ARTICLE IN PRESS

Journal of Biomechanics ∎ (∎∎∎∎) ∎∎∎–∎∎∎

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

A stochastic biomechanical model for risk and risk factors of non-contact anterior cruciate ligament injuries

Cheng-Feng Lin [a], Michael Gross [b], Chuanshu Ji [c], Darin Padua [d], Paul Weinhold [b,e,f], William E. Garrett [g], Bing Yu [b,e,f,*]

[a] Department of Physical Therapy, National Cheng Kung University, Tainan, Taiwan
[b] Center for Human Movement Science, Division of Physical Therapy, The University of North Carolina at Chapel Hill, Chapel Hill, NC, USA
[c] Department of Statistics and Operations Research, The University of North Carolina at Chapel Hill, Chapel Hill, NC, USA
[d] Department of Exercise and Spots Science, The University of North Carolina at Chapel Hill, Chapel Hill, NC, USA
[e] Department of Orthopaedic Surgery, The University of North Carolina at Chapel Hill, Chapel Hill, NC, USA
[f] Department of Biomedical Engineering, The University of North Carolina at Chapel Hill, Chapel Hill, NC, USA
[g] Division of Orthopaedic Surgery, Duke University Medical Center, Durham, NC, USA

ARTICLE INFO

Article history:
Accepted 2 December 2008

Keywords:
Anterior cruciate ligament
Biomechanics
Stochastic
Model
Risk

ABSTRACT

Gender has been identified as a risk factor for non-contact anterior cruciate ligament (ACL) injuries. Although some possible biomechanical risk factors underlying the gender differences in the risk for non-contact ACL injuries have been identified, they have not been quantitatively confirmed yet because of the descriptive nature of the traditional epidemiological methods. The purpose of this study was to validate a stochastic biomechanical model for the risk and risk factors for non-contact ACL injuries. An ACL loading model was developed and instrumented to a Monte Carlo simulation to estimate the ACL injury rate for a stop–jump task in which non-contact ACL injuries frequently occur. Density distributions of independent variables of the ACL loading model were determined from in-vivo data of 40 male and 40 female athletes when performing the stop–jump task. A non-contact ACL injury was defined as the peak ACL loading being greater than 2250 N for males and 1800 N for females. The female-to-male non-contact ACL injury rate ratio was determined as the ratio of the probability of ACL ruptures of females to that of males. The female-to-male non-contact ACL injury rate ratio predicted by the stochastic biomechanical model was 4.96 (SD = 0.22). The predicted knee flexion angle at the peak ACL loading in the simulated injury trials was 22.0 (SD = 8.0) degrees for males and 24.9 (SD = 5.6) degrees for females. The stochastic biomechanical model for non-contact ACL injuries developed in the present study accurately predicted the female-to-male injury rate ratio for non-contact ACL injuries and one of the kinematic characteristics of the injury.

1. Introduction

Anterior cruciate ligament (ACL) tear is one of the most common knee injuries in sports (Griffin et al., 2006). The majority of ACL injuries occur with a non-contact mechanism (Boden et al., 2000) and can be potentially prevented (Griffin et al., 2006). Significant research efforts have been made in the last decade to determine modifiable risk factors of sustaining non-contact ACL injuries so prevention strategies can be developed. Although previous studies have indicated that biomechanical factors such as increased knee valgus moment, quadriceps muscle activation, and proximal anterior tibia shear force may be risk factors for non-contact ACL injuries (Malinzak et al., 2001; Chappell et al., 2002; Ford et al., 2003; Sell et al., 2007), a recent extensive literature review (Yu and Garrett, 2007) failed to find any convincing scientific evidence to support a cause-and-effect relationship between those proposed risk factors and non-contact ACL injuries.

The lack of scientific confirmation of biomechanical risk factors for non-contact ACL injuries is mainly due to a lack of effective and efficient research methods for identifying risk factors. The longitudinal cohort research design is the most commonly used traditional epidemiological method for identifying risk factors for an injury or disease. The studies using the longitudinal cohort design are usually complicated, labor intensive, time consuming, and expensive because of the need to test and follow a large group of subjects for a long time period to obtain a certain number of injury cases (Portney and Watkins, 2000). In addition, the results

* Corresponding author at: Center for Human Movement Science, Division of Physical Therapy, CB# 7135 Bondurant Hall, The University of North Carolina at Chapel Hill, Chapel Hill, NC 27599-7135, USA. Tel.: +1 919 843 8643; fax: +1 919 966 3678.
E-mail address: byu@med.unc.edu (B. Yu).

