene tibial insert contour, and insert constraint. Most knee replacements utilize a fixed bearing[19] TKA (FB-TKA) design in which the polyethylene insert is locked into the tibial baseplate, with nearly all femorotibial interactions occurring along the top surface of the tibial insert. However, instability and polyethylene wear have been established as long-term causes for failure[20]. In response to these concerns, mobile bearing TKAs (MB-TKA) were designed to reduce the shear force transmitted at the bone-implant interface, thus reducing the risk of implant loosening[21]. This design also theoretically helps with self-adjustment of rotational malalignment between the femur and tibia, which may produce more central patellar tracking[22]. A final assumed benefit is the reduction of polyethylene contact stresses and wear that have been associated with aseptic loosening, and osteolysis[23]. In MB-TKA, the lower, “back side” of the tibial insert is allowed to pivot around a central post in the implant tray, allowing contact to be distributed to both the top and under-surface of the tibial insert[21]. Despite these

potential, theoretical advantages[21-23], meta-analyses[7, 21, 22, 24-26] comparing MB-TKA and FB-TKA designs have not shown any significant benefit of the MB-TKA in terms of pain, complications, clinical scores, patient satisfaction, loosening, range of motion, quality of life, or revision rates. Theoretically, the MB-TKA design was a radical and revolutionary concept, though its clinical benefit remains contentious[21]. Gaining knowledge of muscle, ligament, and joint contact forces comparing the two designs may help guide surgical decisions.

Although *in vivo* native knee, MB-TKA, and FB-TKA design contact forces and joint biomechanics have been measured either *ex vivo*, using instrumented knee prostheses, or fluoroscopy imaging systems respectively, it is still challenging to compare the biomechanics of these three cases in the same patient simultaneously. Predictive capabilities of computational musculoskeletal (MS) models enable us to gain such dynamic and objective information on individual patients. MS models have successfully investigated various orthopedic questions, including TKA component malrotation[27, 28], tendon transfer[29-31], osteoarthritis[32, 33], and ligament injuries[34, 35]. However, no previous studies have used a computational MS model to calculate biomechanical differences between native knee, FB-TKA, and MB-TKA designs in a healthy subject. To address these gaps, we developed a computational MS model of one healthy participant with a native knee and virtually replaced it with FB-TKA and MB-TKA designs performing gait and step-up tasks. This approach minimized the biomechanical impact of variations in sex, geometry, implant size, design, positioning, ligament location and tension, and muscle forces found across patients.

The present study's specific aims were to (1) develop a MS model based on subject-specific magnetic resonance (MR) images and kinematic data obtained from motion capture and force plate data; (2) replace the native knee by virtually implanting FB-TKA and MB-TKA designs into the MS model; and (3) assess the differences in knee biomechanics between FB-TKA and MB-TKA designs and their deviations from the native knee.

Methods

Participant. One recreationally active healthy female (age: 25 years; height: 1.68 m; mass: 62 kg) volunteered and provided informed consent to part take in this study. She had no prior or current injury likely to affect her ability to accomplish the essential tasks. The University of Missouri Institutional Review Board approved the study (approval number: 1209961 HS), which was carried out according to approved guidelines.

Instrumentation. We collected motion capture marker trajectories at 100 Hz using an eight-camera motion capture system (VICON, Oxford Metrics Ltd., Oxford, United Kingdom). Ground reaction forces were collected via four AMTI (Advanced Mechanical Technology Inc., Watertown, MA, USA) Optima ground-embedded force plates sampling at 2000 Hz. Surface electromyography (EMG) data were acquired at 1000 Hz from 13 lower-limb muscles on the right leg (identified as the dominant leg) via a wireless EMG systems (Delsys® TrignoTM, Boston, USA). We acquired static MR images (1.5 T, GE Medical Systems, Salt Lake City, USA) of the pelvis, right leg, and right knee cartilage in order to generate 3D models of the knee articulating surfaces, pelvis, femur, tibia, fibula, patella, and foot. When the MR images were made, the participant wore custom motion capture markers visible in the MR images on the pelvis

and right leg. These markers remained on the subject during subsequent tasks to help register the 3D models to the motion capture coordinate system.

Procedures. To model shoe-ground interaction, we 3D scanned shoes (New Balance, MT410BS4) and motion capture markers associated with them that the participant wore during all tasks. We prepared the participant's skin for surface EMG collection by abrasion and sterilization. Then, in accordance with Surface Electromyography for the Non-Invasive Assessment of Muscle (SENIAM) guidelines[36], we applied electrodes to the right limb's gluteus maximus and medius, semitendinosus, biceps femoris, vastus lateralis and medialis, rectus femoris, medial and lateral gastrocnemius, soleus, tibialis anterior, tensor fascia lata and peroneus longus muscles. In order to verify the correct placement of the EMG electrodes EMG-time traces were visually inspected during forceful isometric contractions to inspect for cross-talk. Sixty-one 14 mm retroreflective markers based on the modified Cleveland Clinic marker set[17] were affixed to the participant, including extra markers on the trunk and shoes. The participant completed two tasks: gait and step-up (0.2 m step) on the dominant leg, three trials each.

Data processing. Motion capture marker trajectories were low-pass filtered (6 Hz), zero-lag, 4th order Butterworth filter, which was determined via a residual analysis. EMG data were high-pass filtered (30 Hz), full-wave rectified, and low-pass filtered (6 Hz) using a zero-lag, 4th order Butterworth filter to obtain a linear envelope[37]. EMG data were normalized to the peak amplitude obtained by asking the patient to perform a series of isometric and quasi-static force tasks.

Figure 1: This study employed a native multibody knee model that included 6 degree-of-freedom tibiofemoral and patellofemoral joints. Knee contact forces were computed using impact function. To simulate gait and step-up, the multibody knee, subject-specific geometries of the pelvis, right-limb, and shoes were incorporated into a full-body musculoskeletal model with 44 muscles acting about the hip, knee and ankle.

