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INDUCED ANTERIOR CRUCIATE LIGAMENT CREEP INFULENCES GROUND REACTION FORCES AND MUSCLE ACTIVATION IN WALKING

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by

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Submitted in Partial Fulfillment of the Requirements for the
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Department of Kinesiology
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RESEARCH PAPER APPROVAL

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AN ABSTRACT OF THE RESEARCH PAPER OF

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TITLE: INDUCED ANTERIOR CRUCIATE LIGAMENT CREEP INFULENCES GROUND REACTION FORCES AND MUSCLE ACTIVITATION IN WALKING

MAJOR PROFESSOR: Dr. Michael W. Olson

Prolonged loading of the anterior cruciate ligament (ACL) influences thigh muscle activity during walking. The resulting neuromuscular responses of the thigh musculature and ground reaction forces (GRF) during gait initiation after ACL loading are determined.

PURPOSE: To observe electromyography (EMG) activity of thigh muscles and GRF during gait initiation before and after static loading of the ACL.

METHODS: Eleven healthy individuals (5 male, 6 female; aged 21.6 ± 2.9 years, height 1.69 ± 0.10 m, mass 69.5 ± 12.3 kg) with no history of lower extremity pain/injury participated. Participants were seated while the left knee was flexed to 90º and secured to prevent movement. Maximal voluntary isometric contractions were performed in knee flexion and extension. A padded cuff was then fitted around the proximal lower leg, and a cable was fixed around the pad. The cable ran through a pulley system and loaded the leg for 10 min with a 200 N load for males and 150 N load for females respectively. Gait was initiated with the left leg five times immediately before and after static loading. EMG of rectus femoris (RF), vastus lateralis (VL), vastus medialis (VM), biceps femoris (BF), and semimembranosis (SM) were collected. GRFs were normalized to body weight of the individual and analyzed during the first step. One way analysis of variance (ANOVA) was used to identify changes between pre and post-loading steps during the first 50% of the stance phase. Alpha was set at < 0.05.

RESULTS: Average EMG of RF was increased ($p < 0.05$) but did not change significantly in VL, VM, BF and SM muscles from pre to post walking trials. Peak Fy and Fz were not statistically significant ($p > 0.05$) during the post-walking trials. Peak timing of each muscle during heel contact did not vary significantly after loading the knee joint. No significant difference was found in average Fx, Fy and Fz forces during the post-walking trials. Rate of force development changed, but not uniformly between the participants and the change was not statistically significant ($p > 0.05$).

CONCLUSION: The results of the study conclude that there is a change in the thigh muscle activity from pre to post walking trials after loading the knee joint. Statistically significant change was seen in RF muscle. GRF did not show any significant difference between the trials. Thus the induced ACL creep influences the thigh muscle activity in walking, and active individuals should try to avoid ligamentous creep and include frequent rest periods in order to have a higher level of performance, as well as reduce the risk of injury.

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CHAPTER 1

Introduction

The anterior cruciate ligament (ACL) is a significant contributor to knee joint stability preventing excessive anterior translation of the tibia on the femur, excessive internal rotation of the tibia and knee hyperextension (Markolf, Kochan & Amstutz, 1984). The cruciate ligaments have been identified as the primary stabilizers of anteroposterior translation when the knee is flexed .The ACL forces the tibia to internally rotate during anterior tibial translation, indicating that the ACL primarily restrains internal rotational moments during anteroposterior translation (Dargel et al., 2007). It contributes to sensorimotor control of the knee joint providing proprioception along with the mechanical stability through the mechanoreceptors and associated ligament-hamstrings reflex (Dhillon et al., 2011; Gokeler et al., 2011; Tsuda et al., 2001). The ACL was determined to be the primary restraint to anterior tibial translation at all angles of knee flexion, contributing to 80-85% of the total resistance. It was also found to be the primary restraint to medial tibial translation at full extension and 30º of flexion, and secondary restraint to tibial rotation (internal greater than external) and to varus/valgus angulation (Sakane, Woo, Hilderbrand & Fox, 1996).

