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Computing muscle, ligament, and osseous contributions to the elbow varus moment during baseball pitching

James H. Buffi^{1,2}, Katie Werner⁵, Tom Kepple⁵, and Wendy M. Murray^{1,2,3,4}

¹Department of Biomedical Engineering, Northwestern University, Evanston, IL, USA

²SMPP, Rehabilitation Institute of Chicago, Chicago, IL, USA

³Departments of PM&R and PTHMS, Northwestern University, Chicago, IL, USA

⁴Edward Hines Jr. VA Hospital, Hines, IL, USA

⁵C-Motion, Inc., Germantown, MD, USA

Abstract

Baseball pitching imposes a dangerous valgus load on the elbow that puts the joint at severe risk for injury. The goal of this study was to develop a musculoskeletal modeling approach to enable evaluation of muscle-tendon contributions to mitigating elbow injury risk in pitching. We implemented a forward dynamic simulation framework that used a scaled biomechanical model to reproduce a pitching motion recorded from a high school pitcher. The medial elbow muscles generated substantial, protective, varus elbow moments in our simulations. For our subject, the triceps generated large varus moments at the time of peak valgus loading; varus moments generated by the flexor digitorum superficialis were larger, but occurred later in the motion. Increasing muscle-tendon force output, either by augmenting parameters associated with strength and power or by increasing activation levels, decreased the load on the ulnar collateral ligament. Published methods have not previously quantified the biomechanics of elbow muscles during pitching. This simulation study represents a critical advancement in the study of baseball pitching and highlights the utility of simulation techniques in the study of this difficult problem.

Key Terms

ulnar collateral ligament; musculoskeletal model; valgus torque; dynamic simulation

3. Introduction

Failure of the ulnar collateral ligament (UCL) in the elbow is a common injury among baseball pitchers.³ Damage to this ligament generally requires surgery and more than 10 months of recovery and rehabilitation.¹⁷ Severe injuries to the osseous articulation of the

Corresponding Author: Wendy M. Murray, Contact Address: Sensory Motor Performance Program Suite 1408B, Rehabilitation Institute of Chicago, 345 E Superior Street, Chicago, IL 60611, USA, Contact Telephone: 312-238-6965, Contact Fax: 312-238-2208, w-murray@northwestern.edu.

Conflicts of Interest

None.

elbow are also prevalent in the youth baseball-playing population and require extensive rehabilitation before a return to competition is possible.²² The high risk of elbow injury is a result of the valgus moment placed on the elbow joint by the baseball pitching motion.¹¹ During the pitching motion, the body forcibly rotates the upper arm toward home plate while the inertias of the forearm, hand, and ball resist the rotation, creating an elbow valgus moment. The reported valgus load in professional pitchers is greater than the maximum voluntary isometric moment an average young-adult male can generate in elbow flexion.^{3,19}

A valgus moment at the elbow is countered by a varus moment that is primarily distributed among the muscle-tendon units crossing the medial side of the joint, the osseous articulation, and the UCL.^{6,13} For the UCL to produce a varus moment it must support an elevated level of tension relative to its typical state when the elbow is unloaded in varus-valgus and this tension puts the ligament in danger of failing.²⁸ Research in cadavers has shown that tension in the UCL can be reduced by increasing tension in the medial elbow muscles.^{24,37,40} Examinations of elbow physiology *in vitro* have also revealed that muscle contraction generates joint compression.^{9,13} Under a valgus load, the moment generated by the osseous articulation is a function of the compression force between the radius and humerus.^{9,13} Increased compression can be dangerous, especially for young pitchers with immature skeletons, because it has been shown to result in bone impairments such as Panner's disease and osteochondritis dissecans in over 6% of the youth pitcher population.²²

Despite the potentially pivotal roles of the medial elbow muscles in regulating both UCL and bone compression injury risk, to our knowledge, the individual force and moment contributions of these muscles during baseball pitching have never been quantified. Importantly, an analysis of expert pitchers using electromyography revealed that several elbow muscles approach 100% activation during the phases of the pitching motion when the largest joint accelerations occur (the arm cocking and acceleration phases).¹⁴ Cadaver studies have allowed researchers to generate hypotheses about the underlying muscle forces and moments during these phases,^{24,37,40} but such *in vitro* studies are unable to replicate the complex biomechanics of the task. Even in living subjects, it is currently impossible to measure individual muscle forces or moments directly during pitching because of the invasiveness of the necessary equipment (e.g. Kursa et al. 2005).²³ Consequently, muscle-driven dynamic simulations of the pitching motion are required to fully understand individual muscle contributions and injury implications.

Most existing dynamic simulations of pitching only include inverse analyses that quantify net elbow joint moments from kinematic data recorded by optical tracking protocols.^{3,11,12,32} Traditional inverse dynamic analyses do not account for the dynamics associated with muscle activation and contraction. Furthermore, calculating only net elbow joint moments does not yield conclusive information about injury risk. For example, it has been shown at the knee that the loading levels on different structures can vary considerably under the same net moment due to variability in muscle-tendon contributions.²⁵ To fully understand the risk of injury to specific elbow structures, it is necessary to improve our understanding of elbow muscle function in the complex, dynamic pitching motion.

The goal of this study was to compute the elbow varus counter moments supported by each of the medial muscle-tendon units, the osseous articulation moment, and the load on the UCL during a baseball pitch. Knowledge of the muscles that have the greatest varus moment-generating capacities, as well as the resulting osseous and UCL loading, can inform interventions in pitchers that target specific muscles for training to improve performance and reduce the risks of severe injuries. To achieve our goal, we developed a forward dynamic simulation framework that used a scaled, whole-body, musculoskeletal model to recreate a pitching motion recorded in a high school pitcher. Within this framework, we examined the extent to which the medial elbow muscle-tendon actuators were capable of generating varus moments and regulating injury risk during the complex baseball pitching task.

