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Original article

Biomechanical characteristics of an anterior cruciate ligament injury in javelin throwing

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Abstract

Purpose: The purpose of this study was to understand the mechanism of an anterior cruciate ligament (ACL) injury in javelin throwing and javelin throwing techniques relevant to this ACL injury.

Methods: The patient in this study was an elite female javelin thrower who completed the first three trials and sustained a non-contact ACL injury on her left knee in the fourth trial of javelin throwing during a recent track and field meet. Three-dimensional kinematic data were collected in the injury and non-injury trials. The kinematic data of 52 male and 54 female elite javelin throwers were obtained from a javelin throwing biomechanical database.

Results: The patient had greater forward center of mass velocity and less vertical center of mass velocity after the first 25% of the delivery phase in the injury trial compared to non-injury trials. The patient had less left knee flexion angle and angular velocity but similar left knee valgus and internal rotation angles during the first 21% of the delivery phase in the injury trial compared to non-injury trials. The video images showed an obvious tibia anterior translation at the 30% of the delivery phase in the injury trial. The left knee flexion angle and angular velocity at the time of the left foot landing and the maximal left knee flexion angle during the delivery phase were not significantly correlated to the official distance for 52 male and 54 female elite javelin throwers.

Conclusion: The ACL injury in this study occurred during the first 30% of the delivery phase, most likely during the first 25% of the delivery phase. A stiff landing of the left leg with a small knee flexion angle was the primary contributor to this injury. Javelin throwers may have a soft left leg landing with a flexed knee, which may help them prevent ACL injuries without compromising performance.

Keywords: ACL injury; Biomechanics; Injury mechanism; Injury prevention; Risk factors

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1. Introduction

Anterior cruciate ligament (ACL) rupture is one of the most common knee injuries in sports[1] and has caused significant functional impairments to patients and financial burdens to the society.[2] The majority of ACL injuries are non-contact injuries in nature,[3–8] which indicates that the excessive loading that leads to ACL injuries is likely caused by inappropriate movement patterns. ACL injuries, therefore, may be prevented through training designed to improve movement patterns associated with ACL injury mechanisms and risk factors.[9,10] Understanding the mechanisms and risk factors of ACL injury is essential for preventing ACL injuries, because it will allow us to target the key elements associated with ACL injuries for intervention.

Tremendous efforts have been made to understand ACL injury mechanisms and risk factors for developing effective ACL injury prevention strategies in the last 2 decades.[11–14] Arguments regarding ACL injury mechanisms and risk factors, however, still exist. While evidence has suggested sagittal plane loading including increased anterior shear force and decreased knee flexion angle as the most important loading mechanism for the ACL,[11,12,14] some investigators believe that a frontal plane "valgus collapse" may be the major mechanism for ACL injury, especially in women.[6,15,16] Quantitative biomechanical analysis of ACL injury cases is an effective way to understand ACL injury mechanisms and risk factors. Collecting valid biomechanical data in ACL injury cases, however, is difficult. Several investigators have attempted to obtain kinematic data

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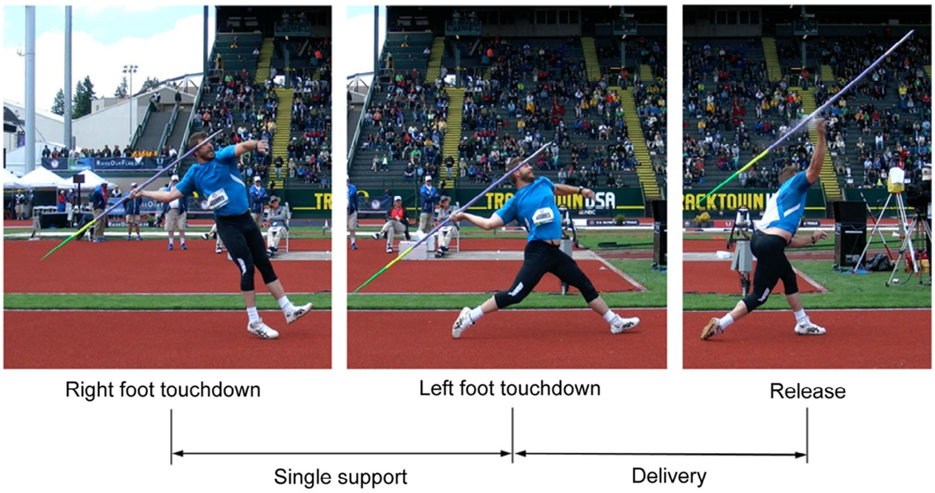

