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Original Article

Fatigue influences lower extremity angular velocities during a single-leg drop vertical jump

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Abstract. [Purpose] Fatigue alters lower extremity landing strategies and decreases the ability to attenuate impact during landing. The purpose of this study was to reveal the influence of fatigue on dynamic alignment and joint angular velocities in the lower extremities during a single leg landing. [Subjects and Methods] The 34 female college students were randomly assigned to either the fatigue or control group. The fatigue group performed single-leg drop vertical jumps before, and after, the fatigue protocol, which was performed using a bike ergometer. Lower extremity kinematic data were acquired using a three-dimensional motion analysis system. The ratio of each variable (%), for the pre-fatigue to post-fatigue protocols, were calculated to compare differences between each group. [Results] Peak hip and knee flexion angular velocities increased significantly in the fatigue group compared with the control group. Furthermore, hip flexion angular velocity increased significantly between each group at 40 milliseconds after initial ground contact. [Conclusion] Fatigue reduced the ability to attenuate impact by increasing angular velocities in the direction of hip and knee flexion during landings. These findings indicate a requirement to evaluate movement quality over time by measuring hip and knee flexion angular velocities in landings during fatigue conditions.

Key words: Injury prevention, Lower extremity kinematics, Landings during fatigue condition

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INTRODUCTION

In sports like basketball and soccer, fatigue inevitably occurs during a game or practice session. Certain lower extremity injuries, such as non-contact anterior cruciate ligament (ACL) injuries, have a higher risk of occurrence in the final phase of games, e.g., between 76 and 90 min of a soccer game[1–3)]. Altered movement patterns caused by fatigue may explain non-contact ACL injuries, as well as other injuries that occur during sporting activities. Indeed, fatigue is one of the main causative factors of sports injuries[4–6)].

In contrast, to explain its effect on the lower extremities, evidence indicates that fatigue induces malalignment of the lower extremities and knee joints stability during sporting activities[7–9)]. Furthermore, while athletes run in fatigue conditions, fatigue causes an increase in impact acceleration, which is regarded as the rapid deceleration of the tibia, during landings[10)]. Moreover, lower extremity angles and the ground reaction force during landings are affected by fatigue[11, 12)]. These findings indicate that fatigue alters lower extremity landing strategies and the ability to attenuate impact during landing. Thus, the

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influence of fatigue during landing, an important movement, requires appropriate evaluation to consider measures to prevent lower extremity injuries.

Researchers are interested in lower extremity kinematics during landings. Therefore, most clinicians use video analysis to evaluate lower extremity angles and alignments at specific times[13–15]. However, lower extremity injuries may cause changes to angles and velocities over time, such as the direction change phase in sidestepping movements or the landing phase of a jump[16, 17]. It is also known that knee joint angular velocity variation during the landing phase is higher in females than in males[18–20] and some studies have measured knee angular velocity to evaluate capacity to attenuate impact, which may be associated with knee injury risk[6, 18, 19]. Accordingly, lower extremity angular velocity during landing reflects the ability to attenuate impact because it results from rapid angular variations in the lower extremities during landing. Therefore, lower extremity angles and angular velocities should be evaluated to identify biomechanical factors associated with lower extremity injuries.

Evidence indicates that fatigue causes changes in lower extremity kinematics, including angles and alignments during landing[4, 7, 8]. Fatigue may also alter lower extremity angular velocities and the ability to attenuate impact during landing. In our previous study, it has reported that fatigue might decrease the ability to attenuate impact by increasing the peak angular velocity in the direction of knee flexion during a single-leg jump landing[21]. However, no studies to date have addressed the angular velocity variables of lower extremities, including hip and ankle joints, during fatigue conditions. Thus, the purpose of this study was to evaluate the influence of fatigue on the lower extremity alignments and angular velocities during landing from a single-leg drop vertical jump. The research hypothesis was that landings during fatigue condition decrease the ability to attenuate impact in the lower extremity, i.e., angular velocities in the lower extremity would increase significantly in landings during fatigue condition compared with non-fatigue condition.

