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Original article

Biomechanical risk factors of non-contact ACL injuries: A stochastic biomechanical modeling study

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Abstract

Background: Significant efforts have been made to identify modifiable risk factors of non-contact anterior cruciate ligament (ACL) injuries in male and female athletes. However, current literature on the risk factors for ACL injury are purely descriptive. An understanding of biomechanical relationship between risk and risk factors of the non-contact ACL injury is necessary to develop effective prevention programs.
Purpose: To compare lower extremity kinematics and kinetics between trials with and without non-contact ACL injuries and to determine if any difference exists between male and female trials with non-contact ACL injuries regarding the lower extremity motion patterns.
Methods: In this computer simulation study, a stochastic biomechanical model was used to estimate the ACL loading at the time of peak posterior ground reaction force (GRF) during landing of the stop-jump task. Monte Carlo simulations were performed to simulate the ACL injuries with repeated random samples of independent variables. The distributions of independent variables were determined from *in vivo* laboratory data of 40 male and 40 female recreational athletes.
Results: In the simulated injured trials, both male and female athletes had significantly smaller knee flexion angles, greater normalized peak posterior and vertical GRF, greater knee valgus moment, greater patella tendon force, greater quadriceps force, greater knee extension moment, and greater proximal tibia anterior shear force in comparison to the simulated uninjured trials. No significant difference was found between genders in any of the selected biomechanical variables in the trials with simulated non-contact ACL injuries.
Conclusion: Small knee flexion angle, large posterior GRF, and large knee valgus moment are risk factors of non-contact ACL injury determined by a stochastic biomechanical model with a cause-and-effect relationship.

Keywords: Anterior cruciate ligament; Risk factors; Stochastic biomechanical model

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1. Introduction

Anterior cruciate ligament (ACL) rupture is one of the most common injuries in sports.[1–3] The majority of ACL injuries occur with non-contact mechanisms, that is, no physical contact on the knee was involved when an injury occurs.[1,4,5] The non-contact nature of the ACL injuries indicates that the risk of ACL injury can be reduced through appropriate neuromuscular training to modify lower extremity biomechanics in athletic tasks, especially for landing tasks.[6–9] To reduce the risk of non-contact ACL injury, modifiable risk factors, especially motor control related lower extremity biomechanics, have to be identified.

As an attempt to reduce the risk of non-contact ACL injuries, tremendous efforts have been made to identify modifiable risk factors. Several studies demonstrated that female athletes on average had smaller knee flexion angle, greater knee valgus angle, greater ground reaction forces, greater proximal tibial anterior shear force, and greater knee extension moment during landing of selected athletic tasks compared to their male counterparts.[10–14] Authors of these studies proposed that small knee flexion angle, large knee valgus angle, and great ground reaction force in landing tasks were risk factors for non-contact ACL injury because of the significantly higher risk of non-contact ACL injury for female athletes in comparison to male athletes.[10–14] An epidemiological study showed that nine female athletes who had non-contact ACL injuries had significantly smaller maximum knee flexion angle, greater maximum ground reaction force, and greater maximum knee valgus moment of the ground reaction force in pre-injury drop landing test than did 196 female athletes who did not injure their ACLs after a 2-year follow-up.[15] This study demonstrated that maximum knee valgus angle and moment of the ground reaction force were significant predictors of non-contact ACL injury, and thus proposed that knee valgus angle and moment were risk factors for non-contact ACL injury.

Although previous studies provided significant information of differences in lower extremity biomechanics between genders in athletic tasks and between injured and uninjured female athletes, the results of these studies are purely descriptive and unable to establish biomechanical relationships between risk and risk factors of the injury.[16,17] These studies, therefore, failed to confirm that those proposed risk factors indeed affect the risk of ACL injury. This limitation of the current literature on the risk factors for non-contact ACL injury prevents us from further understanding of the non-contact ACL injury and developing effective prevention strategies.