Fig. 1. Javelin throwing.

from videographic records of ACL injury cases.[3–7,17] The kinematic data collected in these studies, however, are questionable due to either the two-dimensional (2D) nature of the data or a lack of calibration of cameras when attempting to reduce three-dimensional (3D) data. Krosshaug and Bahr[18] have developed a model based manual image-matching technique to reconstruct 3D body movements from uncalibrated video cameras. This method has been applied to collect kinematic data from videographic records of ACL injury cases in football,[19] skiing,[20] team handball, and basketball.[8] However, even using multiple cameras, the minimal root-mean square errors were still 7.5° ± 12.4° for knee flexion angle, 3.9° ± 9.6° for knee valgus/varus angle, and 7.5° ± 13.9° for knee internal/external rotation angle, respectively,[18] which significantly downgraded the validity of the findings in these studies. To truly understand the biomechanical characteristics of ACL injury cases, motion data need to be collected using valid measurements with minimal errors.

An elite woman javelin thrower sustained a non-contact ACL injury during a recent track and field meet. The injury occurred well within the views of two video camcorders calibrated for the direct linear transformation (DLT) procedure[21] for quantitative video analysis. The DLT procedure has been shown as a reliable and valid measurement with high accuracy.[22,23] This injury case, therefore, provided a unique opportunity for understanding the mechanism and risk factors of ACL injury. The purposes of this study were (1) to understand the mechanism of this ACL injury through kinematic comparisons between the injury and non-injury trials of this female thrower in the same competition; and (2) to understand the relationship between javelin throwing technical factors relevant to this ACL injury and javelin throwing performance. The results of this study would provide significant information for understanding the specific mechanism and risk factors of ACL injury in javelin throwing as well as general mechanisms and risk factors of ACL injury in other sports. The results of this study would also provide information for the feasibility to modified javelin throwing techniques to prevent ACL injury without compromising performance in javelin throwing.

2. Methods

2.1. *Study design*

The current study included two components. The first component was a case study to compare selected biomechanical variables between the injured and non-injured trials of a female javelin thrower. The second component was a cross-sectional study with elite javelin throwers to determine the effect of identified risk factors on performance of javelin throwing.

2.2. *Javelin throwing*

For a right-hand thrower, javelin throwing starts with an approach run followed by a delivery strike (Fig. 1), in which the thrower releases the javelin for the longest official distance. The delivery strike can be divided to a single support phase and a delivery phase. The single support phase starts with the right foot landing and ends with the left foot landing. The delivery phase starts with the left foot landing and ends with the release of the javelin. The official distance is the distance between the nearest mark made by the javelin in the throwing sector and the front edge of the foul arch (Fig. 2).

2.3. *Subjects*

The patient in this study was an elite female javelin thrower who competed in the women's javelin throw final of a recent

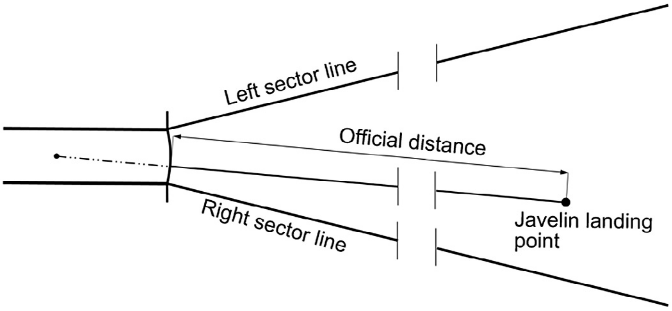

Fig. 2. Official distance in javelin throwing.