4. Materials and Methods

To compute varus muscle-tendon moments, joint compression force, and the UCL load at the elbow during a baseball pitch, we implemented a forward dynamic simulation framework (Figure 1) that reproduced joint kinematics and ground reaction forces measured from a single high school baseball pitcher throwing a fastball. The inputs to this framework were experimental kinematics and ground reaction forces recorded simultaneously during a pitch, as well as a generic, whole body, musculoskeletal model with a defined set of muscle-tendon and torque actuators. The details of the experimental data, model, and simulation methods are provided in more detail below.

Experimental Data

The experimental data analyzed in this study were collected in a separate study by Alderink, Kepple et al. (2008).² As they describe, experimental motion analysis data were recorded from six male high school baseball pitchers as they each threw a fastball using a 7-camera Vicon 612 motion capture system and three AMTI force platforms, sampled at 250 Hz and 1000 Hz, respectively. None of the six subjects had a history of arm injury and each could throw at least 80 mph (35.8 m/s) under game conditions. From the six available data sets, we randomly chose a single, exemplary set for analysis in this study. The subject of the exemplary data set was right handed with a height of 1.85 m and a mass of 81.8 kg.

Musculoskeletal Model

The generic, whole-body model implemented in this study included 12 segments and 30 DOFs (Figure 2), and was designed to represent the anthropometry of a 50th percentile young-adult male (height = 1.77 m, mass = 75 kg). The model was defined by updating the right arm of a validated, open-source, whole-body model of a 50th percentile male that was originally designed to simulate lower-extremity tasks.¹⁶ Only the right arm was updated because the subject we analyzed was right-handed and our biomechanical analysis is focused on the elbow joint. To update the whole-body model, the three-dimensional surface geometry, mass, and inertial properties of the ulna, radius, eight carpal bones, five metacarpal bones, and 14 phalanges were redefined using data files available from a validated, open-source, dynamic model of the upper limb that is also representative of a 50th percentile male.³⁵ In addition, a baseball was added to the model by defining a static homogeneous transformation between the modeled right middle metacarpal and a spherical

mass, such that the hand and ball were coincident. The baseball had dimensions specified by the official major league baseball rules¹ and the inertia of a solid sphere.

Overall, the model of the right upper limb used in this study included six kinematic DOF: three at the shoulder (adduction, flexion, and rotation), two at the elbow (flexion and varus), and one at the forearm (pronation). The kinematic description of the right shoulder joint was the same as the open-source model for lower-extremity tasks,¹⁶ but the order of the three shoulder rotations were updated to match typical descriptions of shoulder kinematics in pitching literature (adduction, flexion, and then rotation).¹¹ Joint kinematic constraints for elbow flexion and forearm pronation were identical to the open-source upper-limb model³⁵ and were defined by humeroulnar hinge and radioulnar hinge joints, respectively. A varus-valgus DOF (Figure 3) was added to the elbow by defining an additional humeroulnar hinge joint at the lateral edge of the proximal surface of the ulna.⁶ The direction of the varus-valgus axis of rotation was defined to be the cross-product of the forearm pronation axis and the elbow flexion axis. Kinematic degrees of freedom for the wrist were eliminated, with the position of the wrist fixed at the mean flexion and radial deviation angles calculated during the arm cocking and acceleration phases of the subject's motion. The hand was fixed in a neutral grip posture as described in the open-source upper-limb model.³⁵ From examination of the raw motion data, it was determined that wrist and hand motion during this phase of the pitching motion was minor and these simplifications were made to substantially increase our computation efficiency.

Actuator Set

Fourteen muscle-tendon actuators capable of generating elbow varus moments were included in the biomechanical model (Table 1).⁶ The 3-D coordinates of muscle origins and insertions, as well as intermediate muscle points and the geometries of muscle wrapping were defined from the open-source upper-limb model.³⁵ Because there is evidence that all four compartments of flexor digitorum superficialis (FDS) extend to the humerus via a common, fibrous connection,⁵ we moved the origins and extended the tendon slack lengths for the actuators representing the index and middle finger FDS compartments such that all four originated on the medial epicondyle of the distal humerus (Table 2). Our simulations implemented the dynamic, Hill-type, muscle-tendon model described by Millard et al. (2013),²⁹ available in OpenSim 3.1, to calculate muscle forces from muscle excitations. In addition to the muscle-tendon actuators, ideal torque actuators were added at all 30 DOFs of the generic model.

Pre-processing of model and experimental data

Within our simulation framework (Figure 1), three pre-processing steps were completed to produce both a scaled model and a set of joint angles representing the recorded motion. These three steps were designed to scale the model to match the height and mass of the subject (scaling), compute joint angles from the marker trajectories (inverse kinematics), and adjust the mass distribution of the scaled model such that both the model and the joint angles were dynamically consistent with the recorded ground reaction forces (residual reduction).

Scaling—Using a tool provided in OpenSim 3.1, the generic model was scaled to match the height and mass of the subject. The masses of all segments were scaled by a single factor such that the mass proportions in the generic model were maintained and the total mass of all segments in the model was equal to the measured mass of the subject. The dimensions of the model segments were scaled using static pose data, available from the experimental data set. Virtual markers were placed on the generic model with the same marker placements used in the subject, and then the segment dimensions were scaled so that the distances between the virtual markers matched the distances between the experimental markers.

Inverse kinematics—The scaled musculoskeletal model and the experimental marker trajectories describing the pitching motion were used to compute the joint angles that allowed the model to best reproduce the recorded motion, via an inverse kinematics algorithm in Visual3D.²⁷ The output joint angles were low-pass filtered with a cutoff frequency of 20 Hz. Because there is less than five degrees of laxity about the elbow varus-valgus axis,³³ we assumed the varus-valgus joint angle was zero throughout the motion.

