

# Predictors of Proximal Tibia Anterior Shear Force during a Vertical Stop-Jump

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**ABSTRACT:** Anterior cruciate ligament (ACL) continues to be a significant medical issue for athletes participating in sports and recreational activities. Biomechanical analyses have determined that anterior shear force is the most direct loading mechanism of the ACL and a probable component of noncontact ACL injury. The purpose of this study was to examine the biomechanical predictors of proximal tibia anterior shear force during a stop-jump task. A biomechanical and electromyographic (EMG) analysis of the knee was conducted while subjects performed a vertical stop-jump task. The task was chosen to simulate an athletic maneuver that included a landing with a sharp deceleration and a change in direction. The final regression model indicated that posterior ground reaction force, external knee flexion moment, knee flexion angle, integrated EMG activity of the vastus lateralis, and sex (female) would significantly predict proximal tibia anterior shear force ( $p < 0.0001$ ,  $R^2 = 0.8609$ ). Knee flexion moment had the greatest influence on proximal tibia anterior shear force. The mathematical relationships elucidated in the current study support previous clinical and basic science research examining noncontact ACL injuries. This data provides important evidence for clinicians who are examining the risk factors for these injuries and developing/validating training programs to reduce the incidence of injury. © 2007 Orthopaedic Research Society. Published by Wiley Periodicals, Inc. J Orthop Res

**Keywords:** ACL; knee; shear force; biomechanics; injury

## INTRODUCTION

Anterior cruciate ligament (ACL) injuries continue to be a significant health concern for young individuals attempting to lead an active, healthy lifestyle.<sup>1-4</sup> Each year, 1 in 1000 individuals between the ages of 15 and 25 will suffer an ACL injury<sup>5</sup> with over 50,000 reconstructive surgeries performed annually.<sup>6</sup> The majority of these injuries occur during participation in sports and recreational activities,<sup>7-10</sup> and are the result of a noncontact mechanism of injury.<sup>7,8,10</sup> Noncontact ACL injury prevention is particularly important to female athletes, as epidemiological research has demonstrated that females are at a significantly higher risk for suffering this injury.<sup>11-14</sup> Injury prevention training programs have been designed to modify the potential risk factors and reduce noncontact ACL injuries<sup>15-20</sup> by attempting to induce neuromus-

cular and biomechanical adaptations that may decrease knee joint loading and ACL strain.

One of the joint forces that can increase ACL strain and lead to ligament rupture is proximal tibia anterior shear force. Although the loading pattern of the knee during noncontact ACL injuries is most likely multidirectional and multiplanar,<sup>8,10,21</sup> proximal tibia anterior shear force is a probable component given that it represents the most direct loading mechanism of the ACL.<sup>22-25</sup> Currently, the *in vivo* biomechanical characteristics that predict an increased proximal tibia anterior shear force are unclear. Measurable *in vivo* biomechanical characteristics that may predict proximal tibia anterior shear force include ground reaction forces, knee joint kinematics, joint resultant moments estimated through inverse dynamics procedures, and myoelectrical activity of the knee musculature measured through surface electromyography (EMG). One study has examined the relationship among knee joint kinematics, knee joint kinetics, and ground reaction forces,<sup>26</sup> and demonstrated that greater ground reaction forces and knee extension moments correlate with greater proximal tibia anterior shear force. The

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current study proposes to examine similar variables with the addition of EMG.

Kinematic observations of the mechanism of injury, kinematic analysis of individuals at risk for noncontact ACL injury, and ACL strain studies have shown that certain movement patterns and joint positions place an individual at greater risk for injury. For example, female athletes participating in high-risk sports (for ACL injury) who land with an increased dynamic knee valgus are at greater risk for injury,<sup>27</sup> which supports previous research that many noncontact ACL injuries occur during landings with the knee in a valgus position.<sup>10,21</sup> Noncontact ACL injuries also typically occur when individuals land with decreased knee flexion.<sup>8,10</sup> This landing position increases ACL strain compared to larger flexion angles.<sup>23–25,28,29</sup> In addition, the majority of evidence indicates that females, who are at greater risk for ACL injury, perform dynamic sports tasks with both increased knee valgus angles<sup>30–34</sup> and decreased knee flexion angles.<sup>30–32,34–36</sup>