national track and field meet. She was a right-hand thrower with a standing height of 1.83 m and a body mass of 82 kg without any lower extremity injury before the ACL injury recorded in this study. During the competition, after successfully completing the first three trials, she fell to the ground after the release of the javelin in Trial 4. After the meet, she was diagnosed with an isolated ACL tear on the left knee, which was confirmed by magnetic resonance imaging. The official distances of Trials 1–4 were 59.09 m, 58.29 m, 59.79 m, and 56.55 m, respectively.

The elite javelin throwers in this study were 52 male and 54 female javelin throwers in our javelin throwing biomechanical database. These javelin throwers were right-hand throwers and competed in the 2007–2014 USA Track and Field Outdoor National Championships. The trial with the longest official distance of each thrower was used for analysis. The mean official distance of the trials used in this study was 50.61 m ranging from 42.16 to 66.67 m for female throwers, and 71.39 m ranging from 59.95 to 91.29 m for male throwers, respectively. We were not able to obtain demographic information and injury history of these athletes because we were not allowed to perform those measurements and questioning in actual competitions.

All data collections were approved by the Biomedical Internal Review Board of The University of North Carolina at Chapel Hill. Consent and permission were obtained from the USA Track & Field Association to conduct the study. A written consent form was also signed by the patient.

2.4. Data collection

All the data in this study were collected and reduced using the same procedures. Two high definition (HD) video camcorders (JVC GC-PX10; JVC, Tokyo, Japan) were set for recording javelin throwing in track and field meets for biomechanical analysis of throwing techniques. The resolution of each camcorder was 1920 × 1080 pixels. The frame rate and shutter speed were set at 59.94 frames/s and 1/1000 s, respectively, for each camcorder. The camcorders were set for the DLT procedure[21] as described in the literature.[24] One camcorder was placed at the right side of the runway with the other behind the runway (Fig. 3). The camcorders were calibrated using a calibration frame with 24 calibration points (Peak Performance; Englewood, CO, USA) placed at three locations on the runway (Fig. 3). A calibration frame of 2.5 × 2 × 2 m (long × wide × high) with 24 calibration points was placed at three locations to form a calibration volume of 5 × 2 × 2.5 m (long × wide × high) that covered the entire space in which all athletes performed the delivery as we defined. Five markers were placed at known positions on the runway to establish a global reference frame with the *x*-axis pointing toward the throwing direction of the runway, the *y*-axis pointing toward the left side of the throwing direction of the runway, and the *z*-axis pointing upwards (Fig. 3).

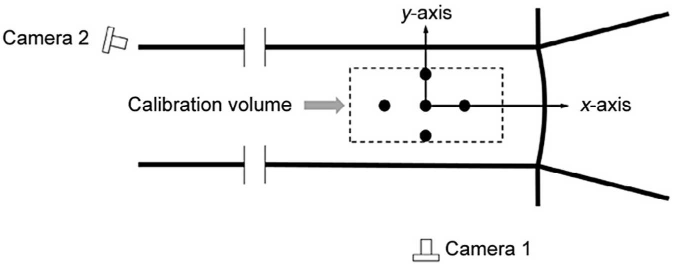

Fig. 3. Camera placements, calibration volume, marker placements (black dots) for establishing global reference frame.

2.5. Data reduction

The calibration points and global reference frame markers were manually digitized from each camcorder in the calibration trials. Twenty-one body landmarks,[25] the front edge of the javelin grip, and the tip and tail of the javelin were also manually digitized in every frame from two frames before the right foot landing to two frames after the release of the javelin for each trial from the view of each camcorder. All digitizations were performed using the Peak Motus system (Peak Performance) and a 24″ HD computer screen with a resolution of 1920 × 1200 pixels. The person who performed the manual digitizing was unaware of the ACL injury while digitizing the video images.