In walking anterior shear forces are applied by the patellar tendon, gastrocnemius and tibiofemoral contact force while hamstrings and the resultant ground reaction forces apply posterior shear forces (Shelburne, Torry & Pandy, 2005). Peak ACL force occurs in the early stance mainly due to anterior pull of the patellar tendon on the tibia and also due to the balanced muscle, joint-contact and ground reaction forces. ACL forces are smaller in the late stance because the posterior component of the GRF nearly equals the sum of the anterior shear forces by patellar tendon, gastrocnemius and tibiofemoral contact forces. Adduction moment in the knee is mainly resisted by the lateral knee structures (Shelburne et al. 2005) while the ACL helps to protect the knee from excessive valgus forces (Flaxman, Andrew & Daniel, 2012).

ACL rupture is the most common injury sustained by active individuals and the relative injury is sex-specific. An injury or tear can occur when hit hard on the side of the knee such as during a football tackle, overextension of knee or rapid stopping or changing the direction while running, landing from a jump, or turning or side stepping. The ligament is prone to tear or sustain injury in sports like basketball, football, soccer and skiing (Griffin et al, 2008). Common injury is in non-contact sports like soccer in which sudden deceleration, landing and pivoting manoeuvers are repeatedly performed (Yu & Garrett, 2007). The highest incidence is in individuals 15-25 years old who participate in pivoting sports. Seventy percent of ACL injuries occur in noncontact situations (Griffin et al, 2008). ACL injuries often occur with other injuries along with tears to the medial collateral ligament (MCL) and the shock-absorbing cartilage in the knee (lateral meniscus) (Shelbourne & Nitz, 1991). It leads to reduced knee stability altering the knee joint biomechanics.

ACL is the primary restraint to anterior tibial translation (Shelburne et al. 2005). Injury to the ACL results in reduced external rotation and anterior translation of tibia during the terminal swing phase of walking (Andriacchi & Dyrby, 2004). Muscle activation patterns are altered in chronic ACL deficient knees with earlier onset of gastrocnemius while walking (Lindstrom, Fellander-Tsai, Wredmark & Henriksson, 2010). An increase in biceps femoris and decrease in quadriceps and gastrocnemius muscle activity was recorded during swing to stance transition phase of gait in ACL injured subjects compared to normal individuals (Limbird, Shiavi, Frazer & Borra, 2005). No significant change in ground reaction forces was recorded after ACL injury while walking (Lindstrom et al., 2010; Shelburne et al., 2005).

Prolonged static loads applied to the ACL and associated viscoelastic structures result in unbalanced muscular activation which puts individuals at increased exposure to injury (Chu, LeBlanc, Ambrosia, Baratta & Solomonow, 2003). Male and female athletes exhibit decrements in proprioceptive ability and alterations in muscular activity subsequent to muscular fatigue of the knee joint (Susan, Scott & Freddie, 1999). If static tension is applied to a ligament over a period of time, the ligament will exhibit creep (stretch) such that when the tension is removed, the ligament will not retract immediately to its original length nor develop the same tension at a given length as before the creep developed. Ligamentous creep is associated with desensitization of the reflex arc initiated by the mechanoreceptors in the ligament diminishing the reflexive muscle activity of agonists and antagonists of the knee joint and exposing the joint to further instability and potential injury (Chu et al, 2003). Females are expected to be exposed to higher risk than males (Longpre, Potvin & Maly, 2013). ACL creep may develop a neuromuscular disorder consisting of spasms, increased muscle activity and increased force in the agonist muscles without compensation from the antagonists.

Chu et al. (2003) reported that creep developed in the ACL due to static anterior load on the proximal tibia evokes a neuromuscular disorder consisting of spasms and increased contractile ability of the flexors and extensors in torque. Spasm is defined as a non-volitional random and unpredictable muscular activity which indicates tissue damage (Chu et al, 2003). It indicates that some type of injury or damage is elicited in the ACL due to creep developed within. Increased muscle activity, as recorded by electromyography (EMG) and corresponding force in the flexors and extensors is not accompanied by compensation from the corresponding antagonist co-activation. The increased force and EMG activity is also more prominent in women than in men (Chu, et.al. 2003).