Stochastic biomechanical modeling is a biomechanical modeling paradigm to determine probability of random outcomes of human motion through repeated random sampling, and is an ideal tool for determining risks and risk factors of acute musculoskeletal injuries. This method has been applied in studies on a variety of musculoskeletal injuries.[18–23] A stochastic biomechanical model for the risk and risk factors of non-contact ACL injury was recently developed.[24] This model was designed to estimate the ACL loading at the peak impact posterior ground reaction force during landing of the stop-jump task as previous studies demonstrated that peak ACL loading occurs at the peak impact posterior ground reaction forces during landing.[25,26] A previous study demonstrated that this model accurately estimated the female-to-male non-contact ACL injury rate ratio of collegiate basketball players and injury characteristics.[24] These results support the validity of the model and the application of the model as an evaluation tool in research and clinical practice in the prevention of non-contact ACL injury.

As a continuation of the previous study, the purposes of this study were to determine biomechanical risk factors of the non-contact ACL injury in a stop-jump task through Monte Carlo simulations with the stochastic biomechanical model developed in our previous study, and to compare (1) lower extremity kinematics and kinetics between trials with and without non-contact ACL injuries, and (2) lower extremity kinematics and kinetics in trials with non-contact ACL injuries between male and female recreational athletes. The stop-jump trials with and without non-contact ACL injuries were simulated using a stochastic biomechanical model.[24] We hypothesized that the landings of the stop-jump trials with non-contact ACL injuries would have significantly smaller knee flexion angle, shorter distance between center of pressure (COP) to the ankle joint center, greater ground reaction forces and knee moments and quadriceps muscle force, and lower hamstring and gastrocnemius muscle forces at the time of peak impact posterior ground reaction force in comparison to those without non-contact ACL injuries. The biomechanical relationships of these lower extremity kinematics and kinetics with ACL loading have been demonstrated in the literature.[27] We also hypothesized that the above described lower extremity kinematics and kinetics of female recreational athletes at the time of peak impact posterior ground reaction force in the landing of the stop-jump trials with non-contact ACL injuries would be significantly different in comparison to those of male recreational athletes. These two hypotheses were tested using the same sample of subjects and experimental data obtained in our previous study.[24]

2. Methods

2.1. Study subjects

A total of 40 male and 40 female recreational athletes without known history of lower extremity disorders were recruited as the subjects for this study. A recreational athlete was defined as a person who played sports or exercise at least three times a week for a total of at least 6 h per week without following a professionally designed training program. The mean age, body mass, and height of the male subjects were 22.34 $\pm$ 3.09 years, 78.7 $\pm$ 9.4 kg, and 1.78 $\pm$ 0.06 m, respectively. The mean age, body mass, and height of the female subjects were 23.20 $\pm$ 2.74 years, 60.0 $\pm$ 11.1 kg, and 1.63 $\pm$ 0.07 m, respectively. Subjects were excluded from the study if they had a history of musculoskeletal injury or any disorder that interfered with motor function. The use of human subjects in this study was approved by the University

Biomedical Institutional Review Board. A written informed consent was obtained from each subject before data collection.

2.2. Data collection

Each subject was asked to perform five successful trials of a stop-jump task that consisted of an approach run up to five steps followed by a two-footed landing, and two-footed vertical takeoff for maximum height.[28] A successful trial was defined as a trial in which the subject performed the stop-jump task as asked and all the data were collected. The subject was asked to perform the stop-jump task naturally as they did for a jump shot or grabbing a rebound in basketball, and at the maximum approach speed with which they felt comfortable to perform the task. The specific techniques of the stop-jump task were not demonstrated to subjects to avoid coaching bias.

Passive reflective markers were placed on the critical body landmarks as described in a previous study.[28] A videographic and analog acquisition system with eight video cameras (Peak Performance Technology, Inc., Englewood, CO, USA) and two force plates (Bertec Corp., Worthington, OH, USA) was used to collect three-dimensional (3-D) coordinates of reflective markers at a sample rate of 120 frames/s and ground reaction forces at a sample rate of 2000 samples/channel/s. A telemetry electromyographic (EMG) data acquisition system (Konigsburg Instruments, Pasadena, CA, USA) was used to collect EMG signals for the vastus medialis, rectus femoris, vastus lateralis, semimembranosus, biceps femoris, medial, and lateral head of gastrocnemius muscles at a sample rate of 2000 samples/channel/s. The videographic, force plate, and EMG data collections were temporally synchronized.