Residual Reduction—To ensure that center of mass accelerations of the scaled model were dynamically consistent with the ground reaction forces in a forward dynamic simulation of the recorded motion, a residual reduction algorithm (RRA)^{8,34,42} was completed in OpenSim 3.1. RRA first computed the excess “residual” forces necessary to be applied to the center of mass to achieve dynamic consistency between the ground reaction forces and the accelerations specified by the joint angles from inverse kinematics. Based on user-defined weights for the joint angles, the algorithm made small adjustments to the mass distribution of the scaled model and the joint angles to minimize the necessary residuals. When DOFs of the model are assigned different weights from each other, the algorithm limits the relative adjustment allowed for the DOF with the higher weight. Of the 30 DOFs in the model, the right shoulder and elbow DOFs were assigned the highest weights and the DOFs of the pelvis and lower extremities were assigned the lowest. Upon completion of the RRA process for the subject, the residual forces necessary for dynamic consistency were less than 2% of the subject’s body weight, which is comparable to results for dynamic simulations of running.¹⁵ Residual forces of small magnitude indicate that the predicted center of mass accelerations are consistent with the applied ground reaction forces.

Forward Dynamic Simulation and Sensitivity Study

After pre-processing, the resulting joint angles and scaled model with adjusted mass proportions were used to calculate muscle forces and moments in multiple forward dynamic simulations (Figure 1). From the results of each forward simulation, the elbow joint compression force and the torque load on the UCL³⁰ were computed in two successive post-processing steps. Four simulations of the same motion were completed because documentation of elbow muscle properties and coordination patterns in pitchers is limited, as are precise estimates of activity levels of individual muscles. Thus, we evaluated the sensitivity of the loading of critical elbow structures to the contribution of muscles across a wide range of potential muscle performance levels. In total, we completed four forward dynamic simulations and associated post-processing analyses of the pitching motion,

varying both the muscle-tendon force-generating parameters in the model's actuator set and the muscle excitation constraints.

Sensitivity study variable: Actuator Set—Across the four forward dynamic simulations we completed in our sensitivity study, three different versions of the model's actuator set were implemented. First, we implemented a standard set of muscle-tendon force-generating parameters for the 14 Hill-type actuators as previously published in the open-source upper-limb model.³⁵ The parameters describing muscle architecture were reflective of a 50th percentile young-adult male and based on anatomical and strength data collected in living subjects^{19,20}. The parameters describing the dimensionless muscle and tendon force development curves (that are ultimately scaled by the architectural parameters) and muscle activation dynamics were defined by Millard et al. (2013).²⁹ Because finger and wrist muscle-tendon units are known to have stiffer tendons than typical muscles, we increased the tendon stiffness and strain at maximum isometric force for FCR, FCU, and the four compartments of FDS from typical values for tendons to reflect existing cadaveric data.^{26,41} Next, we defined a second actuator set in which the force-generating parameters for the 14 Hill-type actuators were adapted. The second actuator set addressed the possibility that the muscles in an average 50th percentile male may be weaker and less effective at generating force at high velocities than an experienced baseball pitcher. Through a literature search, we identified and modified specific muscle model parameters associated with both speed and magnitude of force generation that may be different in baseball pitchers compared to average males. In the second actuator set, we: (i) further increased the tendon stiffness at maximum isometric muscle force by 70%;⁴ (ii) decreased the activation time constant to 0.005 s;³⁶ (iii) increased the maximum isometric muscle force by 29%;²¹ and (iv) increased the maximum eccentric force-velocity multiplier from 1.4 to 1.8.⁴³ Both the parameters chosen to be altered and the process we used are comparable to those implemented in a previous approach to modeling musculoskeletal adaptations that occur in aging adults,³⁸ Finally, because current inverse dynamic analyses of pitching do not take into account the forces developed by muscles when evaluating the effect of the net elbow valgus load on injury risk,³ we implemented a third actuator set in which we removed the 14 Hill-type actuators from the biomechanical model. In this case, only the ideal torque actuators produced the movement at all 30 DOFs.

Sensitivity study variable: Excitation Constraints—Three of the four forward simulations we completed were performed using the three previously described actuator sets, with muscle excitations constrained to stay within experimentally-measured levels observed in expert pitchers.¹⁴ Torque control signals were unconstrained. We performed the fourth simulation with the standard actuator set again included, but excitations for all 14 muscle-tendon actuators were constrained to be 100% throughout the simulated motion.

CMC-Driven Forward Dynamics—For each given set of actuators and excitation constraints, the forward dynamic simulation was produced using an algorithm known as computed muscle control (CMC)³⁹, as implemented in OpenSim 3.1. Because the highest risk for elbow injury is known to occur during the arm cocking and acceleration phases of the standard pitching motion, which is the period from leading foot plant to ball release¹¹,

this is the only aspect of the pitching motion we simulated using CMC. We also completed an inverse dynamic analysis of the full pitching motion to verify that the largest net valgus moments did occur during the arm cocking and acceleration phases.

To calculate actuator control signals that produced each forward simulations of the motion, the outputs from the pre-processing steps (e.g., the scaled model and joint angles, see Figure 1), the ground reaction forces, the actuator set of interest, and the desired excitation constraints were input into the CMC tool. The equations of motion for the whole-body model were generated by CMC in each time step of the motion (Equation 1).

$$\ddot{\mathbf{q}} = [\mathbf{m}(\mathbf{q})]^{-1} [\boldsymbol{\tau}^M + \boldsymbol{\tau}^T + \mathbf{C}(\mathbf{q}, \dot{\mathbf{q}}) + \mathbf{G}(\mathbf{q}) + \boldsymbol{\tau}^F] \quad (1)$$

$$\boldsymbol{\tau}^M = \mathbf{R}(\mathbf{q}) \mathbf{F}^M(\mathbf{u})$$

$$\boldsymbol{\tau}^T = \mathbf{f}(\mathbf{v})$$

$\mathbf{q}$, $\dot{\mathbf{q}}$, and $\ddot{\mathbf{q}}$ are calculated from the joint angle inputs and are each a 30×1 vector, representing the acceleration, velocity, and position, respectively, of the model's 30 generalized coordinates in the current time step. Both the 30×30 mass matrix, $\mathbf{m}(\mathbf{q})$, and the 30×1 gravity vector, $\mathbf{G}(\mathbf{q})$, depend on the coordinate positions in the current time step. The Coriolis and centrifugal torques are defined in a 30×1 vector, $\mathbf{C}(\mathbf{q}, \dot{\mathbf{q}})$, which depends on both position and velocity. The torques caused by the ground reaction forces are represented by the 30×1 vector $\boldsymbol{\tau}^F$.