Musculoskeletal modelling. In order to perform MS simulations in the multibody dynamic analysis program ADAMS (MSC Software Corporation, Santa Ana, CA), we used a previously validated 42 degree-of-freedom (DOF) full-body MS model with 44 musculotendon actuators (right lower limb) and 28 torque actuators (upper body and left limb)[38, 39]. The joints were modeled as follows: hip = 3-DOF ball and socket; ankle (talocrural) joint = pin joint; head-trunk segment = single rigid segment articulating with the pelvis via a 3-DOF ball and socket joint; upper limb = 3-DOF ball and socket (shoulder) and single-DOF (elbow and radioulnar). The subtalar, metatarsophalangeal, and wrist joints were locked[38]. The generic model was scaled in 3-dimensions to the participant's individual bone geometry for the pelvis and right lower limb, and anthropometry for the rest of the segments was determined during a static trial. The generic pelvis and right lower limb were replaced by the participant's bone geometries. Inverse kinematics analyses[40] calculated muscle-tendon lengths for the right lower

limb and joint angles for the upper body and left lower limb. During inverse kinematics, experimental motion capture marker positions drove motion constraints connected to three-axis springs between model and experimental marker positions. Muscle forces were obtained via an EMG-driven feedforward with feedback trim motor control strategy[40]. Experimental EMG data were used to generate individual muscle forces by taking into account the activation dynamics[41-43] and physiological force-length-velocity properties[38, 44] of the musculotendinous units. The predictive capacity of the model was corroborated to knee contact forces in a participant with an instrumented total knee replacement[40].

Table 1 Knee Ligaments used in the computational model

ligament names	number of ligament fibers (arrangement)
anterior cruciate ligament (ACL)	8 (4 anteromedial, 4 anterolateral)
posterior cruciate ligament (PCL)	8 (4 posteromedial, 4 anterolateral)
superficial medial collateral ligament (sMCL)	6 (3 proximal, 3 distal)
deep medial collateral ligament (dMCL)	2
posterior oblique ligament (POL)	3
medial posterior capsule (MPC)	3
lateral posterior capsule (LPC)	3
oblique popliteal ligament (OPL)	2
lateral collateral ligament (LCL)	1
anterolateral ligament (ALL)	1
fabellofibular ligament (FFL)	1

Native knee model. The right-knee model (Fig. 1) was: tibiofemoral (TF) and patellofemoral (PF) joints = 6-DOF joints; cartilage surfaces = high-resolution triangulated meshes; and contact between the articulating surfaces (cartilage-cartilage and menisci-cartilage) used an ADAMS default algorithm which computed contact force as a function of penetration velocity and depth at contact locations, based on contact parameters established in the literature[45-47]. To model menisci structural properties, we used existing methods[48, 49] to discretize the medial and lateral menisci radially, resulting in 25 and 31 elements, respectively. The adjacent elements were connected

using a linear stiffness matrix, resulting in multibody representation of deformable meniscal geometries[45-47]. The origins and insertions of the major knee ligaments (Table 1) came from segmented MR images when available, and scaling and projecting the ligament origins and insertions from the model presented in Kia et al[47]. The structural representation of each ligament fiber was described using a tension-only, nonlinear force-elongation relationship. The ligament force-elongation response in the toe region,[50-54] linear region,[50-54] and reference strains were adapted from Kia et al.'s work[47], which corroborated experimental measurements of ligament forces to model predictions. The patellar tendon's ligament force-elongation response in the toe region and linear region was obtained from literature[55, 56]. The reference strains for the patellofemoral ligaments and patellar tendon were determined from attachment distances obtained in the MR images.

Figure 2: The two knee models (a: fixed bearing and b: mobile bearing) showing the ligaments and implanted TKA components

Fixed and mobile bearing knee modeling. To construct the computerized arthroplasty models (Fig. 2), two fellowship-trained knee arthroplasty surgeons assisted in performing

two virtual surgeries using reverse engineering software (Geomagic, Morrisville, NC, USA). The precise 3D component geometries of a modern knee replacement system (ATTUNE®, DePuy Synthes, Warsaw, IN, USA) were incorporated by scanning components on a laser 3D scanner (NextEngine, Inc., Santa Monica, CA, USA) with an accuracy of +/- 0.125 mm.

We virtually implanted the femoral component using a measured-resection technique that resected 9mm from the distal femur aligned perpendicular to the femoral mechanical axis (axis connecting the center of the femoral head to the center of the knee). The femoral sizing and bone cuts were performed using a posterior-referencing technique referenced to the posterior condylar axis with external rotation being set to 3° about the femur's distal diaphysis. A size 5 narrow femoral component fit well and was selected for both the FB-TKA and MB-TKA models.

The virtual tibial resection was performed for each model incorporating a 5° posterior slope, and the proximal tibial cut was aligned perpendicular to the tibia's mechanical axis (knee center to ankle center). We set the depth of resection to restore the original joint line. Size 4 tibial baseplates were selected for both models and subsequently positioned and rotated with reference to the tibial tubercle. The tibial base plate was constrained to the tibia with a fixed joint positioned at the bone-component interface and oriented parallel to the tibial coordinate system. A revolute joint was established at the center of the MB-TKA post for that model. For the patellar component of both models, a simulated free-hand resection of 9.5 mm was made parallel to the osteochondral junction. A medialized dome design patella (size 35, thickness: 9.5 mm) fit well and was selected

for both. The medial lateral location of the patella implant apex was verified to match native anatomy.

We used the same ligaments as the native knee model, except for the POL, OPL, and ACL, which were deactivated. The tibial insert was separated into medial and lateral components, with contacts created between each geometry and the femoral component. The contact force in the medial and lateral tibial inserts and patella were predicted by means of the ADAMS default IMPACT function. The contact parameters we used were previously defined by Guess et al[57].

