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Abstract

The effects of training on biomechanical risk factors for anterior cruciate ligament (ACL) injuries have been investigated, but the effects of detraining have received little attention. The purpose of this study was to evaluate the effects of a one-month postseason break on knee biomechanics and lower extremity electromyography (EMG) during a stop-jump task. A postseason break is the phase between two seasons when no regular training routines are performed. Twelve NCAA female volleyball players participated in two stop-jump tests before and after the postseason break. Knee kinematics, kinetics, quadriceps EMG, and hamstring EMG were assessed. After one month of postseason break, the players demonstrated significantly decreased jump height, decreased initial knee flexion angle, decreased knee flexion angle at peak anterior tibial resultant force, decreased prelanding vastus lateralis EMG, and decreased prelanding biceps femoris EMG as compared with prebreak. No significant differences were observed for frontal plane biomechanics and quadriceps and hamstring landing EMG between prebreak and postbreak. Although it is still unknown whether internal ACL loading changes after a postseason break, the more extended knee movement pattern may present an increased risk factor for ACL injuries.

Disciplines

Biomechanics | Exercise Science | Expeditionary Education | Kinesiology | Motor Control

Comments

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Original Research

The effects of postseason break on knee biomechanics and lower extremity EMG in a stop-jump task: implications for ACL injury

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Abstract

The effects of training on biomechanical risk factors for anterior cruciate ligament (ACL) injuries have been investigated, but the effects of detraining have received little attention. The purpose of this study was to evaluate the effects of a one-month postseason break on knee biomechanics and lower extremity electromyography (EMG) during a stop-jump task. A postseason break is the phase between two seasons when no regular training routines are performed. Twelve NCAA female volleyball players participated in two stop-jump tests before and after the postseason break. Knee kinematics, kinetics, quadriceps EMG, and hamstring EMG were assessed. After one month of postseason break, the players demonstrated significantly decreased jump height, decreased initial knee flexion angle, decreased knee flexion angle at peak anterior tibial resultant force, decreased pre-landing vastus lateralis EMG, and decreased pre-landing biceps femoris EMG as compared to pre-break. No significant differences were observed for frontal plane biomechanics and quadriceps and hamstring landing EMG between post-break and pre-break. Although it is still unknown whether internal ACL loading changes after a postseason break, the more extended knee movement pattern may present an increased risk factor for ACL injuries.

Introduction

Anterior cruciate ligament (ACL) injuries commonly occur during sports-related activities, with an annual incidence rate of 1 per 3000 people (Miyasaka et al., 1991). Seventy to eighty percent of ACL injuries occur in noncontact maneuvers that involve sudden deceleration (Boden et al., 2000). The likelihood of sustaining an ACL injury is greater for females than for males (de Loes et al., 2000; Prodromos et al., 2007).

Previous studies have shown that females tend to have higher anterior tibial resultant forces (Chappell et al., 2002; Yu et al., 2006), lower knee flexion angles (Chappell et al., 2007; Malinzak et al., 2001; McLean et al., 2004; McLean et al., 2005; Salci et al., 2004; Yu et al., 2006), higher knee abduction angles (Ford et al., 2003; Malinzak et al., 2001; McLean et al., 2004; McLean et al., 2005), lower hip flexion angles (Chappell et al., 2007; Landry et al., 2007; McLean et al., 2004; McLean et al., 2005; Pollard et al., 2007; Salci et al., 2004; Yu et al., 2006), higher knee extension moments (Chappell et al., 2002; Yu et al., 2006), higher quadriceps electromyography (EMG), and lower hamstring EMG (Chappell et al., 2007; Landry et al., 2007; Malinzak et al., 2001; Rozzi et al., 1999; Sigward & Powers, 2006b) than males during athletic tasks. Hewett et al. (2005) found knee abduction angles, knee abduction moments, and side-to-side differences prospectively predicted ACL injuries in young female athletes. Cadaver simulation studies have demonstrated that anterior shear forces applied at the proximal end of the tibia were the primary contributor to ACL loading, and knee valgus, varus, and internal rotation moments contributed to ACL loading when the anterior shear forces were applied (Berns et al., 1992; Markolf et al., 1995). In addition, ACL loading resulting from a constant anterior shear force increased as the knee flexion angles decreased (Berns et al., 1992; Jordan et al., 2007;

Markolf et al., 1995). These in vitro and in vivo studies suggest that differences in biomechanical traits induce higher ACL loading in females as compared to males.