Knee joint resultant moments, as estimated through inverse dynamics, can provide valuable insight into the loading patterns of the knee, especially when combined with EMG data. In the sagittal plane, a net external knee flexion moment typically exists throughout the majority of the stance phase of a stop-jump task and represents a net internal quadriceps moment<sup>34,37</sup> or internal quadriceps requirement. *In situ* and *in vivo* ACL strain increases<sup>38,39</sup> under this condition (quadriceps loading) with greater increases observed at low flexion angles.<sup>38</sup> In the frontal plane, knee joint moment (valgus or varus) can increase ACL strain when combined with a proximal anterior tibial force.<sup>40,41</sup> Valgus moment, as estimated through inverse dynamics, has also been implicated as a predictor of ACL injury in female athletes.<sup>27</sup> Hewett et al.<sup>27</sup> demonstrated that female athletes who perform jump-landing tasks with a greater knee valgus moment are more likely to suffer an injury than those who perform the same task with less valgus moment.

The identification of neuromuscular and biomechanical characteristics that can predict dangerous loading patterns may provide important evidence that support the use of proximal tibia anterior shear force and other biomechanical variables for future prospective studies and development of injury prevention programs. The purposes of this study was to determine if a select group of neuromuscular and biomechanical characteristics are able to significantly predict proximal tibia anterior shear force. Those characteristics included knee

flexion angle, knee valgus angle, external knee flexion moment, external knee valgus moment, integrated EMG (IEMG) of the vastus lateralis and semitendinosus, and sex. We hypothesized that an equation based on these variables would be able to significantly predict proximal tibia anterior shear force.

## MATERIALS AND METHODS

### Subjects

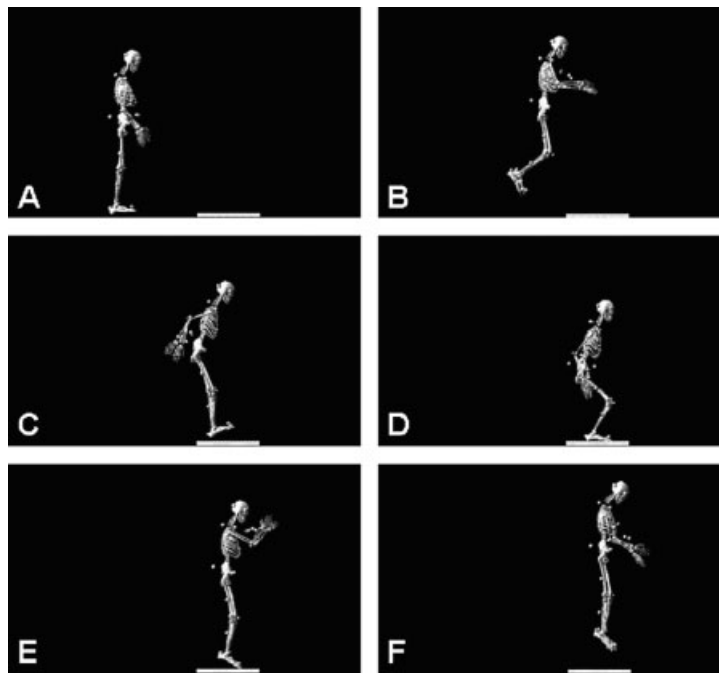
Thirty-six healthy high school basketball players (19 males, 17 females) participated. All subjects were currently participating in organized basketball at least three times per week at the time of testing. Subject demographics are presented in Table 1. Subjects were excluded from the study if they had a history of serious musculoskeletal injury, any musculoskeletal injury within the past 6 months, or suffer from any disorder that interfered with sensory input, musculoskeletal function, or motor function. All subjects provided written informed consent in accordance with the University's Institutional Review Board prior to participation.