The two camcorders were calibrated using the digitized 2D coordinates and known 3D coordinates of calibration points with a mean calibration error less than 5 mm. The digitized 2D coordinates of body landmarks and javelin marks in each frame were corrected for the time difference due to the progressive scan of high definition images. The corrected 2D coordinates of the body landmarks and javelin marks from the two camcorders were synchronized using a critical event method described by Dapena and Chung.[26] Three critical events were visually identified for the synchronization: (1) the right foot landing defined as the first frame in which any part of the right foot was seen on the ground, (2) the left foot landing defined as the first frame in which any part of the left foot was seen on the ground, and (3) the release of the javelin defined as the first frame in which the javelin and right hand were seen separated (Fig. 1). Real-life 3D coordinates of digitized body landmarks and javelin marks were then obtained from the synchronized 2D coordinates and calibration parameters of each camcorder. The real-life 3D coordinates were then filtered using a Butterworth fourth-order zero-lag low-pass filter with a cut-off frequency of 7.14 Hz.[27]

Locations of the whole body center of mass (COM) were determined for each thrower using the segmentation method[25] and the segment inertia parameters and relative locations of the segment COM obtained from the literature.[28] COM velocities were calculated as the time derivatives of COM locations. Segment reference frames of the left thigh and left shank were established for calculating knee joint angles. All reference frames were defined following the right-hand rule. The longitudinal axes of the thigh, shank, and foot were defined as the unit vectors parallel to the line connecting the hip and knee joint centers, the line connecting knee and ankle joint centers, and the line connecting heel and toe, respectively. For the left thigh, the flexion–extension axis was defined as a unit vector perpendicular to the plane determined by the longitudinal axes of the thigh and shank. The valgus–varus axis was defined as a unit vector perpendicular to the flexion–extension and longitudinal axes of the thigh and within the plane determined by the longitudinal axes of the thigh and shank. The internal–external rotation axis was defined as a unit vector parallel to the longitudinal axis of the thigh. For the left shank, the flexion–extension axis was defined as a unit-vector perpendicular to the plane determined by the longitudinal axes of the shank and foot. The valgus–varus axis was defined as a unit-vector perpendicular to the flexion–extension and longitudinal axes of the shank and within the plane determined by the longitudinal axes of the shank and foot. The internal–external rotation axis was defined as a unit vector parallel to the longitudinal axis of the shank. Left knee joint angles were calculated as the Cardan angles of the shank reference frame relative to the thigh reference frame with a rotation order of extension–flexion, varus–valgus, and internal rotation–external rotation. Knee joint angular velocities were reduced as the time derivatives of knee joint angles.

The linear and angular kinematics of the three non-injured trials of the patient were time normalized as the percentage of the duration of the delivery phase, re-sampled from 0 to 100% of the delivery phase, and then averaged at each of the 101 normalized time point. The 95% confidence interval (95%CI) for the mean was also calculated at each of the 101 normalized time points for each kinematic variable for the non-injured trails. The kinematic data of the injured trial of the patient were also time normalized. The knee flexion angle and angular velocity at the time of the left foot landing and maximum knee flexion angle during the delivery phase were identified for the patient and 106 elite javelin throwers.

To evaluate the reliability of the linear and angular kinematics reduced in this study, the injured trial was repeatedly digitized. The linear and angular kinematics were re-reduced. The coefficient of multiple correlation (CMC)[29] and mean error of each kinematic variable reduced from repeatedly digitized coordinates were determined.

2.6. Data analysis

Tibia anterior translation was qualitatively evaluated by an experienced orthopedic surgeon by analyzing the position of the tibia relative to the patella in video images. The kinematics of injured trials were compared to the corresponding means of non-injured trials. The magnitude of a kinematic variable of the injured trial at a given normalized time point was considered significantly different from the corresponding mean of non-injured trials if the magnitude of the kinematic variable of the injured trial was greater than the upper bound of the 95%CI for the mean of the non-injured trials or smaller than the lower bound of the 95%CI for the mean of the non-injured trials at the given normalized time point.

