Kinematics and Kinetics Predictor of Proximal Tibia Anterior Shear Force during Single Leg Drop Landing

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ABSTRACT

Purpose: The purpose of this study was to investigate the kinematic and kinetic variables, which predict anterior tibia shear force during single-leg landing in female athletes.

Methods: Forty-three subjects (mean and standard deviation for age 21.12 ± 2.00 y, height 168.58 ± 7.62 cm, and weight 60.27 ± 7.80 kg) participated in this study. Kinematic and kinetic variables of lower extremity and trunk during single-leg landing were collected by 5 Vicon cameras and Kistler force plate. Stepwise multiple regression and Pearson correlation were used to identify predictor variables of anterior shear force ($P \leq 0.05$).

Results: Peak of extensor moment ($P = 0.004$, $r = -0.394$) and maximum knee flexion ($P = 0.007$, $r = -0.370$) were the best predictors that explained 30% of the variance of the shear force data. Therefore, rise in maximum extensors moment of knee and knee maximum flexion causes increase and decrease in anterior shear force, respectively. In addition, a significant relationship between trunk flexion ($P = 0.039$) and knee flexion angular velocity ($P = 0.048$) at the moment of initial contact with the anterior shear force.

Conclusion: On the basis of previous research, and the relationship between clinical findings, the noncontact of anterior cruciate ligament injury during landing was confirmed. These results can be used in prospective studies examining modifiable noncontact risk factors of ACL injury.

1. Introduction

Anterior cruciate ligament (ACL) injuries are serious concerns for physically active children and adolescents [1]. Female athletes participating in jumping, cutting, and pivoting team sports such as football, basketball, and volleyball are often claimed to have a 4–6 times higher ACL risk injury compared to their male counterparts [2]. At least 70% of ACL injuries are noncontact in nature [1] . Previous descriptive studies of noncontact ACL injury mechanisms have indicated that injuries occur shortly after initial contact via a landing or deceleration motion with minimal or no contact in 70% of cases [3, 4]. Most noncontact ACL injuries occur during sport activities involving single-leg landings [5].

Single-leg landing is a common athletic maneuver performed during sports such as basketball, volleyball, soccer, and badminton [6]. In a jump landing event, the landing phase is more stressful to ACL than the takeoff phase [6]. Epidemiological research has shown female athletes to be a high risk population for ACL injury [7, 8]. Yet the literature lacks a clear and definitive consen-

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sus on female ACL injury [9]. 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 [10]. For example, the literature indicates that landing with increased knee valgus angles [3, 11], decreased knee flexion angles [3, 9], and decreased hip flexion angles [12, 13] increases ACL strain. In fact, a common movement pattern in noncontact ACL injuries includes decrease in knee flexion, hip flexion, and trunk flexion accompany with increase in knee valgus and tibia rotation [11, 14, 15]. Quadriceps contraction in 0-30 degree knee flexion produce proximal tibia anterior shear force that increases ACL strain [16, 17]. Also knee valgus and tibia rotation increase ACL strain but, this strain is lesser than shear force [18]. The majority of studies confirmed that females perform landing with high knee valgus [19], less knee flexion [20, 21], and high proximal anterior shear force [13, 22].

Many studies are performed on determining effective biomechanical factors and injurious forces and how to decrease ACL loading [19-23]. These studies only compared groups or conditions, but did not actually examine the relationships among lower extremity kinematics and kinetics. The researchers who investigated lower extremity kinematics and kinetics relationships only discussed on some of them with conflicting results. One study has examined the relationship among knee joint kinematics, knee joint kinetics, and ground reaction forces and demonstrated that greater ground reaction forces and knee extension moments correlated with greater proximal tibia anterior shear force [13].

The other one examined similar variables with the addition of EMG, and indicated that an increasing posterior ground reaction force, knee flexion moment, and IEMG of vastus lateralis would all predict an increase in proximal tibia anterior shear force [10]. In another study, although interpreted females exhibiting high knee extensor moment and knee shear force during drop landing, they could not find significant relationship among knee extensor moment and proximal knee shear force with anterior and lateral translations [24]. Thus, the purpose of this study was to determine if biomechanical variables are able to predict proximal tibia anterior shear force during one leg drop landing.

