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Three Dimensional Knee Joint Kinematics and Lower Limb Muscle Activity of Anterior Cruciate Ligament Deficient Knee Joint Participants Wearing a Functional Knee Brace During Running

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# Three Dimensional Knee Joint Kinematics and Lower Limb Muscle Activity of Anterior Cruciate Ligament Deficient Knee Joint Participants Wearing a Functional Knee Brace During Running

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Master's thesis submitted to the
Faculty of graduate and postdoctoral studies

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**Acknowledgements**

*"The noblest pleasure is the joy of understanding"*
*Leonardo DaVinci*

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**Three Dimensional Knee Joint Kinematics and Lower Limb Muscle Activity of Anterior Cruciate Ligament Deficient Knee Joint Participants wearing a Functional Knee Brace during Running**


## Abstract

*Background.* Knee braces have been found to provide limited stability to the ACL deficient knee in situations where the knee is loaded during sporting movements. Different adaptation strategies have been found between patients that can cope with the injury and patients that cannot. One of the expected changes can be muscle activation characteristics of the injured knee during strenuous activity with and without a functional knee brace.

*Methods.* Three dimensional (3D) kinematic and electromyography (EMG) data were collected from each participant for ten consecutive gait cycles during running on a treadmill under both braced and unbraced conditions. Participants were administered the "Knee Outcome Survey Activities of Daily Living Scale" to distinguish functional and non-functional candidates.

*Findings.* No significant differences on 3D kinematics and EMG data were noted between functional and non-functional participants, thus data analysis focused on comparisons of bracing conditions for one combined group. Bracing significantly increased total range of motion in the sagittal plane ($p<0.05$) and reduced total range of motion in the transverse plane ($p<0.05$). Muscle activity at heel-strike showed a consistent trend to increase for the hamstrings and decrease for the quadriceps under the braced when compared to the unbraced condition.

*Interpretations.* Our findings indicated that bracing the ACL deficient knee altered kinematics of the injured leg while running. Tendencies towards reductions in quadriceps and increases in hamstrings activity at heel-strike provides added stability to the injured knee. These changes to bracing further support the mechanical and the proprioceptive contributions of the functional knee brace to protect the ACLD knee.

Keywords: Anterior cruciate ligament, Functional knee brace, Electromyography, Proprioception

**Three Dimensional Knee Joint Kinematics and Lower Limb Muscle Activity of Anterior Cruciate Ligament Deficient Knee Joint Participants wearing a Functional Knee Brace during Running**

## Introduction

Functional knee braces were developed to compensate for ligament deficiency but their efficiency is unclear. The effects of functional knee braces on the anterior cruciate ligament deficient (ACLD) limb are well documented, but there is controversy as to their ability to stabilize the affected knee. The majority of studies have found that the added stability provided by functional knee braces is limited to conditions that do not accurately reflect the physiological and mechanical needs found in a sports setting (Beck et al., 1986; Beynnon et al., 1992; Branch et al., 1989; Cawley et al., 1991; DeVita et al, 1991; Vailas & Pink, 1993). Bracing the ACLD knee was found to cause only minor alterations in the knee joint kinematics when landing from a one-legged jump (Ramsey et al., 2001). Braces were also ineffective in preventing excessive anterior translation of the tibia in ALCD knees during the transition from nonweightbearing to weightbearing (Beynnon et al., 2003).

During walking, ACLD patients showed less knee extension during the support phase than normal or reconstructed patients (Boerboom et al., 2001; Knoll et al., 2004). ACLD patients also showed less maximum internal knee rotation during walking when compared to ACL reconstructed and control groups (Georgoulis et al., 2003). The increased laxity in the joint requires the patient's body to compensate for the ACLD by also altering muscle recruitment patterns, such as the hamstrings and quadriceps to adequately stabilize the knee during such activities (Branch et al., 1989; Boerboom et al., 2001; Kalund et al., 1990; Liu et al., 2000; Rudolph et al., 2001; More et al., 1993; Smith et al., 2003). Specifically, the phenomenon of "quadriceps avoidance gait" was observed in the injured leg of ACLD patients during gait (Berchuck et al., 1990; Limbird et al., 1998). This reduced quadriceps activity during the early stages of stance has since been questioned and found to be inconsistent (Rudolph et al., 2001; Timoney et al., 1993). This lack of consistent findings regarding the involvement of quadriceps and hamstrings of ACLD knees may be attributed to the fact that investigators failed to account for patients able to efficiently cope with ACLD (Chmielewski et al., 2001; Rudolph et al., 2001). With the use of questionnaires and functional tests, these authors identified "copers" and "non-copers" (Irrgang et al., 1998). They found that "non-copers" had less knee range of motion and demonstrated a "stiffening strategy" when compared with copers and healthy participants during jogging.

Bracing the ACLD knee has also been found to significantly alter EMG activity of the involved limb during various functional activities (Branch et al., 1989; Németh et al., 1997; Ramsey et al., 2003; Smith et al., 2003). Changes in muscle activation patterns triggered by the brace could result from the added mechanical stability and thus, causing a different need for stabilization through muscle contractions (Branch et al., 1989). However, these changes could more accurately be explained by skin pressure produced with the brace therefore producing a proprioceptive mechanism influencing muscle contraction, particularly agonist/antagonist cocontractions (Németh et al., 1997; Ramsey et al., 2003). It has also been shown that sensorimotor performance (reproduction of knee joint positions) of ALC reconstructed knees was increased when patients were wearing either a functional or a placebo knee brace when compared to the same task without the use of a brace (Wu et al., 2001). The added stability provided by the brace could thus be more a consequence of its proprioceptive properties, and less attributed to the mechanical stabilization.

To truly understand the role and effects of the functional knee brace, it is important to experiment with it under conditions similar to those in which it is actually being used in a sports setting. Braces are often used in settings involving combinations of jumping, running, and cutting movements. Considering the fact that its mechanical properties may play a secondary role to its proprioceptive effects in ACLD knee stabilization during activities such as jumping and alpine skiing, knee bracing might not be as efficient at protecting the knee joint as previously thought (Németh et al., 1997; Ramsey et al., 2003). Additionally, muscular activity changes of ACLD limbs caused by functional bracing during running remain unclear. The purpose of this investigation is therefore to examine the muscular activity and 3D knee joint kinematic changes of ACLD participants in the involved leg under bracing condition during running. Furthermore, discrepancies in these adaptations between “functional” and “non-functional” patients will attempt to characterize these two specific compensation strategies. Our hypotheses are that bracing alters 3D knee joint kinematics, decreases muscle activity in the hamstring muscles, thus reducing the need for agonist stabilization of the ACLD knee joint, and changes the onset of muscle contraction of the involved lower limb for knee stabilization.

**Methods**

Athletes who participate in rigorous sporting activities in which considerable stress is applied to the knee joint frequently use functional knee braces. Understanding the effects of the functional knee brace under these conditions is crucial in determining their efficiency. Consequently, it is the purpose of this study to examine the effects of functional knee braces on the anterior cruciate ligament deficient (ACLD) knee joint during running.

*Participants*

A group of 11 ACLD male participants (mean: 33.5, ± 7.7 yrs, 89.1, ± 12.5 kg, 185.7, ± 7.8 cm) having no prior history of knee pathology aside from the previously mentioned ligament tear were selected for the experimentation. Participants were recruited among patients who were diagnosed as having an ACL tear by an experienced orthopaedic surgeon at the Carleton University Sports Medicine Clinic in Ottawa, Ontario. All the ACLD participants were clinically evaluated with the KT 1000 arthrometer (MEDmetric Corporation, San Diego, USA) and compared with the contralateral leg. All participants obtained a minimum laxity score of +3. Participants were tested 22.75 (± 26.7) months after the ACL injury occurred. All participants were scheduled to undergo ACL reconstructive surgery within the 6 months following the testing session. At the time of the testing session, all participants exhibited full knee range of motion and no pain during walking. Participants were excluded if they had used a functional knee brace on the injured knee prior to the experimentation. All participants read and signed an informed consent form in accordance with the Research Ethics Board of the University of

Ottawa. Only male participants were included in this study because of both anatomical and biomechanical differences between genders. It has been shown that females have different knee stabilization strategies than males (Wojtys et al,. 2003a; Wojtys et al., 2003b; Zeller et al., 2003).

*Functional tests and Questionnaire*

Prior to the testing session, the "Knee Outcome Survey Activities of Daily Living Scale" (Irrgang et al., 1998) consisting of multiple-choice questions (Appendix A) and a series of hop tests were administered to determine the level of deficiency. This information was also used to differentiate "functional" (F) and "non-functional" (NF) participants according to their activity level following the injury. The terms "functional" and "non-functional" are used to describe the two groups of patients because they offer a clearer description of their physical state as it relates to their ability to pursue normal daily activities. Participants were considered "functional" if they scored over 80% in both the questionnaire and hop tests (limb symmetry). The "functional ACL deficient" and "non-functional ACL deficient" groups consisted of five and six participants respectively.

*Motion recordings*

Four high-speed digital video cameras (JVC GR-DVL9800) were positioned on the same side of the injured leg of the participant to record all trials. The cameras recorded at a speed of 60 Hz and were zoomed to include only the ACL deficient limb in the field of view (Figure 1). The filmed volume was approximately 1.5 m x 1.0 m x 0.75 m. A calibration frame (1.5 m x 1.5 m x 1.0 m) equipped with

22 reflective markers was used and each camera was centered on this apparatus to ensure that every marker was in the field of view before experimentation.

Before every testing session, the cameras recorded the calibration frame to allow calibration of the volume where the movement occurred. A PC computer equipped with the SIMI* Motion system (SIMI* Reality Motion Systems GmbH) was used to acquire and analyze video images of the injured limb while the participants were running on the treadmill. Prior to the experimental trials, a standing joint coordinate system (JCS) trial (Grood and Suntay, 1983) was also taken to allow for knee joint 3D kinematics (Figure 2). The JCS is defined as a series of eight markers placed on anatomical landmarks on the participant's injured leg to define the anatomical axes of the knee joint. During all trials, nine reflective markers (three markers per segment) remained on the participant's injured limb, which were visible at all times in at least two camera views. A sound signal generated by the footswitch of the EMG system and transmitted directly into the video cameras was used at the beginning of all trials to synchronize all instruments and to provide an accurate indication of heel-strike.

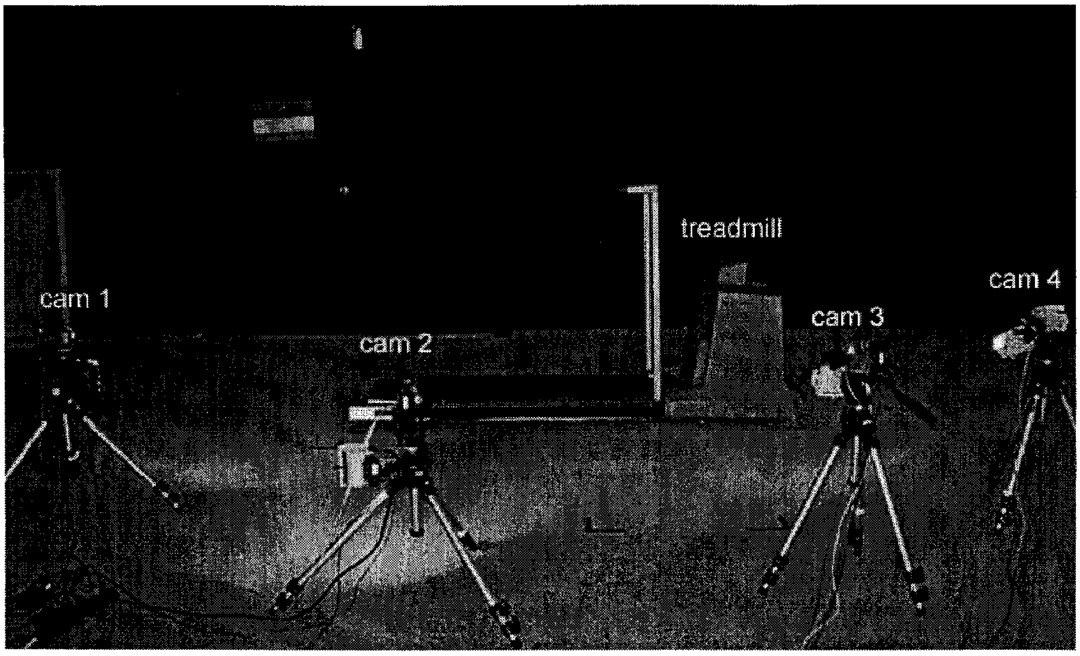


Figure 1. Treadmill and camera setup for 3D kinematics.