Pearson's correlation coefficients were calculated for the 52 male and 54 female elite javelin throwers, respectively, to determine the correlations of the left knee flexion angle and angular velocity at the time of the left foot landing and the maximum knee angle during the delivery phase with the official distance. A type I error rate of 0.05 was chosen as the indication of statistical significance. Coefficients of determination (R^2) was defined as weak if R^2 was less than or equal to 0.09, as moderate if R^2 was greater than 0.09 and less than or equal to 0.25, and as strong if R^2 was greater than 0.25.[30]

3. Results

The CMCs of the kinematics used in this study were no less than 0.9668. The mean errors were 0.04 and 0.03 m/s, respectively, for the horizontal and vertical COM velocities. The mean errors were 0.44°, 1.23°, and 1.47°, respectively, for the knee flexion/extension, valgus/varus, and internal/external rotation angles. The mean error was 13.71°/s for the knee flexion angular velocity (Table 1).

The video images showed that the patella was apparently behind the anterior edge of the tibia plateau at the 30% of the delivery phase in the injury trial, corresponding to 49.5 ms after the left foot landing (Fig. 4). The patella was apparently in front of the anterior edge of the tibia plateau at the 30% of the delivery phase and during the entire delivery phase of each non-injured trial (Fig. 4).

The COM horizontal velocity of the injured trial was greater than the upper bound of the 95%CI for the mean of the non-injured trials after the first 25% of the delivery phase (Fig. 5A). The COM vertical velocity of the injured trial was less than the lower bound of the 95%CI for the mean of the non-injured trials after the first 23% of the delivery phase (Fig. 5B).

The knee flexion angle of the injured trial was less than the lower bound of the 95%CI for the mean of the non-injured trials from 11% to 56% of the delivery phase (Fig. 5C). The knee flexion angle reached its maximum at 77% of the delivery phase (0.1271 s) in the injured trial and at 50% of the delivery phase (0.0914 ± 0.0090 s) in non-injured trials (Fig. 5C). The knee valgus angle of the injured trial was greater than the upper bound of the 95%CI for the mean of the non-injured trials after 21% of the delivery phase (Fig. 5D). The knee valgus angle reached its maximum at 45% of the delivery phase (0.0743 s) in the injured trials and at 18% of the delivery phase (0.0300 ± 0.0204 s) in non-injured trials (Fig. 5D). The knee internal rotation angle of the injured trial was greater than the

Table 1
CMC and mean errors of kinematic data.

	COM horizontal velocity	COM vertical velocity	Knee flexion angle	Knee valgus/varus angle	Knee internal/external rotation angle	Knee flexion angular velocity
CMC	0.9988	0.9995	0.9961	0.9730	0.9668	0.9830
Mean error[a]	0.04	0.03	0.44	1.23	1.47	13.71

[a] COM velocities are in m/s, joint angles are in degree, joint angular velocity is in °/s.
Abbreviations: CMC = coefficient of multiple correlations; COM = center of mass.


Trial 1 Trial 2 Trial 3 Trial 4

Fig. 4. Comparison of tibia anterior translation between the injured trial and non-injured trials. All pictures were captured at 30% of the delivery phase.