2.3. Data reduction

The raw 3-D coordinates of the reflective markers during each stop-jump trial were filtered through a Butterworth low-pass digital filter at a cutoff frequency of 10 Hz. The 3-D coordinates of lower extremity joint centers were estimated from the 3-D coordinates of the reflective markers. Lower extremity kinematics and kinetics were reduced for each trial as described in the previous study.[28]

Raw EMG signals were rectified and band-pass filtered at 20 Hz and 400 Hz, and then low-pass filtered at 10 Hz to obtain linear envelope EMGs.[29] The linear envelope EMGs were normalized to the corresponding linear envelope EMG for the associated maximal voluntary contraction. The normalized linear envelope EMG of the semimembranosus and biceps femoris muscles were averaged to represent the activation of the hamstring muscles. The normalized linear envelope EMG of the medial gastrocnemius and lateral gastrocnemius muscles were averaged to represent the activation of the gastrocnemius muscles.

A stochastic biomechanical model of ACL loading[24] was used to simulate non-contact ACL injuries. The total ACL loading was decomposed into three components in the model: loading due to the anterior draw force at the proximal tibia, loading due to knee valgus-varus moment, and loading due to knee internal−external rotation moment.[20] The model expressed each of these three components as a function of lower extremity kinematics and kinetics (Table 1), and knee joint anatomy and biomechanics.[24] Monte Carlo simulations with the stochastic biomechanical model of ACL loading were performed to simulate the density distribution of ACL loading, which is a function that describes the relative likelihood for this random variable to occur at a given point. In a Monte Carol simulation, the distributions of independent variables of the stochastic biomechanical model were determined based on the experimental data. ACL loading was repeatedly estimated from the independent variables randomly sampled based on their distributions. The density distribution of ACL loading was obtained after a certain number of iterations of the simulation.[24]

A non-contact ACL injury was defined as an ACL loading at the time of peak impact posterior ground reaction force during the landing of the stop-jump task equal to or greater than the strength of the ACL. The strength of the ACL was set at 2250 N for males and 1800 N for females.[30] The number of iterations in each Monte Carlo simulation was arbitrarily set at 100,000 to ensure that a sufficient number of simulated injuries occurred for statistical analysis. The number of simulated non-contact ACL injuries and the values of randomly sampled independent variables in each simulation were recorded. Ten Monte Carlo simulations were performed for each gender to estimate variations of the lower extremity kinematics and kinetics in non-contact ACL injuries. A recent study demonstrated that this model accurately estimated the female-to-male non-contact ACL injury rate ratio in basketball and injury characteristics, which supports the validity of this model.[24]

2.4. Data analysis

The lower extremity biomechanical variables at the peak impact posterior ground reaction force obtained from the experiment that served as independent variables for the stochastic biomechanical model (Table 1) were compared between genders. Those variables with normal distributions were compared by independent *t* tests (Table 1), while those with gamma distributions were compared by Mann−Whitney tests (Table 1). To test the first hypothesis, independent *t* tests

Table 1
Biomechanical variables obtained from experiment that served as independent variables of the stochastic biomechanical model of anterior cruciate ligament loading.

Variable	Distribution
Knee flexion angle (degree)	normal
COP to ankle distance (m)	normal
Posterior ground reaction force (BW)	gamma
Vertical ground reaction force (BH)	gamma
Knee varus-valgus moment (BH.BW[a])	normal
Knee internal−external rotation moment (BH.BW[a])	normal
Hamstring muscle force (BW)	gamma
Gastrocnemius muscle force (BW)	gamma

Abbreviations: BH = body height; BW = body weight; COP = center of pressure.
[a] BH.BW: moment normalized to body height (m) and body weight (N).

were performed to compare the lower extremity biomechanical variables at the peak impact posterior ground reaction force between simulated injured and uninjured trials for each gender. To test the second hypothesis, independent *t* tests were performed to compare the lower extremity biomechanical variables at the peak impact posterior ground reaction force in the simulated injured trials between genders. A Type I error rate of 0.05 was chosen as an indication of statistical significance for all statistical analyses. All statistical analyses were performed using SYSTAT computer program package (Systat Software Inc., Chicago, IL, USA).