### Data Collection and Reduction

Anthropometric measurements were recorded for each subject. They included height and weight, segmental lengths, and circumferences of the thighs and shanks, diameters of the ankles and knees, feet length and width, lateral malleoli height, and pelvic width. The stop-jump task was then demonstrated to each of the subjects. The technique for the stop-jump task consisted of the following: (1) an initial starting point measured as 40% of the subject's height from the edge of the force plates, (2) a two-legged broad jump with a two-legged landing on the force plates (one foot on each plate), and (3) immediate jump for maximum vertical height (Fig. 1). To promote natural performance of the task, subjects were provided the following instructions: (1) begin each jump at the designated starting point, (2) land with one foot on each force plate, and (3) then immediately jump off the force plates for maximum height. All subjects were allowed to practice

**Table 1.** Descriptive Data (Means and Standard Deviations) for All of the Subjects, the Male Subjects, and the Female Subjects

| Variable       | Group             |                   |                     |
|----------------|-------------------|-------------------|---------------------|
|                | Total<br>(n = 36) | Males<br>(n = 19) | Females<br>(n = 17) |
| Age (years)    | 16.1 ± 1.3        | 16.3 ± 1.5        | 15.9 ± 1.1          |
| Body mass (kg) | 68.2 ± 10.4       | 72.1 ± 9.4        | 63.8 ± 10.0         |
| Height (m)     | 1.75 ± 0.09       | 1.80 ± 0.08       | 1.70 ± 0.07         |



**Figure 1.** Vertical stop-jump task: (A) start, (B) approach to force plates, (C) initial contact, (D) peak knee flexion, (E) end contact, (F) apex of vertical jump.

the jump until they were comfortable with the task (approximately three to five trials).

After demonstration and practice of the vertical stop-jump task, subjects were prepped for EMG analysis. Surface EMG activity was collected bilaterally on the vastus lateralis (VL) and semitendinosus (ST). Surface electrodes were placed over the appropriate muscle belly in line with the direction of the fibers with an interelectrode distance of approximately 20 mm. Electrode sites were shaved, abraded, and cleaned with isopropyl alcohol to reduce impedance. Electrode placement sites were based on Delagi et al.<sup>42</sup> A single ground electrode was placed over the anterior aspect of the tibia just distal and medial to the tibial tuberosity. All electrode sites were located via palpation of each subject's anatomy and were confirmed following application of electrodes through visual inspection of signals on the oscilloscope during standardized manual muscle testing.<sup>43</sup> Surface EMG signals were collected at 1200 Hz via an eight channel telemetric system (Noraxon USA Inc., Scottsdale, AZ). Electromyographic signals were recorded using silver-silver chloride, pregelled bipolar surface electrodes (Medicotest, Inc., Rolling Meadows, IL).

Electromyographic data during a 5-s maximum voluntary isometric contraction (MVIC) were collected for the knee flexors and extensors utilizing the Biodex System 3 Multi-Joint Testing and Rehabilitation System (Biodex Medical Inc., Shirley, NY). This data were processed and used for normalization of the corresponding muscle's EMG activity during the dynamic task. Subjects were seated in the chair and secured with straps around the torso, pelvis, and thigh of the leg performing

the MVIC. The axis of the dynamometer was positioned so it was aligned with the axis of rotation of the knee being tested, which was positioned in 60° of flexion. The order of MVIC data collection was the same for each subject (knee extensor data collected first).

Subjects were prepped for the biomechanical analysis of the stop-jump task. A total of 15 retroreflective markers were utilized for data collection of three dimensional (3D) coordinate data during the vertical stop-jump task. The marker system used was based on Kadaba et al.,<sup>44</sup> as developed at the Helen Hayes Hospital in New York. Retroreflective markers were placed bilaterally over the second metatarsal head, posterior aspect of the heel, lateral malleolus, femoral epicondyle, anterior superior iliac spine, and the L5–S1 disc space. The markers were secured to the subject with double-sided tape. Four other markers were attached to wands and secured bilaterally with straps, prewrap, and athletic tape to the lateral aspect of the subject's thigh and shank. Careful attention was paid to marker placement and attachment as to not interfere with the EMG electrodes.

Three dimensional coordinate data were collected and calculated using a 3D optical capture system (Vicon, Centennial, CO). This motion analysis system included six high-speed (120 Hz) optical cameras (Pulnix Industrial Product Division, Sunnyvale, CA) instrumented and synchronized using Peak Motus software (version 7.2, Vicon). Ground reaction force data during the jump tasks were collected at 1200 Hz utilizing two force plates (Kistler Corporation, Worthington, OH) that were flush with the surrounding surface of a custom-built flooring

system. Following the retroreflective marker setup, subjects were allowed to practice the stop-jumps a second time (approximately three to five trials). Subjects performed a total of five jumps with at least 30 s of rest between each jump. The first three successful jumps were utilized for data processing. A successful jump was defined as a jump that began at the proper starting point with a two-legged landing with one foot on each force plate followed by a vertical jump.