Outcome variables. Based on benefits proposed in the literature, we evaluated outcome variables of interest for MB-TKA over FB-TKA. A major proposed benefit the reduction of shear force transmitted in the bone-implant interface[21]. We explored forces at the fixed joint at the tibial bone-implant interface in the anteroposterior, mediolateral, and superioinferior directions for both TKA models. Another proposed benefit is the self-correction of tibiofemoral rotational malalignment, which may result in better patellar tracking[22]. We evaluated this by exploring the tibiofemoral rotational kinematics and quadriceps (sum of rectus femoris, vastus medialis, lateralis, and intermedius) muscle forces as predicted by the native knee two TKA models. A final proposed benefit is the reduction of polyethylene contact stresses[23]. We explored the tibiofemoral contact forces, area, and pressure for both TKA models (see supplementary figures: 7-13 for knee kinematics, knee ligament forces, and detailed right lower limb muscle forces for comparison between the three knee models). Following previous research, our analysis focused not only on the stance phase during gait but also from the beginning to the end of weight acceptance during step-up[12, 14].

Figure 3: Model Comparisons. Predicted (black) and experimental kinematics (dashed/colored) during gait and step-up cycles. Native Knee (A-G), fixed bearing (H), mobile bearing (I). Literature reference kinematics, magenta dashed line, Chen et al.[12]; blue dashed line, Kozanek et al.[16]; red dashed and dotted line, Okamoto et al.[58]. The output plots show the mean (solid line) $\pm$ 1 standard deviations (shaded region)

Model Verification. To furnish assurance in our simulations, we performed qualitative assessments between the model-based predicted knee kinematics and fluoroscopy data in the literature for gait in the native knee and step-up tasks in the native knee, FB-TKA and MB-TKA designs. These qualitative assessments were conducted across the entire stance phase during gait and the weight acceptance phase during step-up to ensure they were within 2 SD of published data[12, 16, 58], as recommended by Hicks et al[59].

Results.

Model Verification. Overall, agreement between model-derived knee angles, translations and published data from fluoroscopy studies were good to excellent. The native model-predicted knee angles for step up and gait were within 1 and 2 SD of published

fluoroscopy data[12, 16, 58], respectively. Native knee mediolateral translation for step up was within 1 SD, and anteroposterior translation for gait and step up were within 2-3 SD. FB-TKA and MB-TKA model-derived knee angles for step up were within 2 SD.

Bone-implant Interface Forces. When comparing the two TKA designs using the two knee models, interface forces in the anteroposterior direction (Fig. 4A, 4D) were higher in FB-TKA during gait with average root-mean-square difference RMSD [max] of 0.16 BW [0.25] and similar during step-up with RMSD [max] of 0.03 BW [0.05]. However, interface forces in the mediolateral direction (Fig. 4B, 4E) were higher in MB-TKA during step-up with RMSD [max] of 0.17 BW [0.28] and similar during gait with RMSD [max] of 0.05 BW [0.08]. Interface forces in the superioinferior direction (Fig. 4C, 4F) were higher in FB-TKA during gait with RMSD [max] of 0.07 BW [0.16] and higher in MB-TKA during step-up with RMSD [max] of 0.09 BW [0.27].

Figure 4: Bone-implant interface forces in the anteroposterior, mediolateral, and superioinferior directions as predicted by the fixed bearing (blue) and mobile bearing (green) designs: (A-C) stance phase during gait, (D-F) weight bearing phase during step-up. (mean, solid line; 1SD, shaded)

Knee Rotation and Quadriceps Muscle Forces. Although all three knee models followed a similar rotational pattern during gait (Fig. 5A), the MB-TKA design more closely followed the rotational changes of the native knee with lower RMSD [max] of 1.7° [2.7] compared to the FB-TKA design with RMSD [max] of 2.7° [4.2]. The knee rotations (Fig. 5C) during step-up were similar between the two TKA knee models, with RMSD [max] of 0.5° [0.3]. However, the knee rotations of the two TKA knee models compared to the native knee differed considerably, with RMSD [max] $> 3.2^{\circ}$ [3.1].

Quadriceps forces (Fig. 5B, 5D) generally were higher in the TKA models than in the native knee, particularly during initial contact, midstance during gait, and at 53% of the step-up cycle with RMSD [max] of Native vs FB-TKA: gait: 0.16 BW [0.29], step-up: 0.33 BW [0.43], and Native vs MB-TKA: gait: 0.11 BW [0.18], step-up: 0.32 BW [0.42]. FB-TKA demonstrated higher quadriceps forces than MB-TKA throughout stance phase of the gait cycle, with differences becoming negligible toward terminal stance with RMSD [max] of MB-TKA vs FB-TKA: gait: 0.08 BW [0.21], and step-up: 0.04 BW [0.10].

Figure 5: Knee axial rotation and Quadriceps (sum of rectus femoris, vastus medialis, lateralis, and intermedius) muscle forces as predicted by the native knee (red), fixed bearing (blue), and mobile bearing (green) designs: (A, B) stance phase during gait, (C, D) weight bearing phase during step-up. (mean, solid line; 1SD, shaded)

Tibiofemoral Contact Pressure. When comparing the the two TKA designs, the MB-TKA knee model predicted slightly greater average medial contact pressure with RMSD [max] 4.6 MPa [11.5] during gait (Fig. 6). During the more demanding step-up

simulations, MB-TKA's increase in medial femorotibial pressures became more pronounced with RMSD [max] 10.3 MPa [46.4] greater than its fixed-bearing counterpart. In addition, the predicted average contact pressures for the lateral compartment in the replacement models were higher for MB-TKA during both gait and step-up activities with RMSD [max] for gait: 7.4 MPa [12.6], and step-up: 15.3 MPa [68.6].

Figure 6: Articular contact pressure in the medial and lateral compartments predicted by fixed bearing (blue) and mobile bearing (green) designs: (A, B) stance phase during gait, (C, D) weight bearing phase during step-up. (mean, solid line; 1SD, shaded)

Discussion. This study's unique computational MS model allowed us to compare the healthy native knee to MB-TKA and FB-TKA conditions, an analysis that is not possible in other experimental designs. Our approach minimized the biomechanical impact of variations in sex, geometry, implant size and positioning, ligament location and tension,

and muscle forces found across patients. To the authors' knowledge, no previous studies have calculated differences in biomechanics between native knee, FB-TKA, and MB-TKA designs using a computational MS model.