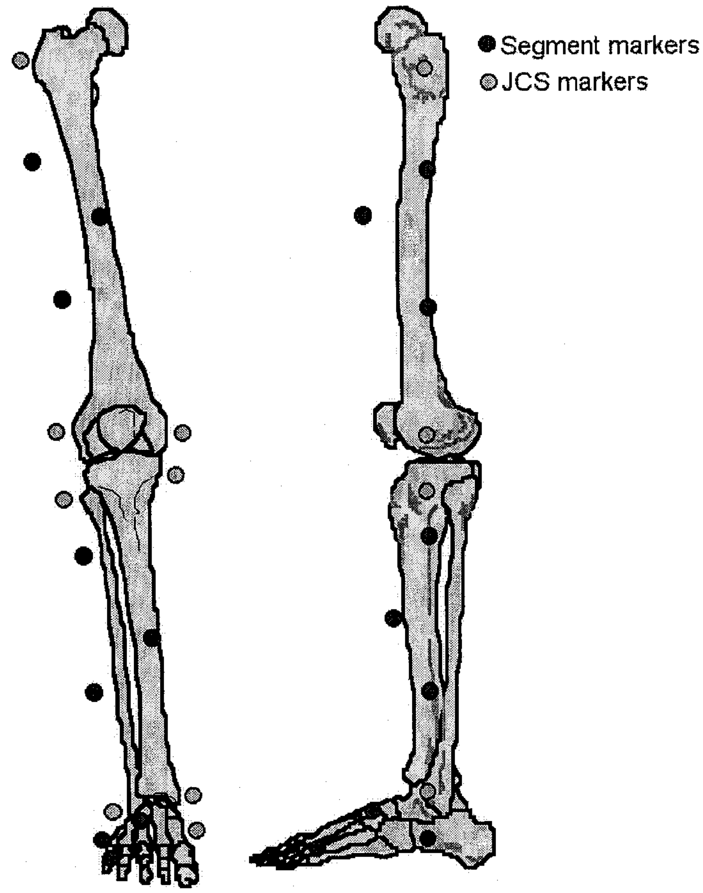


Figure 2. Joint coordinate system (JCS) and skin markers: black dots represent segment markers and grey dots represent JCS markers.

*Electromyography*

In addition to 3D kinematic data, electromyography signal (Bortec Biomedical Ltd., Calgary, Canada) was used to record muscle activity on both lower limbs of the participants during all trials. The surface EMG electrodes (Blue Sensor Ag/AgCl) were positioned on the vastus lateralis and medialis, biceps femoris, semitendinosis, and the lateral and medial gastrocnemius muscles. Electrode leads were connected to two separate eight-channel boxes fitted onto a belt worn around the participant's waist. The signal was collected at 1000 Hz and the amplifier gain was set at 1000. The SIMI software acquired the 16 channels of EMG signal.

*Protocol*

Prior to the running trials, the DonJoy 4titude knee brace was fitted to each participant according to the manufacturer's guidelines and specifications prior to the testing session. Anthropometric measurements of the lower limbs were taken to ensure that the right brace size was selected for the participant. The same investigator was responsible for brace application and adjustment for every participant. Participants were then allowed a warm-up period of five minutes on the treadmill in the braced condition to determine a comfortable running speed. The participants were then required to perform a six-minute running period on the treadmill under both conditions (braced and unbraced). Data were collected for a period of ten seconds during the last minute of the six-minute running period. A total of ten running cycles were used for data analysis. A footswitch placed under the heel of the foot of the injured limb was used to synchronize the running cycle by means of sending a signal to the data acquisition system and the audio channel of each video camera.

Each participant was tested during a single session and used their own running shoes. Dark clothing was provided to facilitate marker identification during the digitizing process.

*Data processing and analysis*

3D kinematics and EMG data were processed and analyzed for ten consecutive running cycles (heel-strike to heel-strike). All ten consecutive cycles were averaged for every participant in both conditions. Raw EMG data were processed using the SIMI* Motion system (SIMI* Reality Motion Systems GmbH). EMG signals were baseline corrected and low pass filtered (2nd order Butterworth) at 6 Hz to produce a linear envelope. From the linear envelope EMG (LE EMG), the amplitudes at heel-strike, and the area under the curves (ILE EMG) were obtained for each muscle and compared across conditions. From the raw EMG data, the onset and activation periods (5% false alarm) were determined for each muscle with custom-made software in MatLab (The Mathworks, Inc. USA) and compared across conditions. The process for detecting the onset of the muscle contraction consisted of selecting first a section of the running cycle where the muscle was not active in order to determine the noise level and the threshold. Then the onset of a contraction was identified when the signal was two standard deviations higher of the threshold. The activation period (time during the muscle activity was present) was determined similarly using the noise level but the software selected all the data points where the activity was two standard deviations higher than the threshold. The result was a time period where the muscle was considered to be active during the running cycle.

Kinematic data were processed with the SIMI* Motion system (SIMI* Reality Motion Systems GmbH) to obtain 3D coordinates of individual segment and JCS markers. The data were then converted into segmental data using custom-made software in MatLab (The Mathworks, Inc. USA) to obtain the 3D kinematics of the injured knee. Peak angular values (flexion/extension, abduction/adduction, internal and external rotations), time to peak angles, and total ranges of motion were compared between conditions.

Because the patients had a clinically diagnosed anterior cruciate deficiency, maximum voluntary contraction (MVC) could not be used to normalize the EMG signals across the participants. Therefore, the peak value in the non-braced condition was used for each muscle to normalize the signals and allow for comparison across conditions. This technique prevented comparisons between participants for variables involving the EMG signal. For this reason, statistical analysis focused on the braced and unbraced conditions.

*Statistics*

Statistical analysis of both the EMG and 3D kinematic data focused on comparisons between the braced and unbraced conditions. From the experimental protocol for the EMG signal, the comparison between subjects was inappropriate. Moreover the initial statistical analysis on the 3D kinematic data could not significantly differentiate the functional and non-functional participants. Therefore all subsequent statistical analyses focused on within participant comparisons. One-way Anova analyses with a p value of 0.05 were used to compare between conditions for both the EMG and 3D kinematic data. Variables compared for the kinematic data were taken from the sagittal (flexion/extension), frontal (abduction/adduction) and transverse planes (internal/external rotation). Variables compared for the EMG data included both amplitude and timing characteristics of the signal. Amplitude parameters were characterized by peak amplitude at heel-strike and LE EMG area (ILE EMG) for both experimental conditions. Timing values included the onset of the muscle contraction, the activation period of the contraction, and the time of the peak amplitude during the running cycle. One-way Anova analyses with a p value of 0.05 were used because there was one independent variable (braced and unbraced conditions) under which the dependant variables were observed (EMG and 3D kinematic variables). In the case of the EMG, data from each muscle was compared separately for the dependant variables across both conditions.

**Results**

During the experimental sessions, no participants experienced discomfort during either the hop tests or the running trials. All participants stated that they were comfortable with the functional brace on before running in that condition. Problems were encountered with the EMG system during the trials of participants 1, 4, and 6. These participants were therefore excluded from the study. Problems were also encountered with the JCS of participant 7 resulting in major errors in the kinematic values. This participant was subsequently excluded from the kinematics analysis but was included for the EMG. A total of 11 participants were included in the study.

A comfortable running speed was determined by the patients and was measured at 2.18 (0.21) m/s in the unbraced condition and 2.12 (0.24) m/s in the braced condition. Differences in running speeds between both conditions were not significant. Table 1 shows the participant information and the results for the "Knee Outcome Survey Activities of Daily Living Scale" and hop test scores. Participants scoring over 80% in both the survey and hop tests were considered to have "functional" ACLD knees. Some participants were not comfortable with the hop tests and therefore did not perform them. They were automatically placed in the NF group. We have decided to provide the participant's information in order to better describe our sample (Table 1).

*Kinematics*

Statistical analysis of the kinematics data did not show any significant differences between the F and NF experimental groups. Therefore, they were considered as one group of ACLD participants (n=11) and the comparisons were focused on the braced and unbraced conditions for both kinematic and EMG data.

*Flexion/extension*

The tibiofemoral flexion/extension pattern varied only slightly during the running cycle between the braced and unbraced conditions. In the braced condition, there was a slight decrease in extension angle. In both conditions, the shapes of the curves were very consistent both in angle amplitudes and timing (Figure 3).

Table 1. Participants information and knee outcome survey of activities of daily living and hop tests scores.

| Participants | Age (yrs) | Weight (cm) | Height (kg) | Time since injury (months) | Survey score | Hop test* | Group** |
|---|---|---|---|---|---|---|---|
| S1 | 32 | 99.8 | 189.0 | 10 | 90 | 94.5 | F |
| S2 | 39 | 97.5 | 204.2 | 60 | 65 | N/A*** | NF |
| S3 | 34 | 113.4 | 180.3 | 72 | 98 | 98 | F |
| S4 | 25 | 90.7 | 185.9 | 7 | 85 | 90 | F |
| S5 | 27 | 77.1 | 179.8 | 6 | 95 | 98 | F |
| S7 | 42 | 82.6 | 185.9 | 6 | 70 | N/A*** | NF |
| S7 | 38 | 78.9 | 179.8 | 9 | 75 | 99 | NF |
| S8 | 28 | 102.1 | 192.0 | 6 | 76 | 76 | NF |
| S9 | 32 | 81.6 | 180.3 | 8 | 71 | 88 | NF |
| S10 | 23 | 74.8 | 176.8 | 60 | 99 | 84 | F |
| S11 | 48 | 81.6 | 189.0 | 5 | 86 | 74 | NF |
| Avg. | 33.45 | 89.10 | 185.73 | 22.6 | | | |
| Std. | 7.70 | 12.47 | 7.82 | 26.8 | | | |

*Hop test scores indicate % of limb symmetry

** Participants were considered to be functional (F) if their scores from both the survey and the hop tests were greater than 80%

*** N/A indicates that participants were not willing to perform the hop tests

*Abduction/adduction*

The abduction/adduction pattern produced the most variation between the two conditions (Figure 4). During the braced condition, the peak abduction value of the ACLD knee, and the total abduction/adduction range of motion during the running cycle were significantly reduced (Table 2). Although the overall shapes of the curves are similar, the angular values were smaller with the brace.

*Internal/external rotation*

The internal/external rotation patterns were again similar in shape but differed in amplitude between conditions (Figure 5). In the unbraced condition, the ACLD knee was rotated internally during the swing phase as the involved leg is preparing for heel strike. This was prevented in the braced condition, thus preventing the involved knee from being internally rotated in anticipation for shock absorption. The brace significantly reduced the total range of motion throughout the running cycle (Table 2). Total range of motion was measured as the difference between peak internal rotation and peak external rotation angles. Peak angles were calculated by averaging the maximum values taken from each participant.

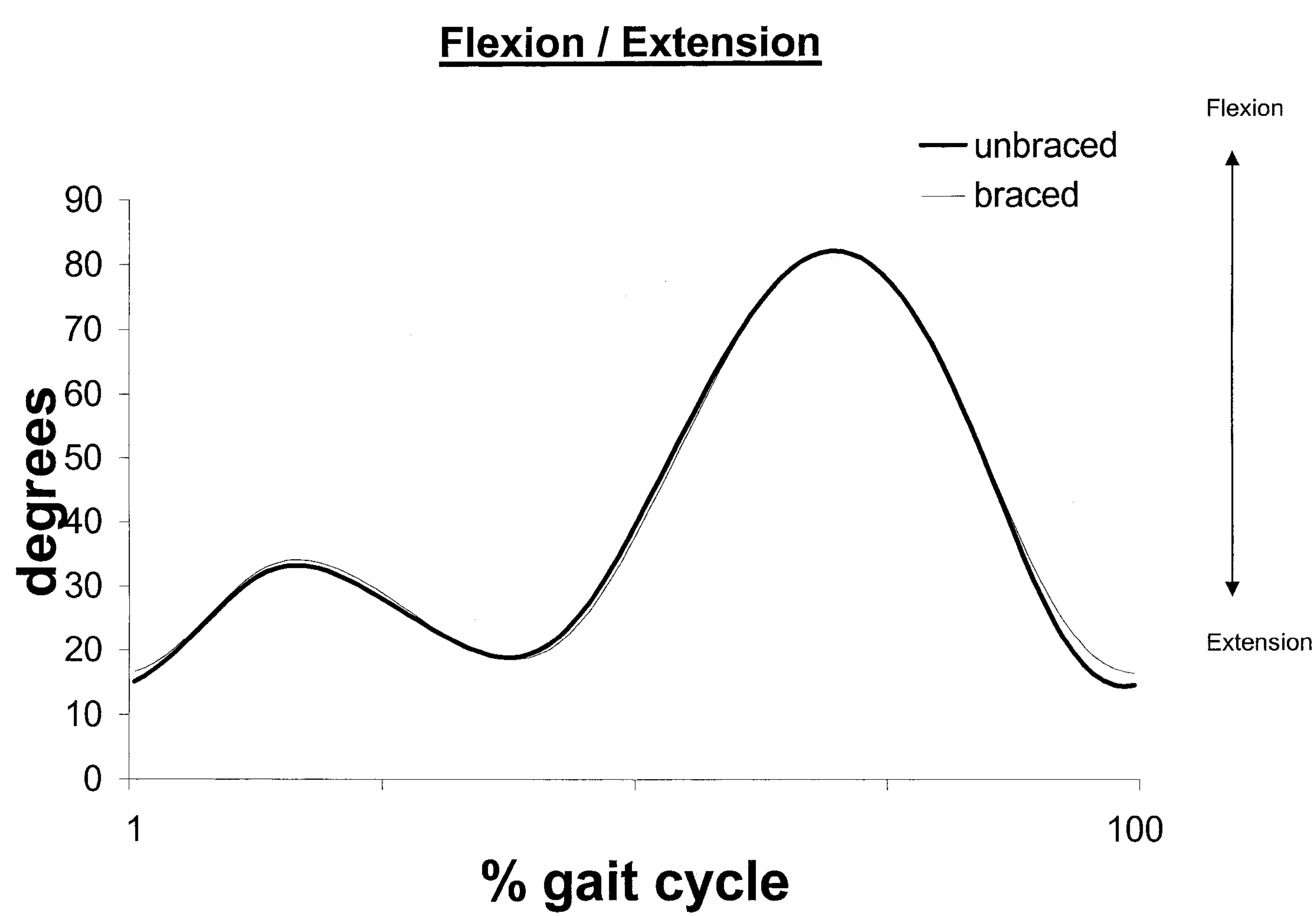


Figure 3. Sagittal plane knee joint angle in the braced and unbraced conditions.