0021-9290/$ - see front matter © 2008 Elsevier Ltd. All rights reserved.
doi:10.1016/j.jbiomech.2008.12.005

of the studies using the traditional cohort design are descriptive and lack cause-and-effect relationships between risk factors and the risk of an injury or disease (Portney and Watkins, 2000).

Stochastic biomechanical modeling is an effective and efficient research method for investigating the random outcomes of human movement (Hughes and An, 1997). This method allows investigators to determine the risk for an injury without following subjects to obtain actual injury cases. This enables the execution of the study to be less complicated, less labor intensive, less time consuming, and less expensive in comparison to traditional epidemiological methods. Also, stochastic biomechanical modeling method allows investigators to identify risk factors with cause-and-effect relationships to the injury in the absence of any observed injuries.

Stochastic biomechanical modeling method has been successfully applied to studies on the variation of human movements and prevention of a variety of musculoskeletal system injuries (Davidson et al., 2004; Langenderfer et al., 2006; McLean et al., 2004; Mirka and Marras, 1993; Santos and Valero-Cuevas, 2004; Valero-Cuevas et al., 2003). Stochastic biomechanical modeling methods have also been applied to recent research pertaining to ACL injury prevention. McLean et al. (2004) estimated the variation of ACL loading in a sidestep cut task using a stochastic biomechanical model. Garrett and Yu (2004) also examined the effects of proximal tibia anterior shear force and knee valgus–varus and internal–external rotation moments on ACL loading using a stochastic biomechanical modeling approach.

The purpose of this study was to validate a stochastic biomechanical model to predict the risk (injury rate or probability for injury) and risk factors (factors that contribute to the risk) for non-contact ACL injuries. We hypothesized that the female-to-male non-contact ACL injury rate ratio estimated using the stochastic biomechanical model proposed in this study would be similar to that determined using traditional epidemiological methods. We also hypothesized that the injury characteristics estimated by the Monte Carlo simulation using the stochastic biomechanical model proposed in this study would be similar to those reported in the literature.

2. Materials and methods

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 (Table 1). A recreational athlete was defined as a person who played basketball, soccer, volleyball, and lacrosse at least 3 times per week for a total of at least 6 h per week without following a professionally designed training program. The use of human subjects was approved by the Biomedical Internal Review Board of the University.

Each subject was asked to perform five successful trials of a stop–jump task that consisted of an approach run of 4–5 steps with maximum effort followed by a two-footed landing, and an immediate two-footed vertical jump for maximum height (Yu et al., 2006). The subjects were asked to perform the stop–jump task as they would naturally do for a jump shot or grabbing a rebound in basketball. The specific techniques of the stop–jump task were not demonstrated to subjects to avoid a coaching effect. A videographic and analog acquisition system with 6 video cameras (Peak Performance Technology, Inc., Englewood, CO, USA) and 2 force plates (Bertec Corporation, Worthington, OH, USA) were used to collect 3-D coordinates of reflective markers on critical body landmarks at a sampling rate of 120 frames/s and ground reaction forces during each trial at a sampling rate of 2000 samples/channel/s.

A telemetry electromyographic (EMG) data acquisition system (Konigsburg Instruments, Pasadena, CA, USA) was used to collect EMG signals of the semimembranosus, biceps femoris, and medial and lateral heads of the gastrocnemius at a sampling rate of 2000 samples/channel/s. Pregelled silver/silver-chloride surface electrodes (Ambu Inc., Glen Burnie, MD, USA) were placed on the skin over the belly of each muscle. A ground electrode was placed over the tibial tuberosity. Each subject's skin over the muscle bellies was shaved and cleaned with isopropyl alcohol before electrode placement. Three trials of maximum voluntary contraction (MVC) were performed for the hamstring muscle group for 5 s with the hip and knee flexed at 90°. Three trials of MVC were also performed for the gastrocnemius muscle group for 5 s with the knee fully extended and the ankle flexed at 90°.

Table 1
Subject age, body mass, and standing height.

	Age (years)	Body mass (kg)	Standing height (m)
Male (SD)	22.4 (3.1)	78.8 (9.4)	1.78 (0.06)
Female (SD)	23.2 (2.7)	60.1 (11.5)	1.63 (0.07)

The raw 3-D coordinates of the 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 the lower extremity joint centers were estimated from the 3-D coordinates of the reflective markers. The knee joint angle, tibia tilt angle (the angle between the line connecting the knee and ankle joint centers and the vertical), location of center of pressure (COP), knee joint resultant anterior shear force, and valgus–varus and external–internal rotation moments due to ground reaction forces at the time of peak posterior ground reaction force were determined for each trial as described in a previous study (Yu et al., 2006).