The purpose of this study is to observe the EMG activity of thigh muscles and ground reaction forces (GRF) during gait initiation before and after static loading of the ACL. Our hypothesis is the creep developed in the ACL due to prolonged static load will have pronounced impact on the reflex activation of the associate musculature in a manner that may increase the risk of injury. Our basic assumption is by altering the mechanical behavior of the ACL the knee stability and sensory-motor responses from muscles surrounding the knee will be modified. Rotational forces and tibial translation are not being measured which is a potential limitation as ACL creep causes tibial translation and the main function of ACL is to limit tibial translation. This study assists in determining the neuromuscular changes in the knee joint due to ACL creep which will be helpful in knowing the effects of load on knee joint.

CHAPTER 2

Methods

Participants

Eleven healthy individuals (5 male, 6 females, aged 21.6± 2.9 years, height 1.69 ± 0.10 m, mass 69.5± 12.3 kg) with no history of lower extremity pain/injury were recruited for this study. Participants were excluded if they had any current or previous incidents of upper extremity, back, or lower extremity pain/dysfunction within the past 12 months. The procedures were reviewed and approved by the human subjects committee of Southern Illinois University Carbondale (SIUC). All participants agreed to perform the procedures and signed a written informed consent form prior to participation.

Instrumentation

Isokinetic dynamometer. A Biodex System 3 dynamometer (Biodex Medical Systems, Inc., Shirley, NY, USA) was used to collect reaction moment data and control the movement velocity of the left knee while a Biodex left knee attachment unit was secured to the axis. The dynamometer was calibrated before and after data collection and was within the manufacture's specifications. Data were collected at a rate of 100 Hz and saved for future processing. Isometric mode was used for the protocol. Maximum voluntary isometric contractions (MVIC) for knee flexion and extension movements were recorded with knee at 60º and the chair back was held straight to maintain the hip at 90º on the dynamometer.

Kinematics. An infrared motion capture system (Qualisys AB, Gothenburg, Sweden) was used to collect kinematics data (100Hz) from retro-reflective markers adhered to the specific bony landmarks on each participant. Markers of 14 mm diameter were affixed on the skin. On left side of the body markers were placed over the shoulder, posterior superior iliac spine (PSIS), anterior superior iliac spine (ASIS), greater trochanter, femoral condyle, fibular head, medial condyle of tibia, medial and lateral malleoli, fifth metatarsal and heel. On right side of the body markers were placed over the shoulder, ASIS and PSIS. Residuals were calculated to be less than 0.97mm during calibration trials. These kinematics data were synchronized with EMG recordings during the reflexive response trials using the Qualisys Track Manager (QTM) software interfaced with a USB 2533 A/D board (Measurement Computing, Inc., Norton, MA, USA).

Electromyography (EMG). Surface EMG was collected from rectus femoris (RF), vastus lateralis (VL), vastus medialis (VM), biceps femoris (BF) and semimembranosis (SM) of left lower limb using a MA-300 system (Motion Lab Systems, Baton Rouge, LA, USA). The skin was abraded and cleaned with alcohol pads prior to electrode placement. Pre-gelled Ag–AgCl electrodes (Biopac Systems, Inc., Goleta, CA, USA) were positioned at a distance of 2.0 cm center to center from the 1.0 cm^2 collection area of each electrode and aligned parallel along the length of the respective muscle. A ground electrode was positioned on the skin over the left iliac crest. Surface EMG signals were band-pass filtered 20-500 Hz with a common mode rejection ratio of $>$100dB at a frequency of 60 Hz, an input impedance of $>$100 MΩ, and amplified up to 1000 times. Data were collected at a rate of 1200 Hz using a 12 bit A/D board and saved for future processing.

Procedures

Protocol: Pre and post-walking trials were conducted before and after loading the knee joint respectively to record the effect of ACL creep on walking.

Pre-walking trials. Pre-walking trials were recorded before MVIC protocol. Participants initiated walking gait from a static standing position and began the gait cycle with the left leg landing on the force platform. The participants walked at their normal walking speeds. A total of five walking trials were recorded with each trial lasting for 10 seconds.