3. Results

The experimental results showed that female recreational athletes had significantly smaller knee flexion angle at the time of peak posterior ground reaction force in comparison to male recreational athletes ($p = 0.004$) (Table 2). The experimental results also showed that female recreational athletes had significantly greater peak posterior ground reaction forces ($p = 0.031$), hamstring and gastrocnemius muscle forces at the time of the peak posterior ground reaction force ($p = 0.033$, $p = 0.006$) (Table 2).

Monte Carlo simulation results showed that both male and female recreational athletes had smaller knee flexion angle at the time of the peak posterior ground reaction force in the simulated injured trials than in the simulated uninjured trials ($p = 0.001$ for males, $p = 0.011$ for females) (Table 3). Both male and female recreational athletes had greater normalized peak posterior and vertical ground reaction forces, knee valgus moment, patella tendon force, quadriceps force, knee extension moment, and proximal tibia anterior shear force in the simulated injured trials than in the simulated uninjured trials ($p \leq 0.025$ for males, $p \leq 0.045$ for females) (Table 3). No significant differences were found in the distance between COP and ankle joint center, normalized knee internal rotation moment, and normalized hamstring and gastrocnemius forces between simulated injured trials and uninjured ($0.439 \geq p \geq 0.077$ for males, $0.444 \geq p \geq 0.077$ for females) (Table 3). No significant differences were found in any of the lower extremity biomechanical variables in the simulated injured trials between male and female recreational athletes ($0.481 \geq p \geq 0.118$) (Table 4).

Table 2
Comparison of experimental gender differences in selected lower extremity biomechanical variables that served as independent variables (mean ± SD) of the stochastic biomechanical model.

Variable	Male	Female	*p* value
Knee flexion angle (degree)	36.70 ± 9.66	32.51 ± 8.26	0.004
COP to ankle distance (m)	0.04 ± 0.03	0.04 ± 0.03	0.051
Posterior ground reaction force (BW)	0.68 ± 0.42	0.81 ± 0.41	0.031
Vertical ground reaction force (BH)	1.99 ± 1.07	2.02 ± 0.98	0.714
Knee varus-valgus moment (BH.BW[a])	0.01 ± 0.05	0.02 ± 0.05	0.105
Knee int−ext rotation moment (BH.BW[a])	0.01 ± 0.04	0.02 ± 0.05	0.141
Hamstring muscle force (BW)	0.27 ± 0.15	0.34 ± 0.18	0.033
Gastrocnemius muscle force (BW)	0.67 ± 0.28	0.86 ± 0.36	0.006

Abbreviations: BH = body height; BW = body weight; COP = center of pressure.
[a] BH.BW: moment normalized to body height (m) and body weight (N).

4. Discussion

The results of this study partially support the first hypothesis of this study that the lower extremity kinematics and kinetics at the peak time of peak posterior ground reaction force in the landings of the stop-jump trials in which non-contact ACL injury occurred were significantly different in comparison to those in which the injury did not occur. The results of this study demonstrate that simulated trials with non-contact ACL injuries on average had smaller knee flexion angle and greater knee valgus moment at the peak impact posterior ground reaction force, and greater peak impact posterior ground reaction force during the landing of the stop-jump task in comparison to simulated trials without injuries. Considering the biomechanical relationships of the ACL loading with these lower extremity kinematics and kinetics in our stochastic biomechanical model, the results confirmed that these lower extremity kinematic and kinetic variables are risk factors for non-contact ACL injury. The results of this study also showed that recreational athletes had significantly greater patella tendon force, quadriceps muscle force, knee extension moment, and proximal tibia anterior shear force in the simulated trials with injuries than in the simulated trials without injuries. These differences, however, are due to the differences in peak impact posterior ground reaction force between simulated injured and uninjured trials, and therefore, should not be considered as separate risk factors.