In the CMC tool, the vector of moments applied to the model by the 14 muscles ($\boldsymbol{\tau}^M$) and the torques applied by the ideal torque actuators ($\boldsymbol{\tau}^T$) in each time step are initially unknowns. The muscle moments are functions of their current moment arms, $\mathbf{R}(\mathbf{q})$, and muscle forces, $\mathbf{F}^M(\mathbf{u})$. At each time step, CMC solves a static optimization problem that calculates the control signals for the 14 muscle-tendon actuators ($\mathbf{u}$) and the 30 idealized torque actuators ($\mathbf{v}$) which, when transformed to muscle forces and ideal torques, generate the desired joint accelerations. The computed control signals in each time step minimize the error between the modeled joint accelerations and the desired joint accelerations specified by the input joint angles. The CMC tool integrates the equations of motion using a 5th-order Runge-Kutta-Feldberg integrator.

Muscle control signals, in the form of muscle excitations, were transformed to muscle activations and then forces via the activation and contraction dynamics described by Millard et al. (2014).²⁹ Torque actuator control signals were transformed to joint torques by multiplying the control signals by user-defined maximum torques.

At the elbow joint, the static optimization resolved redundancy between the 14 Hill-type and ideal torque actuators at the three elbow DOFs by minimizing the sum of the square of the 14 muscle and torque actuator control signals.⁷ In the optimization, each muscle-tendon and ideal torque actuator was weighted by the maximum moment or torque it could produce. The maximum moment for a given muscle was defined by biomechanical factors that influence the muscle's moment generating potential at that time step: moment arm, optimal isometric force, fiber length, fiber velocity, and activation.²⁹ The maximum torques for the ideal torque actuators at the elbow DOFs were defined via a trial-and-error approach. Ultimately,

we specified the minimum torques possible that also enabled completion of the simulation. At joints with only torque actuators, CMC simply calculated the torque control signals that produced the desired joint acceleration at each time step.

Post-Processing: Computation of the Elbow Compression Force and Torque Load on UCL³⁰

Upon completion of the forward simulations, each of the four output data sets (including model states, forces and moments over the time, t , of the simulated motion) was used to compute the resultant joint compression force profile in the elbow, and then to determine the consequent moment necessary to be supported by the UCL and associated ligamentous structures.

Joint Compression Force—Using the joint reaction analysis tool in OpenSim 3.1, the reaction force and moment vectors necessary to maintain the kinematic constraints imposed by the varus-valgus joint definition were calculated for each simulation over time. The force and moment balances were completed at the varus-valgus center of rotation. The component of the resultant force vector parallel to the longitudinal axis of the radius and perpendicular to the varus-valgus axis at each time step was defined to be the elbow joint compression force ($F_{jc}(t)$).

Torque Load on the UCL and Associated Ligamentous Structures—For each simulation in the sensitivity study, we used the output data from the CMC-driven forward dynamic simulation and the associated joint reaction analysis to evaluate a varus-valgus moment balance at the elbow and determine the consequent moment necessary to be supported by the elbow ligamentous structures (referred to as the UCL load). This was done in MATLAB (MathWorks; Natick, MA).

Each varus-valgus moment balance was constructed by solving a system of two equations: the first defined by the output of the CMC-driven simulation and the second defined by elbow varus-valgus musculoskeletal anatomy. From the forward dynamic simulation results, we specified that the sum of the individual Hill-type muscle-tendon moments (M_{Hill}) and the supplemental torque (T) from the ideal varus-valgus actuator were equal and opposite to the net moment imposed by the subject's motion (M_{valgus}) over time. That is,

$$M_{valgus}(t) = \sum_{i=1}^{14} M_{Hill}^i(t) + T(t) \quad (2)$$

where $M_{Hill}^i(t)$ was computed by multiplying a given muscle's force output at that time step by its varus moment arm. In the simulation without Hill-type actuators, the valgus load was only countered by torque actuators (i.e. $M_{Hill}^i(t=0)$ and $M_{valgus}(t) = T(t)$ for all t).

Next, from musculoskeletal anatomy, we assumed that, under valgus loading, the counter varus moment was distributed among the UCL and associated ligamentous structures (M_{UCL}), the osseous articulation (M_{art}), and the 14 individual muscle-tendon units (M_{mt} ; Equation 3).³⁰

$$M_{valgus}(t) = M_{UCL}(t) + M_{art}(t) + \sum_{i=1}^{14} M_{mt}^i(t) \quad (3)$$

An offset from the joint center to the center of pressure for the contact between the humerus and the forearm transforms the joint contact force ($F_{jc}(t)$) to the osseous articulation moment (Equation 4).

$$M_{art}(t) = F_{jc}(t) * d_{art} \quad (4)$$

It has been shown that the center of pressure in the elbow under a valgus load is located near the center of the contact area between the radius and the humerus.⁹ In our simulations, the offset distance (d_{art}) from the varus-valgus joint center to the center of pressure was defined to be half the length of the vector from the joint center to the lateral edge of the radius, perpendicular to the varus-valgus axis of rotation. For each time step in which the pitching motion imposed a valgus load at the elbow joint, we solved Equations (2) and (3), to calculate the moment supported by the UCL (assuming $M_{mt}^i = M_{Hill}^i$). Thus,

$$M_{UCL}(t) = T(t) - M_{art}(t) \quad (5)$$

Under varus loading ($M_{valgus} < 0$), the moment supported by the UCL was defined to be zero.