MVIC. The participant was seated comfortably on the specialized chair and trunk and thigh of the left lower limb were secured to the chair. This chair was used to reduce potential movement of the pelvis and extremities during each protocol. Velcro straps were used to secure each participant's trunk to the chair to minimize extraneous movement. The left lower limb was secured to a cushioned metal arm attached to the dynamometer. The height of the chair and the knee attachment were adjusted to align the lateral epicondyle parallel to the axis of rotation of the dynamometer. The limb weight was measured using the dynamometer. Participants performed three trials of knee flexion and extension contractions alternatively at 60º while seated in an erect trunk position. Verbal encouragement was given to attain maximum effort. MVIC' s were performed for six seconds with 60 seconds rest between maximal efforts. EMG of left RF, VL, VM, BF and SM were recorded for the MVIC trials for six seconds. A 10 minute rest period was provided between the end of the last MVIC and the loading protocol.

Loading protocol. After the 10 minute rest period the left knee was positioned at 90º of flexion and secured to the dynamometer throughout the protocol. The participants were instructed to fold their upper limbs across the chest and avoid any body or limb movement throughout the protocol. A padded cuff was then fitted around the proximal lower leg, and a cable was fixed around the pad. The cable ran through a pulley system and loaded the leg for 10 minutes with a 200 N load for men and 150N load for women respectively. The participants were seated comfortably and were asked to report any discomfort during the protocol. After 10 minutes the load was released and the straps were removed.

Post-walking trials. The participants performed post-walking trials immediately after the loading protocol. The markers and EMG electrode positions were checked and fixed before the walking trials. The participants walked at their normal walking speeds placing their left foot on the force platform at the beginning of the gait cycle. Each trial was recorded for ten seconds for a total of five trials.

Data Processing

Surface EMG signals collected during the MVIC trials were rectified and smoothed with a low pass fourth-order, zero-lag Butterworth filter with a cut off frequency of 4 Hz. The maximum EMG value attained for RF, VL, VM, BF and SM muscles during the MVIC trials for each respective muscle was used for normalization of EMG during the walking trials.
During the 10 minute loading protocol, EMG signals were collected for the first 30 seconds of each minute for 10 minutes and normalized with the maximum EMG of MVIC trials. The means of the normalized EMG of each 30 second time period were then calculated and compared with each other. Torque data collected by the isokinetic dynamometer were filtered with a low pass fourth-order Butterworth filter with a cut off frequency of 2 Hz. Motion capture data were filtered using a low pass Butterworth filter with a cut off frequency of 4 Hz.

Average EMG amplitude and timing of peak EMG amplitude were calculated for the duration while the foot was on the force platform. The vertical force (Fz) is used for determining the contact of the foot with the force platform, and hence latency between contact and peak activity is evaluated between pre and post gait parameters.

The forces were normalized to the body weight of the individual. The average force reading of the medio-lateral force (Fx), anteroposterior force (Fy) and Fz were calculated. Peak

Fy and Fz were determined. Rate of force development (RFD) was calculated for Fy at push off and Fz at foot contact and push off respectively.

$$RFD = \frac{\Delta F}{\Delta t} = \frac{F_{final} - F_{initial}}{t_{final} - t_{initial}}$$

Statistical Analysis

One way analysis of variance (ANOVA) was performed on the dependent variables (average muscle EMG, delay of peak EMG amplitude, force peaks, RFD) between pre and post-walking conditions. Signs of the force data were modified to correspond with the orientation of the room coordinate system. The EMG of each muscle during the loading protocol was analyzed and compared using one way ANOVA. Normalized average of Fx, Fy, and, Fz forces and peaks of Fy and Fz forces were analyzed by one way ANOVA. The level of significance was set at $p < 0.05$.

CHAPTER 3

Results

Electromyography

Average EMG between the subjects showed a significant increase in RF muscle activity ($F_{1, (118)} = 5.805$, $p=0.018$) after loading the knee joint. The means of normalized average EMG (Table 1 and Figure 1) indicate that there was no significant change ($p > 0.05$) in VL, VM, BF and SM activity from pre to post-walking trials.

Table 1

Means (±sd) of average normalized EMG for pre and Post-walking trails.

Condition	RF	VL	VM	BF	SM
Pre-walk	0.044(0.02)	0.142(0.12)	0.090(0.10)	0.145(0.15)	0.153(0.15)
Post-walk	**0.065**(**0.05**)	0.188(0.19)	0.142(0.20)	0.159(0.17)	0.168(0.14)

Note: Bold indicates significant difference from pre-walking trials.