Raw analog data from the force plates were used to calculate the ground reaction force data for each jump trial and were filtered using a fourth-order Butterworth filter (zero phase shift) at a cutoff frequency of 100 Hz. The raw coordinate data were also filtered with a fourth-order Butterworth filter (zero phase shift) with an optimized cutoff frequency (typically 5 Hz).<sup>45</sup> Raw analog data from the force plates were used to calculate the ground reaction force data for each jump trial. All joint kinematic and kinetic calculations were performed in the Kinematic module of the Peak Motus software package (Vicon, Centennial, Englewood, CO). Joint kinematic calculations were based on Vaughan et al.<sup>46</sup> An inverse dynamics procedure was used to calculate the joint resultant moments and forces and is briefly described here.

Resultant joint forces and moments were calculated based on body segment parameters (measured and estimated),<sup>46–48</sup> linear kinematics, centers of gravity, angular kinematics, and ground reaction forces based on Greenwood.<sup>49</sup> Joint resultant forces were calculated based on the acceleration and mass of each segment that is determined by first calculating the change in linear momentum. Joint resultant moments were calculated in a similar manner. They were calculated by first determining the rate of change in angular momentum, which was based on the moments of inertia, segmental angular velocities, and segmental angular accelerations. These calculations were first performed distally then through an inverse dynamics procedure that was calculated proximally through the kinetic chain. The joint resultant moment and forces calculated using this procedure were the estimated external moments and forces and were based on the ground reaction forces and segment inertial forces.<sup>50</sup> In addition, the proximal tibia anterior shear force includes all the soft tissue forces and joint contact forces at the knee, and does represent the shear force transmitted to the ACL or the shear force applied by the patellar tendon. Joint resultant forces were normalized to body weight and joint resultant moments were normalized to body weight\*height.<sup>50,51</sup>

Joint kinematic data, joint kinetic data, and ground reaction force data were exported to Matlab (Release 12, The MathWorks, Natick, MA) for identification of the variables of interest. The ground reaction force data were used to calculate the maximum posterior ground reaction force (maximum deceleration force) during the initial stance phase of the stop-jump tasks. This point was then identified in the joint kinetic and kinematic data to determine proximal tibia anterior shear force, knee flexion/extension moment, knee flexion/extension angle,

and the knee valgus/varus angle at the point of maximum deceleration. Data were averaged across three trials.

Raw analog data from the MVIC, synchronized raw analog data from the stop-jump trials, and the ground reaction force data from the stop-jump trials were imported into Matlab for data processing. The mean value of each MVIC was used for normalization of the EMG during the stop-jump trials.<sup>52</sup> Both the MVIC and trial EMG data were processed with a linear envelope prior to normalization using a Butterworth filter (fourth-order, zero-phase shift, cutoff frequency of 12 Hz). The point of peak posterior ground reaction force (maximum deceleration of the body) was identified in each jump trial using the ground reaction force data. From this reference point, the IEMG was calculated for each muscle for the 150 ms prior to maximum deceleration of the body. Data for each EMG variable was averaged across the same stop-jump trials used in the kinematic and kinetic analysis.

## Data Analysis

A stepwise multiple regression model were fit using Stata (Stata 8; Stata Corporation, College Station, TX) to determine which neuromuscular and biomechanical variables significantly predict proximal tibia anterior shear force at the time of maximum deceleration (peak posterior ground reaction force). The predictor variables included knee flexion angle, knee valgus angle, knee flexion moment, knee valgus moment, IEMG of the vastus lateralis and semitendinosus, and sex. The response variable was proximal tibia anterior shear force at the time of maximum deceleration. Pairwise correlations were also performed to further examine the relationships between the biomechanical predictor variables and the response variable. Finally, the normalized beta coefficients for the predictor variables were estimated to assess the relative predictive power of each of the predictor variables. An alpha level of 0.05 was selected to determine if predictor variables would be included in the final equation, for determining the significance of the model in predicting the response variable, and for determining if the pairwise correlations were significant.