5. Results

When we examined the elbow varus-valgus load imposed by the pitching task over all phases of the motion using an inverse dynamic analysis, we found that the subject's motion imposed the largest valgus moments during the arm cocking and acceleration phases of the motion (Figure 4). During these phases, the maximum valgus load approached 115 Nm. In contrast, the valgus load imposed by the motion did not exceed 37 Nm from the beginning of the windup through the end of the stride phase. The overall length of the full pitching motion from the start of the windup to release was 6.376 s. The period from arm cocking through acceleration lasted 0.192 s.

In our simulation framework, the standard Hill-type muscle-tendon actuators (designed to represent a 50th percentile young adult male) made substantial contributions to the total varus moment that countered the valgus load imposed by the simulated pitching motion. In response to the imposed load, all 14 of these Hill-type actuators were activated (Figure 5). Except for a 0.04 s period when the actuators representing the triceps exerted the largest varus moment, the sum of the moments generated by the four compartments of FDS was the largest contribution to the total muscle-tendon varus moment (Figure 6). The time period when the triceps produced the greatest varus moment corresponded to peak valgus loading. The flexor-pronator muscles (FDS, PT, FCR, and FCU) and brachialis did not generate their peak moments until a short time after the peak valgus load.

In repeated simulations, modified actuator sets and excitation patterns substantially increased the muscle-tendon contributions to the elbow varus moment computed during the simulated pitching motion (Figure 7). For example, when we simulated the motion with the 14 standard Hill-type actuators constrained to be fully activated throughout the cocking and acceleration phase of the pitching motion, the total muscle-tendon varus moment at the time of the peak valgus load was increased by 53% relative to the moment produced when excitations were prescribed to stay within the bounds observed experimentally. When muscle excitations were prescribed to stay within the experimentally reported bounds but the standard actuators were replaced by augmented actuators designed to address the possibility that muscles in baseball pitchers generate more power than the 50th percentile male, the muscle-tendon varus moment at the time of peak valgus loading was increased by 64% at the same time point.

In all cases, regardless of the included excitation constraints or force-generating parameters, the muscle-tendon actuators did not generate an elbow varus moment that was sufficient to completely balance the imposed valgus load (Figure 7). Because of this, a supplemental torque was required from the ideal torque actuator acting about the varus-valgus degree of freedom in order to simulate the recorded pitching motion and run to completion. In each simulation analyzed here, the combined contributions from the 30 ideal torque actuators (including the varus-valgus torque actuator) and the 14 muscle-tendon actuators resulted in simulated motions that tracked the experimental data within 0.2 degrees (RMS Error; Figure 8).

Across simulations, we observed a considerable tradeoff between the joint compression force and the load on the UCL. As muscle-tendon actuator output increased, our simulation framework predicted that the joint compression force increased and the UCL load decreased (Figure 9). Comparing the three simulations in which muscle excitations were not constrained to be 100%, the simulation including the adapted force-generating parameters produced the largest joint compression force and the smallest UCL load. Conversely, the compression force was the smallest (close to 0 N) and the UCL was predicted to support the entire valgus load in the simulation with only torque actuators. Furthermore, for the last 0.02 s of this torque-actuated simulation, a tensile force was calculated rather than the compression force seen using the muscle-actuated models.

6. Discussion

The purpose of this study was to compute the muscle-tendon contributions to the elbow varus moment, the elbow joint compression force, and the consequent UCL load during a baseball pitching motion. We achieved this by developing a forward dynamic simulation framework that allowed us to generate multiple simulations which reproduced joint kinematics and ground reaction forces measured in a high school pitcher. Using this framework, we simulated the measured pitching motion during the phases of the motion associated with high elbow injury risk and confirmed that it corresponded to peak valgus loading (Figure 4). We then repeated the simulation using different sets of actuators and excitation constraints to evaluate the potential for muscle-tendon units to mitigate elbow injury risk in pitching, assuming a broad range of muscle-tendon performance. As expected,

we found that increasing the outputs of the muscle-tendon units both increased the load on the osseous articulation (i.e. increased the joint compression force) and decreased the load on the UCL (Figure 9). Without muscle-tendon actuation, the magnitude of the external load supported by the UCL was computed to be 330% of its reported failure load of 35 Nm,²⁸ highlighting the vital role of muscle-tendon contributions to reducing UCL injury risk.

Regardless of the force-generating parameters and excitation constraints implemented in our sensitivity study, the Hill-type muscle-tendon actuators directly contributed an elbow varus moment that was at least 35% of the total varus moment necessary at the time of the peak valgus load imposed by the pitching motion (Figure 7). For both the standard and adapted actuator sets when excitations were constrained to remain within experimentally observed limits, the three compartments of the triceps made the largest contribution to the net varus moment at this critical time point (Figure 6). Joint kinematics at the elbow at the time of peak loading and musculoskeletal geometry explain this result. Specifically, at this point in the motion, the subject's elbow was flexing (Figure 8) and therefore the triceps compartments were contracting eccentrically, compared to the flexor-pronator muscles, which were contracting concentrically. Muscles are able to generate larger forces during eccentric contractions.⁴³ Similarly, our results are consistent with the flexor-pronator muscles shortening at high speeds at the time of peak valgus loading, limiting muscle output.