Raw EMG signals were processed as described in a previous study (Chappell et al., 2007) to obtain linear envelope EMGs (Gerleman and Cook, 1992). The linear envelope EMGs were normalized to the corresponding linear envelope EMG for the associated MVC. The normalized linear envelope EMG of semimembranosus and biceps femoris muscles were averaged to represent activation of the hamstring muscles. The normalized linear envelope EMG of medial gastrocnemius and lateral gastrocnemius muscles were averaged to represent activation of the gastrocnemius muscles.

A biomechanical model of ACL loading developed by McLean et al. (2004) was modified to estimate ACL loading from given lower leg kinematics and kinetics (Appendix). Peak ACL loading occurs at the time of peak vertical ground reaction force in a hop–stop task (Lamontagne et al., 2005) similar to the stop–jump task in this study. The peak posterior ground reaction force occurred less than 0.001 s earlier than the peak vertical ground reaction forces in the stop–jump task performed by the subjects in this study (Yu et al., 2006). The modified ACL loading model was, therefore, instrumented into a Monte Carlo simulation to estimate the probability that the ACL loading at the time of the peak posterior ground reaction force was greater than the strength of the ACL in the stop–jump task. The density distribution of ACL loading was expressed as a function of density distributions of the independent variables at the time of peak posterior ground reaction force using the ACL loading model through the Monte Carlo simulation. A non-contact ACL injury was defined as an ACL loading at the time of peak 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, respectively (Stapleton et al., 1998). The number of injuries was incremented when a calculated ACL loading was equal to or greater than the strength of the ACL of the given gender. The output of the simulation was the probability of ACL rupture for a given gender.

The density distribution of each independent variable of the ACL loading model at the time of peak posterior ground reaction force was determined for each gender from the in-vivo empirical data of the 40 male and 40 female subjects. Skewness and kurtosis were determined for each independent variable (Table 2). A normality test (Gujarati, 2003) was performed to determine if the density distribution of a given independent variable in the ACL loading model at the time of posterior ground reaction force was a normal distribution or gamma distribution. Density distributions of the hamstring and gastrocnemius forces were determined from the mean and standard deviation of the linear envelope EMGs of hamstring and gastrocnemius muscles at the time of posterior ground reaction force collected in this study and the mean hamstring and gastrocnemius muscle forces in a similar task (Pflum et al., 2004). This was accomplished assuming a linear relationship between muscle force and muscle activation (Hatze, 1981; Kaufman et al., 1991), without considering the length–tension and force–velocity relationships. The cumulative density distribution function was then determined for each independent variable. The cumulative density function is a distribution function that describes the probability distribution of a real-valued random variable or describes the probability that a variable takes on a value less than or equal to a number (Evans et al., 2000). A random number generator was assigned to each independent variable to sample each independent variable randomly from the corresponding cumulative density distribution function.

Among the independent variables of the ACL loading model, the magnitudes of the peak posterior and vertical ground reaction forces are correlated (Yu et al., 2006). Considering this correlation, the vertical ground reaction force at the peak posterior ground reaction force was determined as

$$F_{\mathrm{VGRF}} = 567.884 + 1.406 \times F_{\mathrm{PGRF}} - 232.803 \times G + 0.044 \times F_{\mathrm{PGRF}} \times G + \varepsilon$$
$$(G = 0 \text{ for males}, \quad G = 1 \text{ for females}) \; (r^2 = 0.65;\ p < 0.001)$$

where ε is the regression residual considered as a random variable distributed as a normal distribution with a mean of zero.

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 in each simulation for statistical analysis. The number of simulated non-contact ACL

Please cite this article as: Lin, C.-F., et al., A stochastic biomechanical model for risk and risk factors of non-contact anterior cruciate ligament injuries. Journal of Biomechanics (2009), doi:10.1016/j.jbiomech.2008.12.005

Table 2
Skewness and kurtosis of each independent variable for Monte Carlo simulation.

	Male		Female	
	Skewness	Kurtosis	Skewness	Kurtosis
Knee flexion angle (deg.)	0.25	3.38	0.05	4.81
Tibial tilting angle (deg.)	−0.02	3.63	0.09	2.64
COP to ankle distance (m)	−0.65	3.68	−0.68	3.42
Posterior ground reaction force (BW)	1.09	4.01	0.66	2.94
Knee varus–valgus moment (BH.BW)	0.04	3.02	−0.15	2.62
Knee internal–external rotation moment (BH.BW)	0.20	2.99	−0.07	2.62
Hamstring EMG	0.24	2.28	0.11	2.20
Gastrocnemius EMG	0.58	2.55	0.40	2.23

BH.BW: moment normalized to body height (m) and body weight (N).

Table 3
Normality test and types of distributions of independent variables for male subjects.