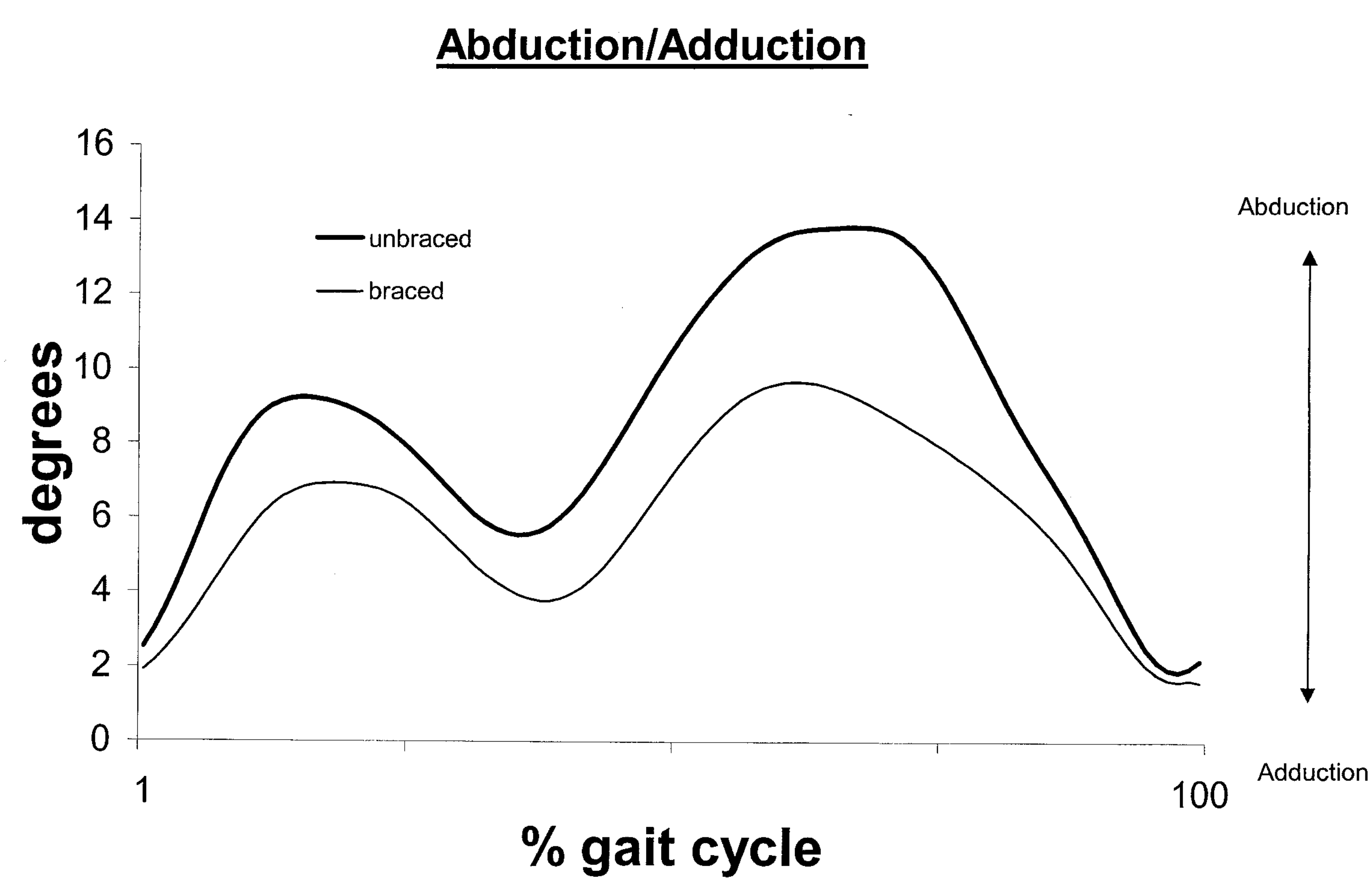


Figure 4. Frontal plane knee joint angle in the braced and unbraced conditions.

Table 2. Kinematics data in the sagittal, frontal, and transverse planes over the duration of the running cycle, n=11 (numbers inside the parenthesis are standard deviations).

| | UB | B |
|---|---|---|
| *Sagittal plane* | | |
| Peak flexion (deg) | 82.4 (8.0) | 83.4 (7.1) |
| Peak extension (deg) | 13.2 (5.5) | 14.2 (4.6) |
| Total flex/ext ROM (deg) | 69.2 (8.9) | 68.7 (6.4) |
| Flex/ext angle at heel-strike (deg) | 14.9 (6.3) | 16.6 (5.3) |
| *Frontal plane* | | |
| Peak abduction (deg) | 15.3 (4.2)* | 10.2 (2.9)* |
| Peak adduction (deg) | 1.2 (2.7) | 1.0 (2.4) |
| Total abd/add ROM (deg) | 14.7 (4.7)* | 9.2 (2.8)* |
| Abd/add angle at heel-strike (deg) | 2.5 (1.9) | 1.9 (1.8) |
| *Transverse plane* | | |
| Peak external rot. (deg) | -10.8 (5.9) | -9.7(5.4) |
| Peak internal rot. (deg) | 5.1 (8.0) | 1.2 (3.0) |
| Total int/ext rot. ROM (deg) | 15.9 (5.6)* | 10.9 (4.8)* |
| Int/ext rot. angle at heel-strike (deg) | -1.7 (8.8) | -4.8 (6.3) |

** A negative value indicates an internal rotation position of the knee

* Significant difference in the angular value between bracing conditions ($p<0.05$)

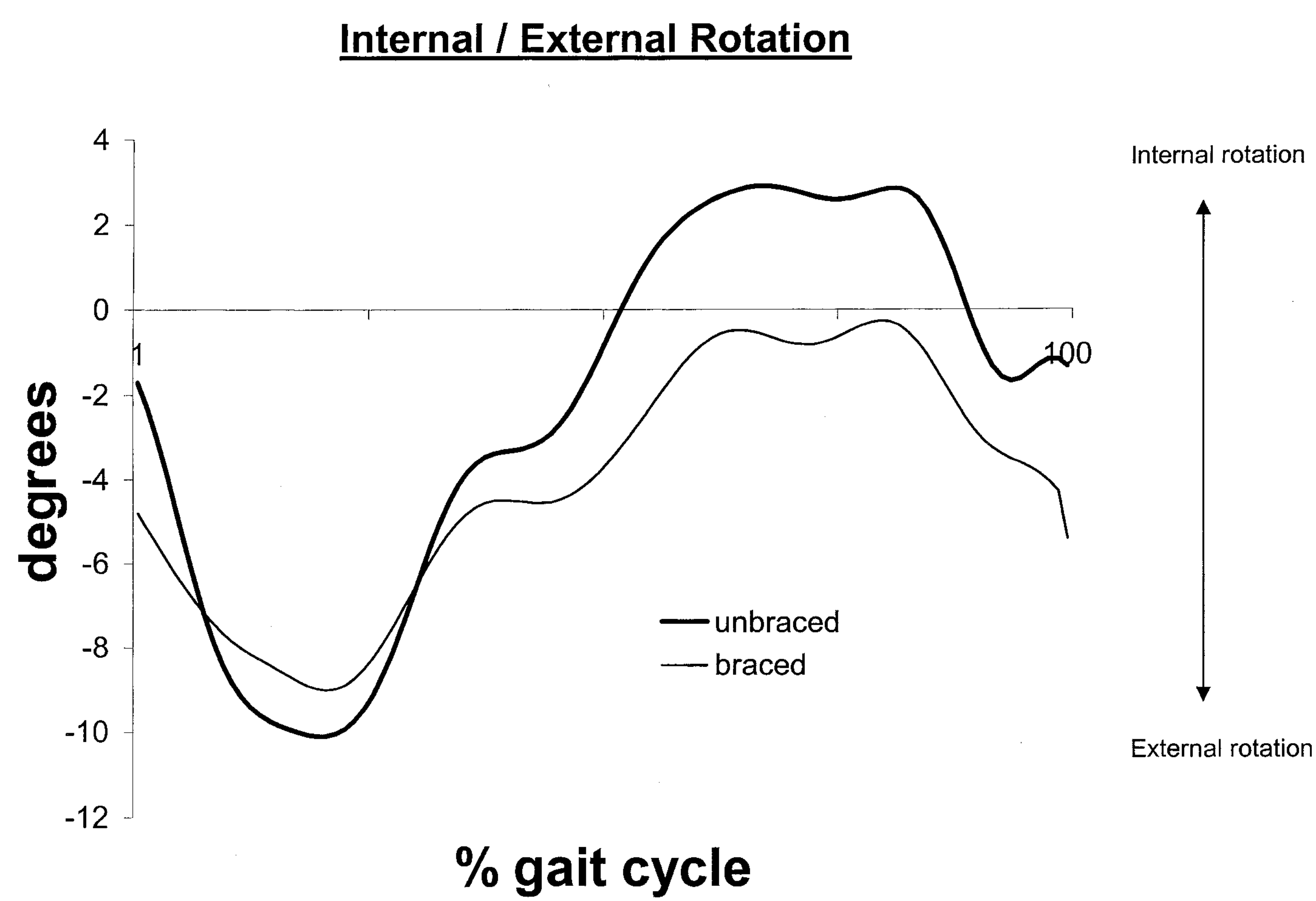


Figure 5. Transverse plane knee joint angle in the braced and unbraced conditions.

*Electromyography*

Although tendencies can be observed in EMG parameters as a result of bracing the ACLD knee, there were no significant differences between bracing conditions for any of the muscles measured. Table 3 shows the data from EMG activity variables for all muscles in the two conditions. Although no significant differences were found, activity of the semitendinosis muscle was altered the most by the brace during running (Figure 6). Tendencies showed that bracing reduced and delayed the peak activity in late swing but increased the activity at heel strike. The rest of the muscles were only slightly affected by the brace. Although not significant, interesting increases in activity are noted for the gastrocnemius muscle group. Peak amplitude differences were calculated to determine the direction of the effect caused by the brace. A reduction in amplitude was observed only for the vastus lateralis muscle while the brace increased the activity of all other muscles. Timing variables in the EMG signal were also affected by the brace (Table 4). The activation period (period of time where the muscle was considered to be active during the running cycle) showed a tendency towards a shorter period of EMG activation as a result of bracing for the hamstrings and gastrocnemius muscles. The onsets of the muscle contractions seem to be delayed by the brace in all but the medial gastrocnemius and vastus medialis muscles, but no significant changes were noted.

Table 3. ILE EMG amplitude at heel strike and peak amplitude difference in the braced and unbraced conditions for each muscle over the duration of the running cycle, n=11 (numbers inside the parenthesis are standard deviations).

| | | ILE EMG | Amp. at HS | | Peak amplitude difference* |
|---|---|---|---|---|---|
| Vast. Lat. | UB | 2.60 (0.32) | 0.73 (0.19) | | |
| | B | 2.61 (0.51) | 0.69 (0.22) | | -0.08 (0.30) |
| Vast. Med. | UB | 2.60 (0.34) | 0.75 (0.20) | | |
| | B | 2.78 (0.54) | 0.67 (0.15) | | 0.05 (0.20) |
| Bic. Fem. | UB | 3.70 (0.10) | 0.40 (0.29) | | |
| | B | 3.82 (0.78) | 0.49 (0.31) | | 0.07 (0.10) |
| Semitend. | UB | 3.19 (0.61) | 0.28 (0.26) | ** | |
| | B | 3.46 (1.13) | 0.42 (0.30) | | 0.03 (0.19) |
| Gast. Lat. | UB | 3.62 (1.13) | 0.47 (0.26) | | |
| | B | 3.90 (1.50) | 0.45 (0.25) | | 0.08 (0.30) |
| Gast. Med. | UB | 3.11 (0.70) | 0.39 (0.24) | | |
| | B | 3.41 (0.86) | 0.49 (0.30) | | 0.10 (0.18) |

* Peak amplitude difference was calculated from peak amplitudes in the unbraced and braced conditions during the running cycle.