Immediately following peak valgus loading, when the subject was nearing ball release in his motion, our simulations predicted that the four compartments of FDS generated the largest varus moments while the triceps contributions were diminished (Figure 6). After the time of peak loading, the subject stopped flexing at his elbow and began extending (Figure 8). Therefore, during this time period, the flexor-pronator muscles were primarily contracting eccentrically while the triceps compartments were contracting concentrically. Of the flexor-pronator muscles in the forearm, FDS generated the largest varus moments after peak valgus loading for two reasons. First, the combined volume of the four FDS compartments is greater than the volume of each of the other flexor-pronator muscles examined in this study.²⁰ Second, FDS originates on the medial epicondyle of the humerus⁵ which allowed it to operate with a large varus moment arm throughout the subject's pitching motion. It is important to note that we assumed that all four compartments of FDS originate on the humerus and can generate varus moments. This assumption requires further examination and validation. Additionally, although the palmaris longus muscle is capable of generating a varus moment, it was excluded because it is reported to be absent in 15–20% of the population.¹⁰ As illustrated in this manuscript, our forward dynamic simulation framework enables the ability to evaluate and analyze a range of assumptions regarding muscle paths and force-generating abilities. The results of our sensitivity study highlight the critical role of muscles in mitigating injury risk. As such, there is an urgent need for more detailed studies of the quantitative anatomy of the baseball pitching population.

The relative muscle-tendon tendon moments we calculated broadly agree with experimental results in cadavers. A cadaveric study examining the effect of muscle-tendon tension on UCL strain reported that FCR, FCU, and FDS are likely to be important elbow stabilizers during dynamic tasks.²⁴ Our simulation results are consistent with this conclusion, as each

of these muscle-tendon units were activated and made a substantial contribution to the total varus moment (Figure 5). Similarly, a study in cadavers completed in 2009 revealed that combined tension in the biceps, triceps, and brachialis reduced varus-valgus laxity, indicating that these muscles are also important elbow stabilizers.³⁷ In our simulations, this result was reinforced (Figure 6). The total moments we computed in our dynamic simulations also agree with existing inverse dynamic analyses. For example, the maximum valgus load we observed (Figure 4) is similar in magnitude and timing to the approximate 100 Nm peak loading level reported by Anz et al. (2010)³ for pitchers at risk for elbow injury.

To our knowledge, we have presented the first quantifications of individual muscle-tendon varus moments during the pitching motion. We accomplished this by implementing a forward dynamic simulation framework. This is in contrast to the current literature, in which dynamic simulations of pitching are generally limited to inverse analyses that only calculate the net elbow varus-valgus loads at specific instances in the motion.^{3,11,12,31,32} In addition, none of these previous studies have examined muscle-tendon contributions to the net varus-valgus moments. Hirashima and Ohtsuki (2008) proposed an alternative forward dynamic approach in which the pitching motion was simulated in the upper extremity by applying net muscle torques at each joint.¹⁸ Importantly, this approach still lacked a mechanism for examining individual muscle moments. Our simulation framework reveals which muscles have the greatest moment-generating capacities during pitching, and therefore can inform potential interventions in pitchers that target specific muscles for training to improve performance.

An important limitation to this study was the absence of lateral elbow muscles and palmaris longus in our simulations. Any moments generated by lateral muscle-tendon units during the simulated motion would have exacerbated the valgus loading imposed by the pitching motion. However, the increased joint compression caused by lateral elbow muscle contraction would have also subsequently increased the varus osseous articulation moment. Thus, it is unclear whether the inclusion of lateral elbow muscles would have increased or decreased the predicted load on the UCL. Our simulations represented an idealized circumstance for the medial elbow muscles, in which there is no possibility that the medial muscle contributions to the varus moment would be diminished by valgus moments generated by the antagonistic lateral muscles.

Another limitation was our fixation of the joints in the hand and wrist. Though motions at these joints during the examined phases of the pitching motion were small in the raw kinematic data, allowing motion in these joints could influence the simulated activation patterns in the muscles. A third limitation to our modeling approach was our assumption that the line of action of the joint compression force was through the center of the contact area between the radius and humerus. The true line of action may have a different direction. Error in this line of action could affect the magnitudes of the osseous articulation and UCL loads but would likely not change the effects of increasing muscle output on these loads. A fourth limitation was our assumption of no varus-valgus deviation during the pitching motion. Accounting for varus-valgus deviation could change the absolute magnitudes of the elbow varus-valgus moments. However, we do not expect it would substantially impact the relative

magnitudes of the muscle moments because relative moment arm sizes and muscle-tendon velocities should remain largely unchanged. Further study of these limitations is warranted and our simulation framework will enable such evaluation.

Overall, we have developed a unique dynamic simulation framework that represents a critical advancement in the study of baseball pitching biomechanics. Our simulation framework allows, for the first time, the quantification of elbow injury risk accounting for the actions of individual muscles. Our results demonstrate that it is essential to consider active muscle-tendon contributions when attempting to reduce the risks for severe elbow injuries in baseball pitchers.