SUBJECTS AND METHODS

Thirty-four females (age: 20.7 ± 1.8 years old; height: 159.9 ± 5.6 cm; weight: 52.7 ± 5.9 kg) without a history of orthopedic hip, knee and ankle surgery participated in this study. The dominant foot was right in 30 subjects and left in 4 subjects; dominance was determined as preferred side to kick a ball[22]. All participants granted written informed consent for participation before testing and were randomly assigned to either the fatigue group (N=17; age: 21.2 ± 2.1 years old; height: 160.5 ± 5.0 cm; weight: 52.9 ± 7.1 kg) or control group (N=17; age: 20.2 ± 1.4 years old; height: 159.3 ± 6.3 cm; weight: 52.6 ± 4.5 kg). This study followed the Declaration of Helsinki and was approved by the Ethics Committee at the Saitama Medical University, Saitama, Japan (M-54).

A three-dimensional motion analysis system employing eight cameras (Vicon MX system, Vicon Motion Systems, Oxford, UK) was used to record lower extremity kinematic data and the center of mass during single-leg drop vertical jumps. Kinematic data were sampled at 240 Hz with a 16 Hz low-pass filter and a fourth-order zero lag Butterworth filter. Thirty-five reflective markers were placed on specific anatomical landmarks according to the Plug-in-Gait full body model, whisch is widely employed by researchers who use VICON[23]. Two force plates (MSA-6 Mini Amp, AMTI, MA, USA) recorded ground reaction forces during landing, from each single-leg drop vertical jump, at a 1,200 Hz sampling rate.

All participants wore closely fitted, dark shorts to aid data collection. They performed a single-leg drop vertical jump on their dominant foot (referred to as the “pre-trial”), which involves a first landing after dropping from a 40 cm. A single-leg drop vertical jump consists of 1st landing after dropping down from a 40 cm box and then a second landing after a maximum vertical jump rebounding from the drop. All participants were shown the testing sequence by assistant researchers, and several practice trials were conducted to enable them to correctly perform the required task. Subsequent trials were repeated until data from five successful trials were achieved; these were excluded if the person lost their balance during the landing. After the pre-trial, all participants were required to use a bike ergometer (COMBI, Japan), with participants in the fatigue group pedaling at 100 W per minute for 5 minutes or until they exceeded 17 (very hard) on the Borg scale[24, 25]. A poster showing the Borg scale was placed in front of the bike ergometer, and level of fatigue was recorded at 30 second intervals. In contrast, all participants in the control group pedaled a bike ergometer without load (less than 10 W per minute) for 5 minutes, with fatigue level was verified in the same way. After using the bike ergometer, all participants repeated the second single-leg drop vertical jump trial (referred to as the “post-trial”) according to the pre-trial procedure.

Landing from a single-leg drop vertical jump was defined as the period from initial ground contact to takeoff during the first landing. Two force plates were used to determine initial ground contact and takeoff, with the sampling rate set at 1,200 Hz. Initial ground contact was defined as the moment when the force plate data indicated that vertical ground reaction force exceeded 10 N, whereas takeoff was defined as the moment when the force plate data indicated that vertical ground reaction force was <10 N. All outcome measures were analyzed and computed as the average of three pre-trials and post-trials. The angles and angular velocities of hip flexion, knee flexion, and ankle dorsiflexion analyzed during all first landings. Hip flexion, knee flexion, and ankle dorsiflexion angles were calculated as positive values from filtered three-dimensional coordinate data; these parameters were defined as angular displacements from static anatomical positions. To decrease data fluctuation, hip flexion, as well as knee flexion, and ankle dorsiflexion angular velocities, were calculated by differentiating respective mean angles over five frames using a moving average. Peak hip flexion, knee flexion, and ankle dorsiflexion angles as well as angular velocities were determined from each maximum value during the first landings of the pre-trials and

Table 1. Peak angles and angular velocities of lower extremity, and ratios of each variable in the post-trials to those in the pre-trials