** Trend towards an increase in LE EMG amplitude at heel strike in the braced condition (p=0.16).

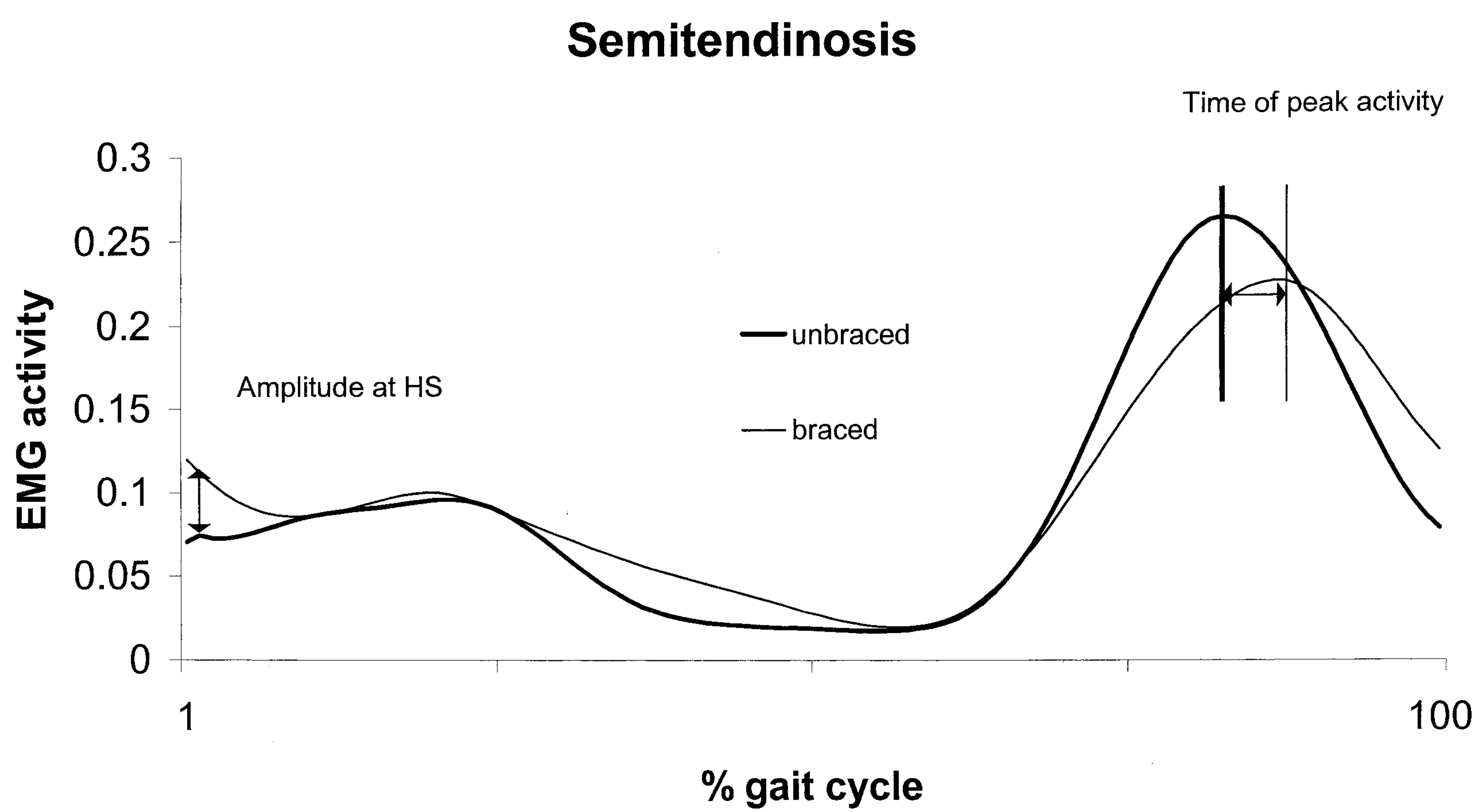


Figure 6. LE EMG profile of the semitendinosis muscle during running for the braced and unbraced conditions showing the differences in amplitude at heel-strike and in timing of the peak activity during the gait cycle.

Table 4. EMG timing parameters under the braced and unbraced conditions for each muscle over the duration of the gait cycle (100%), n=11 (numbers inside the parenthesis are standard deviations)

| | | Onset (%) | Activation period (%) | Peak time (% of gait cycle) | Absolute time (s) |
|---|---|---|---|---|---|
| Vast. Lat. | UB | 84.0 (7.0) | 36.0 (5.0) | 6.8 (5.8) | 0.719 (0.023) |
| | B | 86.0 (5.0) | 36.0 (8.0) | 10.7 (8.8) | 0.711 (0.027) |
| Vast. Med. | UB | 88.0 (6.0) | 31.0 (8.0) | 6.8 (5.6) | 0.719 (0.023) |
| | B | 88.0 (5.0) | 33.0 (4.0) | 8.7 (3.2) | 0.711 (0.027) |
| Bic. Fem. | UB | 71.0 (10.0) | 49.0 (14.0) | 90.5 (12.6) | 0.719 (0.023) |
| | B | 76.0 (7.0) | 45.0 (17.0) | 91.7 (12.7) | 0.711 (0.027) |
| Semitend. | UB | 71.0 (7.0) | 44.0 (14.0) | 85.5 (7.2)* | 0.719 (0.023) |
| | B | 74.0 (9.0) | 41.0 (13.0) | 92.4 (13.5) | 0.711 (0.027) |
| Gast. Lat. | UB | 91.0 (12.0) | 49.0 (13.0) | 18.1 (7.8) | 0.719 (0.023) |
| | B | 92.0 (12.0) | 48.0 (14.0) | 21.5 (13.7) | 0.711 (0.027) |
| Gast. Med. | UB | 89.0 (11.0) | 48.0 (12.0) | 16.9 (7.6) | 0.719 (0.023) |
| | B | 88.0 (7.0) | 47.0 (11.0) | 15.6 (7.2) | 0.711 (0.027) |

* Trend towards a delay in peak activity time in the braced condition (p=0.15).

**Discussion**

Several studies have investigated the effects of functional bracing on the ACLD knee (Beynnon et al., 2003; Boerboom et al., 2001; Ramsey et al., 2001; Ramsey et al., 2003;). The present investigation aimed to identify changes in muscular activation during functional knee bracing with participants with ACL deficiency while running on a treadmill. During the running movement, and particularly surrounding heel-strike, the ACL is subjected to significant strain. We have studied running because it is an important part of sporting activities and should be considered when trying to understand the effects of the brace on the body in motion.

Scores obtained from the "Knee Outcome Survey Activities of Daily Living Scale" and the hop tests indicate that five and six participants were "functional" and "non-functional" respectively. Statistical comparisons between the two groups did not show any significant differences in 3D kinematic variables. This could explain the lack of statistical differences observed in kinematic and EMG variables between the two conditions. Even if the statistical analysis showed no difference between the two groups, there were certainly interactions, which could taint the findings for the analysis between conditions.

*Kinematics*

Fairly similar flexion/extension knee joint patterns were observed between the braced and non-braced conditions. Bracing the ACLD knee did not alter the flexion/extension pattern during running. Despite the added mechanical restrictions, participants were able to obtain the same range of motion while running with the brace. The high variability observed for the time of peak extension can be explained by the fact that in some participants, peak extension occurred slightly before heel-strike whereas for others, it occurred slightly after. This difference is translated into a time value either very early or very late in the running cycle. Total range of motion was higher than other running investigations (Reinschmidt et al. 1997).

The patterns and amplitudes of abduction/adduction during the running cycle were more affected by the brace. Although these patterns were similar in both conditions, bracing significantly reduced the peak abduction angle, and the total range of motion during running. Brace restrictions account for the differences found between both conditions. Similarity between the shapes of the flexion/extension and abduction/adduction curves brings forward the possibility that the large amplitudes in abduction could be the result of cross-talk which can occur if the joint coordinate system is not in perfect alignment with the limb and the brace.

The internal/external rotation curve for the braced condition showed a significantly lower range of motion than in the unbraced condition. In both conditions, the knee was externally rotated at heel-strike and through the stance phase. In the braced condition, the knee reached values that were very close to neutral, limiting the internal rotation of the articulation. In the unbraced condition, the knee went into approximately three degrees of internal rotation during swing, and then returned to external rotation in preparation for heel-strike. The brace therefore kept the knee from rotating internally during the gait cycle. By preventing internal rotation in late swing, the brace had the effect of placing the knee in a more stable position in preparation for heel-strike. Because injuries to the ACL often occur in situations where landing and pivoting are involved, this alteration of joint position is of particular importance in preventing injuries.

Overall analysis demonstrated that bracing reduced the overall range of motion of the knee joint in the frontal and transverse planes but increases the range of motion in the sagittal plane for ACLD participants during running. The amplitudes of the curves for these movements are slightly higher that previously reported in the literature (Reinschmidt et al. 1997). Bracing the ACLD knee during running had the effect of placing the injured limb in a safer kinematic position, particularly in preparation for and during weightbearing. Bracing also had the effect of increasing the efficiency of the stride by increasing stride length. While ACLD participants have been shown to have a more erect gait cycle (Boerboom et al., 2001; Knoll et al., 2004), our results showed that bracing contributes in restoring the altered gait pattern.

*Electromyography*

Comparisons of muscle activity between the braced and unbraced conditions revealed some changes in timing and amplitude characteristics of the EMG signal. Muscle activity (measured as the area under the LE EMG curve) increased for all the muscles in the braced condition. This increase ILE EMG was not significant most likely because of high variability within participants. As reported by Beynnon et al. (2003), bracing is ineffective in reducing excessive anterior tibial translation in the transition from nonweightbearing to weightbearing. Therefore, agonist/antagonist muscle activity during that period would have to assume the task of providing stability to the tibiofemoral joint. Our results showed a tendency that at heel-strike, the EMG amplitude of the quadriceps (vastus medialis and lateralis) decreased while hamstrings (biceps femoris and semitendinosis) increased in the braced condition. These findings were consistent with results obtained by Limbird et al. (1988) where they found decreases in vastus lateralis and increases in biceps femoris during early stance. Increases in hamstring activity in the braced condition were also noted by Németh et al. (1997) when the participant had a more unstable knee joint. Bracing the ACLD knee therefore seems to provide both a mechanical and a proprioceptive stabilization strategy in situations where the knee is more unstable. The increases in EMG activity caused by bracing were not in accordance with Branch et al. (1989) who reported decreases in activity for both quadriceps and hamstrings during the stance phase of side step cutting. Decreases in hamstrings activity were also reported by Ramsey et al. (2003) as participants were landing from a one-legged jump. Our findings suggest that bracing might play a role in increasing agonist while reducing antagonist muscle contractions in the transition from non-weightbearing to weightbearing. It must be noted that the compared motions are very different from all reported studies.

Timing characteristics of the EMG signals were also affected by bracing conditions. Onset of the raw signal was consistently later in the running cycle for all muscles except gastrocnemius medialis and vastus medialis during the braced conditions. Bracing the ACLD knee therefore seemed to have delayed the onset and reduced the time of muscle contractions. Bracing also appeared to decrease the total contraction time of the hamstrings while increasing the time for the vastus medialis muscle, even though differences were not significant. Although differences were not significant, overall muscle activity (ILE-EMG) of the hamstrings showed an increase in the braced condition, while the period of time where this muscle group was active diminished. This observation is translated into a tendency towards an increase in hamstrings activity during late swing and in preparation for heel strike. In the braced condition, the peak activity showed a consistent tendency to occur later in the running cycle for both the quadriceps and hamstrings muscle groups.

The findings of this study suggested that bracing significantly affected the kinematic profile of ACLD patients during running. Although no significant differences were observed for the EMG variables, tendencies were noted both in terms of muscle activity and timing, especially for the semitendinosis muscle. The functional knee brace therefore seemed to have both a proprioceptive and a mechanical effect on the injured knee because differences were noted in the kinematics data, while clear trends suggesting alterations in EMG data were observed but not significant. The tendencies in EMG activity changes caused by the brace to the semitendinosis muscle were not however in accordance with the added mechanical restrictions expected while wearing a functional knee brace. This led us to believe that the brace had a proprioceptive effect on the injured limb, resulting in added active muscular stability. These findings are therefore in accordance with the phenomenon of a proprioceptive contribution of the functional knee bracing (Németh et al. 1997). We must point out that the sample size might have a large influence in the statistical difference in the EMG findings.

Future studies comprised of a larger sample size would be needed to support these results and to further understand the concept of functionality as related to anterior cruciate ligament deficiency. Investigation into the proprioceptive role of knee braces during sporting activities would also provide key information in understanding their effects on muscle firing patterns related to active joint stabilization.