Knee flexion angle affects ACL loading through its effects on the patella tendon-tibia shaft angle and ACL elevation angle as modeled in the stochastic biomechanical model in this study. The patella tendon-tibia shaft angle is increased as the knee flexion angle is decreased.[31] The anterior draw force applied at proximal tibia is increased as the patella tendon-tibia shaft angle is increased while the quadriceps force remains a constant. The ACL loading is increased as the anterior shear force at proximal tibia is increased. The ACL elevation angle is also increased as the knee flexion angle is decreased.[32] The ACL loading is increased as the ACL elevation angle is increased while the anterior draw force at proximal tibia remains constant. Previous studies repeatedly demonstrate that decreasing knee flexion angle increases ACL loading.[33−36] A small knee flexion angle at landing, therefore, would increase the risk of non-contact ACL injury.

Impact peak posterior ground reaction force affects ACL loading through its effects on the quadriceps force and patella tendon force as modeled in the stochastic biomechanical model. A posterior ground reaction force creates a flexion moment at the knee joint which needs to be balanced by a knee extension moment generated by the quadriceps muscles through the patella tendon. The greater the posterior ground reaction force is, the greater the knee extension moment[28] and thus the greater the quadriceps force and patella tendon force (Table 2). The ACL loading is increased as the patella tendon

Table 3
Biomechanical characteristics (mean ± SD) of simulated injured and uninjured jumps with 2250 N as the strength of ACL for males and 1800 N for females.

Variable	Male			Female		
	Injured	Uninjured	p value	Injured	Uninjured	p value
Knee flexion angle (degree)	22.06 ± 8.27	36.82 ± 9.56	0.001	24.88 ± 5.61	32.88 ± 8.20	0.011
COP to ankle distance (m)	0.01 ± 0.03	0.03 ± 0.03	0.077	0.02 ± 0.03	0.04 ± 0.03	0.077
Posterior ground reaction force (BW)	1.44 ± 0.58	0.67 ± 0.41	0.002	1.45 ± 0.47	0.78 ± 0.38	0.001
Vertical ground reaction force (BW)	2.67 ± 1.09	1.72 ± 0.91	0.025	2.54 ± 0.97	1.76 ± 0.87	0.037
Knee valgus moment (BH.BW[a])	0.07 ± 0.07	0.01 ± 0.05	0.021	0.06 ± 0.05	0.02 ± 0.05	0.045
Knee internal rotation moment (BH.BW[a])	0.02 ± 0.03	0.01 ± 0.04	0.268	0.03 ± 0.04	0.02 ± 0.05	0.315
Hamstring force (BW)	0.26 ± 0.14	0.27 ± 0.15	0.439	0.33 ± 0.16	0.34 ± 0.16	0.444
Gastrocnemius force (BW)	0.67 ± 0.26	0.66 ± 0.27	0.343	0.78 ± 0.32	0.85 ± 0.34	0.322
Patellar tendon force (BW)	13.95 ± 5.30	4.85 ± 3.87	0.001	13.66 ± 4.00	5.63 ± 3.53	0.000
Quadriceps force (BW)	12.20 ± 4.62	4.24 ± 3.39	0.001	11.93 ± 3.50	4.92 ± 3.09	0.000
Knee extension moment (BH.BW[a])	0.38 ± 0.16	0.13 ± 0.12	0.001	0.37 ± 0.12	0.15 ± 0.11	0.000
Proximal tibial anterior shear force (BW)	1.33 ± 0.59	0.51 ± 0.42	0.001	1.28 ± 0.46	0.59 ± 0.38	0.001

Abbreviations: BH = body height; BW = body weight; COP = center of pressure; ACL = anterior cruciate ligament.
[a] BH.BW: moment normalized to body height (m) and body weight (N).

force is increased when the knee flexion angle is less than 60°.[31,37–42] Previous studies demonstrate that the *in vivo* maximum ACL loading in a landing task occurs at time when the peak impact vertical ground reaction force occurs,[25,26] and that the peak impact posterior and vertical forces occur at the same time.[28] Increasing the peak impact posterior ground reaction force, therefore, would also increase ACL loading and thus the risk of non-contact ACL injury.