Variable	Normality test (*p*-value)	Distribution	Mean	SD	Shape	Scale
Knee flexion angle (deg.)	0.13	Normal	36.7	9.7		
Tibia tilt angle (deg.)	0.20	Normal	−5.1	6.5		
COP to ankle distance (m)	0.00	Normal	0.03	0.03		
Posterior ground reaction force (BW)	0.00	Gamma	0.68	0.42	2.66	0.26
Knee varus–valgus moment (BH.BW)	0.06	Normal	0.01	0.05		
Knee internal–external rotation moment (BH.BW)	0.20	Normal	0.01	0.04		
Hamstring muscle force (N)	0.01	Gamma	500	203	6.01	83.13
Gastrocnemius force (N)	0.00	Gamma	200	109	3.35	59.69

BH.BW: moment normalized to body height (m) and body weight (N).

Table 4
Normality test and types of distributions of independent variables for female subjects.

Variable	Normality test (*p*-value)	Distribution	Mean	SD	Shape	Scale
Knee flexion angle (deg.)	0.00	Gamma	32.5	8.3	15.47	2.10
Tibia tilt angle (deg.)	0.20	Normal	−5.85	5.62		
COP to ankle distance (m)	0.00	Normal	0.04	0.03		
Posterior ground reaction force (BW)	0.00	Gamma	0.68	0.42	3.91	0.21
Knee varus–valgus moment (BH.BW)	0.20	Normal	0.02	0.05		
Knee internal–external rotation moment (BH.BW)	0.20	Normal	0.02	0.05		
Hamstring muscle force (N)	0.00	Gamma	500	203	6.09	82.05
Gastrocnemius force (N)	0.00	Gamma	200	96	4.31	46.41

BH.BW: moment normalized to body height (m) and body weight (N).

injuries and the values of randomly sampled independent variables in each simulation were recorded. A total of 10 Monte Carlo Simulations were performed for each gender. Injury rate in each simulation was calculated as the number of simulated injuries divided by 100,000.

The means and standard deviations of the number of simulated non-contact ACL injuries, injury rate, female-to-male injury rate ratio, and the values of independent variables in simulated non-contact ACL injury trials and non-injury trials were calculated. *t*-Tests were performed to compare the variables with Normal distributions while Mann–Whiney tests were performed for variables with Gamma distributions between simulated injury and non-injury trials to reveal biomechanical characteristics of the simulated injury trials. A Type I error rate of 0.05 was chosen as an indication of statistical significance.

3. Results

The density distributions of the knee valgus–varus and internal–external rotation moments, tibia tilt angle, and COP to ankle distance at the time of posterior ground reaction force were distributed as a Normal distribution for both genders (Tables 3 and 4). The density distributions of the peak posterior ground reaction force, and hamstring and gastrocnemius muscle forces at the time of posterior ground reaction force were distributed as Gamma distributions for both genders (Tables 3 and 4). The density distribution of the knee flexion angle was distributed as a Normal distribution for male subjects but as a Gamma distribution for female subjects (Tables 3 and 4).

Table 5
Simulated injury rates and female-to-male injury rate ratio.

Injury rate (SD)		Female-to-male injury rate ratio
Male	Female	
0.0097 (0.0003)	0.0480 (0.0005)	4.96 (0.22)

The injury rate of simulated non-contact ACL injuries during the stop–jump task was 0.0097 (SD = 0.0003) for males and 0.0480 (SD = 0.0005) for females (Table 5). The simulated female non-contact injury rate was significantly greater than male's

Please cite this article as: Lin, C.-F., et al., A stochastic biomechanical model for risk and risk factors of non-contact anterior cruciate ligament injuries. Journal of Biomechanics (2009), doi:10.1016/j.jbiomech.2008.12.005

Table 6
Mean (SD) of biomechanical characteristics of simulated injury and non-injury trials of male subjects.

Variable	Injury	Non-injury	*p*-Value
Knee flexion angle (deg.)	22.0 (8.0)	36.8 (9.6)	0.000
Tibia tilt angle (deg.)	−2.0 (6.2)	−5.1 (6.5)	0.000
COP to ankle distance (m)	0.02 (0.03)	0.04 (0.03)	0.000
Posterior ground reaction force (BW)	1.44 (0.57)	0.67 (0.41)	0.000
Knee valgus moment (BH.BW)	0.07 (0.07)	0.01 (0.05)	0.000
Knee internal rotation moment (BH.BW)	0.02 (0.04)	0.01 (0.04)	0.000
Hamstring muscle force (BW)	0.62 (0.26)	0.67 (0.28)	0.000
Gastrocnemius force (BW)	0.26 (0.14)	0.27 (0.15)	0.023
Sagittal plane loading (N)	1840 (895)	61 (409)	0.000
Non-sagittal plane loading (N)	927 (666)	187 (428)	0.000

BH.BW: moment normalized to body height (m) and body weight (N).