It has been shown that bracing the knee increaseed the ability to reproduce knee joint positions when comparing results to the same task performed without a brace (Wu et al., 2001). Our findings were in accordance with previous studies stating the importance of proprioceptive feedback when considering the effects of functional knee bracing of the ACLD knee (Boerboom et al., 2001; Limbird et al.,1988; Németh et al., 1997; Ramsey et al., 2003).

## Conclusion

Fairly similar flexion/extension knee joint patterns were observed between braced and unbraced conditions. Bracing significantly reduced the peak abduction angle and total range of motion during running. The internal/external rotation angle of the knee joint for the braced condition showed a significantly lower range of motion than in the unbraced condition. Results from the kinematic data from this study differ from previous studies because of the protocol used (JCS) to define the anatomical axes of rotation for the involved limb of the participants. Limitations to this study also include the small number of participants and the normalization protocol of the EMG signal. Bracing the ACLD knee has showed a trend towards increases in muscle activity during the complete running cycle. Increased activity was particularly notable for the hamstrings muscles in the transition from non-weightbearing to weightbearing. A tendency towards a reduction in activity for the quadriceps muscles has also been observed during the same period of the running cycle. These effects contributed to increased knee joint stability by encouraging agonist action of the hamstrings and limiting the antagonist action of the quadriceps muscles. Functional bracing in patients with ACL deficiency may be increasing joint stability through both proprioceptive feedback and mechanical restrictions. These adaptations caused by the brace theoretically place the knee in an optimal situation for shock absorption and active stabilization.

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**APPENDIX A**

## KNEE OUTCOME SURVEY ACTIVITIES OF DAILY LIVING SCALE

*SYMPTOMS*

1. To what degree does pain in your knee affect your daily activity level?

    (5) I never have pain in my knee.
    (4) I have pain in my knee, but it does not affect my daily activity.
    (3) Pain affects my activity slightly.
    (2) Pain affects my activity moderately.
    (1) Pain affects my activity severely.

2. To what degree does grinding or grating of your knee affect your daily activity level?

    (5) I never have grinding or grating in my knee.
    (4) I have grinding or grating in my knee, but it does not affect my daily activity.
    (3) Grinding or grating affects my activity slightly.
    (2) Grinding or grating affects my activity moderately.
    (1) Grinding or grating affects my activity severely.
    (0) Grinding or grating in my knee prevents me from performing all daily activities.

3. To what degree does stiffness in your knee affect your daily activity level?

    (5) I never have stiffness in my knee.
    (4) I have stiffness in my knee, but it does not affect my daily activity.
    (3) Stiffness affects my activity slightly.
    (2) Stiffness affects my activity moderately.
    (1) Stiffness affects my activity severely.
    (0) Stiffness in my knee prevents me from performing all daily activities.

4. To what degree does swelling in your knee affect your daily activity level?

    (5) I never have swelling in my knee.
    (4) I have swelling in my knee, but it does not affect my daily living.
    (3) Swelling affects my activity slightly.

(2) Swelling affects my activity moderately.
(1) Swelling affects my activity severely.
(0) Swelling in my knee prevents me from performing all daily activities.

5. To what degree does slipping of your knee affect your daily activity level?

   (5) I never have slipping of my knee.
   (4) I have slipping of my knee, but it does not affect my daily living.
   (3) Slipping affects my activity slightly.
   (2) Slipping affects my activity moderately.
   (1) Slipping affects my activity severely.
   (0) Slipping of my knee prevents me from performing all daily activities.

6. To what degree does buckling of your knee affect your daily activity level?

   (5) I never have buckling of my knee.
   (4) I have buckling of my knee, but it does not affect my daily activity level.
   (3) Buckling affects my activity slightly.
   (2) Buckling affects my activity moderately.
   (1) Buckling affects my activity severely.
   (0) Buckling of my knee prevents me from performing all daily activities.

7. To what degree does weakness or lack of strength of your leg affect your daily activity level?

   (5) My leg never feels weak.
   (4) My leg feels weak, but it does not affect my daily activity.
   (3) Weakness affects my activity slightly.
   (2) Weakness affects my activity moderately.
   (1) Weakness affects my activity severely.
   (0) Weakness of my leg prevents me from performing all daily activities.

**FUNCTIONAL DISABILITY WITH ACTIVITIES OF DAILY LIVING**

8. How does your knee affect your ability to walk?

   (5) My knee does not affect my ability to walk.
   (4) I have pain in my knee when walking, but it does not affect my ability to walk.
   (3) My knee prevents me from walking more than 1 kilometer.
   (2) My knee prevents me from walking more than ½ kilometer.
   (1) My knee prevents me from walking more than 1 block.

9. Because of your knee, do you walk with crutches or a cane?

    (3) I can walk without crutches or a cane.
    (2) My knee causes me to walk with 1 crutch or a cane.
    (1) My knee causes me to walk with 2 crutches.
    (0) Because of my knee, I cannot walk even with crutches.

10. Does your knee cause you to limp when you walk?

    (2) I can walk without a limp.
    (1) Sometimes my knee causes me to walk with a limp.
    (0) Because of my knee, I cannot walk without a limp.

11. How does your knee affect your ability to go up stairs?

    (5) My knee does not affect my ability to go up stairs.
    (4) I have pain in my knee when going up stairs, but it does not limit my ability to go up stairs.
    (3) I am able to go up stairs normally, but I need to rely on use of a railing.
    (2) I am able to go up stairs one step at a time with use of a railing.
    (1) I have to use crutches or a cane to go up stairs.
    (0) I cannot go up stairs.

12. How does your knee affect your ability to go down stairs?

    (5) My knee does not affect my ability to go down stairs.
    (4) I have pain in my knee when going down stairs, but it does not limit my ability to go down stairs.
    (3) I am able to go down stairs normally, but need to rely on the use of a railing.
    (2) I am able to go down stairs one step at a time with the use of a railing.
    (1) I have to use crutches or a cane to go down stairs.
    (0) I cannot go down stairs.

13. How does your knee affect your ability to stand?

    (5) My knee does not affect my ability to stand. I can stand for unlimited amounts of time.
    (4) I have pain in my knee when standing, but it does not limit my ability to stand.
    (3) Because of my knee I cannot stand for more than 1 hour.
    (2) Because of my knee I cannot stand for more than ½ hour.
    (1) Because of my knee I cannot stand for more than 10 minutes.
    (0) I cannot stand because of my knee.

14. How does your knee affect your ability to kneel on the front of your knee?

    (5) My knee does not affect my ability to kneel on the front of my knee. I can kneel for unlimited amounts of time.
    (4) I have pain when kneeling on the front of my knee, but it does not limit my ability to kneel.
    (3) I cannot kneel on the front of my knee for more than 1 hour.
    (2) I cannot kneel on the front of my knee for more than ½ hour.
    (1) I cannot kneel on the front of my knee for more than 10 minutes.
    (0) I cannot kneel on the front of my knee.

15. How does your knee affect your ability to squat?

    (5) My knee does not affect my ability to squat. I can squat all the way down.
    (4) I have pain when squatting, but I can still squat all the way down.
    (3) I cannot squat more than ¾ of the way down.
    (2) I cannot squat more than ½ of the way down.
    (1) I cannot squat more than ¼ of the way down.
    (0) I cannot squat at all.

16. How does your knee affect your ability to sit with your knee bent?

    (5) My knee does not affect my ability to sit with my knee bent. I can sit for unlimited amounts of time.
    (4) I have pain in my knee when sitting with my knee bent, but it does not affect my ability to sit.
    (3) I cannot sit with my knee bent for more than 1 hour.
    (2) I cannot sit with my knee bent for more than ½ hour.
    (1) I cannot sit with my knee bent for more than 10 minutes.
    (0) I cannot sit with my knee bent.

17. How does your knee affect your ability to rise from a chair?

    (5) My knee does not affect my ability to rise from a chair.
    (4) I have pain when rising from the seated position, but it does not affect my ability to rise from the seated position.
    (2) Because of my knee I can only rise from a chair if I use my hands and arms to assist.
    (0) Because of my knee I cannot rise from a chair.

# FUNCTIONAL TESTS

*HOP TESTS*

*<u>One-legged Hop, 1 hop for distance</u>*

**Healthy leg** __________

**Injured leg** __________

**% of limb symmetry**________

*<u>One-legged Hop, 3 hop for distance</u>*

**Healthy leg** __________

**Injured leg** __________

**% of limb symmetry**________

*<u>One-legged Hop, timed hop over 6 meters</u>*

**Healthy leg** __________

**Injured leg** __________

**% of limb symmetry**________

*<u>One-legged Hop, cross-over for distance</u>*

**Healthy leg** __________

**Injured leg** __________

**% of limb symmetry**________

**AVERAGE % OF LIMB SYMMETRY** ________

# APPENDIX B

## Thesis Proposal

**Background**

Locomotion is one of the most common movements associated with our everyday living. This innate motor pattern takes several forms and functions depending on our needs. With today's society preaching healthy and active living, human locomotion is more than ever encouraged in all its forms may it be walking, running, or playing sports. The study of human locomotion has evolved greatly over time. Advancements in the study of locomotion have been motivated by the demand for knowledge. In this quest for knowledge, we discover new techniques and ideas that allow us to further our understanding. Gait analysis is now an essential tool in diagnosing pathologies and treating patients with prosthesis and orthosis. Injury to the knee has been an increasing phenomenon in the past years. A more active population and an increased participation in sports have caused a subsequent increase in knee injuries. The anterior cruciate ligament (ACL) injury is common with 70% percent of cases occurring in a sports setting (Smith et al. 1993). Injuries to this ligament have been studied extensively and their mechanisms are more defined. There are, however, some uncertainties in the treatment methods and in the body's adaptations to the rupture of this ligament. Functional knee braces have been developed in the hopes of compensating for ligament deficiency but their success is also unclear. The effects of functional knee braces on the anterior cruciate ligament deficient limb are well documented, but there is controversy as to their ability to stabilize the affected knee. Their use has been solicited in both a treatment and a rehabilitative course of action. In a sports setting, the functional knee brace is recommended after treatment of the injury, to further support the injured knee as the athlete often returns to play prematurely. The majority of studies have tested the knee under conditions that do not accurately reflect the physiological and mechanical needs found in a sports setting. It is important, to truly understand the role and effects of the functional knee brace, to experiment with it under conditions similar to those where it is actually being used. A movement that is part of practically every sport setting or rigorous activity is running. Surprisingly, few studies focus their attention of the effects of knee bracing in anterior cruciate deficient patients during running. It is therefore the purpose of the present investigation to quantify the effects of a functional knee brace on the biomechanics of running in patients with recent anterior cruciate ligament deficiency.

**Statement of the Problem**

The identification of participants as "Functional ACL Deficient Knee Joint " and "Non-Functional ACL Deficient Knee Joint" is a recent development in research on the anterior cruciate ligament deficient patient. Patients identified as having functional ACL deficient knee joints experience no apparent disabilities as a result of their ACL deficiency. They are able to take part in the same activities, may they be sports or simply everyday living, as they did before sustaining the injury. On the other hand, patients identified as having non-functional ACL deficient knee joints experience very significant reductions in their ability to function as they did prior to the injury. Their knee stability is greatly reduced and they experience "giving out" of the knee during physical activity. This phenomenon could be the cause of the conflicting results obtained in studies that have failed to categorize the participants.

Other factors such as injury variability and the use of activity involving low physiological demand have also contributed to the inconsistencies in the literature. This study will attempt to categorize the participants as "functional ACL deficient knee joint" and "non-functional ACL deficient knee joint" before experimentation and will analyse the data separately if differences can be found between the two groups. The protocol will involve a dynamic activity and the participants will be at approximately the same post-injury stage of rehabilitation. The results will consequently reflect the actual effect of the functional knee brace on the anterior cruciate ligament deficient knee, eliminating the sources responsible for confusion in the literature.

**Hypothesis**

Our hypotheses are that during running, bracing alters 3D knee joint kinematics, decreases muscle activity in the hamstring muscles, thus reducing the need for agonist stabilization of the ACLD knee joint, and changes the onset of muscle contraction of the lower limb muscles involved in knee stabilization.

**Rationale**

The studies done on functional bracing of the ACL during running were conducted using participants who have evident differences in their injury, treatment method, time since occurrence of injury, and level of activity after sustaining the injury. All of these factors can have a significant influence on the results because the participants are not really comparable. To accurately determine the biomechanical effects of a functional knee brace on the anterior cruciate deficient knee, it is necessary to evaluate participants that are similar in reference to their injury, treatment method, time since occurrence of the injury, as well as activity level. With comparable participants, the effects of functional bracing of the ACL deficient knee will be easier to understand in a very select population of injured participants.

**Limitations**

- Skin movement artefact associated with the use of skin markers introduces some errors in comparison with the use of intracortical pin-mounted markers, which more accurately reflect the actual movements of the skeleton (Reinschmidt et al. 1997). Skin markers accurately reproduce movements such as flexion/extension whereas abduction and rotation movements are participant to a greater risk of error (Reinschmidt et al. 1997). The practicalities of skin markers however largely outweigh the risk of error they are associated with in the case of the present study.
- Participant variability has been a weakness in past biomechanical studies involving ACL injuries. It is rather difficult to obtain an adequate number of participants who are at approximately the same stage in their rehabilitation process. It is therefore important to attempt to minimize inter-participant variability by focusing on participants that are “waiting” for an ACL reconstruction.