Training interventions have been effective in reducing biomechanical risk factors for ACL injuries (Lim et al., 2009; Myer et al., 2005; Myer et al., 2006) as well as reducing the ACL injury rate (Hewett et al., 1999; Mandelbaum et al., 2005). However, detraining effects on biomechanical risk factors for ACL injury have received little attention. Detraining has been defined as the partial or complete loss of training-induced adaptations in response to an insufficient training stimulus (Mujika & Padilla, 2000). Detraining could be induced by onset of illness and injury, postseason break, or retirement.

Although it is unknown whether ACL injuries are more likely to happen after a postseason break, higher overall injury rates during the preseason as compared to the regular season have been documented in 15 NCAA sports (Hootman et al., 2007). The overall injury rate during preseason practice was twice as high as during regular season practice in volleyball (Agel et al., 2007). Athletes beginning preseason training with poor conditioning could be one of the causes of the higher injury rates (Hootman et al., 2007). Postseason breaks are commonly four weeks for highly trained athletes. Four weeks of detraining may result in decreased muscle physiological functions (Mujika & Padilla, 2000) and decreased strength (Mujika & Padilla, 2001) . In addition, decreases in quadriceps isometric EMG were observed in athletes after four weeks of detraining (Häkkinen & Komi, 1983). Few studies have documented the effects of detraining on knee biomechanics. However, recent studies investigating the relationships between strength, kinematics, and kinetics suggested that athletes might change their movement patterns due to detraining effects. Lawrence et al. (2008) found that females with lower hip and quadriceps/hamstring strength demonstrated higher external knee adduction and flexor moments during a single leg dropping task. Claiborne et al. (2006) showed that adults with lower hip abductor, knee flexor, and knee extensor strength demonstrated higher knee valgus motion during a single leg squat. Lower relative strength has been associated with lower knee flexion angles (Chappell et al., 2005). Furthermore, lower knee flexion angles have been associated with higher frontal plane motion and moments (Pollard et al., 2010). Because of the relationships between strength and lower extremity biomechanics, highly trained athletes might change their movement patterns due to decreases in strength after postseason break.

The purpose of this study was to evaluate the effects of a one-month postseason break on the knee biomechanics and lower extremity EMG for a stop-jump task in collegiate female volleyball players. It was hypothesized that after a one-month postseason break, players would demonstrate decreased knee flexion angles, increased knee abduction angles, increased knee extension moments, increased knee adduction moments, and decreased quadriceps and hamstring EMG.

Methods

Subjects

Knee flexion angle at peak anterior tibial resultant force (PATRF) was used for the power analysis since changes in knee flexion angles have been associated with changes in ACL loading (Berns et al., 1992; Jordan et al., 2007; Markolf et al., 1995). Based on a previous study (Herman et al., 2009), the expected change in knee flexion angles between pre-break and post-break was 5º and the standard deviation was 5.5º. To achieve a power of 0.8 at an alpha level of 0.05, twelve subjects were needed. Therefore, twelve NCAA Division I female volleyball players (Table 1) were recruited as subjects on a volunteer basis. All subjects were right leg dominant, based on their preferred kicking leg. Subjects were excluded from this study if they had suffered an ACL injury, meniscus damage, or substantial ligament damage to the knee or ankle; had a lower extremity injury that prevented participation in physical activity for >2 weeks over the previous 6 months; or possessed any condition that prevented them from participating at maximal effort in sporting activities. Informed consent was obtained in accordance with the Iowa State University Institutional Review Board.