The longstanding debate regarding optimal design for knee replacements will likely continue for the foreseeable future. In theory, the mobile-bearing design affords several potential advantages, though these have not been borne out in the form of superior clinical results. Function and outcome studies have shown excellent results for both designs, and neither has demonstrated clinical superiority[24, 60-63]. At present, it remains unclear which design better restores normal knee kinematics during prosthetic replacement. In this study, our results suggest that neither design reproduced the biomechanics of the native knee, and that measurable differences exist between the two designs.

Previous studies comparing the ability of FB-TKA versus MB-TKA to replicate natural knee axial rotation have yielded inconsistent results. We observed greater knee axial rotational with the MB-TKA during gait. This superiority was not seen during step-up, where similar rotation was observed. These results parallel the findings of Sharma *et al.*[64] who created a 3D MS model based on fluoroscopic data. We found that the MB-TKA demonstrated higher degrees of external rotation during loaded knee bending. Similarly, in a knee simulator study, Fisher *et al.*[65] found that, during gait, the MB-TKA design allowed increased axial rotation (7.0 degrees) compared to the FB-TKA design (4.5 degrees). In contrast, other researchers have found little difference in the axial rotation between two designs. Using a cadaveric robotic testing system, Most *et al.*[66] found slightly improved rotational motion (7% greater) with the FB-TKA design but the

trend did not achieve statistical significance. Similarly, D'Lima *et al.*[67] found both replacement designs altered kinematics of the normal knee but observed little difference between the two models.

Another proposed benefit of the MB-TKA design is its ability to increase bearing-surface congruity and reduce contact pressure at the tibiofemoral articulation because flexion is uncoupled from axial rotation. In our study, this proposed benefit was not established. Instead, we predicted higher average contact pressure in gait and step-up simulations. This is consistent with prior research using mechanical testing and computational simulation models[64, 65, 68, 69]. However, some investigators in similar studies have reported decreased stresses with the MB-TKA design[70-72]. The approach used to calculate contact pressure in this study does have its limitations since the average contact pressures calculated were greater than the yield strength of polyethylene (~22 MPa).

Conflicting results from implant retrieval studies further complicate our understanding of how design differences affect contact stresses and subsequent wear. One such study of 48 MB-TKAs by Kelly *et al.*[73] found no improvement in polyethylene wear resistance. Engh *et al.*'s.[74] analysis of 54 MB-TKA and FB-TKA inserts found that the designs exhibited different modes of polyethylene wear, though the overall rate of wear was similar. In another retrieval study of both designs, Lu *et al.*[75] noted an increase in low-grade polyethylene wear with MB-TKA designs (burnishing, abrasion, cold flow) whereas FB-TKA designs displayed higher-grade modes of wear (scratching, pitting, and delamination). In a large retrieval study of 312 implants, Berry *et al.*[76] reported decreased overall lower polyethylene wear in MB-TKA knees, with the lowest wear rates appearing in FB-TKA designs made of cobalt-chrome versus titanium. To date, the

decreased contact stresses and improved polyethylene wear of the MB-TKA design has yet to be clearly demonstrated. Our investigation parallels this finding, as the MB-TKA design did not demonstrate a reduction in contact stresses in either the gait cycle or step-up simulations.

One further proposed advantage of the MB-TKA design is its potential to divert stresses away from the bone-implant interface and thereby decrease the likelihood of component loosening. This has yet to be realized in clinical outcomes. In our investigation, interface forces were largely equivalent for both design models. During the gait cycle, the FB-TKA exhibited higher anterior-posterior forces, but medial-lateral forces remained equivalent. Conversely, during step-up, the MB-TKA displayed higher forces but only in a medial-lateral direction, while anterior-posterior forces remained similar. For both activities, forces directed in a superior-inferior direction were similar. Therefore, although our results showed overall similar interface forces for the two models, directional components of the force vectors are relevant, and they vary with the type of implant. Our findings echo multiple clinical studies demonstrating no difference in the development/progression of component radiolucencies or periprosthetic osteolysis over mid- and long-term follow-up[77, 78]. Multiple meta-analyses have shown similar rates of loosening for both designs[79-81]. This equivalence is further underscored by a Cochrane Review reporting no difference in the rates of component loosening or revision surgery between the two designs[61].

Unlike most outcome variables considered in this investigation, few studies have investigated how knee design impacts knee musculature. In our study, the predicted quadriceps forces during the walking and step-up simulations were similar regardless of

the implant design. This is consistent with the work of Sharma *et al.*[64], who found the forces generated within the quadriceps during deep knee bending to be equivalent in both designs, with less than 1% difference in maximum quadriceps forces noted between models. On the other hand, Innocenti *et al.*'s.[68] 3D computational study comparing quadriceps forces during simulated squatting found that both designs exhibited higher quadriceps forces as flexion increased, with higher forces in the MB-TKA model. Further investigation into the effects of implant design on supporting muscle function is needed.

Limitations. Our study has presented some novel insights, though we recognize and acknowledge the work's limitations. We studied one healthy and active participant, and further research should consider the influence of sex, variations in bony morphology, and relevant pathological populations. Another limitation is the use of origins, insertions, reference strains, and ligament stiffness parameters from published literature to model the knee ligaments, changes in these parameters to describe this subject are unaccounted for. Neither was the effect of ligament balancing on the ligament properties post TKA surgery. Implementation of ultrasound-based in situ ligament strain imaging[82] would improve ligament force predictions. Our approach to estimating muscle forces using an EMG-driven feedforward with feedback trim motor control strategy has limitations. The feedforward control scheme uses experimental EMG data from a normal healthy participant to predict muscle forces for native as well as TKA knee models, but it is known that TKA patients commonly experience neuromuscular adaptations[83] that were not modeled in this study. The maximum EMG signal from various trials was used to normalize EMG data, though the accuracy of the EMG normalization[84] is still under debate. The muscle modeling parameters and architecture parameters were based on a

published generic MS model and assumed linear scaling in 3D. Including MR image-based, subject-specific muscle parameter prediction and more complex muscle-tendon dynamics models would improve muscle force predictions. Use of surrogate contact models[85] or finite element methods[86] to predict refined estimates of deformation, pressure, and stress in order to estimate wear may yield further clinical and component design insight. This study is limited to only one manufacturer and care must be taken not to generalize trends or findings into other manufacturers or design variations.