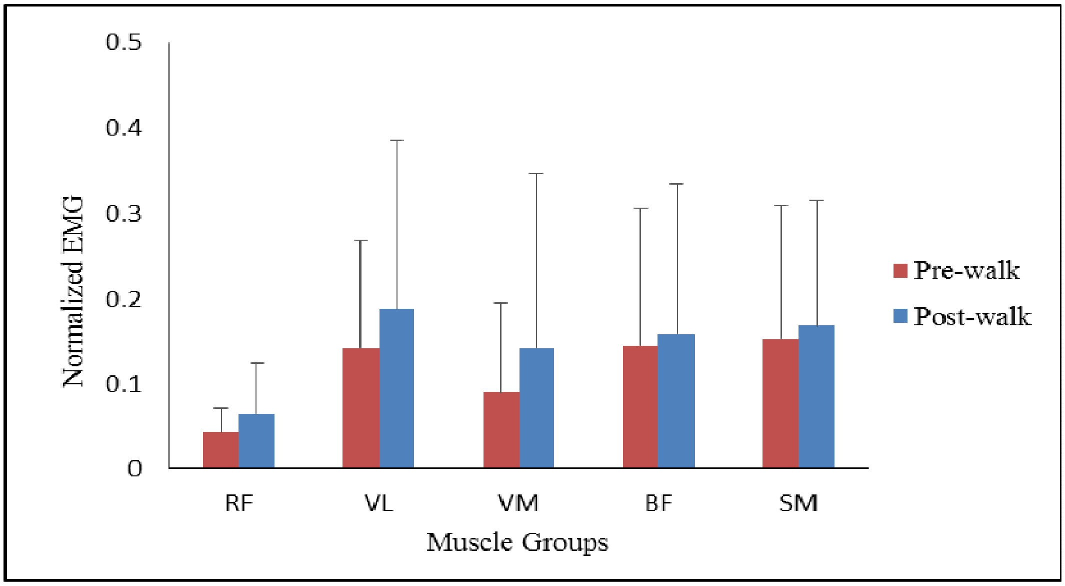

Figure 1. Means (±sd) of normalized average EMG for pre and post-walking trials.

Loading and spasm

The means of normalized average EMG show that there is no significant change ($p > 0.05$) was recorded in the muscles during the 10 minute loading period (Table 2 and Figure 2). Spasms were evident during the loading protocol. SM and RF muscles have undergone spasm in most of the participants compared to other muscles (Figure 3).

Table 2

Means (±sd) of normalized average EMG during the 10 minute loading period.

RF	VL	VM	BF	SM
0.016 (0.01)	0.111 (0.21)	0.032 (0.04)	0.058 (0.07)	0.043 (0.09)

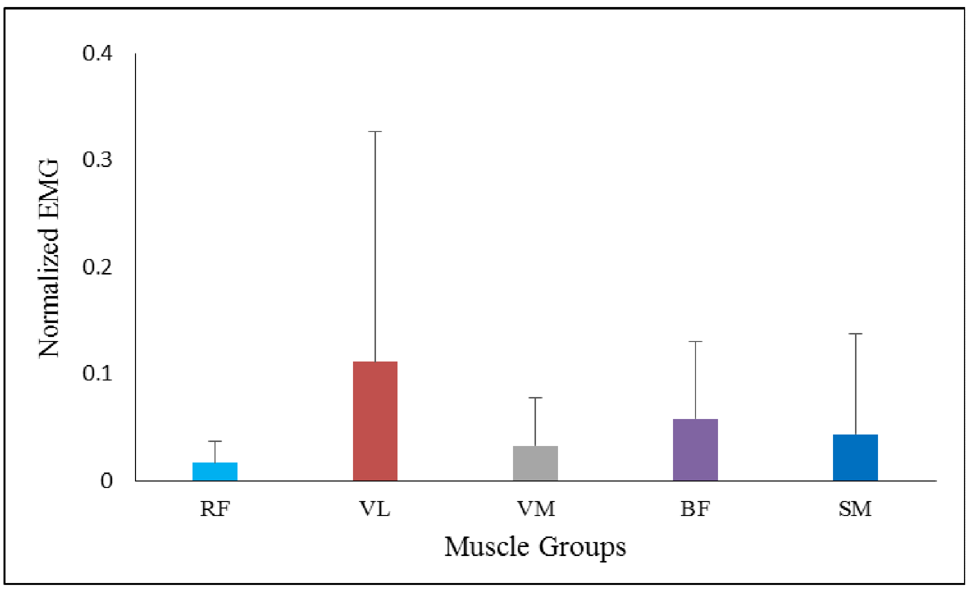

Figure 2. Means (±sd) of normalized average EMG during the 10 minute loading period.