Table 7
Mean (SD) of biomechanical characteristics of simulated injury and non-injury trials of female subjects.

Variable	Injury	Non-injury	*p*-Value
Knee flexion angle (deg.)	24.9 (5.6)	32.9 (8.2)	0.000
Tibial tilt angle (deg.)	−3.4 (5.4)	−6.0 (5.6)	0.000
COP to ankle distance (m)	0.02 (0.03)	0.04 (0.03)	0.000
Posterior ground reaction force (BW)	1.45 (0.47)	0.78 (0.38)	0.000
Knee valgus moment (BH.BW)	0.06 (0.05)	0.02 (0.05)	0.000
Knee internal rotation moment (BH.BW)	0.03 (0.04)	0.02 (0.05)	0.000
Hamstring muscle force (BW)	0.78 (0.33)	0.86 (0.36)	0.000
Gastrocnemius muscle force (BW)	0.33 (0.16)	0.34 (0.17)	0.000
Sagittal plane loading (N)	1773 (604)	347 (445)	0.000
Non-sagittal plane loading (N)	501 (389)	166 (298)	0.000

BH.BW: moment normalized to body height (m) and body weight (N).

($p<0.001$). The female-to-male injury rate ratio was 4.96 (SD = 0.22).

The comparison of biomechanical characteristics at the time of peak posterior ground reaction time between simulated injury and non-injury trials showed that the simulated injury trials had smaller knee flexion angles ($p<0.001$), posteriorly tilted tibial angles ($p<0.001$), COP to ankle joint distance ($p<0.001$), and hamstring and gastrocnemius muscle forces ($p\leqslant 0.028$) than did simulated non-injury trials (Tables 6 and 7). The knee flexion angle at the peak ACL loading in simulated injury trials was 22.0 (SD = 8.0) degrees for males and 24.9 (SD = 5.6) degrees for females. The comparison also showed that simulated injury trials also had greater normalized posterior ground reaction force ($p<0.001$), normalized valgus moment ($p<0.001$), and normalized external rotation moment ($p<0.001$) in comparison to simulated non-injury trials (Tables 6 and 7).

4. Discussion

The results of this study support the validity of the stochastic biomechanical model for the peak ACL loading. The stop–jump task was chosen as a testing task to predict ACL injury rate in this study. This task was frequently performed in basketball games and training, and has been identified as a task in which ACL injuries frequently occur. The estimated female-to-male non-contact ACL injury rate ratio was 4.96 (SD = 0.22), similar to 4.59 (SD = 0.64), the mean female-to-male non-contact ACL injury rate ratio of college basketball players over a 13-year period obtained using traditional epidemiological methods (Agel et al., 2005). These results indicate that the stochastic biomechanical model for peak ACL loading developed in this study accurately estimated at least the injury rate ratio for non-contact ACL injury and achieved the purpose of this study. Therefore, these results support the overall validity of the model.

The validity of the stochastic biomechanical model developed in this study was also supported by some predicted biomechanical characteristics of non-contact ACL injuries. The predicted knee flexion angle at the peak ACL loading in the simulated injury trials was 22.0 (SD = 8.0) degrees for males and 24.9 (SD = 5.6) degrees for females. These results are consistent with those reported in the literature. Boden et al. (2000) reported that the non-contact ACL injuries often occurred immediately after foot contact with the knee positioned in small flexion angles. Olsen et al. (2004) reported that the knee flexion angles in all ACL injury cases they reviewed ranged between 5° and 25°. Cochrane et al. (2007) also reported that most non-contact ACL injuries occurred with the knee flexion angle smaller than 30°. The similarity of the predicted knee flexion angle at injury in this study with those reported in the literature provides further support to the validity of the stochastic biomechanical model developed in this study.

An ACL loading model was developed for the stochastic biomechanical model based on in-vitro data (Markolf et al., 1995), in-vivo data (Li et al., 2005; Nunley et al., 2003) and mathematical analysis data (Imran et al., 2000). ACL loading was assumed to be a linear function of anterior draw force, knee valgus–varus and internal–external rotation moments at a given knee flexion angle. The validity of this assumption is critical for the validity of the ACL loading model. An in-vitro study by Hsieh and Draganich (1998) demonstrated that the knee external flexion moment, quadriceps muscle force, and anterior tibia translation are linearly correlated to each other. An in-vivo study by Fleming et al. (2001) demonstrated that ACL loading is linearly correlated to anterior draw force at the proximal tibia with weight bearing. A finite element modeling study by Bendjaballah et al. (1997) and an in-vitro study by Mazzocca et al. (2003) demonstrated that ACL loading is linearly correlated to knee valgus–varus and internal–external rotation moment as well. These previous studies support the validity of the ACL loading model developed in this study.