- To sufficiently stress the ACL, the participants will have to run at reasonable running speeds without feeling uncomfortable or hesitant. This psychological barrier may prevent the ACL from being stressed enough to allow for distinguishable differences in kinematics.
- The numerous data acquisition devices that will be placed on the participant's involved limb may create discomfort and alter the normal movements of the joints.
- The number of participants

**Review of Literature**

The knee is the largest articulation in the human body and is exposed to considerable amounts of stress during physical activity; therefore its stability is crucial. The articulation of the knee joint can be divided into two distinct articulations. The distal end of the femur articulates with the proximal end of the tibia to form the tibiofemoral joint, and the patella articulates with the patellar surface of the femur to form the patello-femoral articulation. Two restraint systems are responsible for the stability of this articulation. The passive system is composed of the ligaments (medial collateral, lateral collateral, posterior cruciate, anterior cruciate), meniscus (medial and lateral), and joint capsule while the dynamic system is composed of neuromuscular elements.

Three rotational and three translation movements describe kinematically this articulation. The rotational movements are flexion-extension, internal-external rotation, and abduction-adduction. The translation movements are anterior-posterior drawer, medial-lateral shift, and distraction-compression (Reinschmidt et al. 1997). It has been documented that the anterior cruciate ligament (ACL) is the predominant structure involved for knee stabilization and therefore has an important function in the healthy knee (DeVita et al. 1991). ACL injuries are more and more frequent, especially in an athletic setting. Injuries to this ligament often involve rehabilitation or surgery and therefore prevent the athlete from competing for extended periods of time. This review of past literature will focus on the role of the ACL, both in healthy and pathological knees, as well as the effect of wearing a functional knee brace on the participant.

*Knee Joint and Gait Analysis*

Knee function during gait has been studied extensively in many different settings. With our evolving technology, three-dimensional kinematic analysis of the articulation is now possible. Three-dimensional kinematic analysis of the knee provides many advantages compared to two-dimensional kinematic analysis such as the possibility to analyse rotational movements of the articulation. Lafortune et al. (1992) conducted a study aimed at analyzing the three dimensional kinematics of the human knee during walking. To increase accuracy, target markers were fixed to the tibia and femur using intra-cortical traction pins. Three-dimensional coordinates of the markers were measured through the use of high-speed video cameras while the participants were walking at a speed of 1.2 m/s.

The results of this study show that the average pattern of flexion/extension of the tibiofemoral joint was biphasic (slight extension followed by an extension during the stance phase and a large flexion followed by an extension during the swing phase). An initial peak knee flexion of 20° was reached at heel-strike followed by an extension to 1.3° short of full extension during the stance phase. The second knee flexion then reached 35° by toe-off, followed by 60° at full extension. Other motions other than flexion-extension are of relatively small amplitudes. The abduction/adduction pattern showed no movement from heel-strike until shortly before toe-off and the tibiofemoral joint remained abducted approximately 1.2°. From that point until slightly after maximum flexion, the tibiofemoral joint abducted to a mean peak of 6.4°. During the remainder of the swing phase, the joint reassumed 1.2° of abduction. The tibiofemoral joint rotated twice internally during the stance phase, once at heel-strike and then prior to toe-off. The previously mentioned 3D kinematics of the knee joint was obtained from five healthy males (mean age of 27.2 yrs) with no lower-extremity pathologies. It is presumed that participants without an ACL would exhibit larger amplitudes of motion of the tibia in reference to the femur for movements such as abduction/adduction, internal/external rotation, and translation.

The use of skin markers would consequently be less accurate because of the errors associated with skin movement, which are well documented (Reinschmidt et al. 1997). The skin movement artefact during running can be reduced by not placing reflective markers near the bellies of large muscles, which cause skin movement when contracting. In the previously mentioned study, it was found that the mean RMS difference between bone and skin markers were 4.1° in abduction/adduction, 4.4° for internal/external rotation, and 5.3° for flexion/extension of the knee joint. These values reflect a 70.4% difference for abduction/adduction, 63.6% difference in internal/external rotation, and a 20.8% difference for flexion/extension of the knee joint. Errors associated with skin markers are therefore greater for movements that cover a smaller range of motion.

The use of force platforms in analyzing human movement has provided a useful tool for quantifying external forces. Force platforms are used in biomechanical investigations to measure ground reaction forces, which act on the foot during standing, walking or running. Force platforms measure this ground reaction force in 3D with vertical and two shear components acting along the force plate surface (Winter 1990). Furthermore, force platforms also produce torque values about the vertical axis and the location of the center of pressure. The ground reaction forces (kinetic data), combined with kinematic and anthropometric data, allow the researcher to calculate the joint reaction forces and muscle moments (link segment modeling) through inverse dynamics.

As a baseline for comparison to healthy participants, Winter (1983) and Munro et al. (1987) conducted studies using kinetic and kinematic (2D) data on healthy participants performing running trials. Their results can consequently be used as a comparison standard for patients with different pathologies. The study by Winter (1983) was aimed at reporting the patterns of moments of forces and joint powers at the ankle, knee, and hip of healthy adult joggers. For the hip, the beginning of the stance phase showed an extensor moment followed by a flexor moment before mid-stance. This flexor moment is in effect until mid-swing, decelerating the backwards rotation of the thigh and reversing its direction to drive it forward into swing. The extensor moment then returns at the end of swing to decelerate the thigh in preparation for the next foot-strike. The peak extensor moment occurs at 20% of the complete stance phase and reached a value of 30 N·m. The ankle moment pattern is characterized by a large plantar flexor moment shortly after foot-strike, restricting excessive dorsiflexion as the foot is flat on the ground. At mid-stance, the ankle begins plantar flexion, generating 700 W of positive work to produce forward motion. From toe-off and through the entire swing phase, the ankle moment is very low, with sufficient dorsiflexion to ensure that the foot clears the ground. The peak extensor moment at the ankle joint occurs at 60% of the complete stance phase. The knee moment is composed of five power phases, with the peak extensor moment occurring at 40% of the complete stance phase.

The study by Munro et al. (1987), examined the ground reaction force characteristics in 20 healthy male participants performing running trials at different speeds. Several different ground reaction force (GRF) components were analyzed, but only a few are related to the specific topic of the present study. The antero-posterior GRF during running is predominantly biphasic. The initial phase opposes the movement (braking) and the subsequent phase is consistent with the direction of the movement (propulsion). The transition between the braking and propulsion phase occurs at 48% of stance. The vertical GRF for running trials showed the typical double peak pattern. The first "impact peak" is followed by a relative minimum and a subsequent rise to a second "thrust maximum" peak (Munro et al. 1987; Winter 1990). Impact peak values occur between 6% and 17% of total stance time and reach 2.32 times body weight for running speeds of around 5 m/s. Impact peak increases significantly with running speed (Munro et al. 1987). Another important component is the average vertical GRF, which is less participant to intra-individual differences than any other vertical GRF variable. This low variability means that differences observed when comparing participants with a control group point towards a functionally significant difference. This is useful to monitor the effects of an injury to a patient's gait pattern in comparison to data obtained from healthy participants. Average vertical GRF increases significantly with running speed so it is important to control this variable if valid comparisons are to be possible. The medio-lateral GRF characteristics were reported to be very variable between participants by Munro et al. (1987), thus no descriptor variable was established. These findings were supported by past studies and are explained by the small magnitude of this GRF component throughout the gait cycle. The results of this study provide very useful norms for GRF during running in healthy participants.

The results of these studies are an excellent tool in comparing the joint kinetics and kinematics, as well as ground reaction force characteristics, of healthy participants with patients with different pathologies that could produce an abnormality in their gait patterns.

*Role and Function of the ACL: Injury Mechanics*

Inside the combination of stabilizing structures of the knee joint, the ACL serves a highly specialized role in guiding knee motion. Smith et al., (1993) describes the ACL as geometrically complex and consisting of fibre bundles of varying lengths that traverse the knee joint. The ACL limits excessive anterior translation of the tibia as a primary role, and also serves to limit varus-valgus and axial tibial rotations of the knee.

Injury to the ACL is relatively common with approximately 70% of cases being sports related, involving both recreational and high-performance athletes (Smith et al. 1993). To single out a distinct mechanism for this injury is very difficult because the situation is present in both contact and non-contact sports. Several mechanisms have been identified which result in ruptures of the ACL. These mechanisms can be divided into two categories: non-contact, and contact injuries. The non-contact injuries occur most often when the foot is fixed to the ground and the femur rotates on the tibia. Rupture occurs with both internal and external rotation. This happens when the athlete attempts a "cutting manoeuvre", especially when the athlete plants the foot and rotates to the opposite side, causing external rotation of the femur on the tibia. Injury to the ACL is classified as an "all or nothing" phenomenon, the ligament seldom strains without tearing completely. Contact induced ACL injuries occur in team sports (hockey, football, soccer) where the athlete collides with an opponent/team-mate, straining the knee to the point of ACL rupture. A typical mechanism for this type of injury occurs when the lower leg is locked under the athlete while the knee is in external rotation. An external valgus stress then causes the ACL to rupture. Contact injuries also occur typically in downhill skiing when the skier either falls backwards (Barone et al. 1999) or the ski rotates internally when a tip is caught (Johnson et al. 1993).

Treatment of a rupture of the ACL can be oriented either operative or non-operative. Some controversy exists as to which treatment method is more effective. Advances in surgical procedures now allow athletes to return to action as early as four months after the injury.

Strain on the ACL in the healthy and pathological knee has been measured in several experimental conditions. To better understand the factors that induce strain in the ACL, it is necessary to know exactly when and how the ligament is strained during different combinations of muscle contractions of the leg. Beynnon et al. (1995), have conducted an *in vivo* study where Hall effect transducers are surgically positioned on the ACL of healthy participants to measure how the ligament is strained.

The results show that ACL strain was dependant on knee flexion angle during isometric contraction of the quadriceps muscle but was not for isometric contraction of the hamstrings. Isometric contraction of the quadriceps produced significantly higher strain values at 15° and at 30° then at 60° and 90°. Furthermore, the strain values at 15° were significantly higher than at 30°. Compared to the relaxed state, isometric contraction of the quadriceps produced a significant increase in strain at these angles. As for hamstrings isometric contraction, ACL strain values did not depend on knee angle or the magnitude of muscle activity. For all angles of the knee, contraction of the hamstrings did not produce a significantly higher strain value than the relaxed state under a non-weightbearing condition. These findings show that the quadriceps muscles act as an ACL antagonist and that the hamstrings act as ACL protagonists. This statement is supported by the findings of More et al. (1993) who conducted an *in vitro* study aimed at determining the role of the hamstring muscles in an ACL deficient limb. Using an Oxford device to simulate a squatting motion with and without hamstrings load, cadaveric knees were instrumented to measure the kinematics of the knee. Hamstrings load had two significant effects on the kinematics of the knee. Decreases during knee flexion in both anterior translation of the tibia and internal rotation of the tibia were observed.

Fleming et al. (2001) pursued Beynnon's work and conducted similar study, with the addition of weightbearing, which is reported to stiffen or even stabilize the tibiofemoral joint. A differential variable reluctance transducer (DVRT) was arthroscopically applied to the ACL of eleven patients with normal ligament function. A knee-loading fixture was designed to independently apply anterior-posterior shear forces, internal-external torque, and varus-valgus moments to the tibia with and without compressive load to the knee (weightbearing). The results for the mean ACL strain as the knee gradually made the transition from non-weightbearing to the weightbearing condition show that strain changed significantly from –2.0(±1.78)% to 2.1(±1.78)%, with the negative value indicating the absence of strain or loading on the ACL. When the anterior-posterior shear forces were applied, the ACL (non-weightbearing and weightbearing) showed an increase in strain as the anterior shear load increased. During the weightbearing condition, the level of strain on the

ACL was dependant on the level of weightbearing and significantly less than the non-weightbearing condition for anterior shear loads of less than 40 N. The difference was not significant for loads of between 40 N and 130 N. For the internal-external rotational stress test, weightbearing on ACL strain depended on the level of torque applied to the knee. An internal torque of –9 N·m did not produce different ACL strain values for the two conditions whereas an external torque of –9 N·m produced no strain in the non-weightbearing condition, and a significantly higher ACL strain level in the weightbearing condition. Averaging all torque tests produced a significantly higher strain value in the weightbearing condition. For the varus-valgus stress test, weightbearing produced significantly higher ACL strain values. For the range of varus-valgus moments tested, the non-weightbearing ACL wasn't strained. Weightbearing significantly increased ACL strain but the values remained consistent over the range of moments tested. These results show that the effect of weightbearing is not "as protective" as previously mentioned. Increase in weightbearing resulted in an increase in ACL strain for the tests performed in this study.