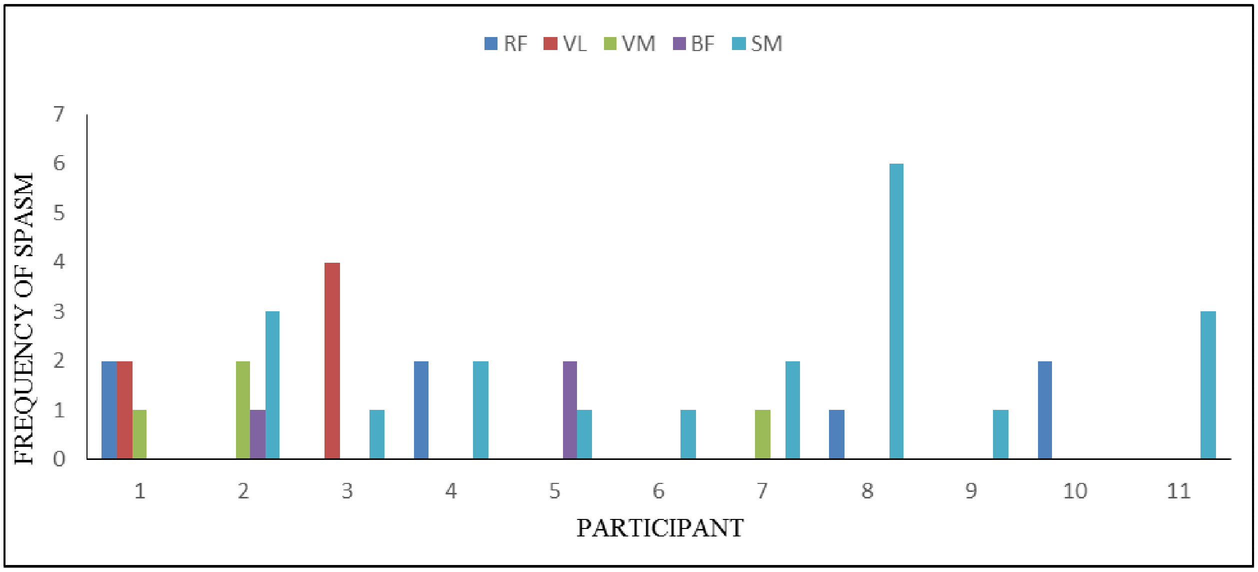

Figure 3. Frequency of spasm in each muscle during the 10 minute loading period per individual.

Peak timing

Peak timing of each muscle at foot contact did not show a significant difference ($p > 0.05$) from pre to post-walking trials (Table 3).

Table 3

Means (±sd) of average peak timing (seconds) of the muscles for pre and post-walking trials.

Condition	RF	VL	VM	BF	SM
Pre-walk	0.054 (0.24)	0.117 (0.07)	0.197 (0.12)	0.208 (0.19)	0.106 (0.05)
Post-walk	0.124 (0.18)	0.156 (0.07)	0.149 (0.11)	0.231(0.22)	0.116 (0.10)

Force data

Average of normalized Fx, Fy and Fz forces did not show a statistically significant change from pre to post-walking trials as shown in Table 4. No significant difference was found in peak Fy and Fz forces during the post-walking trials (Table 5).

Table 4

Means (±sd) of normalized average forces during pre and post-walking trials.

Condition	Average Fx	Average Fy	Average Fz
Pre-walk	-0.068 (0.02)	0.019 (0.01)	0.739 (0.05)
Post-walk	-0.066 (0.02)	0.026 (0.02)	0.719 (0.09)

Table 5

Means (±sd) of normalized peak Fy and Fz forces during pre and post-walking trials.