Conclusion. In Conclusion, this study uses a computational MS model to compare healthy native knee to MB-TKA and FB-TKA designs. When comparing bone-implant interface forces between the two designs, FB-TKA exhibited higher anterior-posterior forces during gait, while MB-TKA displayed higher medio-lateral forces during step-up task. Greater knee axial rotation was observed with the MB-TKA during gait, while this superiority was not observed during step-up task. The predicted quadriceps muscle forces were similar regardless the Implant design. MB-TKA design experienced greater average and peak contact pressure for both tasks.

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References

1. Kurtz S, Ong K, Lau E, Mowat F, Halpern M. Projections of primary and revision hip and knee arthroplasty in the United States from 2005 to 2030. *The Journal of bone and joint surgery American volume* 89(4): 780, 2007
2. Bourne RB, Chesworth BM, Davis AM, Mahomed NN, Charron KDJ. Patient satisfaction after total knee arthroplasty: who is satisfied and who is not? *Clinical orthopaedics and related research* 468(1): 57, 2010
3. Beswick AD, Wylde V, Gooberman-Hill R, Blom A, Dieppe P. What proportion of patients report long-term pain after total hip or knee replacement for osteoarthritis? A systematic review of prospective studies in unselected patients. *BMJ open* 2(1): e000435, 2012

4. Losina E, Walensky RP, Kessler CL, Emrani PS, Reichmann WM, Wright EA, Holt HL, Solomon DH, Yelin E, Paltiel AD, Katz JN. Cost-effectiveness of total knee arthroplasty in the United States: patient risk and hospital volume. *Archives of internal medicine* 169(12): 1113, 2009
5. Naili JE, Wretenberg P, Lindgren V, Iversen MD, Hedström M, Broström EW. Improved knee biomechanics among patients reporting a good outcome in knee-related quality of life one year after total knee arthroplasty. *BMC musculoskeletal disorders* 18(1): 122, 2017
6. Matsuzaki T, Matsumoto T, Kubo S, Muratsu H, Matsushita T, Kawakami Y, Ishida K, Oka S, Kuroda R, Kurosaka M. Tibial internal rotation is affected by lateral laxity in cruciate-retaining total knee arthroplasty: an intraoperative kinematic study using a navigation system and offset-type tensor. *Knee surgery, sports traumatology, arthroscopy : official journal of the ESSKA* 22(3): 615, 2014
7. Kim YH, Kook HK, Kim JS. Comparison of fixed-bearing and mobile-bearing total knee arthroplasties. *Clinical orthopaedics and related research* (392): 101, 2001
8. Kia M, Wright TM, Cross MB, Mayman DJ, Pearle AD, Sculco PK, Westrich GH, Imhauser CW. Femoral Component External Rotation Affects Knee Biomechanics: A Computational Model of Posterior-stabilized TKA. *Clinical orthopaedics and related research* 476(1): 113, 2018
9. Smith JW, Christensen JC, Marcus RL, LaStayo PC. Muscle force and movement variability before and after total knee arthroplasty: A review. *World journal of orthopedics* 5(2): 69, 2014

10. Ali AA, Shalhoub SS, Cyr AJ, Fitzpatrick CK, Maletsky LP, Rullkoetter PJ, Shelburne KB. Validation of predicted patellofemoral mechanics in a finite element model of the healthy and cruciate-deficient knee. *Journal of biomechanics* 49(2): 302, 2016
11. Clary CW, Fitzpatrick CK, Maletsky LP, Rullkoetter PJ. The influence of total knee arthroplasty geometry on mid-flexion stability: An experimental and finite element study. *Journal of Biomechanics* 46(7): 1351, 2013
12. Chen CH, Li JS, Hosseini A, Gadikota HR, Gill TJ, Li G. Anteroposterior stability of the knee during the stance phase of gait after anterior cruciate ligament deficiency. *Gait Posture* 35(3): 467, 2012
13. Akbari Shandiz M, Boulos P, Saevarsson SK, Yoo S, Miller S, Anglin C. Changes in knee kinematics following total knee arthroplasty. *Proceedings of the Institution of Mechanical Engineers Part H, Journal of engineering in medicine* 230(4): 265, 2016
14. Kozanek M, Hosseini A, de Velde SK, Moussa ME, Li JS, Gill TJ, Li G. Kinematic evaluation of the step-up exercise in anterior cruciate ligament deficiency. *Clin Biomech (Bristol, Avon)* 26(9): 950, 2011
15. Shabani B, Bytyqi D, Lustig S, Cheze L, Bytyqi C, Neyret P. Gait knee kinematics after ACL reconstruction: 3D assessment. *Int Orthop* 39(6): 1187, 2015
16. Kozanek M, Hosseini A, Liu F, Van de Velde SK, Gill TJ, Rubash HE, Li G. Tibiofemoral kinematics and condylar motion during the stance phase of gait. *Journal of Biomechanics* 42(12): 1877, 2009
17. Fregly BJ, Besier TF, Lloyd DG, Delp SL, Banks SA, Pandy MG, D'Lima DD. Grand challenge competition to predict in vivo knee loads. *J Orthop Res* 30(4): 503, 2012