The stochastic biomechanical model for risk and risk factors of non-contact ACL injuries developed in this study can be applied in future research to determine biomechanical risk factors. Biomechanical risk factors can be identified through a sensitivity analysis using computer simulations to demonstrate which biomechanical factors have significant effects on the estimated injury rate. The model can also be applied in future research on training programs for preventing ACL injuries. The injury rates of a group of subjects before and after training programs for modifying biomechanical risk factors can be estimated using the stochastic biomechanical model developed in this study. The post training-to-pre training injury rate ratio would be an indication of the effectiveness of the training program. These potential applications of the model developed in this study will significantly lower the cost and shorten the implementation time for future research efforts.

Gastrocnemius and hamstring muscle forces were estimated from EMGs without considering the length–tension and velocity–tension relationships in the stochastic biomechanical model developed in the present study. The lack of consideration of length–tension and velocity–tension relationships might have caused errors in estimated gastrocnemius and hamstring muscle forces. These errors, however, should not significantly affect the estimated variation of ACL loading and female-to-male non-contact ACL injury rate ratio. Previous studies have demonstrated that gastrocnemius and hamstring muscle forces have little effects on ACL loading (More et al., 1993; O'Connor, 1993; Pandy and Shelburne, 1997; Durselen et al., 1995; Pflum et al., 2004).

Although the accurately predicted female-to-male injury rate ratio and selected biomechanical characteristics of non-contact ACL injury demonstrated the overall validity of the stochastic biomechanical model developed in this study, future studies are needed to improve the model. The ACL loading model used in the stochastic biomechanical model in this study was a modification of the model developed by McLean et al. (2004). An ACL anterior shear force share coefficient was added to the model (Eq. (2)), which eliminated the assumption that the ACL is the only structure to bear the anterior shear force at the knee. The ACL elevation angle was also added to the model (Eq. (2)), which eliminated the assumption that ACL loading is the same as the anterior shear force it bears. In addition, cadaver data obtained in previous studies have been included in the model to establish the relationships among biomechanical factors in the model. These modification efforts should have improved the validity of the original ACL loading model. The current ACL loading model, however, assumed that the patella tendon force was the only knee extension moment generator. A knee extension moment sharing coefficient may need to be determined as a function of knee flexion angle to improve the accuracy of estimated patella tendon force and ACL loading. Further, the effect of ACL loading rate on ACL mechanical properties (Danto and Woo, 1993; Lydon et al., 1995; Noyes et al., 1974) may also need to be considered in future studies to determine ACL strength more accurately. Future studies may also proportionally include multiple tasks in the model to improve the accuracy of the predicted injury rate. Finally, the unit of estimated injury rate in this study may need to be converted to that used in traditional epidemiological studies. The unit of the estimated injury rate in this study was in number of injuries per 100,000 stop–jumps while the unit of the injury rate in traditional epidemiological studies is in number of injuries per 1000 exposure hours. Making the estimated injury rate in stochastic biomechanical modeling studies consistent with that in traditional epidemiological studies will allow us to further validate our models.

A stochastic biomechanical model for risk and risk factors of non-contact ACL injuries has been developed. The estimated female-to-male non-contact ACL injury rate ratio was similar to that reported in the literature, which supports the overall validity and application of the model for estimating relative risk of non-contact ACL injuries. The estimated knee flexion angle for the injury was also similar to that reported in the literature, which also supports the validity of estimated biomechanical characteristics of non-contact ACL injuries. Future studies are needed to further validate the estimated injury characteristics of the model.

Conflict of interest statement

None

Acknowledgement

This study was conducted at the Motion Analysis Laboratory of the Center for Human Movement Science in the Division of Physical Therapy at the University of North Carolina at Chapel Hill.

Appendix A. Supporting Information

Supplementary data associated with this article can be found in the online version at doi:10.1016/j.jbiomech.2008.12.005.

References

Agel, J., Arendt, E.A., Bershadsky, B., 2005. Anterior cruciate ligament injury in national collegiate athletic association basketball and soccer: a 13-year review. Am. J. Sports Med. 33, 524–530.

Bendjaballah, M.Z., Shirazi-Adl, A., Zukor, D.J., 1997. Finite element analysis of human knee joint in varus–valgus. Clin. Biomech. 12, 139–148.

Boden, B.P., Dean, G.S., Feagin Jr., J.A., Garrett Jr., W.E., 2000. Mechanisms of anterior cruciate ligament injury. Orthopedics 23, 573–578.