*Role of the ACL in gait*

Being the primary knee stabilizer, ACL deficiency causes significant knee instability (Smith et al. 1993) and therefore affects a patient's gait cycle. Physiological investigations show that ACL deficiency was responsible for significant strength deficits on the involved limb (McHugh et al. 1994). This same study also shows an 8% increase in $VO_2$ for jogging at 160.9 m/min in a group of

ACL deficient patients compared with healthy control patients. These results demonstrate that the economy of jogging is reduced by the injury. The patient's body therefore has to compensate for the ligament deficiency by altering his muscle recruitment, such as the hamstrings (Boerboom et al. 2001; Kalund et al. 1990; Liu et al. 2000; Rudolph et al. 2001; More et al. 1993) to stabilize the knee. The effect of ACL deficiency on the tibiofemoral joint during gait has not been well established. Patients with ACL insufficiencies have typically shown an abnormal gait pattern identified as "quadriceps avoidance gait" (Berchuck et al. 1990). In their study, Berchuck et al. (1990) compared sixteen patients with unilateral ACL deficiency (14 males, 2 females, mean age 26± 9.5 yrs.) with a control group of 10 healthy participants (5 males and 5 females, mean age 26± 5 yrs.). The results show that ACL deficient (ACLD) knees produced a significantly greater extension moment at foot strike compared to the control group during both level walking and jogging. At mid-stance, the peak flexion moment of the affected knees of the patients was significantly smaller than the healthy knees of the control group. Furthermore, ACLD knees showed no large external flexion moment (created by contraction of the quadriceps) between foot strike and mid-stance, which are typically found in healthy participants. Instead, the external extension moment was balanced by a net internal flexion moment produced by the contraction of the hamstrings. Patients with ACL deficiency consequently show a characteristic temporal pattern of flexion-extension moments where no net quadriceps (extension) moment is necessary during the mid-stance of the gait cycle. This pattern was present in both level walking and jogging, with the patients modifying their gait pattern the most during the walking trials. This pattern, or quadriceps avoidance gait, was present in 75% of the ACLD patients in this study.

The hamstring muscles are knee flexors and therefore resist anterior displacement of the tibia in relation to the femur. This muscle group has been identified as an important element in compensating for ACL deficiency by acting as a knee stabilizer (Kalund et al. 1990).

The use of electromyography (EMG) is an important tool in measuring muscle activity. EMG analysis is the measurement of the action potentials conducted by muscle tissue (Lamontagne et al. 2001). This action potential is detected with electrodes either placed inside the muscle tissue or on the skin surface. These electrodes can therefore monitor muscle activity. ACL deficiency was found to have several effects on muscle activity during both walking and running conditions. During walking, the majority of differences between ACL deficient patients and the healthy control group occur mainly during the transition periods of gait, namely loading and stance-to-swing (Limbird et al. 1988; Ciccotti et al. 1994; van Lent et al. 1994; Rudolf et al. 2001; Boerboom et al. 2001). During the loading phase of walking trials, significantly less muscle activity occurs in rectus femoris, vastus lateralis, and gastrocnemius (Limbird et al. 1988) while more activity is present in biceps femoris. This results in a greater generation of posterior force on the tibia in reference to the femur during loading, further supporting the role of the hamstrings in protection of the ACL deficient knee. ACL deficient patients show an additional phase of rectus femoris activity during mid to late stance and less semitendinosis activity during terminal stance. There is consequently an increase in hamstrings activity during loading of the injured leg of the ACL deficient patients compared to the healthy leg of the healthy control group.

For activities such as running, it was found that these differences are still present but less significant (Limbird et al. 1988; Ciccotti et al. 1994; Rudolph et al. 2001). In a study comparing muscle activity of control and ACL deficient groups during several functional activities, Cicotti et al. (1994) found that there was an overall increase in muscle activity in the involved limb of ACL deficient participants compared to the healthy control group. During the early stance phase of running, vastus medialis and vastus lateralis activity was significantly lower in the ACL deficient group. This supports the notion of "quadriceps avoidance gait" previously mentioned by Berchuck et al. (1990). The middle and late stance phase produced a significant increase in vastus lateralis activity in the ACL deficient group. This increase in hamstrings activity further supports the notion that this muscle group acts as an ACL protagonist and helps protect the injured limb by stabilizing it by muscle contraction.

Rudolph et al. (2001) and Chmielewski et al. (2001) have brought an additional element to the study of ACL deficiency following several studies by their research group. They have identified two categories of ACL deficient patients who exhibit significantly different characteristics. The "copers" have no symptoms of knee instability even during activities involving cutting and pivoting movements. The "non-copers" however, exhibit the normal symptoms associated with ACL deficiency. The purpose of their study was to investigate knee stabilization strategies in both groups of ACL patients (copers and non-copers). The participants were comprised of 10 healthy (control group), 11 copers, and 10 non-copers. The participants were evaluated on a series of activities but for the purpose of this review, our focus will remain on the running trials. The participants were free to run at a speed they felt comfortable to them. Both the copers and non-copers run more slowly and had a shorter stride length than the control group of healthy participants. Upon a kinetic and kinematic analysis, it was found that the average GRF was not significantly lower in the non-copers in comparison to the copers. On the other hand, the non-copers had significantly less knee flexion during loading than the other groups. This reduction in peak knee flexion resulted in a lower moment at the knee of the injured leg in non-copers, compared to their non-injured leg and both legs of the copers and control group. Non-copers, as a protection mechanism, produce a significantly lower knee and higher hip contribution for the total support moment. Muscle activity was also analyzed and as previously mentioned, the differences in muscle activation patterns among the three different groups were less during the running trials than in the walking trials. The overall magnitude of muscle activity was found to be significantly higher for the injured leg of the non-copers compared to the other groups. Furthermore, significantly less activity was present in the vastus lateralis and gastrocnemius muscles.

The results of these studies give an important insight on muscle activity adaptations of ACL deficient limbs. There is however some contradiction involving the results and methodologies. It is clear that differences exist between ACL deficient participants and control groups but as identified by Rudolph et al. (2001), it seems that the copers may be the cause of confusion. It is important to consider these patients because they show an adaptation to ACL deficiency which enables them to function better than the non-copers. Further confusion is present where some authors have identified differences using kinetic and kinematic data, but stated muscle activation conclusions without EMG data to truly support it (Berchuck et al. 1990). This notion of "quadriceps avoidance gait" is also questioned by Timoney et al. (1993) who report that this phenomenon was not present in ACL reconstructed patients, but information on participants at the pre-reconstruction level was not available. Some investigators have also used kinetics to infer muscle activity have concluded that ACL deficient participants reduce the use of the quadriceps muscle because of it's role as an ACL antagonist, yet EMG studies have shown the contrary (Ciccotti et al. 1994). It is therefore important to consider all aspects of gait analysis when attempting to analyze the ACL deficient knee. The majority of confusion present in the literature involves muscle activation patterns because of the identification of patients categorised as having a "functional ACLD knee joint". The use of kinematics, GRF data and electromyography provides the researcher with all the tools to identify the alterations in gait, and relate these differences to muscle activity accurately.

*Role and Function of the Functional Knee Brace*

Functional bracing of the knee has been a solution to stabilize the ACL deficient articulation since the early 1970s. Technological advancements have made braces more and more effective over the last decade. The extent of that effectiveness to stabilize the knee is however somewhat controversial. The use of the functional knee brace is also debated. Athletes need additional support following an injury to the knee, particularly the ACL, and a functional knee brace is used to provide that needed support. These braces are prescribed either to compensate for ACL deficiency in patients where reconstructive surgery is not performed, or to surgery recovering patients to provide added support to allow the healing process to progress smoothly. Functional knee braces are manufactured by a multitude of different companies. However, a few characteristics in brace construction including the uprights, the hinge, and the shell/strapping remain consistent among the different models. The most important characteristic that determines brace performance is hinge placement, which can affect the kinematics of the knee (Vailas and Pink 1993; Liggins and Bowker 1991). Proper fitting of the brace to ensure that the hinge axis is aligned with the axis of rotation of the knee is also very important and requires precise fitting. Misalignment of the axis, frequently caused by brace slippage, may cause restraints and alter knee moments and forces, leading to discomfort, unnecessary ligament stress (Vailas and Pink 1993), and premature fatigue in the affected leg muscles. The factor responsible in controlling this brace slippage is the restraining straps and shell. There are two types of fittings available for functional knee braces: custom-fitted and off-the-shelf models. Although it may seem obvious that custom-made braces should offer the best fit and support, it is not always the case (Beck et al. 1986).

Early investigations involving functional knee braces reported improvements in knee stability but also a reduction in overall performance of the limb and athlete (Houston and Goemans, 1982). This study had high participant variability including different types of knee injuries and different types of functional knee braces. The participants were tested on a number of different criteria including isokinetic contractions, angular velocity, vertical velocity, power, and endurance. The participant variability makes the results of this study of little use if not to suggest that braces have a general positive effect for the injured participant.

Beynnon et al. (1992) more recently conducted an *in vivo* study as a continuation of his work the ACL aimed at measuring the effect of a functional knee brace on ACL strain in healthy knees. A Hall-effect strain-transducer was applied to the ACL and the participant was seated on an instrumented table equipped with a modified Knee Signature System (Acufex Microsurgical, Norwood, Massachusetts). The participants were tested with and without the brace. A total of seven functional knee braces were tested during anterior shear loading, internal/external torque of the tibia, isometric contraction of the quadriceps, and active range of flexion/extension of the knee. Aimed at reducing the strain on the ACL in pathological knees, it was the researcher's hypothesis that the braces would exhibit the same protective properties on a healthy knee. It is reported that no evidence was found that indicates an increase in ACL strain as a result of the use of a functional knee brace. This conclusion means that there is no risk of producing additional strain on the ligament by wearing a functional knee brace, therefore the risk of aggravating an injury to the ACL in also minimal.

Nonetheless, this conclusion is contradicted by the findings of Arms et al., (1987) who obtained an increase in ACL strain during passive knee motion for the braced in reference to the unbraced condition. However, these results were obtained with a cadaveric specimen and older brace models, which can introduce some error and challenge the validity of these results. In anterior shear loading and internal torque conditions, Beynnon found that brace performance was relative to the magnitude of the load. No significant differences were found in ACL strain with and without a functional knee brace when anterior shear loads exceeded 180 N. Therefore, the functional knee braces were effective in shielding the ACL only at relatively low loads. As for the internal/external torque condition, it was reported that only four of the seven braces showed significant strain shielding effects on the ACL with loads of 5 N·m of internal torque. Jonsson and Karrholm (1990) obtained similar results, with the knee brace failing to shield the ACL of strain caused by internal torque. Again, with increased loads, brace performance decreased and strain on the ACL was not significantly less than in the unbraced condition. The functional knee braces tested did reduce strain on the ACL, but not in conditions that would be encountered during sporting or rigorous activity (Beck et al. 1986; Branch et al. 1989; Branch et al. 1990; Cawley et al. 1991; Beynnon et al. 1992). As an additional but important remark, there was no apparent advantage of custom braces compared to off-the-shelf models during the study. These findings are further supported by the results of Jonsson and Karrholm (1990) who conducted an *in vivo* study of ACL deficient patients.

With a wide variety of functional knee braces on the market, Beck et al. (1986) conducted a study aimed at comparing seven different models on three ACL insufficient patients. Anterior tibial displacement was tested using the Medmetric KT-1000 Device (KT-1000) and the Stryker Knee Laxity Tester (KLT). The results of this study support the statement that knee brace performance is reduced with increased loading of the knee. It is also interesting to note that of the seven braces tested, three were off-the-shelf and four were custom-fitted. For the anterior displacement test, the DonJoy 4-Point brace, which is an off-the-shelf model, was the most effective in stabilizing the articulation.

Lamontagne et al. (1997) conducted a study aimed at measuring the net passive elastic moment of force of both ACL deficient and normal knees, with and without a functional knee brace (Legend brace: Smith and Nephew DonJoy Inc.). Joint stiffness was calculated using a KIN-COM isokinetic device at a fixed velocity of 5°/s. The results show a significantly greater elastic moment of force between the braced and unbraced conditions. These increases were greater at the more extreme angles of the range of motion, probably the result of restrictions caused by the straps and mechanical properties of the brace. Lamontagne et al. (1997) also measured the net viscoelastic moment of force (angular damping), using the same participant groupings and conditions as stated in the previous study. The viscoelastic moment was measured at five different angles of knee flexion using the suspension method based on the small oscillatory theory and EMG electrodes were placed on the quadriceps and hamstrings muscles to ensure that the muscles were inactive throughout the experimentation. No significant differences were found between the participants (ACL deficient and control) but significant increases in angular damping were noted between the braced and unbraced conditions. The findings of the previous investigations further supports the presence of a certain mechanical and physiological cost of wearing a functional knee brace.