18. Taylor WR, Schütz P, Bergmann G, List R, Postolka B, Hitz M, Dymke J, Damm P, Duda G, Gerber H, Schwachmeyer V, Hosseini Nasab SH, Trepczynski A, Kutzner I. A comprehensive assessment of the musculoskeletal system: The CAMS-Knee data set. *Journal of Biomechanics* 65: 32, 2017
19. Carey BW, Harty J. A comparison of clinical- and patient-reported outcomes of the cemented ATTUNE and PFC sigma fixed bearing cruciate sacrificing knee systems in patients who underwent total knee replacement with both prostheses in opposite knees. *Journal of orthopaedic surgery and research* 13(1): 54, 2018
20. Blunn GW, Joshi AB, Minns RJ, Lidgren L, Lilley P, Ryd L, Engelbrecht E, Walker PS. Wear in retrieved condylar knee arthroplasties. A comparison of wear in different designs of 280 retrieved condylar knee prostheses. *The Journal of arthroplasty* 12(3): 281, 1997
21. Feczko PZ, Jutten LM, van Steyn MJ, Deckers P, Emans PJ, Arts JJ. Comparison of fixed and mobile-bearing total knee arthroplasty in terms of patellofemoral pain and function: a prospective, randomised, controlled trial. *BMC Musculoskeletal Disorders* 18: 279, 2017
22. Haider H, Garvin K. Rotating Platform versus Fixed-bearing Total Knees: An In Vitro Study of Wear. *Clinical orthopaedics and related research* 466(11): 2677, 2008
23. Ghomrawi HM, Kane RL, Eberly LE, Bershadsky B, Saleh KJ. Patterns of functional improvement after revision knee arthroplasty. *The Journal of bone and joint surgery American volume* 91(12): 2838, 2009

24. Bo ZD, Liao L, Zhao JM, Wei QJ, Ding XF, Yang B. Mobile bearing or fixed bearing? A meta-analysis of outcomes comparing mobile bearing and fixed bearing bilateral total knee replacements. *The Knee* 21(2): 374, 2014
25. Apostolopoulos AP, Michos IV, Mavrogenis AF, Chronopoulos E, Papachristou G, Lallios SN, Efstathopoulos NE. Fixed versus mobile bearing knee arthroplasty: a review of kinematics and results. *Journal of long-term effects of medical implants* 21(3): 197, 2011
26. Capella M, Dolfin M, Saccia F. Mobile bearing and fixed bearing total knee arthroplasty. *Annals of Translational Medicine* 4(7), 2016
27. Chen Z, Wang L, Liu Y, He J, Lian Q, Li D, Jin Z. Effect of component mal-rotation on knee loading in total knee arthroplasty using multi-body dynamics modeling under a simulated walking gait. *J Orthop Res* 33(9): 1287, 2015
28. Thompson JA, Hast MW, Granger JF, Piazza SJ, Siston RA. Biomechanical effects of total knee arthroplasty component malrotation: A computational simulation. *Journal of Orthopaedic Research* 29(7): 969, 2011
29. Mogk JP, Johanson ME, Hentz VR, Saul KR, Murray WM. A simulation analysis of the combined effects of muscle strength and surgical tensioning on lateral pinch force following brachioradialis to flexor pollicis longus transfer. *J Biomech* 44(4): 669, 2011
30. Magermans DJ, Chadwick EK, Veeger HE, Rozing PM, van der Helm FC. Effectiveness of tendon transfers for massive rotator cuff tears: a simulation study. *Clinical biomechanics (Bristol, Avon)* 19(2): 116, 2004

31. Murray WM, Hentz VR, Friden J, Lieber RL. Variability in surgical technique for brachioradialis tendon transfer. Evidence and implications. *The Journal of bone and joint surgery American volume* 88(9): 2009, 2006
32. Crossley KM, Dorn TW, Ozturk H, van den Noort J, Schache AG, Pandy MG. Altered hip muscle forces during gait in people with patellofemoral osteoarthritis. *Osteoarthritis and cartilage* 20(11): 1243, 2012
33. Gerus P, Sartori M, Besier TF, Fregly BJ, Delp SL, Banks SA, Pandy MG, D'Lima DD, Lloyd DG. Subject-specific knee joint geometry improves predictions of medial tibiofemoral contact forces. *J Biomech* 46(16): 2778, 2013
34. Donnelly CJ, Lloyd DG, Elliott BC, Reinbolt JA. Optimizing whole-body kinematics to minimize valgus knee loading during sidestepping: implications for ACL injury risk. *J Biomech* 45(8): 1491, 2012
35. Buffi JH, Werner K, Kepple T, Murray WM. Computing muscle, ligament, and osseous contributions to the elbow varus moment during baseball pitching. *Annals of biomedical engineering* 43(2): 404, 2015
36. Hermens HJ, Freriks B, Disselhorst-Klug C, Rau G. Development of recommendations for SEMG sensors and sensor placement procedures. *Journal of electromyography and kinesiology : official journal of the International Society of Electrophysiological Kinesiology* 10(5): 361, 2000
37. Lloyd DG, Besier TF. An EMG-driven musculoskeletal model to estimate muscle forces and knee joint moments in vivo. *Journal of Biomechanics* 36(6): 765, 2003
38. Arnold EM, Ward SR, Lieber RL, Delp SL. A model of the lower limb for analysis of human movement. *Ann Biomed Eng* 38(2): 269, 2010

39. Rajagopal A, Dembia CL, DeMers MS, Delp DD, Hicks JL, Delp SL. Full-Body Musculoskeletal Model for Muscle-Driven Simulation of Human Gait. *IEEE transactions on bio-medical engineering* 63(10): 2068, 2016
40. Razu SS, Guess TM. Electromyography-Driven Forward Dynamics Simulation to Estimate In Vivo Joint Contact Forces During Normal, Smooth, and Bouncy Gaits. *Journal of biomechanical engineering* 140(7): 071012, 2018
41. Thelen DG. Adjustment of muscle mechanics model parameters to simulate dynamic contractions in older adults. *J Biomech Eng* 125(1): 70, 2003
42. Winters JM. An improved muscle-reflex actuator for use in large-scale neuro-musculoskeletal models. *Ann Biomed Eng* 23(4): 359, 1995
43. Zajac FE. Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control. *Crit Rev Biomed Eng* 17(4): 359, 1989
44. Millard M, Uchida T, Seth A, Delp SL. Flexing computational muscle: modeling and simulation of musculotendon dynamics. *J Biomech Eng* 135(2): 021005, 2013
45. Guess TM, Razu S, Jahandar H. Evaluation of Knee Ligament Mechanics Using Computational Models. *J Knee Surg* 29(2): 126, 2016
46. Guess TM, Razu SS, Kuroki K, Cook JL. Function of the Anterior Intermeniscal Ligament. *J Knee Surg* 31(1): 68, 2018
47. Kia M, Schafer K, Lipman J, Cross M, Mayman D, Pearle A, Wickiewicz T, Imhauser C. A Multibody Knee Model Corroborates Subject-Specific Experimental Measurements of Low Ligament Forces and Kinematic Coupling During Passive Flexion. *J Biomech Eng* 138(5): 051010, 2016