Knee valgus moment due to ground reaction force and knee valgus movement also affects ACL loading as modeled in the stochastic biomechanical model. Cadaver studies demonstrate that knee valgus moment significantly increases ACL loading when an anterior draw force is applied at proximal tibia.[36] Computer simulation studies using finite element model also demonstrate that knee valgus moment significantly increases ACL loading,[43,44] or reduces the tolerance of the ACL to anterior draw force.[45] These previous studies combined with the results of the current study suggest that the greater knee valgus moment due to the ground reaction force is a risk factor of non-contact ACL injury, as well. Previous studies, however, also demonstrated that knee valgus moment alone may not be able to cause isolated ACL injury with minimum MCL damage as clinical observations showed.[44,46–48]

Table 4
Comparison of lower extremity biomechanical characteristics (mean ± SD) of simulated injured trials between male and female recreational athletes.

Variable	Male	Female	p value
Knee flexion angle (degree)	22.06 ± 8.27	24.88 ± 5.61	0.192
COP to ankle distance (m)	0.01 ± 0.03	0.02 ± 0.03	0.234
Posterior ground reaction force (BW)	1.44 ± 0.58	1.45 ± 0.47	0.481
Vertical ground reaction force (BW)	2.67 ± 1.09	2.54 ± 0.97	0.391
Knee valgus moment (BH.BW[a])	0.07 ± 0.07	0.06 ± 0.05	0.360
Knee internal rotation moment (BH.BW[a])	0.02 ± 0.03	0.03 ± 0.04	0.268
Hamstring force (BW)	0.26 ± 0.14	0.33 ± 0.16	0.156
Gastrocnemius force (BW)	0.67 ± 0.26	0.78 ± 0.32	0.118
Patellar tendon force (BW)	13.95 ± 5.30	13.66 ± 4.00	0.445
Quadriceps force (BW)	12.20 ± 4.62	11.93 ± 3.50	0.441
Knee extension moment (BH.BW[a])	0.38 ± 0.16	0.37 ± 0.12	0.437
Proximal tibial anterior shear force (BW)	1.33 ± 0.59	1.28 ± 0.46	0.418

Abbreviations: BH = body height; BW = body weight; COP = center of pressure.
[a] BH.BW: moment normalized to body height (m) and body weight (N).

The three risk factors confirmed by the results of this study are consistent with the literature. Several laboratory studies found that female athletes had smaller knee flexion angle, and greater ground reaction forces and knee valgus moment in landing tasks than their male counterparts do when performing athletic tasks.[10,11,14,28,49] A recent epidemiological study also found that the female athletes who injured their ACLs had smaller knee flexion angle, and greater vertical ground reaction force and knee valgus moment in a vertical landing task before the injury in comparison to uninjured female athletes.[15] These studies proposed the small knee flexion angle, and great ground reaction forces and knee valgus moment in landing tasks as risk factors of non-contact ACL injury. These studies, however, did not establish direct biomechanical relationships between the proposed risk factors and the injury as the current study does.

The results of this study showed no significant difference in hamstring muscle force between simulated injured and uninjured trials, which appears to be inconsistent with the literature. Studies repeatedly showed that increasing hamstring muscle force decreases ACL loading,[50,51] which appears to suggest lower hamstring muscle force as a risk factor of non-contact ACL injury. These studies, however, examined the effects of hamstring muscle force on ACL loading by maintaining a constant quadriceps muscle force, which actually decreased knee extension moment. Decreasing knee extension moment means a change in movement. The hamstring muscle force does not always reduce ACL loading if its effect on ACL loading is examined with knee extension moment maintained as a constant. Increasing hamstring muscle force will result in an increase in quadriceps muscle force if the knee extension moment is maintained as a constant. As previously discussed, the patella tendon-tibia shaft angle increases as the knee flexion angle decreases. The hamstring tendon-tibia shaft angle, however, decreases as the knee flexion angle decreases. Increasing hamstring muscle force may increase anterior draw force instead of decreasing it when the knee flexion angle is small. Increasing hamstring muscle force, therefore, is not necessarily protecting the ACL, and may actually increase ACL loading.