## RESULTS

The means and standard deviations for each of the variables are listed in Table 2. The multiple linear regression model is presented in Table 3. Based on this model five of the predictor variables were maintained in the final equation. Those variables were peak posterior ground reaction force, knee flexion/extension moment, knee flexion angle, IEMG activity of the VL, and sex. This model accounts for 86.1% of the variance in the proximal tibia anterior shear force during the vertical stop-jump task ( $p < 0.001$ ). For the individual predictor variables, the coefficients reveal that the greater

**Table 2.** Biomechanical and Neuromuscular Data (Means  $\pm$  Standard Deviations) for the Entire Group, Male Subjects, and Female Subjects

| Variable   | Group              |                    |                      |
|--|--------------------|--------------------|----------------------|
|  | Total ( $n = 36$ ) | Males ( $n = 19$ ) | Females ( $n = 17$ ) |
| Peak posterior ground reaction force (body weight)         | $-0.77 \pm 0.25$   | $-0.83 \pm 0.28$   | $-0.71 \pm 0.19$     |
| Proximal anterior tibia shear force (body weight) at PPGRF | $0.29 \pm 0.22$    | $0.23 \pm 0.18$    | $0.36 \pm 0.25$      |
| Knee flexion moment (body weight * height) at PPGRF        | $-0.043 \pm 0.052$ | $-0.030 \pm 0.055$ | $-0.056 \pm 0.044$   |
| Knee flexion angle (degrees) at PPGRF                      | $29.0 \pm 8.5$     | $29.1 \pm 7.7$     | $28.8 \pm 9.5$       |
| Knee valgus angle (degrees) at PPGRF                       | $0.8 \pm 5.7$      | $1.9 \pm 5.6$      | $0.3 \pm 5.8$        |
| Knee valgus moment (body weight * height) at PPGRF         | $-0.084 \pm 0.062$ | $-0.068 \pm 0.044$ | $-0.101 \pm 0.074$   |
| IEMG activity of the VL (%MVIC*s) Prior to PPGRF (150 ms)  | $0.084 \pm 0.062$  | $0.068 \pm 0.044$  | $0.101 \pm 0.062$    |
| IEMG activity of the MH (%MVIC*s) Prior to PPGRF (150 ms)  | $0.127 \pm 0.225$  | $0.115 \pm 0.271$  | $0.140 \pm 0.161$    |

PPGRF, peak posterior ground reaction force; IEMG, integrated electromyographic; VL, vastus lateralis; MH, semitendinosus.

the peak posterior ground reaction force, knee flexion/extension moment, knee flexion angle, IEMG activity of the VL, and being female would predict higher proximal tibia anterior shear forces. The pairwise correlations between the response variable and the biomechanical predictor variables are listed in Table 4. Proximal tibia anterior shear force was significantly correlated with peak posterior ground reaction force, knee flexion moment, knee flexion angle, knee valgus angle, and the IEMG activity of the VL. Of those variables, only knee flexion moment had a strong correlation.<sup>53</sup> The normalized beta coefficients for the regression model are listed in Table 5. Based on the beta coefficients, knee flexion/extension moment would have the most dramatic effect on proximal tibia anterior shear force. A one standard

deviation increase in knee flexion/extension moment would cause a predicted increase of 0.77 standard deviations in the proximal tibia anterior shear force.

## DISCUSSION

The purpose of this study was to conduct a biomechanical and neuromuscular analysis of males and females performing a stop-jump task and determine what characteristics are able to predict proximal tibia anterior shear force. We hypothesized that an equation based on knee flexion angle, knee valgus angle, knee flexion moment, knee valgus moment, IEMG of the vastus lateralis and semitendinosus, and sex would be able to significantly predict proximal tibia anterior

**Table 3.** Multiple Linear Regression Model Predicting Proximal Tibia Anterior Shear Force

| Multiple Linear Regression Model     |             |        |        |                |              |
|--------------------------------------|-------------|--------|--------|----------------|--------------|
| Source                               | SS          | df     | MS     | Observations   | 72           |
| Model                                | 3.6828      | 5      | 0.7366 | F(5,66)        | 86.68        |
| Residual                             | 0.5952      | 66     | 0.0090 | Prob > F       | $p < 0.0001$ |
| Total                                | 4.2780      | 71     | 0.0603 | $R^2$          | 0.8609       |
|                                      |             |        |        | Adjusted $R^2$ | 0.8503       |
| Predictor Variables                  | Coefficient | t      |        | p-value        |              |
| Peak posterior ground reaction force | -0.2760     | -3.94  |        | 0.000          |              |
| Knee flexion/extension moment        | -3.9683     | -13.15 |        | 0.000          |              |
| Knee flexion angle                   | 0.0034      | 2.00   |        | 0.050          |              |
| IEMG activity of the VL              | 0.5179      | 2.62   |        | 0.011          |              |
| Sex                                  | 0.0593      | 2.40   |        | 0.019          |              |
| Constant                             | -0.2421     | -3.05  |        | 0.003          |              |