2. Materials and Methods

Forty-three female elite basketball and volleyball players with 4 years' experience in Iran national league (18-25 years old) participated in this study. Subjects were free from lower extremity injury in either leg within the past 6 months. Lower extremity injury was defined as any injury resulting in more than one day loss in physical activity or referral to a physician. Subjects would also be excluded if they had a history of surgery to the lower extremity within the past 2 years or a history of ACL surgery or presence of any lower extremity malalignment. These malalignments included hip anteversion, Q angle, tibiofemoral angle, knee recurvatum, tibial torsion, and foot pronation that were evaluated by standard clinical methods [25, 26].

Upon arrival, all subjects read and signed a consent form approved by Kharazmi University. Then, they were gotten familiar with the testing procedures. Demographic information was collected for each subject and a health questionnaire was used to assess lower extremity injury status. Kinematics and kinetics of subjects' dominant leg were collected during the first 6 days of menses to control any potential hormone effects on resulting knee joint neuromechanics (27). Three-dimensional trajectory data were obtained using a 5-camera motion analysis system (Vicon 460 Motion Capture), were sampled at 200 Hz, and digitally recorded. Furthermore, ground reaction forces were collected at 1000 Hz using a calibrated and leveled force plate (Kistler; 9286A) embedded in the floor in ergonomics laboratory of University of Social Welfare and Rehabilitation Sciences. Reflective markers were placed on anatomical landmarks according to the Kadaba marker set [28].

Anatomical landmarks were placed on C7, right and left anterior-superior iliac spines, mid-thigh, lateral femoral epicondyles, mid-shank, lateral malleoli, heel, and the second metatarsal of each subject. First, a stationary trial was taken with each subject in a neutral (standing) position to align her with the global laboratory coordinate system. Each subject's local joint coordinates was aligned to her standing position to control for inter-subject variation in anatomical alignment (i.e., zero-position valgus alignment) during the static trial. Raw marker coordinates were recorded with Workstation software. The dominant leg was defined as the leg used to kick a ball for maximum distance. Vertical GRF was used to identify the time at initial contact with the ground. Initial ground contact was defined as the instant at which the vertical ground reaction force exceeded 30 N.

Then, subjects completed a 5-minute running warm up on a treadmill at a self-selected pace, and were allowed to practice the jump. After demonstration and practice of the one leg drop landing task, subjects performed a

Table 1. Descriptive data for subjects.

Variable	Mean± SD
Age (y)	21.12 ± 2.00
Athletic experience (y)	8.19 ± 2.97
Height (cm)	168.85 ± 7.62
Weight (kg)	60.72 ± 8.70
BMI (% BW)	21.71 ± 2.37

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total 3 correct lands off a 50-cm high platform over a horizontal distance equal to 10 cm, and land with one leg on force plate with at least 30 s of rest between each land. The drop landing consisted of the subject starting on top of the box with her feet positioned 35 cm apart (distance measured between toe markers) [6].

Kinematics and kinetics data were filtered through a low-pass Butterworth digital filter at a cutoff frequency of 10 to 50 Hz, respectively [21]. A Newton-Euler inverse dynamic process was used to estimate the proximal tibia anterior shear force and joint movments in MATLAB. Forces and movments were normalized to body weight (BW %) and multiplied by height (% BW* H). Our laboratory coordinate system was based on our agreement that X, Y, Z axis showed anterioposterior, mediolateral, and vertical axis, respectively. The angles' and moments' sign were determined based on right handed rules, in a way that knee flexion angle, velocity and moment, trunk flexion, and proximal tibia anterior shear force were positive and knee valgus and knee valgus moment, knee extension, and posterior shear force were negative.

Statistical Analysis

All data analyses was performed using SPSS version 20. Kolmogorov–Smirnov test was used in order to check the normal distribution of the variables. A stepwise multiple regression model and Pearson correlation were fitted using SPSS to determine what biomechanical variables significantly predict proximal tibia anterior shear force. Statistical significance was accepted at the level of $\alpha \leq 0.05$.

3. Results

Demographic variables of 43 female basketball (23) and volleyball (20) players with 4 years experience were listed in Table 1.

Mean and standard deviation of variables is listed in Table 2.

The correlations between proximal tibia anterior shear force and independent variables are listed in Table 3. Proximal tibia anterior shear force was significantly correlated with peak knee flexion, peak knee extensor moment, trunk flexion at initial contact time, and knee flexion angular velocity.