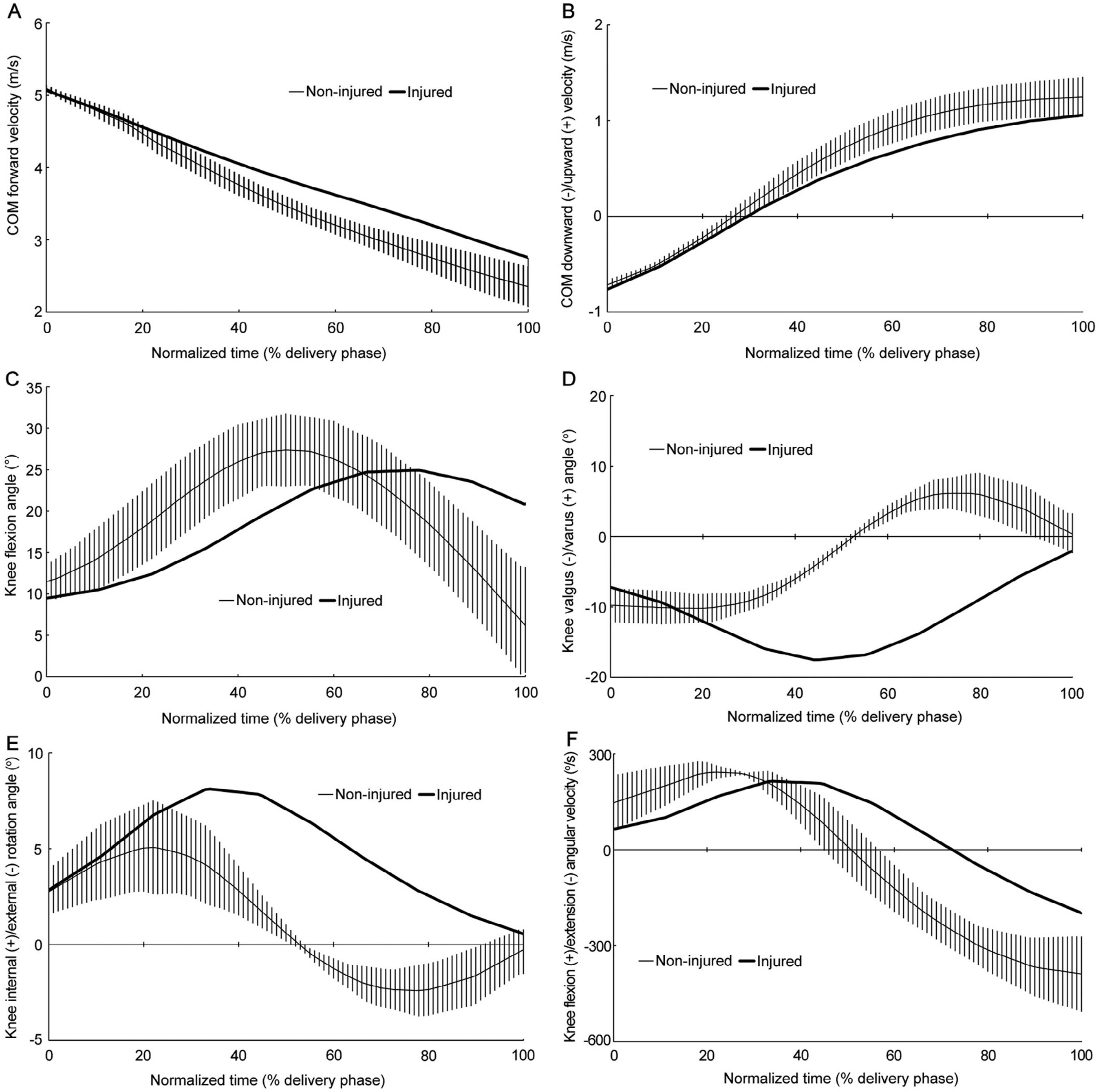

Fig. 5. Comparison of center of mass (COM) forward velocity, COM downward (−)/upward (+) velocity, knee flexion angle, knee valgus (−)/varus (+) angle, knee internal (+)/external (−) rotation angle, and knee flexion (+)/extension (−) angular velocity between the injured trial and non-injured trials. Vertical bars are 95% confidence intervals for the mean of the non-injured trials.

upper bound of the 95%CI for the mean of the non-injured trials after 26% of the delivery phase (Fig. 5E). The knee internal rotation angle reached its maximum at 34% of delivery phase (0.0561 s) in the injured trial and at 22% of the delivery phase (0.0367 ± 0.0008 s) in non-injured trials (Fig. 5E). The knee flexion angular velocity of the injured trial was less than the lower bound of the 95%CI for the mean of the non-injured trials from 2% to 30% of the delivery phase (Fig. 5F). The knee flexion angular velocity reached its maximum at 33% of the delivery phase (0.0561 s) in the injured trial and at 22% of the delivery phase (0.0428 ± 0.0102 s) in non-injured trials (Fig. 5F).