This model with the predictor variables peak posterior ground reaction force, knee flexion/extension moment, knee flexion angle, IEMG activity of the VL, and sex accounted for 86.1% of the variance of the response variable, proximal tibia anterior shear force. The associated  $p$ -value for this model is  $p < 0.0001$ . SS, sum of the squares; df, degrees of freedom; MS, mean squares; IEMG, integrated electromyographic; VL, vastus lateralis.

**Table 4.** Pairwise Correlations between the Response Variable (Proximal Tibia Anterior Shear Force) and the Predictor Variables

| Variable  | Correlation Coefficient | <i>p</i> -value  |
|---|-------------------------|------------------|
| Peak posterior ground reaction force (body weight)        | -0.2360                 | 0.046            |
| Knee flexion moment (body weight * height) at PPGRF       | -0.8986                 | <i>p</i> < 0.001 |
| Knee flexion angle (degrees) at PPGRF                     | 0.4318                  | <i>p</i> < 0.001 |
| Knee valgus angle (degrees) at PPGRF                      | 0.2551                  | 0.031            |
| Knee valgus moment (body weight * height) at PPGRF        | -0.1628                 | 0.172            |
| IEMG activity of the VL (%MVIC*s) Prior to PPGRF (150 ms) | 0.2531                  | 0.032            |
| IEMG activity of the MH (%MVIC*s) Prior to PPGRF (150 ms) | 0.0186                  | 0.877            |
| Sex   | 0.3225                  | 0.006            |

PPGRF, peak posterior ground reaction force; IEMG, integrated electromyographic; VL, vastus lateralis; MH, semitendinosus.

shear force. Our hypothesis was partially supported as the multiple linear regression model indicated that peak posterior ground reaction force, knee flexion/extension moment, knee flexion angle, IEMG activity of the VL, and sex (female) significantly predicted proximal tibia anterior shear force. The results of our analysis have implications for future research related to the examination of risk factors for noncontact ACL injuries and the development of training programs to reduce the incidence of injury.

We chose to investigate proximal tibia anterior shear force and its biomechanical predictors because it is the most direct loading mechanism of the ACL,<sup>22-25</sup> and it can be estimated through inverse dynamics. It is important to note that proximal tibia anterior shear force, as calculated in this study, is a resultant force that includes all of the soft tissues joint contact forces acting at the knee and it does not represent a shear force transmitted to the ACL or the shear force applied by the patellar tendon. Given these limitations, Yu et al.<sup>54</sup> described how proximal tibia anterior shear force (estimated through inverse dynamics) may be an indicator of ACL loading. Their mathematical analysis and simulation indicated that an increase in proximal tibia anterior shear force will increase the knee anterior drawer force, which should

positively correlate to ACL forces. Currently, only a few studies have estimated proximal tibia anterior shear force during a dynamic task.<sup>26,34,37,55-58</sup> This force has been implicated as a potential risk factor for noncontact ACL injury due to the demonstrated differences observed between males and females, with females performing the dynamic sports tasks with significantly greater proximal tibia anterior shear force.<sup>26,34,37</sup>