48. Guess TM, Thiagarajan G, Kia M, Mishra M. A subject specific multibody model of the knee with menisci. *Medical Engineering & Physics* 32(5): 505, 2010
49. Guess TM, Razu S, Jahandar H, Stylianou A. Predicted loading on the menisci during gait: The effect of horn laxity. *J Biomech* 48(8): 1490, 2015
50. Blankevoort L, Huiskes R. Ligament-bone interaction in a three-dimensional model of the knee. *J Biomech Eng* 113(3): 263, 1991
51. Hara K, Mochizuki T, Sekiya I, Yamaguchi K, Akita K, Muneta T. Anatomy of normal human anterior cruciate ligament attachments evaluated by divided small bundles. *Am J Sports Med* 37(12): 2386, 2009
52. Petersen W, Zantop T. Anatomy of the anterior cruciate ligament with regard to its two bundles. *Clin Orthop Relat Res* 454: 35, 2007
53. Robinson JR, Bull AM, Amis AA. Structural properties of the medial collateral ligament complex of the human knee. *J Biomech* 38(5): 1067, 2005
54. Woo SL, Abramowitch SD, Kilger R, Liang R. Biomechanics of knee ligaments: injury, healing, and repair. *J Biomech* 39(1): 1, 2006
55. Yanke A, Bell R, Lee A, Shewman EF, Wang V, Bach BR, Jr. Regional mechanical properties of human patellar tendon allografts. *Knee surgery, sports traumatology, arthroscopy : official journal of the ESSKA* 23(4): 961, 2015
56. Yanke AB, Bell R, Lee AS, Shewman E, Wang VM, Bach BR, Jr. Central-third bone-patellar tendon-bone allografts demonstrate superior biomechanical failure characteristics compared with hemi-patellar tendon grafts. *The American journal of sports medicine* 41(11): 2521, 2013

57. Guess TM, Stylianou AP, Kia M. Concurrent prediction of muscle and tibiofemoral contact forces during treadmill gait. *J Biomech Eng* 136(2): 021032, 2014
58. Okamoto N, Nakamura E, Nishioka H, Karasugi T, Okada T, Mizuta H. In vivo kinematic comparison between mobile-bearing and fixed-bearing total knee arthroplasty during step-up activity. *The Journal of arthroplasty* 29(12): 2393, 2014
59. Hicks JL, Uchida TK, Seth A, Rajagopal A, Delp SL. Is my model good enough? Best practices for verification and validation of musculoskeletal models and simulations of movement. *J Biomech Eng* 137(2): 020905, 2015
60. Bailey O, Ferguson K, Crawford E, James P, May PA, Brown S, Blyth M, Leach WJ. No clinical difference between fixed- and mobile-bearing cruciate-retaining total knee arthroplasty: a prospective randomized study. *Knee surgery, sports traumatology, arthroscopy : official journal of the ESSKA* 23(6): 1653, 2015
61. Hofstede SN, Nouta KA, Jacobs W, van Hooff ML, Wymenga AB, Pijls BG, Nelissen RG, Marang-van de Mheen PJ. Mobile bearing vs fixed bearing prostheses for posterior cruciate retaining total knee arthroplasty for postoperative functional status in patients with osteoarthritis and rheumatoid arthritis. *The Cochrane database of systematic reviews* (2): Cd003130, 2015
62. Jacobs WC, Christen B, Wymenga AB, Schuster A, van der Schaaf DB, ten Ham A, Wehrli U. Functional performance of mobile versus fixed bearing total knee prostheses: a randomised controlled trial. *Knee surgery, sports traumatology, arthroscopy : official journal of the ESSKA* 20(8): 1450, 2012

63. Mahoney OM, Kinsey TL, D'Errico TJ, Shen J. The John Insall Award: no functional advantage of a mobile bearing posterior stabilized TKA. *Clinical orthopaedics and related research* 470(1): 33, 2012
64. Sharma A, Komistek RD, Ranawat CS, Dennis DA, Mahfouz MR. In vivo contact pressures in total knee arthroplasty. *The Journal of arthroplasty* 22(3): 404, 2007
65. Fisher J, McEwen H, Tipper J, Jennings L, Farrar R, Stone M, Ingham E. Wear-simulation analysis of rotating-platform mobile-bearing knees. *Orthopedics* 29(9 Suppl): S36, 2006
66. Most E, Li G, Schule S, Sultan P, Park SE, Zayontz S, Rubash HE. The kinematics of fixed- and mobile-bearing total knee arthroplasty. *Clinical orthopaedics and related research* (416): 197, 2003
67. D'Lima DD, Trice M, Urquhart AG, Colwell CW, Jr. Comparison between the kinematics of fixed and rotating bearing knee prostheses. *Clinical orthopaedics and related research* (380): 151, 2000
68. Innocenti B, Pianigiani S, Labey L, Victor J, Bellemans J. Contact forces in several TKA designs during squatting: A numerical sensitivity analysis. *Journal of biomechanics* 44(8): 1573, 2011
69. Jennings LM, Bell CI, Ingham E, Komistek RD, Stone MH, Fisher J. The influence of femoral condylar lift-off on the wear of artificial knee joints. *Proceedings of the Institution of Mechanical Engineers Part H, Journal of engineering in medicine* 221(3): 305, 2007
70. Buechel FF, Sr. Long-term followup after mobile-bearing total knee replacement. *Clinical orthopaedics and related research* (404): 40, 2002