According to the variables' sign and the agreement base on our laboratory coordinate system and right handed rules, proximal tibia anterior shear force is positive and peak knee extensor moment is negative. Thus, a negative correlation among them was expressed that in

Table 2. Means and standard deviations for biomechanical data.

Variable	Mean± SD
Peak proximal tibia anterior shear force from IC to peak knee flexion (body weight)	0.25 ± 0.13
Knee flexion at IC (degree)	11.11 ± 5.28
Knee valgus at IC (degree)	-0.07 ± 0.06
Knee flexion angular velocity at IC (degree/s)	219.38 ± 76.06
Peak knee extensor moment from IC to peak knee flexion (BW×H)	-28.01 ± 4.75
Knee flexion moment at IC (BW×H)	4.73 ± 2.73
Trunk flexion at IC (degree)	11.18 ± 1.02
Peak knee flexion (degree)	53.84 ± 10.83
Knee valgus moment at IC (BW×H)	0. 21 ± 0.01
Peak posterior ground reaction from IC to peak knee flexion (BW)	-0.43 ± 0.10

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IC= initial contact, moments were normalized to (BW×H)
The variable signs were determined based on right hand rules and according to laboratory coordination.

Table 3. Correlation between the Investigated variables and Peak proximal tibia shear force.

Variable	Peak proximal tibia shear force from initial contact to peak knee flexion	
	R	P value
Knee flexion at IC (degree)	-0.17	0.141
Knee valgus at IC (degree)	-0.19	0.105
Knee flexion angular velocity at IC (degree/s)	-0.26	0.048*
Peak knee extensor moment from IC to peak knee flexion (BW×H)	-0.39	0.004*
Knee flexion moment at IC (BW×H)	-0.23	0.073
Trunk flexion at IC (degree)	-0.27	0.039
Peak knee flexion (degree)	-0.37	0.007*
Knee valgus moment at IC (BW×H)	-0.02	0.457
Peak posterior ground reaction from IC to peak knee flexion (BW)	-0.29	0.080

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which increase in peak knee extensor movment causes an increase in proximal tibia anterior shear force.

The multiple regression model is presented in Table 4. Based on this model, two of the predictor variables were maintained in the final equation. Those variables were peak knee extensor moment (P = 0.009, Adjusted R2:0.149, F(1,41):7.55) and peak knee flexion (P = 0.001, Adjusted R2:0.278, F(2,40):8.81). First model accounts for 15.5% (R2= 0.155) of the variance in the in proximal tibia anterior shear force during one-leg drop landing. The second model with added peak knee flexion can account for 15.1% of the variance in the in the proximal tibia anterior shear force. Thus, the other variables included EMG variables and other biomechanical variables likely account for 70% of residual variance in the proximal tibia anterior shear force during performing this task.

4. Discussion

The purpose of this study was to conduct biomechanical analysis of female athletes performing a drop landing task and determine what characteristics are able to predict proximal tibia anterior shear force. Our hypothesis was partially supported as the multiple stepwise regression models indicated that peak knee extensor moment and peak knee flexion angle significantly predicted proximal tibia anterior shear force. Furthermore the results of Pearson Production Moment showed a significant relationship between knee angular velocity and trunk flexion with proximal tibia shear force.

We chose to investigate proximal tibia anterior shear force and its biomechanical predictors because it is the most direct loading mechanism of the ACL [29] and can be estimated through inverse dynamics. Yu et al. described how proximal tibia anterior shear force (estimated through inverse dynamics) may be indicative of ACL loading [13].

According to results of the regression equation, peak knee extension moment was the best predictor of proximal tibia shearing force. This result is in agreement with Yu et al. study. They indicated that there is a significant relationship between peak knee extensor moment and peak tibia shear force [13]. Shelburne et al. also reported that ACL is affected by a lot of load when shearing force is in anterior direction [30]. This result indicates that the quadriceps muscles play a significant role in the ACL loading as literature shows [31, 32]. Furthermore, knee extension moment is also an indicator of ACL loading because patella tendon force is the result of quadriceps muscle contraction and quadriceps are the major knee extension muscles [13].