	Fatigue group (N=17)		Control group (N=17)	
	Pre-trial [b] Post-trial [b]	The ratio (%) [a]	Pre-trial [b] Post-trial [b]	The ratio (%) [a]
Hip flexion				
Peak angle	41.1 ± 12.6		41.8 ± 9.3	
	41.2 ± 13.3	100.1 ± 6.7	40.6 ± 10.3	96.6 ± 8.6
Peak angular velocity	216.7 ± 44.6		238.9 ± 51.8	
	231.6 ± 42.5	107.8 ± 9.7 *	238.4 ± 56.1	99.6 ± 9.2
Knee flexion				
Peak angle	60.8 ± 8.2		60.0 ± 9.9	
	60.6 ± 8.1	99.8 ± 4.9	60.1 ± 8.7	100.7 ± 6.2
Peak angular velocity	419.2 ± 53.4		450.6 ± 57.6	
	443.5 ± 52.4	106.2 ± 7.3 *	452.2 ± 49.9	100.8 ± 6.6
Ankle dorsiflexion				
Peak angle	30.1 ± 6.9		29.2 ± 4.9	
	30.5 ± 7.2	101.2 ± 5.9	31.0 ± 4.7	107.0 ± 13.5
Peak angular velocity	674.6 ± 22.3		715.3 ± 42.0	
	655.8 ± 88.8	98.0 ± 7.6	692.0 ± 142.2	97.2 ± 9.8

[a]The ratios (%) of each kinematic variable in the post-trials to those in the pre-trials were presented alongside.
[b]Angles and angular velocities of hip, knee flexion, and ankle dorsiflexion are shown in each (deg) and (deg/s).
*Significant difference between the ratios of fatigue and control groups (p<0.05).

post-trial. The vertical position of the center of mass was calculated from filtered 3D coordinate data during all first landings; to decrease data fluctuation, vertical position of the center of mass was calculated by differentiating mean vertical position of the center of mass over five frames using a moving average. Minimum vertical position and vertical velocity, which were defined as the lowest position and peak velocity in the direction of the ground, were determined from each minimum value during all first landings. The ratios (%) of all kinematic variables in the post-trials to those in the pre-trials were calculated.

Hip flexion, knee flexion, ankle dorsiflexion angle, and angular velocity at specific times were analyzed the moment of the initial ground contact, 40 milliseconds after the initial ground contact, and the peak vertical ground reaction. The moment of 40 milliseconds after the initial ground contact was chosen because some reports had shown ACL strain reached its peak value within approximately 40 milliseconds after initial ground contact[26, 27].

Data were analyzed using the SPSS software (version 19.0), and unpaired t-tests were used to compare changes in lower-extremity kinematics and center of mass between the fatigue and control groups. A p value of <0.05 was considered to indicate significant difference.

RESULTS

Ratios of the vertical position of the center of mass during the post-trials to each variable during the pre-trials did not differ significantly between the fatigue group (99.5 ± 2.4%; pre-trial 831.3 ± 40.1 mm, post-trial 827.3 ± 47.2 mm) and the control group (100.4 ± 1.4%; pre-trial 816.7 ± 45.5 mm, post-trial 819.8 ± 46.2 mm). Ratios of the vertical velocity of the center of mass during the post-trials to each variable during the pre-trials did not differ between the fatigue group (100.8 ± 5.5%; pre-trial 1,966.8 ± 228.9 mm/s, post-trial 1,978.4 ± 229.5 mm/s) and the control group (99.0 ± 4.2%; pre-trial 1,969.0 ± 175.0 mm/s, post-trial 1,945.7 ± 161.9 mm/s). Findings were similar for the ratios of post-trial of the vertical position and velocity center of mass to their pre-trial counterparts.

The ratios of hip flexion, knee flexion, ankle dorsiflexion angles, and angular velocities during the post-trials to each variable during the pre-trials are shown in Table 1. The ratio of hip flexion angular velocity during the post-trials to that during the pre-trials increased significantly in the fatigue group. Furthermore, the ratio of knee flexion angular velocity during the post-trials to that during the pre-trial also increased significantly in the fatigue group (p<0.05); however, the ratio of ankle dorsiflexion angular velocity did not differ between groups during either set of trials. In contrast, peak hip flexion, as well as knee flexion and ankle dorsiflexion angles did not alter after the fatigue protocol; thus, hip and knee flexion angular velocities increased significantly in the fatigue group compared to the control group, whereas all lower extremity joint angles were unchanged.