Chappell, J.D., Yu, B., Kirdendall, D.T., Garrett, W.E., 2002. A comparison of knee kinetics between male and female recreational athletes in stop-jump tasks. Am. J. Sports Med. 30, 261–267.

Chappell, J.D., Creighton, R.A., Giuliani, C., Garrett, W.E., Yu, B., 2007. Kinematics and EMG of landing preparation in vertical stop-jump: risk for non-contact ACL injury. Am. J. Sports Med. 5, 235–241.

Cochrane, J.L., Lloyd, D.G., Buttfield, A., Seward, H., McGivern, J., 2007. Characteristics of anterior cruciate ligament injuries in Australian football. J. Sci. Med. Sport 10, 96–104.

Danto, M.I., Woo, S.L.Y., 1993. The mechanical properties of skeletally mature rabbit anterior cruciate ligament and patellar tendon over a range of strain rates. J. Orthop. Res. 11, 58–67.

Davidson, P.L., Chalmers, D.J., Wilson, B.D., 2004. Stochastic-rheological simulation of free-fall arm impact in children: application to playground injuries. Comput. Methods Biomech. Biomed. Eng. 7, 63–71.

Durselen, L., Claes, L., Kiefer, H., 1995. The influence of muscle forces and external loads on cruciate ligament strain. Am. J. Sports Med. 23, 129–136.

Evans, M., Hastings, N., Peacock, B., 2000. Statistical Distributions, Third ed. Wiley, New York, pp. 6–8.

Ford, K.R., Myer, G.D., Hewett, T.E., 2003. Valgus knee motion during landing in high school female and male basketball players. Med. Sci. Sports Exercise 35, 1745–1750.

Fleming, B.C., Renstrom, P.A., Beynnon, B.D., Engstrom, B., Peura, G.D., Badger, G.J., Johnson, R.J., 2001. The effect of weight bearing and external loading on anterior cruciate ligament strain. J. Biomech. 34, 163–170.

Garrett, W.E., Yu, B., 2004. Gender differences in lower extremity kinetics and probability for non-contact ACL injuries in a stop-jump task. In: Proceedings of the American Academy of Orthopaedic Surgery, San Francisco, CA.

Gerleman, D.G., Cook, T.M., 1992. Instrumentation. In: Soderberg, G.L. (Ed.), Selected topics in surface electromyography for use in the occupational setting: expert perspectives. DHHS (NIOSH) Publication, pp. 51–53 (No. 91–100).

Griffin, L., Albohm, M., Arendt, E., Bahr, R., Beynnon, B., DeMaio, M., Dick, R., Engebretsen, L., Garrett, W., 2006. Understanding and preventing noncontact anterior cruciate ligament injuries. A review of the Hunt Valley II Meeting, January 2005. Am. J. Sports Med. 34, 1512–1532.

Gujarati, D.N., 2003. Basic Econometrics, Fourth ed. McGraw-Hill Companies, USA.

Hatze, H., 1981. Estimation of myodynamic parameter values from observations on isometrically contracting muscle groups. Eur. J. Appl. Physiol. Occup. Physiol. 46, 325–338.

Hughes, R.E., An, K.N., 1997. Monte Carlo simulation of a planar shoulder model. Med. Biol. Eng. Comput. 35, 544–548.

Hsieh, L.F., Draganich, L.Y., 1998. Increasing quadriceps loads affect the lengths of the ligaments and the kinematics of the knee. J. Biomech. Eng. 120, 750–756.

Imran, A., Huss, R.A., Holstein, H., O'Connor, J.J., 2000. The variation in the orientations and moment arms of the knee extensor and flexor muscle tendons with increasing muscle force: a mathematical analysis. Proc. Inst. Mech. Eng. H 214, 277–286.

Kaufman, K.R., An, K.N., Litchy, W.J., Chao, E.Y., 1991. Physiological prediction of muscle forces—II. Application to isokinetic exercise. Neuroscience 40, 793–804.

Lamontagne, M., Benoit, D. L., Ramsey, D. K., Caraffa, A., Cerulli, G., 2005. What can we learn from in vivo biomechanical investigations of lower extremity? In: XXIII International Symposium of Biomechanics in Sports, Beijing, China, pp. 49–56.

Langenderfer, J.E., Carpenter, J.E., Johnson, M.E., An, K.N., Hughes, R.E., 2006. A probabilistic model of glenohumeral external rotation strength for healthy normals and rotator cuff tear cases. Ann. Biomed. Eng. 34, 465–476.

Li, G., Defrate, L.E., Rubash, H.E., Gill, T.J., 2005. In vivo kinematics of the ACL during weight-bearing knee flexion. J. Orthop. Res. 23, 340–344.