For the correlational analysis for the elite javelin throwers, the mean left knee flexion angle at the time of the left foot landing was 22° ± 8° ranging 3°–37° for the female elite javelin throwers, and 18° ± 9° ranging 6°–33° for the male elite javelin throwers. The left knee flexion angle at the time of the left foot landing was not significantly correlated to the official distance for the elite javelin throwers ($R^2 = 0.04$, $p = 0.978$ for females; $R^2 = 0.01$, $p = 0.989$ for males).

The mean left knee flexion angular velocity at the time of the left foot landing was 122.6° ± 145.5°/s with a minimum of −228.6°/s and a maximum of 394.2°/s for the female elite javelin throwers, and 155.8° ± 136.4°/s with a minimum of −89.4°/s and a maximum of 463.5°/s for the male elite javelin throwers. The left knee flexion angular velocity at the time of the left foot landing was not significantly correlated to the official distance for elite javelin throwers either ($R^2 = 0.02$, $p = 0.953$ for females; $R^2 = 0.01$, $p = 0.992$ for males).

The mean maximal left knee flexion angle during the delivery phase was 32° ± 8° ranging 17°–48° for the female elite javelin throwers, and 35° ± 16° ranging 7°–72° for the male elite javelin throwers. The maximal left knee flexion angle during the delivery phase was not significantly correlated to the official distance for elite javelin throwers either ($R^2 = 0.01$, $p = 0.989$ for females; $R^2 = 0.06$, $p = 0.971$ for males).

4. Discussion

The kinematic data reduced from manually digitized data in this study were highly reproducible. The CMC and mean error were used as measures for correspondence and agreement, respectively, to evaluate the reliability of kinematic data in this study. The results showed excellent correspondence and agreement of the kinematics reduced from repeatedly manual digitization.

The tibia had an excessive anterior translation in the injured trial. Literature has demonstrated that the patella should be in front of the anterior edge of the tibia when the knee flexion angle is less than 70°.[31] The results of this study showed that the patella was apparently behind the front edge of the tibia plateau at 30% of the delivery phase when knee flexion angle was less than 13° in the injured trial. These results combined with the literature[31] and confirmed ACL injury together suggest that an excessive tibia anterior translation likely occurred at 30% of the delivery phase. Considering the excessive tibia anterior translation occurred after the ACL was injured, we believe that the ACL injury actually occurred before 30% of the delivery phase.

The ACL injury in this study most likely occurred during the first 25% of the delivery phase. Javelin throwers decrease COM horizontal velocity and increase COM vertical velocity during the delivery phase.[32] For a right-handed thrower, the backward horizontal ground reaction force for decreasing COM horizontal velocity and the upward vertical ground reaction force for increasing COM vertical velocity are mainly generated by the left leg during the delivery phase.[32] The results of this study showed that the patient did not decrease COM horizontal velocity and increase COM vertical velocity after the first 25% of the delivery phase in the injured trial as much as she did in non-injured trials. These results suggest that the patient's left leg likely failed to generate as much backward horizontal ground reaction force and upward vertical ground reaction force after the first 25% of the delivery phase in the injured trial as she did in non-injured trials, and that the ACL injury most likely occurred before 25% of the delivery phase. The decreased official distance in the injured trial compared to the non-injured trials was likely the consequence of the ACL injury and compromised function of the left leg. The above discussed timing of this ACL injury is consistent with those reported in previous studies. Krosshaug et al.[6] estimated that ACL injuries occurred during the first 50 ms after initial foot contact with the ground based on qualitative inspections of video images of 39 injury cases in basketball. Koga et al.[8] estimated that ACL injuries occurred approximately 40 ms after initial foot contact with the ground in 10 injury cases in team handball and basketball. An understanding of the timing of ACL injury is important for understanding the injury mechanisms and risk factors.