The ratios of hip flexion, knee flexion, ankle dorsiflexion angles, and angular velocities during the post-trials to each

Table 2. Angles of lower extremity and ratios of each variable in the post-trials to those in the pre-trials at specific times

	Fatigue group (N=17)		Control group (N=17)	
	Pre-trial (deg) Post-trial (deg)	The ratio (%) [a]	Pre-trial (deg) Post-trial (deg)	The ratio (%) [a]
Initial ground contact				
Hip flexion angle	21.7 ± 8.4		21.4 ± 6.4	
	21.0 ± 8.2	97.8 ± 16.1	21.0 ± 6.7	99.2 ± 12.2
Knee flexion angle	18.2 ± 6.3		18.5 ± 7.1	
	17.1 ± 6.0	94.5 ± 11.8	18.4 ± 7.4	101.0 ± 16.2
Ankle dorsiflexion angle	−16.3 ± 6.1		−15.1 ± 4.9	
	−15.1 ± 5.8	93.5 ± 15.2	−13.3 ± 4.3	90.9 ± 20.1
40 ms after initial ground contact				
Hip flexion angle	25.8 ± 9.0		25.8 ± 7.0	
	25.3 ± 8.8	99.2 ± 13.1	25.3 ± 7.4	109.1 ± 13.4
Knee flexion angle	29.0 ± 6.7		30.1 ± 7.1	
	28.1 ± 6.6	96.9 ± 5.6	29.6 ± 7.4	98.6 ± 7.4
Ankle dorsiflexion angle	2.7 ± 5.4		5.4 ± 3.1	
	3.5 ± 5.3	36.1 ± 409.6	6.3 ± 4.1	130.9 ± 75.9
The moment of peak ground reaction force				
Hip flexion angle	30.5 ± 8.9		30.6 ± 7.5	
	29.5 ± 9.4	96.1 ± 7.2	29.1 ± 7.7	95.3 ± 12.0
Knee flexion angle	38.8 ± 8.2		39.3 ± 7.7	
	36.5 ± 8.6	94.3 ± 10.2	37.0 ± 7.7	94.7 ± 14.1
Ankle dorsiflexion angle	15.6 ± 4.4		16.4 ± 5.0	
	14.1 ± 6.6	89.1 ± 29.1	15.1 ± 5.1	94.0 ± 23.7

[a]The ratios (%) of each kinematic variable in the post-trials to those in the pre-trials were presented alongside

variable during the pre-trials at the moment of initial ground contact, 40 milliseconds after the initial ground contact, and the peak ground reaction force, are shown in Tables 2 and 3. First, results show that angles, and angular velocities of lower extremity joints upon initial ground contact were not altered by the fatigue protocol. Second, 40 milliseconds after the initial ground contact, the ratio of hip flexion angular velocity during the post-trials to each variable during the pre-trials increased significantly in the fatigue group in comparison with the control group ($p<0.05$), whereas knee flexion and ankle dorsiflexion angular velocities were unchanged following the fatigue protocol. Finally, at the peak vertical ground reaction force, lower extremity joint angles and angular velocities were also unchanged following the fatigue protocol.

DISCUSSION

The purpose of this study was to reveal influences of fatigue on the lower extremity alignments, and angular velocities, during landing from a single-leg drop vertical jump. One of the key findings of this study is that it is important to select appropriate evaluation parameters and timing during landing for assessing effects of fatigue.

Lower extremity kinematics and ground reaction force during landings alter under fatigue conditions[5, 6, 11)], and reveal hip or knee angles, moments, and alignments of lower extremities during landing during fatigue conditions. Moreover, modification of these parameters by fatigue may a biomechanical factor leading to lower extremity injuries, including ACL injuries. Although evidence indicates that fatigue alters lower extremity kinematics, and that fatigue is a risk factor for incurring such injuries, most studies measured hip and knee angles[6, 11, 19)], or moments[5)], during landings. However, lower extremity injuries, including ACL injuries, may change lower extremity angles or velocities over time, including the phase of changing direction in sidestepping, or landing from a jump. Furthermore, rapid changes in lower extremity alignment occur frequently in many situations during sporting activities, and these findings indicate that measuring angle and angular velocity during fatigue is useful for preventing lower extremity injuries.