In addition, the results of the present study are in agreement with C Sell et al. findings. They reported that knee flexion/extension moment would significantly predict proximal tibia anterior shear force [10]. Body acceleration in landing is really high and quadriceps eccentrically contracts to support the individual's acceleration and weight. Thus, the quadriceps force can apply a proximal tibia anterior shear force via the extensor mechanism (quadriceps tendon and patellar ligament) [10]. Furthermore, because these movments were calculated via inverse dynamics and without knowledge of the muscle

Table 4. Multiple regression model for predicting peak proximal tibia shear force.

Models			df	SS	MS	B	T	R	R^2	Adjusted R^2	F	P value	SD
1	Peak knee extensor moment from IC to peak knee flexion (body weight×height)	regression	1	0.69	0.69								
		residual	41	3.75	.092	-0.39	-2.74	0.39	0.155	0.149	7.55	0.009	0.30
		Total	42	4.45									
2	Peak knee extensor moment from IC to peak knee flexion (body weight×height)	regression	2	1.39	0.68	-0.41	-3.12	0.55	0.306	0.278	8.81	0.001	0.28
	Peak knee flexion (degree)	residual	40	3.09	0.77	-0.38	-2.94						
		Total	42	4.45									

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forces, it would be difficult to determine whether the increased internal quadriceps movment that predicts greater proximal tibia anterior shear force is due to an increased quadriceps force and or a decreased hamstrings force.

The results of this study showed that peak knee flexion angle could predict proximal tibia shear force. This result is in agreement with C Sell et.al and Yu et.al findings [10, 13]. Reduction in anterior shear forces requires recruiting the specific muscle of lower extremity and keeping more knee flexion angle [33]. Particularly, the quadriceps and gastrocnemius muscles act as antagonist of the ACL and can increase the anterior shear force [33-36]. Hamstring muscles act as agonist of the ACL and can decrease the anterior shear force [33-37]. Ability of these muscles for impact on the ACL loading can be adjusted by knee flexion [38-40]. The ability of quadriceps muscles for creating anterior shear force increases in low knee flexion angle and the ability of hamstring to neutralize this force is reduced [10, 41].

The results also showed that there is a significant negative relationship between knee flexion angular velocity and tibia shear force. This result indicates that decline in knee flexion angular velocity relates to rise in anterior tibia shear force. Furthermore the results of the previous studies showed that there is relationship between high vertical ground reaction force and low knee angular velocity at initial contact [13, 42].

The finding of this study showed a significant negative relationship between trunk flexion and tibia shear force at initial contact. Trunk flexion potentially reduces the quadriceps force requirement and subsequent load placed on the ACL immediately after ground contact, which is when ACL injury reportedly occurs. Trunk flexion during landing also produces greater knee and hip flexion compared to a more erect or trunk-extended landing posture, placing the lower extremity in a position associated with low ACL injury risk [43]. The results of this study are in agreement with Blackburn et al. and Klaus et al. [43-45] findings. Kulas et al. reported that increased hamstring activity by trunk flexion would lead to decrease in tibia shear force [45].

The relationship between peak posterior ground reaction force and tibia shear force was not statistically significant, however, the significance of this hypothesis has been expected with regard to previous literature. According to P value (P = 0.08), it could be possible that in larger sample size, this variable becomes statistically significant. Since this variable is one of the effective factors in calculating the shear force, the relationship between these variables is not unexpected [10, 13]. In other words, one reason for this discrepancy may be the nature of the task evaluated. In this study, the drop landing task has been evaluated, while in previous studies, the stop jump task has been examined [10, 13].

Also, the finding of this study showed that there are no significant relationships between knee valgus, knee valgus moment, knee flexion at initial contact and proximal tibia shear force. This results are in agreement with C Sell et al. study [10]. The only difference was knee flexion. C sell et al. reported that knee flexion at peak posterior shear force could predict proximal tibia shear force in regression analysis [10]. However, in the present study, peak knee flexion could predict shear force. This difference may be related to the skill of our subjects. In this research, we used players in national league so maybe, safe strategy of landing was instructed to them and in performance just peak extensor moment and maximum knee flexion base on imbalance muscle contraction can predict variation of anterior shear force.

In sum, the result of the present study showed that peak extensor moment and peak knee flexion in one-leg drop landing were the best predictors of tibia anterior shear force; increase in peak extensor moment and peak knee flexion decrease tibia anterior shear force. There are negative significant relationships between knee flexion angular velocity and trunk flexion at initial contact with proximal tibia anterior shear force. In addition, these results can assist to coaches and physical therapists to design and make correction injury, prevention, and exercises programs.

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