Instrumentation

Three-dimensional marker coordinates were recorded using eight infrared video cameras (Oxford Metrics Ltd, Oxford, UK) at a sampling rate of 160 Hz. Ground reaction forces were collected by a force platform (AMTI, MA, USA) at a sampling rate of 1600 Hz. Muscle electrical activities were collected by a surface EMG capture system (Delsys Inc., MA, USA) with a bandwidth from 20-450 Hz at a sampling rate of 1600 Hz. Marker coordinates, ground reaction forces, and EMG signals were synchronized using a Vicon Nexus data acquisition system (Oxford Metrics Ltd, Oxford, UK).

Experimental Procedure

After informed consent was obtained, the subject’s injury history was recorded. Once the subject met the inclusion criteria, anthropometric parameters were measured (Vaughan et al., 1992). During the test, all subjects wore spandex shorts, t-shirts, and personal shoes and socks. Subjects conducted stretching exercises and ran on a treadmill at a self-selected speed for warm-up.

A total of 21 retroreflective markers were used (Figure 1a, b). Markers were attached on the spinous process of the right and left acromioclavicular joints, fifth cervical vertebra, upper edge of sternum, right and left anterior superior iliac spines, right and left posterior superior iliac spines, and right and left greater trochanters. On the right side of the body, markers were placed on the anterior and lateral mid-thigh, medial and lateral femoral condyle, anterior shank below tibial tuberosity, posterior mid-shank, medial and lateral malleolus, lateral foot, dorsal foot and heel. Surface electrodes were aligned with the muscle belly of the vastus medialis oblique (VM), vastus lateralis (VL), biceps femoris (BF), and semitendinosus (ST) of the right leg after the subject's skin was cleaned (Cram et al., 1998). A common ground electrode was placed on the right tibial tuberosity.

Isometric maximum voluntary contractions (MVC) were conducted for the quadriceps and hamstring muscle groups. MVC tests for the quadriceps were performed while the subject was in a sitting position on a secured table with the hip and knee flexed at 90°. An investigator held the participant's lower anterior tibia proximal to the ankle joint. The subject was instructed to extend her knee as hard as she could for five seconds. MVC tests for the hamstring muscles were performed while the subject was in a prone position on a mat table with the knee flexed at 90°. An investigator held the participant's lower posterior shank proximal to the ankle joint. The subject was instructed to flex her knee as hard as she could for five seconds.

Prior to performing the vertical stop-jump tasks, the subjects were asked to stand upright with their feet placed shoulder width apart for a static video capture. The vertical stop-jump task consisted of an approach run followed by a one-footed takeoff, a two-footed landing, and a two-footed takeoff while raising her arms (Chappell et al., 2002). The approach was set at two steps, but the distance of approach was not restricted due to variations in step length. Subjects were instructed to jump vertically as high as possible, but no other technique instructions were given to avoid changing natural jump preferences. Subjects were allowed to practice until they were comfortable with the task, then each subject performed five successful trials of the vertical stop-jump task. A successful trial meant that the subject performed a vertical stop-jump with her right foot landing on the force plate. A trial was excluded when the subject did not meet the requirements of a vertical stop jump, her right foot did not land on the force plate, or markers were not properly tracked during data collection.

After the pre-break tests, subjects were asked to record any injury that occurred and self-selected exercise they performed during the postseason break. After the one-month postseason break, participants returned to the lab. Injuries and self-selected exercise during the postseason break were recorded and the same data collection procedure was repeated.