The inclusion of peak posterior ground reaction force in the final model supports previous analyses of the noncontact mechanism of injury,<sup>8,10,21</sup> which revealed that ACL injuries occur during different sports maneuvers, but characteristic among them is a sharp deceleration of the body, which is represented by a posteriorly directed ground reaction force. In our study, peak posterior ground reaction force occurred  $0.028 \pm 0.19$  s after initial foot contact. This was prior to peak vertical ground reaction force ( $0.062 \pm 0.041$  s after initial contact) and peak knee flexion angle ( $0.172 \pm 0.040$  s). The regression equation indicated that proximal tibia anterior shear force would increase as the posterior ground reaction force increased. Yu et al.<sup>26</sup> also demonstrated this relationship during a similar stop-jump task. The relationship between posterior ground reaction force and proximal tibia anterior shear force is a debated topic.<sup>54,59</sup> We agree with the assessment that posterior ground reaction force creates an external flexion moment at the knee, which would need to be counteracted by an internal quadriceps force.<sup>54</sup> The quadriceps would have to contract to control knee flexion, which would result in a anteriorly directed force at the proximal tibia due to the effect of the patellar ligament.<sup>54</sup>

Similar to posterior ground reaction forces, our analysis also indicated that an increase in external knee flexion moment would also predict an increase in proximal tibia anterior shear forces. The landing of the stop-jump task and subsequent spike in

**Table 5.** Normalized Beta Coefficients for the Predictor Variables

| Variable                             | Beta Coefficient |
|--------------------------------------|------------------|
| Peak posterior ground reaction force | -0.2099          |
| Knee flexion/extension moment        | -0.7658          |
| Knee flexion angle                   | 0.1177           |
| IEMG activity of the VL              | 0.1310           |
| Sex                                  | 0.1213           |

IEMG, integrated electromyographic; VL, vastus lateralis.

posterior ground reaction force creates an external knee flexion moment. The external knee flexion moment as measured through inverse dynamics equates to a net internal quadriceps moment (quadriceps force). The quadriceps force can apply a proximal tibia anterior shear force via the extensor mechanism (quadriceps tendon and patellar ligament).<sup>54</sup> Without knowledge of the muscle forces, it is difficult to determine if the increased internal quadriceps moment that predicts greater proximal tibia anterior shear force is due to an increased quadriceps force and/or a decreased hamstrings force. The results of the EMG analysis may provide some insight into the basis of these differences. In the present study, an increase in IEMG of the VL would predict greater proximal tibia anterior shear force. The IEMG of the ST was not included in the final regression equation, and based on the results of this study, does not influence proximal tibia anterior shear force as estimated through inverse dynamics. Previous biomechanical analyses of cadaveric knees have demonstrated that an increased quadriceps force will increase the amount of anterior tibial translation and proximal tibia anterior shear force.<sup>23–25,28,29</sup> The results of the current *in vivo* study support this previous *in vitro* research. In the final regression model, both a greater external knee flexion moment and IEMG of the VL would predict a greater proximal tibia anterior shear force.

The contrasting evidence between cadaveric studies and the relationship between knee flexion angle and proximal tibia anterior shear force in this study may be due to the lack of an established relationship between ACL strain, which increases at knee angles close to full extension, and proximal tibia anterior shear force during a dynamic task.<sup>23–25,28,29</sup> These individuals measured ACL strain during static positioning of cadaveric knees. It is not clear what the *in vivo* strain is during dynamic jumping and landing activities. Only one published article has measured *in vivo* strain during a similar task (one-legged jump landing).<sup>60</sup> This was a case study, and did not include calculation of proximal tibia anterior shear force. Further research is necessary to establish the relationship between knee flexion angle and proximal tibia anterior shear force.

We acknowledge that the current study has certain limitations. The accuracy of skin-based marker systems in estimating joint kinematics and joint kinetics has been questioned during gait.<sup>61–64</sup> Although careful consideration and attention was given to marker attachment, the errors due to skin movement that have been reported during gait may

be increased during the high-speed athletic tasks in this study. Although the regression analysis performed in the current study is only a mathematical analysis of the relationship of biomechanical variables estimating knee joint kinematics, resultant knee joint forces and moments, EMG activity of the knee musculature, and ground reaction forces, our model supports the reported anatomical and physiological implications for these relationships.

## CONCLUSION

The results of our analysis of biomechanical predictors of proximal tibia anterior shear force indicate that an increasing posterior ground reaction force, knee flexion moment, and IEMG of the VL would all predict an increase in proximal tibia anterior shear force and potentially an increase in ACL forces. These mathematical relationships support the previous clinical and basic science research examining the potential mechanism of noncontact ACL injury. These results provide clinicians important evidence to include these predictor variables as well as proximal tibia anterior shear force as part of future prospective studies examining risk factors for noncontact ACL injury and the validation of training studies designed to reduce injury.

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