These previous studies have investigated the effect of functional knee braces on the knee under different testing conditions where the participants were not involved in activities that would be considered typical physiological loading such as running. The results also demonstrate that brace performance is reduced by an increase in joint loading (Beck et al. 1986; Branch et al. 1989; Branch et al. 1990; Cawley et al. 1991; Beynnon et al. 1992). To better understand the effect of the functional knee brace on the stability of an ACL deficient knee, it is valuable to include actual physiologic activity in the experimental protocol. Ramsey et al. (1998) obtained results from a study where seven ACL deficient participants were an *in vivo* three-dimensional kinematic analysis was performed during a *One Legged Jump.* The jump length was to the participant's discretion, with sufficient stress on the ACL resulting. The participants were required to jump from their uninjured limb and to land with their injured limb on the force platform. Steinmann traction pins were surgically implanted into the femur and tibia posterio-laterally with the knee flexed at 45°. Target markers were placed on the pins for kinematic analysis. Kinetic results focusing on peak vertical GRF showed consistency between conditions, but varied slightly between participants. Kinematic data showed that tibiofemoral rotations and translations were consistent across participants, with the most significant differences occurring in the braced and unbraced conditions. The participants therefore showed differences in the way they responded to the braced and unbraced conditions. This phenomenon is possibly due to the fact that some participants were "copers" and others "non-copers". The "copers" would not show variability between the bracing conditions while the "non-copers" would because their limb has not adapted as well to the deficiency and the brace would provide the needed support. This past study was more representative of an actual setting because the participant was performing an exercise that genuinely stressed his ACL.

*Role and Function of the Knee Brace during Gait*

Functional knee bracing is a popular solution for added knee stability for athletes participating in sports. With the ever-increasing speed and size of the athletes, injury to the ACL is becoming more and more frequent especially in sports involving cutting maneuvers and contact. It is therefore important to study the effects of functional knee braces in the conditions in which they are used by the patients. Early studies by Cook et al. (1989) and Marans et al. (1990) have taken a participantive approach as part of their research protocol to determine the participant's impressions on the performance of different functional knee braces for the ACL. The participants would comment on the comfort, level of support, incidences of giving out feeling of the knee, and magnitude of perceived added stability. These results provide useful insight on the participant's impressions of the functional knee braces but do not necessarily reflect the brace's performance.

Cook et al. (1989) also performed a series of static and dynamic tests using the KT-1000, an isokinetic device, and force platform data. Conclusions from this study state that participants demonstrated improved running and cutting performance while wearing their braces. In comparing peak torque of the quadriceps muscles, it was found that one-third of the participants did achieved

80% of the torque achieved with the uninjured limb. That one-third of the participants demonstrated significant increases in force during brace wear for running and cutting maneuvers. These conclusions are somewhat confusing because it is not clear where this data was obtained, creating inconsistency between the participantive and objective approaches.

Cutting manoeuvres have been frequently stated as an injury mechanism for ACL injury. Branch et al. (1989) conducted a study aimed at determining if functional knee bracing of the ACL deficient knee altered muscle firing amplitude, duration, and timing during cutting. The study compared the performances of 10 ACL deficient participants with a control group of 5 healthy participants, using two different functional knee braces. Significant differences were found in comparing the unbraced control participants with the unbraced ACL deficient participants. Peak EMG increased 32% in the lateral hamstrings of the ACL patients during the pre-cut swing phase. During the ACL patients' stance phase, peak quadriceps activity decreased 11% while the peak EMG activity of the medial hamstrings increased by 36%. This peak in EMG activity of the medial hamstrings also occurred 53% (15 ms) later in stance phase in the participants than in the controls. These findings seem to support the fact that the hamstrings adapt to ACL deficiency, creating a protection mechanism during loading. In comparing the braced and unbraced trials of the ACL deficient participants, it was found that during the swing phase of the braced trials, only the medial hamstrings demonstrated a significant alteration in peak EMG activity. The braced trials showed a decrease in peak EMG activity of 27% compared to the unbraced participants. During the stance phase, both quadriceps and medial hamstrings activity was significantly reduced by the brace. Peak quadriceps activity dropped by 12% and peak hamstrings activity dropped by 15%. This indicates that the brace increased the support of the ACL deficient knee, reducing the need for muscle contraction to stabilize the articulation. It is however not clear what the specific role of the brace is during the activity. It is speculated that the brace either acts mechanically in reducing the need for muscle stabilization during the movement or acts by altering the joint position during the movement, stabilizing the articulation without contribution from the muscles.

Biomechanical investigations of the effects of functional knee braces during running are not very abundant and the conclusions are contradictory. The role of the knee brace in static protocols and in dynamic tests using isokinetic dynamometers are well documented. These studies have reported positive contributions of functional knee braces but at loads that do not reflect those encountered during sporting or strenuous activity (Beck et al. 1986; Branch et al. 1989; Branch et al. 1990; Cawley et al. 1991; Beynnon et al. 1992). Performance of knee braces was also significantly reduced when loads increase to levels encountered during activity. In an attempt to fill this void in functional knee brace research, DeVita et al. (1991) conducted a study aimed at assessing the biomechanical effects of a functional knee brace on joint moments of force and powers in the lower extremity during the stance phase of running. This study was performed using five participants with a prior history of ACL pathologies including reconstructed and resected ligaments. A control group of five healthy participants were also tested for comparison. Two-dimensional kinematic data was obtained using a single high-speed video camera at 100 Hz and kinetic data were measured with an AMTI force platform sampling at a rate of 1000 Hz. Five trials were performed by the ACL group under both the braced and unbraced condition. The control group only performed the trials in the unbraced condition. The brace used for all the trials was of the uniaxial hinge, post and strap design. Results for the kinematic data shows no significant differences between the braced and unbraced conditions in the three different parameters. When comparing the ACL group to the control group, differences were found in the hip and knee kinematics. The healthy runners had 8 more degrees of hip flexion throughout the stance phase and the knee was flexed 10° more at heel-strike as well as 12° more at maximum knee flexion. This resulted in the ACL participants walking with a more erect body position than the control group during the stance phase.

The ACL participants showed similar GRF patterns with and without the functional knee brace. The maximum impact force was highest in the ACL unbraced condition; with the control group and braced conditions being 28% and 20% lower respectively. Joint moments of force and powers showed no significant differences between the braced and unbraced conditions for the ACL participants. Significant differences were however obtained between the healthy runners and ACL participants in both braced and unbraced conditions. The extensor angular impulse was 59% larger at the hip of the unbraced ACL participants compared to the healthy group. Impulse values at the ankle joint were significantly higher both the braced and unbraced ACL participants compared to the control group. In contrast, the impulse values at the knee of the control group were approximately twice as important, being 241% and 227% higher than the unbraced and braced ACL conditions. Moment patterns showed that the healthy runners performed 321% more negative work and 191% more positive work at the knee joint. These results are further supported by a subsequent study by DeVita et al. (1996) in which he found that ACL deficient patients used greater extensor torques at the hip and ankle and lower extensor torques and joint power at the knee during gait compared to healthy participants. This can be explained by the fact that ACL deficient patients reduce work at the knee and increase the work at the hip and ankle to protect the unstable knee. In conclusion, the functional knee brace did not affect the kinetics or kinematics of the lower limb significantly in the previously injured participants. This lack of differences between the braced and unbraced conditions can be attributed to several factors such as inter-participant variability. The type of ACL injuries and treatment methods differed greatly among the participants tested. The presence of "copers" and "non-copers" may also have contributed to masking differences between the bracing conditions. The participants were also tested on average 2.8 years postoperatively, meaning that neuromuscular accommodation due to the wearing of a knee brace were present. Testing on ACL deficient participants in a relatively early post-injury period may provide the researcher with a more accurate representation of the effects of bracing.

DeVita et al. (1998) also found reduction in extensor moments at the knee and increases in extensor moments at the hip and ankle in ACL reconstructed participants during walking with a functional knee brace. However, although still significantly different, the general form of the knee moment curve in the braced ACL reconstruction group was closer to the healthy moment curve compared to with the unbraced ACL group. The brace had the effect of reducing the knee moment throughout the stance phase. It is suggested by DeVita that the brace may have an additional neuromuscular effect on the affected knee, on top of the known kinetic and kinematic changes.

## Methodology

To measure the effects of a functional knee brace on the biomechanics of running in patients who have been diagnosed with anterior cruciate ligament deficiency, the experimental protocol will consist of measuring 15 ACL deficient knee joint participants with and without a functional knee brace. The following sections describe the research protocol to be used.

### *Participants*

A group of 15 anterior cruciate deficient participants having no prior history of knee pathologies aside from the previously mentioned ligament tear will be selected for the experimentation. Participants for the study will be selected among patients who have suffered a diagnosed rupture of the ACL by an orthopaedic surgeon experienced with ACL injuries. These patients will be scheduled for ACL reconstructive surgery. A priority will be assigned to patients who have an isolated ACL rupture without any meniscal or collateral ligament damage. Only male participants will be considered for the study. Participant will be excluded of the research program if he or she has used a functional knee brace to treat the injured knee prior to the experimentation. An inquiry into their activity level after the injury will attempt to recognize possible participants that would have a "functional ACL deficient knee joint". Participants will be separated into functional ACL deficient and non-functional ACL deficient knee joint groups according to their activity level following the injury. Data from the two separate groups will then be analyzed separately and compared.

*Instrumentation*

Four high-speed digital video cameras (JVC GR-DVL9600) will be used to record all trials. The cameras will be set to record at a speed of 60 Hz and will be zoomed to include only the ACL deficient limb in the picture. The filmed volume will be approximately 1.5m x 1.5m x 1.0m. A calibration frame (1.5m x 1.5m x 1.0m) equipped with 22 reflective markers will be used and each camera will be centered on this apparatus to ensure that every marker is in the field of view before experimentation. Before every trial, the cameras will record the calibration frame to allow calibration of the area where the movement will occur. A computer equipped with the SIMI* Motion system (SIMI* Reality Motion Systems GmbH) will be used to perform the 3D motion analysis. During all trials, the 9 reflective markers on the participants injured limb will be visible at all times in all four camera views. A sound signal generated by the footswitch of the EMG system and transmitted directly into the video cameras was used at the beginning of all trials to synchronize all instruments and to provide an accurate indication of heel-strike.

In addition to kinematic data, electromyography will used to record muscle activity on the involved leg during all trials. The surface EMG electrodes will be positioned on the vastus medialis, vastus lateralis, lateral and medial hamstrings, medial and lateral gastrocnemius. Electrode placement and recording parameters will follow guidelines suggested by Lamontagne (2001). Analysis of the EMG signal will be performed using the SIMI* Motion system (SIMI* Reality Motion Systems GmbH) and custom-made software in MatLab (The Mathworks, Inc.).

The DonJoy 4titude knee brace will be fitted to each participant according to the manufacturer's guidelines and specifications. Anthropometric measurements will be taken to ensure that the right brace size is selected for the participant. The same researcher will do brace application for every participant according to the correct strap sequence suggested by the company. An extension stop of 10° at the polycentric uniaxial hinge will be set to prevent hyperextension of the injured knee.

*Protocol*

To accurately reproduce the lower limb, reflective markers will be placed on the participant to allow the use of a joint-coordinate system for calibration of the lower limb. Prior to the first trial, the participant will be required to stand still in the position previously occupied by the calibration frame to be recorded with all the joint-coordinate markers inside the field of view of all cameras. Before the experimentation, nine reflective markers will be positioned in a non-collinear pattern on the participant's segments (3 on the thigh, 3 on the leg and 3 on the foot) to allow for 3D reconstruction of the lower limb.

Following marker, EMG electrodes, and brace placement, participants will be allowed run on the treadmill to get accustomed to the equipment and to determine a comfortable running speed. The participant will then be required to perform a 6 minute running period on the treadmill under both conditions (braced and unbraced). Data will be collected for a period of ten seconds at approximately the 1 minute stage of the 6 minute running period. A total of ten gait cycles will be used for data analysis. All data acquisition instruments will be synchronized by an audio signal generated by the footswitch of the EMG system. Data acquisition will be executed for every series during ten consecutive gait cycles at approximately the halfway mark of the three-minute running period. All data acquisition instruments will be synchronized by an audio signal. After every series of running periods, a standing still trial will be performed in an attempt to monitor brace migration cause by the dynamic activity.

Each participant will be tested during a single session and will be using their running shoes. Dark clothing will be provided to facilitate marker identification during the digitizing process. Trial condition will be determined randomly before preparation begins for the participant.

*Data Analysis*

Captured video trials from all cameras, calibration frame, and joint coordinate recordings for every participant will be processed using the SIMI* Motion system. For both the kinematics and electromyography data, the total gait cycle of the involved limb will be analyzed with comparisons being made between conditions and participants. Absolute angular and linear data will be compared for the duration of the gait cycle. Statistical analysis will focus on descriptive statistics because of the small population size. Mean kinematic values and standard deviations will be used to compare the different conditions and participants. Mean EMG activity and the linear envelope of the EMG signal will be analysed for vastus lateralis, vastus medialis, lateral and medial hamstrings, medial and Lateral gastrocnemius. Muscle activation timing patterns will be analysed across participants and conditions.

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