The results of this study show that peak hip flexion and knee flexion angular velocities increased significantly after the fatigue protocol, suggesting that fatigue decreased capacity to perform deceleration movements in the hip and knee joints during landings from jumps. Other investigators have suggested that knee kinematics play an important role in attenuating impact during running or jump landings[6, 28, 29)]; indeed, these results, indicate that fatigue may induce the knee joint to decrease impact attenuation during landings because knee flexion angular velocity increased significantly after the fatigue

Table 3. Angular velocities of lower extremity and ratios of each variable in the post-trials to those in the pre-trials at specific times

	Fatigue group (N=17)		Control group (N=17)	
	Pre-trial (deg/s)		Pre-trial (deg/s)	
	Post-trial (deg/s)	The ratio (%) [a]	Post-trial (deg/s)	The ratio (%) [a]
Initial ground contact				
Hip flexion angular velocity	95.8 ± 55.9		104.6 ± 67.3	
	98.0 ± 54.2	99.7 ± 37.0	99.8 ± 56.4	118.4 ± 75.5
Knee flexion angular velocity	318.1 ± 67.6		340.6 ± 89.7	
	311.7 ± 69.4	97.9 ± 7.0	321.7 ± 81.9	95.9 ± 13.4
Ankle dorsiflexion angular velocity	600.4 ± 136.1		672.5 ± 139.2	
	589.8 ± 106.4	99.2 ± 6.6	642.4 ± 162.1	95.7 ± 13.1
40 ms after initial ground contact				
Hip flexion angular velocity	186.7 ± 47.8		205.0 ± 62.6	
	201.0 ± 48.0	109.1 ± 37.0 *	205.9 ± 61.8	100.8 ± 8.2
Knee flexion angular velocity	408.3 ± 56.3		429.0 ± 65.3	
	425.9 ± 62.1	104.4 ± 7.0	434.1 ± 57.6	101.7 ± 6.3
Ankle dorsiflexion angular velocity	601.9 ± 102.2		600.6 ± 99.5	
	583.9 ± 75.1	97.7 ± 7.5	579.3 ± 65.0	97.5 ± 9.3
The moment of peak ground reaction force				
Hip flexion angular velocity	207.9 ± 47.2		216.9 ± 63.9	
	228.1 ± 41.8	111.4 ± 12.7	233.6 ± 53.6	114.0 ± 39.6
Knee flexion angular velocity	389.8 ± 59.4		387.8 ± 77.6	
	427.7 ± 55.5	110.6 ± 10.7	432.0 ± 47.9	115.5 ± 27.2
Ankle dorsiflexion angular velocity	405.1 ± 103.0		368.2 ± 96.5	
	428.2 ± 78.2	108.7 ± 17.2	411.6 ± 59.4	121.7 ± 50.5

[a]The ratios (%) of each kinematic variable in the post-trials to those in the pre-trials were presented alongside.
*Significant difference between the ratios of fatigue and control groups ($p<0.05$)

protocol. Moreover, we showed that hip flexion angular velocity increased significantly during landings after the fatigue protocol, and that fatigue might induce the hip and knee joints to decrease impact attenuation during landings. In contrast, peak ankle dorsiflexion angular velocities did not change after the fatigue protocol; thus, on the basis of these findings, angular velocity of knee and hip flexion during fatigue conditions should be analyzed to evaluate their capacity for impact attenuation by the lower extremities during landings. Hip and knee angular velocities altered by fatigue may be important parameters to measure to prevent knee injuries.

In this study, we measured hip flexion, knee flexion, and ankle dorsiflexion angles as well as angular velocities, at the moment of initial ground contact, 40 milliseconds after initial ground contact, and peak ground reaction force. The moment of initial ground contact represents the initial phase when contraction of the lower extremity muscles occurs during the landing phase. Furthermore, during the phase between initial ground contact and the peak ground reaction force, the ground reaction force in the vertical direction rapidly increased from the moment of initial ground contact during landings. Subsequently, it is likely that the ground reaction force continues to decrease until the takeoff from a vertical jump. For these reasons, fatigue may alter lower extremity kinematics upon initial ground contact with increasing muscle contractions of the lower extremity and the vertical ground reaction force. However, lower extremity kinematic values were unchanged by fatigue condition at the moment of initial ground contact and peak ground reaction force.