71. Callaghan JJ, Squire MW, Goetz DD, Sullivan PM, Johnston RC. Cemented rotating-platform total knee replacement. A nine to twelve-year follow-up study. *The Journal of bone and joint surgery American volume* 82(5): 705, 2000
72. Stukenborg-Colsman C, Ostermeier S, Hurschler C, Wirth CJ. Tibiofemoral contact stress after total knee arthroplasty: comparison of fixed and mobile-bearing inlay designs. *Acta orthopaedica Scandinavica* 73(6): 638, 2002
73. Kelly NH, Fu RH, Wright TM, Padgett DE. Wear damage in mobile-bearing TKA is as severe as that in fixed-bearing TKA. *Clinical orthopaedics and related research* 469(1): 123, 2011
74. Engh GA, Zimmerman RL, Parks NL, Engh CA. Analysis of wear in retrieved mobile and fixed bearing knee inserts. *The Journal of arthroplasty* 24(6 Suppl): 28, 2009
75. Lu YC, Huang CH, Chang TK, Ho FY, Cheng CK, Huang CH. Wear-pattern analysis in retrieved tibial inserts of mobile-bearing and fixed-bearing total knee prostheses. *The Journal of bone and joint surgery British volume* 92(4): 500, 2010
76. Berry DJ, Currier JH, Mayor MB, Collier JP. Knee wear measured in retrievals: a polished tray reduces insert wear. *Clinical orthopaedics and related research* 470(7): 1860, 2012
77. Abdel MP, Tibbo ME, Stuart MJ, Trousdale RT, Hanssen AD, Pagnano MW. A randomized controlled trial of fixed- versus mobile-bearing total knee arthroplasty: a follow-up at a mean of ten years. *The bone & joint journal* 100-b(7): 925, 2018
78. Rahman WA, Garbuz DS, Masri BA. Randomized Controlled Trial of Radiographic and Patient-Assessed Outcomes Following Fixed Versus Rotating Platform Total Knee Arthroplasty. *The Journal of arthroplasty* 25(8): 1201, 2010

79. Carothers JT, Kim RH, Dennis DA, Southworth C. Mobile-bearing total knee arthroplasty: a meta-analysis. *The Journal of arthroplasty* 26(4): 537, 2011
80. Moskal JT, Capps SG. Rotating-platform TKA no different from fixed-bearing TKA regarding survivorship or performance: a meta-analysis. *Clinical orthopaedics and related research* 472(7): 2185, 2014
81. Pijls BG, Plevier JWM, Nelissen R. RSA migration of total knee replacements. *Acta orthopaedica* 89(3): 320, 2018
82. Gijsbertse K, Sprengers A, Naghibi Beidokhti H, Nillesen M, de Korte C, Verdonschot N. Strain imaging of the lateral collateral ligament using high frequency and conventional ultrasound imaging: An ex-vivo comparison. *Journal of biomechanics* 73: 233, 2018
83. Kuntze G, von Tscharnner V, Hutchison C, Ronsky JL. Alterations in lower limb multimuscle activation patterns during stair climbing in female total knee arthroplasty patients. *Journal of neurophysiology* 114(5): 2718, 2015
84. Sinclair J, Taylor PJ, Hebron J, Brooks D, Hurst HT, Atkins S. The Reliability of Electromyographic Normalization Methods for Cycling Analyses. *Journal of human kinetics* 46: 19, 2015
85. Smith CR, Vignos MF, Lenhart RL, Kaiser J, Thelen DG. The Influence of Component Alignment and Ligament Properties on Tibiofemoral Contact Forces in Total Knee Replacement. *J Biomech Eng* 138(2): 021017, 2016
86. Halloran JP, Petrella AJ, Rullkoetter PJ. Explicit finite element modeling of total knee replacement mechanics. *Journal of Biomechanics* 38(2): 323, 2005

Supplementary Material

Figure 7: Tibiofemoral rotations in the sagittal, coronal, and transverse planes as predicted by the three knee models: stance phase during gait, weight bearing phase during step-up. (mean, solid line; 1SD, shaded)

Figure 8: Tibiofemoral translations in the anteroposterior and mediolateral directions as predicted by the three knee models: stance phase during gait, weight bearing phase during step-up. (mean, solid line; 1SD, shaded)

Figure 9: Medial and lateral collateral ligaments tension and posterior cruciate ligament tension predicted by the three knee models: stance phase during gait, weight bearing phase during step-up. (mean, solid line; 1SD, shaded)

Figure 10: Major right lower-limb muscle forces as predicted by the three knee models at stance phase during gait, (mean, solid line; 1SD, shaded)

Figure 11: Major right lower-limb muscle forces as predicted by the three knee models at stance phase during gait, (mean, solid line; 1SD, shaded)

Figure 12: Major right lower-limb muscle forces as predicted by the three knee models during step-up cycle, (mean, solid line; 1SD, shaded)

Figure 13: Major right lower-limb muscle forces as predicted by the three knee models during step-up cycle, (mean, solid line; 1SD, shaded)

VITA

Swithin Samuel Razu was born on 23rd april, 1983 in Bangalore, India to Dr. I John Mohan Razu and Ms. Pauline Razu. He completed his high school in Bangalore, India and received his bachelor's degree from Visvesvaraya Technological University in mechanical engineering in 2005. He then proceeded to pursue his graduate studies from Missouri University of Science and Technology in the department of mechanical engineering with an emphasis on demand distribution modeling and its effects on concept selection. He then proceeded to pursue his PhD in biological engineering at the University of Missouri, Columbia. His PhD thesis was to develop a musculoskeletal model and corroborate model predictions to experimentally measured in vivo knee contact forces, in order to study the biomechanical consequences of two different total knee arthroplasty designs.