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References

1. Official Baseball Rules. [Accessed March, 27, 2014] 2014. http://mlb.mlb.com/mlb/downloads/y2013/official_baseball_rules.pdf
2. Alderink, A.; Kepple, T.; Siegel, KL.; Razzook, A.; Stanhope, S. Sources of Forward Ball Velocity in a Pitched Baseball. Paper presented at: 32nd Annual Meeting of the American Society of Biomechanics; 2008; Ann-Arbor, MI.
3. Anz AW, Bushnell BD, Griffin LP, Noonan TJ, Torry MR, Hawkins RJ. Correlation of Torque and Elbow Injury in Professional Baseball Pitchers. *Am J Sports Med.* 2010; 38:1368–1374. [PubMed: 20400752]
4. Arampatzis A, Karamanidis K, Morey-Klapsing G, De Monte G, Stafilidis S. Mechanical properties of the triceps surae tendon and aponeurosis in relation to intensity of sport activity. *J Biomech.* 2007; 40:1946–1952. [PubMed: 17101142]
5. Bogumill GP. Functional anatomy of the flexor tendon system of the hand. *Hand Surgery.* 2002; 7:33–46. [PubMed: 12365048]
6. Buchanan TS, Delp SL, Solbeck JA. Muscular resistance to varus and valgus loads at the elbow. *J Biomech Eng-T Asme.* 1998; 120:634–639.
7. Crowninshield RD, Brand RA. A Physiologically Based Criterion of Muscle Force Prediction in Locomotion. *J Biomech.* 1981; 14:793–801. [PubMed: 7334039]
8. Delp SL, Anderson FC, Arnold AS, et al. OpenSim: open-source software to create and analyze dynamic simulations of movement. *IEEE Trans Biomed Eng.* 2007; 54:1940–1950. [PubMed: 18018689]
9. Duggan JP, Osadebe UC, Alexander JW, Noble PC, Lintner DM. The impact of ulnar collateral ligament tear and reconstruction on contact pressures in the lateral compartment of the elbow. *J Shoulder Elb Surg.* 2011; 20:226–233.
10. Eric M, Krivokuca D, Savovic S, Leknan I, Vucinic N. Prevalence of the palmaris longus through clinical evaluation. *Surg Radiol Anat.* 2010; 32:357–361. [PubMed: 19816651]
11. Fleisig GS, Andrews JR, Dillman CJ, Escamilla RF. Kinetics of Baseball Pitching with Implications About Injury Mechanisms. *Am J Sports Med.* 1995; 23:233–239. [PubMed: 7778711]
12. Fleisig GS, Kingsley DS, Loftice JW, et al. Kinetic comparison among the fastball, curveball, change-up, and slider in collegiate baseball pitchers. *Am J Sports Med.* 2006; 34:423–430. [PubMed: 16260466]
13. Fornalski S, Gupta R, Lee TQ. Anatomy and biomechanics of the elbow joint. *Sports Med Arthrosc.* 2003; 11:1–9.
14. Hamilton CD, Glousman RE, Jobe FW, Brault J, Pink M, Perry J. Dynamic stability of the elbow: electromyographic analysis of the flexor pronator group and the extensor group in pitchers with valgus instability. *J Shoulder Elb Surg.* 1996; 5:347–354.

15. Hamner SR, Delp SL. Muscle contributions to fore-aft and vertical body mass center accelerations over a range of running speeds. *J Biomech.* 2013; 46:780–787. [PubMed: 23246045]
16. Hamner SR, Seth A, Delp SL. Muscle contributions to propulsion and support during running. *J Biomech.* 2010; 43:2709–2716. [PubMed: 20691972]
17. Hariri S, Safran MR. Ulnar Collateral Ligament Injury in the Overhead Athlete. *Clin Sports Med.* 2010; 29:619–644. [PubMed: 20883901]
18. Hirashima M, Ohtsuki T. Exploring the mechanism of skilled overarm throwing. *Exerc Sport Sci Rev.* 2008; 36:205–211. [PubMed: 18815490]
19. Holzbaur KRS, Delp SL, Gold GE, Murray WM. Moment-generating capacity of upper limb muscles in healthy adults. *J Biomech.* 2007; 40:2442–2449. [PubMed: 17250841]
20. Holzbaur KRS, Murray WM, Gold GE, Delp SL. Upper limb muscle volumes in adult subjects. *J Biomech.* 2007; 40:742–749. [PubMed: 17241636]
21. Hortobagyi T, Katch FI. Eccentric and Concentric Torque-Velocity Relationships during Arm Flexion and Extension - Influence of Strength Level. *Eur J Appl Physiol O.* 1990; 60:395–401.
22. Kobayashi K, Burton KJ, Rodner C, Smith B, Caputo AE. Lateral compression injuries in the pediatric elbow: Panner's disease and osteochondritis dissecans of the capitellum. *J Am Acad Orthop Sur.* 2004; 12:246–254.
23. Kursk K, Diao E, Lattanza L, Rempel D. In vivo forces generated by finger flexor muscles do not depend on the rate of fingertip loading during an isometric task. *J Biomech.* 2005; 38:2288–2293. [PubMed: 16154416]
24. Lin F, Kohli N, Perlmutter S, Lim D, Nuber GW, Makhosous M. Muscle contribution to elbow joint valgus stability. *J Shoulder Elb Surg.* 2007; 16:795–802.
25. Lipps DB, Oh YK, Ashton-Miller JA, Wojtys EM. Effect of Increased Quadriceps Tensile Stiffness on Peak Anterior Cruciate Ligament Strain during a Simulated Pivot Landing. *J Orthopaed Res.* 2014; 32:423–430.
26. Loren GJ, Lieber RL. Tendon Biomechanical Properties Enhance Human Wrist Muscle Specialization. *J Biomech.* 1995; 28:791–799. [PubMed: 7657677]
27. Lu TW, O'Connor JJ. Bone position estimation from skin marker co-ordinates using global optimisation with joint constraints. *J Biomech.* 1999; 32:129–134. [PubMed: 10052917]
28. McGraw MA, Kremchek TE, Hooks TR, Papangelou C. Biomechanical evaluation of the docking plus ulnar collateral ligament reconstruction technique compared with the docking technique. *Am J Sports Med.* 2013; 41:313–320. [PubMed: 23193145]
29. Millard M, Uchida T, Seth A, Delp SL. Flexing Computational Muscle: Modeling and Simulation of Musculotendon Dynamics. *J Biomech Eng-T Asme.* 2013; 135
30. Morrey BF, An KN. Articular and Ligamentous Contributions to the Stability of the Elbow Joint. *Am J Sports Med.* 1983; 11:315–319. [PubMed: 6638246]
31. Nissen CW, Solomito M, Garibay E, Ounpuu S, Westwell M. A biomechanical comparison of pitching from a mound versus flat ground in adolescent baseball pitchers. *Sports health.* 2013; 5:530–536. [PubMed: 24427428]
32. Nissen CW, Westwell M, Ounpuu S, Patel M, Solomito M, Tate J. A Biomechanical Comparison of the Fastball and Curveball in Adolescent Baseball Pitchers. *Am J Sports Med.* 2009; 37:1492–1498. [PubMed: 19448049]
33. Pomianowski S, O'Driscoll SW, Neale PG, Park MJ, Morrey BF, An KN. The effect of forearm rotation on laxity and stability of the elbow. *Clin Biomech.* 2001; 16:401–407.
34. Remy CD, Thelen DG. Optimal estimation of dynamically consistent kinematics and kinetics for forward dynamic simulation of gait. *J Biomech Eng.* 2009; 131:031005. [PubMed: 19154064]
35. Saul KR, Hu X, Goehler CM, et al. Benchmarking of dynamic simulation predictions in two software platforms using an upper limb musculoskeletal model. *Comput Methods Biomech Biomed Engin.* 2014:1–14.
36. Scoville CY, Ronsky JL. Sensitivity of a Hill-based muscle model to perturbations in model parameters. *J Biomech.* 2006; 39:2055–2063. [PubMed: 16084520]