Condition	Peak Fy	Peak Fz
Pre-walk	0.276 (0.04)	1.007 (0.06)
Post-walk	0.276 (0.04)	1.017 (0.07)

Rate of force development (RFD). RFD for Fy and Fz was not statistically significant ($p > 0.05$) from pre to post-walking trials (Table 6).

Table 6

Means (±sd) of non-normalized average RFD of Fy and Fz for pre and post-walking trials.

Condition	Fy at push off	Fz at foot contact	Fz at push off
Pre-walk	- 721.72 (256.12)	4851.42 (1256.49)	2739.50(705.13)
Post-walk	-678.97 (206.64)	5202.437 (2503.42)	2934.85(678.22)

Note: Units of non-normalized forces - Newtons

CHAPTER 4

Discussion

The purpose of this study was to observe the EMG activity of thigh muscles and GRF's during gait initiation before and after static loading of the ACL. The data from this study show that RF activity increased after loading indicating an effect of the ACL creep on the thigh muscles but the effect was not profound in VL, VM, BF and SM during the post-walking trials. Chu et al. (2003) showed an increased EMG in quadriceps during MVIC' s after loading the knee joint while in the present study we found a significant increase only in RF muscle activity. The sensory and the mechanical properties of the ligament were altered resulting in ACL creep and desensitizing the sensory receptors located in the ligament. The increase in the muscle activity can be attributed to the ligamentous creep that reduced the neural inhibition of the thigh muscles (Chu et al., 2003). This can be a considered as a potential mechanism of neuromuscular disorder which may influence the incidence of injury.

During the loading protocol the participants avoided any limb or body movement for ten minutes maintaining the left knee at 90° of flexion. The load was limited to 150-200 N (Chu et al., 2003) for safety and ethical reasons such that it will not subject the individual to injury but will elicit some response. During normal daily and sport activities knee joint sustains much larger loads than the loads selected for this study. The hamstrings/quadriceps muscles contract to minimize the stress in the ACL ligament (Dhillon et al., 2011, Chu et al., 2003). The spasms recorded from the thigh muscles, evident in this study, were potentially present to minimize the stress of ACL during the loading period. SM and RF muscles have undergone spasm in most of the participants compared to other muscles.

The body weight of an individual mainly transmits through the knee joint while walking and so we analyzed the effect of ACL creep in walking. Peak timing of each muscle did not change significantly during the post-walking trials indicating that passive loading did not alter the latency between foot contact and peak activity.

No significant differences were found in the force data between pre and post-walking trials. Average and peak Fx, Fy and Fz were not influenced by passive knee loading during the post-walking trials. This finding is also in agreement with the findings of Shelburne et al., 2005 where the patellofemoral and tibiofemoral joint forces in normal and ACL deficient knees were not very different while walking as the ACL does not resist much of the varus moment applied by the GRF. Rotational forces and tibial translation were not measured which is a potential limitation as ACL creep causes tibial translation and the main function of ACL is to limit tibial translation.The rate of force development for Fy ad Fz was not altered indicating no difference in the force transmission over time from pre to post-walking trials.

CHAPTER 5

Conclusion

In summary the results of the present study determined the neuromuscular changes in the ACL creep and the effects of load on the knee joint. Modifications in the EMG data from the thigh muscles were detected presenting a potential evidence of neuromuscular alteration to compensate for increased compliance of the ACL. The creep in the ACL did influence the muscle activity and GRF' s while walking but further data are needed to support our hypothesis to evaluate the risk of injury. Further studies are needed to evaluate the effects of ACL creep in walking as gait is a complex motion. In conclusion, the induced ACL creep influences the thigh muscle activity in walking, and active individuals should try to avoid ligamentous creep in order to have a higher level of performance, as well as reduce the risk of injury. So the work and sport activities should be scheduled minimizing the static joint loading and emphasizing sufficient rest periods to allow recovery of ligamentous creep and return to balanced muscular activation and co-activation.

REFERENCES

Andriacchi, T.P., & Dyrby, C.O. (2005). Interactions between kinematics and loading during walking for the normal and ACL deficient knee. *Journal of Biomechanics, 38,* 293-298.

Chu, D., LeBlanc R., Ambrosia P.D., Ambrosia R.D., Baratta R.V., & Solomonow, M. (2003). Neuromuscular disorder in response to anterior cruciate ligament creep. *Clinical Biomechanics, 18,* 222-230.