A stiff landing with small knee flexion angle and angular velocity[33,34] was likely a significant contributor to the ACL injury in this study. The results of this study showed that the patient had smaller left knee flexion angle and angular velocity from 11% to 30% of the delivery phase in the injured trial compared to the non-injured trials. These results suggest that the left leg landing was stiffer in the injured trial compared to the non-injured trials. The left leg landing in javelin is similar to the landing in stop jump tasks, during which the landing leg experiences impact posterior and vertical ground reaction forces.[12,34–36] Previous studies have shown that the smaller the knee flexion angle and angular velocity at landing are, the greater the peak impact ground reaction force and anterior shear force at the proximal end of the tibia would be.[12,34–36] Previous studies have also shown that ACL loading is increased as the knee flexion angle is decreased when the anterior shear force at the knee is constant.[37–39] Two studies have shown that maximum ACL strain occurs when the knee flexion angle reaches minimal during both walking and landing.[40,41] The results of this study combined with those of previous studies suggest that a stiff landing with a small knee flexion angle was a significant contributor to the ACL injury in this study.

Knee valgus and knee internal rotation motions were not likely significant contributors to the ACL injury in this study. The results of this study showed that the patient's left knee valgus and internal rotation angles in the injured trial differed from the corresponding mean of the non-injured trials after 21% of the delivery phase and reached their maximum after 34% of the delivery phase. These results indicated that knee valgus and internal rotation angles were likely post injury events as a previous study indicated.[42] A study also showed that increased dynamic knee valgus actually decreased ACL length.[43] These results combined together do no support knee valgus and internal rotation motions as significant contributors to the ACL injury in this study.

The ACL injury in this study might be preventable without compromising performance. A left foot landing with straight and stiff knee was recommended in the traditional javelin throwing technique.[32] The results of this study showed that the left foot landing with a straight and stiff knee was likely the cause of the ACL injury. The analysis of elite javelin throwers' techniques, however, revealed that the all throwers flexed their left knee after the landing for delivery. The left knee flexion angle and angular velocity at the left foot landing and the maximum knee flexion angle during the delivery phase had essentially no correlation with the official distance. These results suggest that there is no need for javelin throwers to have a stiff landing with essentially a straight knee, and that javelin throwers can have a soft left leg landing with a flexed knee to reduce the risk for ACL injury,[36] and still have a long official distance. Future studies to identify the training effect of landing techniques on ACL injury rates in javelin throwers are warranted.

The current study was limited to one ACL injury in javelin throwing. The findings from this single injury case provided significant information for understanding mechanisms of non-contact ACL injury, but may not represent injury mechanisms of all non-contact ACL injuries. The current study was also limited to kinematic analysis. Kinetic variables such as ground reaction forces and knee joint resultant moments may provide additional information for understanding injury mechanisms.[12] In addition, only 3D coordinates of the joint centers were obtained in this study. The definitions of thigh and shank segment reference frames were different from those in laboratory studies,[34,36] which may result in overestimated knee valgus and internal rotation angles in this study, but should not affect the conclusions of this study because joint angles in the injured and non-injured trials were calculated using the same method.

5. Conclusion

The results of this study warrant the following conclusions: (1) the ACL injury in this study appeared to occur before the first 30% of the delivery phase, and mostly likely during the first 25% of the delivery phase; (2) a stiff left leg landing with a small knee flexion angle was the primary contributor to the ACL injury; (3) increased knee valgus and internal rotation motions appeared to be post injury events; and (4) javelin throwers may have a soft left leg landing with a flexed knee, which may help them prevent ACL injuries without compromising performance.

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