Data reduction

The video coordinates and force plate data collection were time-synchronized to 1600 Hz using linear interpolation. The coordinate data were filtered using a fourth–order, zero-phase-shift Butterworth filter at a low-pass cutoff frequency of 12 Hz. The hip joint center was determined using the method of Bell et al. (1990). The knee joint center was defined as the midpoint between the medial and lateral femoral condyles, and the ankle joint center was defined as the midpoint between the medial and lateral malleoli. Joint centers were estimated during the static standing trial and recreated during the stop-jump tests using singular value decomposition (Soderkvist & Wedin, 1993). Tibial reference frames were determined from the coordinates of ankle joint center, knee joint center, and anterior shank markers. Thigh reference frames were determined using knee joint center, hip joint center, and lateral femoral condyle markers. For both the tibia and femur, the vertical axis was defined by using proximal and distal joint centers. The second axis was then defined by the cross product of the vertical axis and an intermediate axis defined by one joint center and the additional marker. The third axis was defined by the cross product of the vertical axis and the second axis (Grood & Suntay, 1983). The Cardan joint angles were calculated in a flexion–extension, abduction–adduction, and internal–external rotation order. Knee flexion, adduction, and internal rotation were denoted as positive joint angles.

An inverse dynamics approach was used to calculate the three-dimensional knee joint resultant moments and resultant forces. Segment masses, center of mass locations, and segment moments of inertia were based on Vaughan et al. (1992). Segment angular velocities and accelerations were determined using the methods described by Amirouche (1992). Knee joint resultant forces and moments were transferred to the tibial reference frame and expressed as internal loading. Joint resultant forces were normalized to body weight, and joint resultant moments were normalized to the product of body weight and height. By using similar data collection and reduction methods, a previous study found kinematic and kinetic variables had excellent reliability in the sagittal plane and moderate to excellent reliability in the frontal and transverse planes during landing tasks (Ford et al., 2007).

The initiation of landing was identified by the time when the vertical ground reaction force first increased above 20N. The toe off event was identified by the time when the vertical ground reaction force decreased below 20N after landing. The landing phase was defined as the first 20% of the entire stance phase (Chappell et al., 2005). Jump height was calculated by using the vertical coordinate data of the right anterior superior iliac spine marker. Subject approach speeds were calculated by determining the difference in position for the midpoint of the right and left posterior superior iliac spine markers in the time period from one frame before toe contact to one frame after toe contact. PATRF was calculated and further used as a critical time point for knee loading. Knee flexion and abduction angles at the initiation of landing and at PATRF, maximum knee flexion and knee abduction angles during the stance phase, knee extension and adduction moments at PATRF, and maximum knee extension and adduction moments during the landing phase were determined for each trial.

EMG data for each muscle were filtered at a low-pass cutoff frequency of 450 Hz and a high-pass cutoff frequency of 20 Hz. Filtered EMG data were then rectified and filtered at a cutoff frequency of 10 Hz to calculate the EMG linear envelope. MVC amplitudes were calculated for each muscle by determining the maximum one second averages from the EMG linear envelopes. For the stop-jump tasks, EMG linear envelopes of each muscle group were normalized as a percentage of the MVC. EMG data for the 50 ms before landing were averaged for each muscle to represent pre-landing muscle activities (Nagano et al., 2007). EMG data for the first 20% of the stance phase were averaged to represent the muscle activities during landing (Sigward & Powers, 2006a). All of the kinematic, kinetic, and EMG data calculations were performed in MATLAB 7.4.0 (MathWorks Inc., PA, USA).

Data Analysis

Data were averaged across five trials for each subject during each test. Because of a relatively small sample size, 2-tailed Wilcoxon signed-rank tests were used to compare stop-jump variables between pre-break and post-break. A Type I error rate of 0.05 was selected as an indication of statistical significance. Statistical analyses were conducted in SPSS 16.0 (SPSS, IL, USA).