37. Seiber K, Gupta R, McGarry MH, Safran MR, Lee TQ. The role of the elbow musculature, forearm rotation, and elbow flexion in elbow stability: an in vitro study. *J Shoulder Elb Surg.* 2009; 18:260–268.
38. Thelen DG. Adjustment of muscle mechanics model parameters to simulate dynamic contractions in older adults. *J Biomech Eng-T Asme.* 2003; 125:70–77.
39. Thelen DG, Anderson FC, Delp SL. Generating dynamic simulations of movement using computed muscle control. *J Biomech.* 2003; 36:321–328. [PubMed: 12594980]
40. Udall JH, Fitzpatrick MJ, McGarry MH, Leba TB, Lee TQ. Effects of flexor-pronator muscle loading on valgus stability of the elbow with an intact, stretched, and resected medial ulnar collateral ligament. *J Shoulder Elb Surg.* 2009; 18:773–778.
41. Ward SR, Loren GJ, Lundberg S, Lieber RL. High stiffness of human digital flexor tendons is suited for precise finger positional control. *J Neurophysiol.* 2006; 96:2815–2818. [PubMed: 16870841]
42. Xiang Y, Arora JS, Abdel-Malek K. Optimization-based prediction of asymmetric human gait. *J Biomech.* 2011; 44:683–693. [PubMed: 21092968]
43. Zajac FE. Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control. *Crit Rev Biomed Eng.* 1989; 17:359–411. [PubMed: 2676342]

Figure 1.
A flow chart depicting the simulation framework and sensitivity study.

Figure 2.

The whole-body musculoskeletal model. It was created by adding segments from the right arm of an existing upper-extremity model to an existing model of the whole body designed for simulating lower-extremity tasks. The green sphere represents a baseball and is welded to the right hand of the model.

Figure 3.
The varus-valgus axis of rotation. The orange arrow indicates the direction of valgus moments and the green arrow indicates the direction of varus moments.

Figure 4.

The net moment about the elbow varus-valgus axis across the four phases of the pitching motion from the windup to ball release. A positive moment means the motion is imposing a valgus load, and the model is generating a varus moment to counter it. The blue moment profile from 0 s to 6.184 s has been compressed to emphasize the moment profile from 6.184 s to 6.376 s. The two diagonal black dashes along the horizontal axis depict the transition between these two time periods.

Figure 5.

The muscle activation profiles of the standard Hill-type actuators over the course of the simulated arm cocking and acceleration phase of the pitching motion. The activations for the biceps actuators were excluded because the biceps generated a negligible varus moment.

Figure 6.

The varus moments generated by individual muscle-tendon actuators during the simulated arm cocking and acceleration phase of the pitching motion.

Figure 7.

The total varus moment generated by the standard Hill-type actuators with activations allowed to vary (solid black line), with activations constrained to be 100% (perforated black line), and by the adapted Hill-type actuators with activations allowed to vary (dotted black line). The solid red line represents the magnitude of the imposed valgus load.

Figure 8.

Joint angles of the elbow and shoulder during the arm cocking and acceleration phase of the pitching motion. The elbow angles include flexion, pronation, and varus-valgus and the shoulder angles include adduction, flexion, and rotation. The solid green lines represent the experimental joint angles calculated from marker trajectories, while the thick dotted black lines represent the simulated joint angles produced by computed muscle control. The thin, vertical, dotted lines indicate the time of peak valgus loading. In this figure, the results are shown for the simulation completed using the standard Hill-type muscle-tendon actuators designed to represent an average young adult male.

Figure 9.

The predicted joint compression force (left plot) and load on the UCL (right plot) for the simulations with no muscles (dotted lines), standard muscle-tendon actuators (solid lines), and adapted muscle-tendon actuators (dashed lines). The UCL load for the adapted muscle-tendon actuators is not visible because in this simulation the UCL was predicted to be unloaded.

Table 1

The medial elbow muscles

Muscle Groups*	Muscle-Tendon Actuators
Anconeus (Anc)	Anconeus
Biceps (Bic)	Biceps Long Head
	Biceps Short Head
Brachialis (Bra)	Brachialis
Flexor Carpi Radialis (FCR)	Flexor Carpi Radialis
Flexor Carpi Ulnaris (FCU)	Flexor Carpi Ulnaris
Flexor Digitorum Superficialis (FDS)	Index Flexor Digitorum Superficialis
	Middle Flexor Digitorum Superficialis
	Ring Flexor Digitorum Superficialis
	Little Flexor Digitorum Superficialis
Pronator Teres (PT)	Pronator Teres
Triceps (Tri)	Triceps Lateral Head
	Triceps Long Head
	Triceps Medial Head

* The medial elbow muscles with varus moment arms that were represented by Hill-type muscle-tendon actuators in the forward dynamic simulations.

Table 2

The modifications to the paths of the FDS compartments

FDS compartment	Updated coordinates of origin on humerus (x, y, z) *	Original tendon slack length (m)	Updated tendon slack length (m)
Index	(0.00421, -0.27598, -0.03864)	0.275	0.321
Middle	(0.00421, -0.27598, -0.03864)	0.295	0.340
Ring	(0.00479, -0.2788, -0.03731)	0.328	0.328
Little	(0.00421, -0.27598, -0.03864)	0.339	0.339

* Presented in the coordinate frame of the humerus from the open-source lower-limb model¹⁶