Lydon, C., Crisco, J.J., Panjabi, M., Galloway, M., 1995. Effect of elongation rate on the failure properties of the rabbit anterior cruciate ligament. Clin. Biomech. 8, 428–433.

Malinzak, R.A., Colby, S.M., Kirkendall, D.T., Yu, B., Garrett, W.E., 2001. A comparison of knee joint motions patterns between men and women in selected athletic maneuvers. Clin. Biomech. 16, 438–445.

Markolf, K.L., Burchfield, D.M., Shapiro, M.M., Shepard, M.F., Finerman, G.A., Slauterbeck, J.L., 1995. Combined knee loading states that generate high anterior cruciate ligament forces. J. Orthop. Res. 13, 930–935.

Mazzocca, A.D., Nissen, C.W., Geary, M., Adams, D.J., 2003. Valgus medial collateral ligament rupture causes concomitant loading and damage of the anterior cruciate ligament. J. Knee Surg. 16, 148–151.

McLean, S.G., Huang, X., Su, A., Van Den Bogert, A.J., 2004. Sagittal plane biomechanics cannot injure the ACL during sidestep cutting. Clin. Biomech. 19, 828–838.

Mirka, G.A., Marras, W.S., 1993. A stochastic model of trunk muscle coactivation during trunk bending. Spine 18, 1396–1409.

More, R.C., Karras, B.T., Neiman, R., Fritschy, D., Woo, S.L., Daniel, D.M., 1993. Hamstrings—an anterior cruciate ligament protagonist. An in vitro study. Am. J. Sports Med. 21, 231–237.

Noyes, F.R., Delucas, J.L., Torvik, P.J., 1974. Biomechanics of anterior cruciate ligament failure: an analysis of strain-rate sensitivity and mechanisms of failure in primates. J. Bone Joint Surg. Am. 56, 236–253.

Nunley, R.M., Wright, D.W., Renner, J.B., Yu, B., Garrett, W.E., 2003. Gender comparison of patella tendon–tibial shaft angle with weight-bearing. Res. Sports Med. 11, 173–185.

O'Connor, J.J., 1993. Can muscle co-contraction protect knee ligaments after injury or repair? J. Bone Joint Surg. Br. 75, 41–48.

Olsen, O.E., Myklebust, G., Engebretsen, L., Bahr, R., 2004. Injury mechanisms for anterior cruciate ligament injuries in team handball: a systematic video analysis. Am. J. Sports Med. 32, 1002–1012.

Pandy, M.G., Shelburne, K.B., 1997. Dependence of cruciate-ligament loading on muscle forces and external load. J. Biomech. 30, 1015–1024.

Pflum, M.A., Shelburne, K.B., Torry, M.R., Decker, M.J., Pandy, M.G., 2004. Model prediction of anterior cruciate ligament force during drop-landings. Med. Sci. Sports Exercise 36, 1949–1958.

Portney, L.G., Watkins, M.P., 2000. Foundations of Clinical Research: Applications to Practice. Prentice Hall Health, Upper Saddle River, NJ.

Santos, V., Valero-Cuevas, F.J., 2004. A Bayesian approach to biomechanical modeling to optimize over large parameter spaces while considering anatomical variability. In: Proceedings of the 26th Annual International Conference of the IEEE Engineering in Medicine and Biology Society, vol. 6, pp. 4626–4629.

Sell, T.C., Ferris, C.M., Abt, J.P., Tsai, Y.S., Myers, J.B., Fu, F.H., Lephart, S.M., 2007. Predictors of proximal tibia anterior shear force during a vertical stop-jump. J. Orthop. Res. 25, 1589–1597.

Stapleton, T.R., Waldrop, J.I., Ruder, C.R., Parrish, T.A., Kuivila, T.E., 1998. Graft fixation strength with arthroscopic anterior cruciate ligament reconstruction. Two-incision rear entry technique compared with one-incision technique. Am. J. Sports Med. 26, 442–445.

Valero-Cuevas, F.J., Johanson, M.E., Towles, J.D., 2003. Towards a realistic biomechanical model of the thumb: the choice of kinematic description may be more critical than the solution method or the variability/uncertainty of musculoskeletal parameters. J. Biomech. 36, 1019–1030.

Yu, B., Garrett, W.E., 2007. Mechanisms of non-contact ACL injuries. Br. J. Sports Med. 41 (Suppl. 1), i47–i51.

Yu, B., Lin, C.F., Garrett, W.E., 2006. Lower extremity biomechanics during the landing of a stop–jump task. Clin. Biomech. 21, 297–305.

Please cite this article as: Lin, C.-F., et al., A stochastic biomechanical model for risk and risk factors of non-contact anterior cruciate ligament injuries. Journal of Biomechanics (2009), doi:10.1016/j.jbiomech.2008.12.005