The results of this study also showed no significant differences in the distance between the COP and ankle joint center, knee internal−external rotation moment, and the gastrocnemius muscle force between simulated injured and uninjured trials. These non-significant results were likely due to low sensitivities of the ACL loading from these variables. They may biomechanically affect ACL loading but their effects may be relatively small and not obvious when other variables that influence ACL loading are influenced.

The results of this study do not support the second hypothesis of this study that the lower extremity kinematics and kinetics of female recreational athletes at the peak posterior ground reaction force in the landing of the stop-jump trials in which non-contact ACL injury occurred were significantly different in comparison to those of male recreational athletes. The results of this study showed no significant differences in the lower extremity kinematics and kinetics at the peak impact posterior ground reaction force in the simulated injured trials between male and female recreational athletes. These results suggest that the risk factors of non-contact ACL injury are similar for both genders, which do not support the hypothesis that mechanisms and risk factors of non-contact ACL injury are different for different genders.[17] Future studies may be needed to further test this hypothesis. The similarity of risk factors for ACL injuries between genders taken together with considerably higher risk for ACL injury in female athletes supports previous studies that demonstrate female athletes are more likely to land with these risk factors being present.

The results of this study provide significant information for developing prevention strategies for non-contact ACL injury. The results indicate that training programs should be focused on increasing knee flexion angle and reducing peak impact ground reaction force and knee valgus moment during landing tasks. To achieve these objectives, athletes should be trained to flex not only the knee but also the hip before the landing tasks. A previous study demonstrates that the knee flexion angular velocity at the initial foot contact with the ground of the stop-jump task negatively correlated to the peak impact vertical ground reaction force while the hip flexion angular velocity at the same time negatively correlated to the peak impact posterior ground reaction force.[28] These results indicate that flexing the knee may assist in reducing peak impact vertical ground reaction force while flexing the hip may assist in reducing peak impact posterior ground reaction force. A recent study demonstrates that flexing the hip not only assists in increasing knee flexion angle, but also assists in reducing knee valgus moment by reducing ground reaction forces and knee valgus angle in landing tasks.[52] The details of how hip flexion assists in reducing knee valgus angle, however, are not clear.

Although the current study established the biomechanical relationships between risk factors and non-contact ACL injury through stochastic biomechanical modeling, the results are limited to the stop-jump task because only the stop-jump task was included in the model. Non-contact ACL injuries frequently occur not only in stop-jump tasks but also in cutting and vertical landing tasks. In comparison to the stop-jump task in the model in this study, side-cutting task may have greater knee valgus-varus and internal−external rotation moments than the stop-jump task does while the vertical landing task may have less posterior ground reaction force but greater vertical ground reaction force than the stop-jump task does. Including these tasks in the future studies may improve our understanding of the risk factors of non-contact ACL injury. Also, the current study only compared the lower extremity kinematics and kinetics between simulated injured and uninjured trials. Future studies are needed to determine the sensitivities of the probability of non-contact ACL injury to each of the lower extremity kinematics and kinetics to further understand the risk factors of non-contact ACL injury and possible differences in risk factors between genders. Further, the stochastic biomechanical model used in this study limited the simulation of ACL loading to the time of peak impact posterior ground reaction force. More sophisticated models may be needed in future studies to understand the neuromuscular control related to the lower extremity biomechanics associated with the injury.

5. Conclusion

A validated stochastic biomechanical model of the risk and risk factors were used to simulate non-contact ACL injuries with biomechanical relationships between the injury and lower extremity kinematics and kinetics. The results confirmed that small knee flexion angle and great peak impact posterior ground reaction force and knee valgus moment are risk factors of non-contact ACL injury in the stop-jump task. Not all gender differences in lower extremity motion patterns are necessarily risk factors of non-contact ACL injury. No gender differences were found in the risk factors of non-contact ACL injury in the stop-jump task.

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