Krosshaug et al. used video analysis to show that most ACL injuries occur approximately 25–50 milliseconds after initial ground contact during landings[17)], and Koga et al. have suggested that it is likely that the majority of ACL injuries occur within 40 milliseconds after initial ground contact[30)]. Indeed, other evidence shows that ACL is strained during the initial phase of landings, approximately 40 milliseconds after initial ground contact, during landings[26, 27)]. According to the results of this study, 40 milliseconds after initial ground contact, hip flexion angular velocity increased significantly after the fatigue protocol, whereas other kinematics variables were unchanged. Thus, we conclude that fatigue does not affect knee kinematics, 40 milliseconds after initial ground contact, the point at which most ACL injuries are considered to occur. In contract, hip flexion angular velocity was significantly affected by fatigue condition 40 milliseconds after initial ground contact. Evidence indicates that increased hip flexion angle during landings is a biomechanical risk factor for ACL injuries[17)]. However, we were unable to demonstrate that fatigue affect hip flexion angle or the position of the center of mass during landings, considering that hip flexion angle and the position of the center of mass did not change after fatigue in this study. Furthermore, ACL injuries are caused by the combined movement of lower extremity angles in the sagittal plane and their variables in the frontal or horizontal planes, including hip adduction or knee abduction angles[15]. Therefore, it may be necessary to consider the influence of hip flexion velocity on ACL injuries by measuring knee abduction and hip adduction angles or knee external or internal angles, or both.

This study shows that fatigue decreased the ability to attenuate impact by increasing angular velocity in the direction of hip and knee flexion during a single-leg jump landing. Moreover, 40 milliseconds after initial ground contact, hip flexion angular velocity increased during fatigue, whereas knee and ankle kinematics remained unchanged. These results indicate a requirement to evaluate movement quality over time by measuring peak hip and knee flexion angular velocities during the landing phase during fatigue conditions. In addition, it may be necessary to evaluate these parameters at each peak variable and at specific times such as 40 milliseconds after initial ground contact during the landing phase during fatigue conditions. These findings further suggest that measuring hip and knee angular velocity during landings might be useful for efforts to prevent knee injuries during fatigue conditions.

The results of this study indicate that fatigue decreases the ability to attenuate impact by increasing angular velocity in the direction of hip and knee flexion during single-leg jump landing. Furthermore, it may be necessary to evaluate knee and hip angular velocities at their peak values and a specific time such as 40 milliseconds after initial ground contact, during the landing phase under fatigue conditions. These findings suggest that measuring hip and knee flexion angular velocities during fatigue conditions may serve as important evaluation parameters to prevent knee injuries, including ACL injuries.

REFERENCES

1) Hawkins RD, Fuller CW: A prospective epidemiological study of injuries in four English professional football clubs. Br J Sports Med, 1999, 33: 196–203. [Medline] [CrossRef]
2) Hawkins RD, Hulse MA, Wilkinson C, et al.: The association football medical research programme: an audit of injuries in professional football. Br J Sports Med, 2001, 35: 43–47. [Medline] [CrossRef]
3) Price RJ, Hawkins RD, Hulse MA, et al.: The Football Association medical research programme: an audit of injuries in academy youth football. Br J Sports Med, 2004, 38: 466–471. [Medline] [CrossRef]
4) Chappell JD, Herman DC, Knight BS, et al.: Effect of fatigue on knee kinetics and kinematics in stop-jump tasks. Am J Sports Med, 2005, 33: 1022–1029. [Medline] [CrossRef]
5) McLean SG, Fellin RE, Suedekum N, et al.: Impact of fatigue on gender-based high-risk landing strategies. Med Sci Sports Exerc, 2007, 39: 502–514. [Medline] [CrossRef]
6) Moran KA, Marshall BM: Effect of fatigue on tibial impact accelerations and knee kinematics in drop jumps. Med Sci Sports Exerc, 2006, 38: 1836–1842. [Medline] [CrossRef]
7) Kellis E, Kouvelioti V: Agonist versus antagonist muscle fatigue effects on thigh muscle activity and vertical ground reaction during drop landing. J Electromyogr Kinesiol, 2009, 19: 55–64. [Medline] [CrossRef]
8) Kernozek TW, Torry MR, Iwasaki M: Gender differences in lower extremity landing mechanics caused by neuromuscular fatigue. Am J Sports Med, 2008, 36: 554–565. [Medline] [CrossRef]
9) Ortiz A, Olson SL, Etnyre B, et al.: Fatigue effects on knee joint stability during two jump tasks in women. J Strength Cond Res, 2010, 24: 1019–1027. [Medline] [CrossRef]
10) Mizrahi J, Verbitsky O, Isakov E, et al.: Effect of fatigue on leg kinematics and impact acceleration in long distance running. Hum Mov Sci, 2000, 19: 139–151. [CrossRef]
11) Madigan ML, Pidcoe PE: Changes in landing biomechanics during a fatiguing landing activity. J Electromyogr Kinesiol, 2003, 13: 491–498. [Medline] [CrossRef]
12) Quammen D, Cortes N, Van Lunen BL, et al.: Two different fatigue protocols and lower extremity motion patterns during a stop-jump task. J Athl Train, 2012, 47: 32–41. [Medline]
13) Boden BP, Dean GS, Feagin JA Jr, et al.: Mechanisms of anterior cruciate ligament injury. Orthopedics, 2000, 23: 573–578. [Medline]
14) Hewett TE, Torg JS, Boden BP: Video analysis of trunk and knee motion during non-contact anterior cruciate ligament injury in female athletes: lateral trunk and knee abduction motion are combined components of the injury mechanism. Br J Sports Med, 2009, 43: 417–422. [Medline] [CrossRef]
15) Olsen OE, Myklebust G, Engebretsen L, et al.: Injury mechanisms for anterior cruciate ligament injuries in team handball: a systematic video analysis. Am J Sports Med, 2004, 32: 1002–1012. [Medline] [CrossRef]
16) Cochrane JL, Lloyd DG, Buttfield A, et al.: Characteristics of anterior cruciate ligament injuries in Australian football. J Sci Med Sport, 2007, 10: 96–104. [Medline] [CrossRef]
17) Krosshaug T, Nakamae A, Boden BP, et al.: Mechanisms of anterior cruciate ligament injury in basketball: video analysis of 39 cases. Am J Sports Med, 2007, 35: 359–367. [Medline] [CrossRef]
18) Decker MJ, Torry MR, Wyland DJ, et al.: Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. Clin Biomech (Bristol, Avon), 2003, 18: 662–669. [Medline] [CrossRef]
19) Gehring D, Melnyk M, Gollhofer A: Gender and fatigue have influence on knee joint control strategies during landing. Clin Biomech (Bristol, Avon), 2009, 24: 82–87. [Medline] [CrossRef]
20) Yu B, Lin CF, Garrett WE: Lower extremity biomechanics during the landing of a stop-jump task. Clin Biomech (Bristol, Avon), 2006, 21: 297–305. [Medline] [CrossRef]
21) Tamura A, Akasaka K, Otsudo T, et al.: Fatigue alters landing shock attenuateion during a single leg vertical drop jump. Orthop J Sports Med, 2016, 4:

2325967115626412 [CrossRef]. [Medline]
22) Ford KR, Myer GD, Hewett TE: Valgus knee motion during landing in high school female and male basketball players. Med Sci Sports Exerc, 2003, 35: 1745–1750. [Medline] [CrossRef]
23) Wernli K, Ng L, Phan X, et al.: The relationship between landing sound, vertical ground reaction force, and kinematics of the lower limb during drop landings in healthy men. J Orthop Sports Phys Ther, 2016, 46: 194–199. [Medline] [CrossRef]
24) Borg GA: Psychophysical bases of perceived exertion. Med Sci Sports Exerc, 1982, 14: 377–381. [Medline] [CrossRef]
25) Willson JD, Binder-Macleod S, Davis IS: Lower extremity jumping mechanics of female athletes with and without patellofemoral pain before and after exertion. Am J Sports Med, 2008, 36: 1587–1596. [Medline] [CrossRef]
26) Shin CS, Chaudhari AM, Andriacchi TP: The influence of deceleration forces on ACL strain during single-leg landing: a simulation study. J Biomech, 2007, 40: 1145–1152. [Medline] [CrossRef]
27) Withrow TJ, Huston LJ, Wojtys EM, et al.: The effect of an impulsive knee valgus moment on in vitro relative ACL strain during a simulated jump landing. Clin Biomech (Bristol, Avon), 2006, 21: 977–983. [Medline] [CrossRef]
28) Derrick TR: The effects of knee contact angle on impact forces and accelerations. Med Sci Sports Exerc, 2004, 36: 832–837. [Medline] [CrossRef]
29) Derrick TR, Dereu D, McLean SP: Impacts and kinematic adjustments during an exhaustive run. Med Sci Sports Exerc, 2002, 34: 998–1002. [Medline] [CrossRef]
30) Koga H, Nakamae A, Shima Y, et al.: Mechanisms for noncontact anterior cruciate ligament injuries: knee joint kinematics in 10 injury situations from female team handball and basketball. Am J Sports Med, 2010, 38: 2218–2225. [Medline] [CrossRef]