Dargel, J., Gotter, M., Mader, K., Pennig, D., Koebke. J., & Wiethoff R.S. (2007). Biomechanics of the anterior cruciate ligament and implications for surgical reconstruction. *Strategies in Trauma and Limb reconstruction, 2(1),* 1-12.

Dhillon M.S., Bali K., & Prabhakar., S. (2011). Proprioception in anterior cruciate ligament deficient knees and its relevance in anterior cruciate ligament reconstruction. *Indian Journal of Orthopaedics, 45(4),* 294-300.

Gokeler A., Benjaminse, A., Hewett, T.E., Lephart, S.M., Engebresten, L., Ageberg, E., & Dijkstra, P.U. (2011). Proprioceptive deficits after ACL injury: are they clinically relevant? *British Journal of Sports Medicine, 46(3),* 180-192.

Griffin, L.Y., Aqel, J., Albohm, M.J., Arendt, E.A., Dick, R.W., Garrett, W.E., & Wojtys, E.M. (2000). Noncontact anterior cruciate ligament injuries: risk factors and prevention strategies. *Journal of American Academy of Orthopaedic Surgeons, 8,* 141-150.

Limbird, T.J., Shiavi, R., Frazer, M., & Borra, H. (2005). EMG profiles of knee joint musculature during walking: Changes induced by anterior cruciate ligament deficiency. *Journal of Orthopedic Research, 6(5),* 630-638.

Lindstrom, M., Fellander-Tsai, L., Wredmark, T., & Henriksson, M. (2010). Adaptations of gait and muscle activation in chronic ACL deficiency. *Journal of Knee Surgery, Sports Traumatology, Arthroscopy, 18(1),* 106 -114.

Longpre, H.S., Potvin, J.R., & Maly, M.R. (2013). Biomechanical changes at the knee after lower limb fatigue in healthy young women. *Clinical Biomechanics, 28(4),* 441-447.

Lloyd, D.G., Buchanan, T.S., & Besier, T.F. (2005). Neuromuscular biomechanical modelling to understand knee ligament loading. *Medicine and Science in Sports and Exercise, 37(11),* 1939-1947.

Markolf, K.L., Kochan, A., & Amstutz, H.C. (1984). Measurement of knee stiffness and laxity in patients with documented absence of the anterior cruciate ligament. *Journal of Bone and Joint Surgery, 66(2)*, 242-252.

Rozzi, S.L., Lephart, S.M., Gear, W.S., & Fu, F.H. (1999). Knee joint laxity and Neuromuscular characteristics of male and female soccer and basketball players. *The American Journal of Sports Medicine, 27(3),* 312-319.

Sakane M., Woo S., Hiderbrand K.A., & Fox R.J. (1996). The contribution of the anterior cruciate ligament to knee joint kinematics: Evaluation of its in situ forces using a robot/universal force-moment sensor test system. *Journal of Orthopaedic Science, 1(5),* 335-347.

Shelbourne, K.D., & Nitz, P.A. (1991). The O'Donoghue triad revisited: Combined knee injuries involving anterior cruciate and medial collateral ligament tears. *American Journal of Sports Medicine, 19,* 474-477.

Shelbourne, K.B., Torry, M.R., & Pandy, M.G. (2005). Muscle, ligament and joint-contact forces at the knee during walking. *Medicine and Science in Sports and Exercise, 37(11),* 1948-1956.

Smith, A.J., Flaxman, T.E., Speirs A.D., & Benoit D.L. (2012). Reliability of knee joint muscle activity during eight bearing force control. *Journal of Electromyography and Kinesiology: Official journal of the International Society of Electrophysiological Kinesiology, 22(6),* 914-922.

Tsuda, E., Okamura, Y., Otsuka, H., Komatsu, T., & Tokuya, S. (2001). Direct evidence of the anterior cruciate ligament-hamstring reflex arc in humans. *American Journal of Sports Medicine, 29(1),* 83-87.

Yu, B., & Garrett,W.E. (2007). Mechanisms of non-contact ACL injuries. *British Journal of Sports Medicine, 41,* 47-51.

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Induced Anterior Cruciate Ligament Creep Influences Ground Reaction Forces and Muscle Activity in Walking

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