Results

During the competition season, training time was 20 hours/week (12-13 hours practice, 1-2 hours strength training, and 6 hours game competition). During the one-month postseason break, athletes performed self-selected training with a mean training duration of 2.62 ± 1.53 hours/week (1.33 hours cardiovascular, 0.92 hours strength training, and 0.36 hours volleyball playing). No significant differences were observed between pre-break and post-break for height (p=0.91) or mass (p=0.33).

Post-break jump heights significantly decreased as compared to pre-break (p<0.01, Table 2). In addition, post-break knee flexion angles at initial foot contact with the ground (p=0.05, Figure 2) and knee flexion angles at PATRF (p=0.02) were significantly reduced as compared to pre-break. No significant differences were observed between pre-break and post-break for approach speeds (p=1.00), PATRF during landing (p=0.56), maximum knee flexion angles during stance phase (p=0.06), knee extension moments at PATRF (p=0.32), maximum knee extension moments during landing (p=0.78), knee abduction angles at initial foot contact with ground (p=0.85), knee abduction angles at PATRF (p=0.73), maximum knee abduction angles during stance phase (p=0.44), knee adduction moments at PATRF (p=0.66), and maximum knee adduction moments during landing (p=0.80).

Post-break values for pre-landing VL EMG (p=0.05, Figure 3) and pre-landing BF EMG (p<0.01, Figure 4) were significantly reduced as compared to pre-break. No significant differences were observed between pre-break and post-break for pre-landing VM EMG (p=0.24), pre-landing ST EMG (p=0.90), landing VM EMG (p=0.52), landing VL EMG (p=0.24), landing ST EMG (p=0.41), and landing BF EMG (p=0.21).

Discussion

The purpose of this study was to investigate the effects of a one-month postseason break on the knee biomechanics and lower extremity EMG in a stop-jump task in female volleyball players. During the postseason break, subjects conducted self-selected exercise and the training duration was reduced 87% as compared to the competition season.

In the current study, jump height decreased after the postseason break. Marques and González-Badillo (2006) observed significantly decreased ball throwing velocity, but unchanged countermovement jump heights after seven weeks of detraining in male professional handball players. In addition, no significant changes were found in jump heights after six weeks of detraining in recreational strength-trained men (Kraemer et al., 2002). The discrepancy between the current study and previous studies could be caused by the differences between sports and athletic populations. For highly trained athletes, eccentric force and sport-specific power may suffer significant declines after four weeks of training cessation (Mujika & Padilla, 2000; Mujika & Padilla, 2001). Häkkinen and Komi (1983) found that quadriceps strength and isometric EMG decreased at 4-week and 8-week time points of detraining, with the changes in strength significantly correlated with isometric EMG. The authors suggested that a reduction in neural activity mainly contributed to decreased performance during the early detraining period. Although strength was not tested before and after the detraining period in the current study, reductions in pre-landing muscle EMG suggested that athletes might experience changes in neural control which could cause changes in jump-task performance.

Post-break knee flexion angles were reduced at initial foot contact and at PATRF as compared to pre-break. Based on the potential effects of strength on lower extremity biomechanics (Claiborne et al., 2006; Herman et al., 2009), it is speculated that decreased muscle strength after detraining precipitated a more extended posture to prevent lower extremity collapse during the landing. Given that muscle strength and explicit motor control were not measured in the current study, we could not confirm these plausible explanations. However, these decreases in knee flexion angles could provide some insight into ACL injury risks posed by postseason breaks. Markolf et al. (1995) reported anterior shear force was the most direct ACL loading of cadaver knees, and the loading increased as knee flexion angle decreased. With a given quadriceps contraction force, decreasing knee flexion angles increased ACL loading because of the increased patella tendon-tibia shaft angle and ACL elevation angle (Li et al., 2005; Nunley et al., 2003). Less knee flexion during athletic tasks has been repeatedly found in females compared to males. Salci et al. (2004) found female volleyball players demonstrated lower knee flexion angles during spike and block landings. Female recreational and collegiate athletes also demonstrated lower knee flexion angles in comparison to men during side-cutting, cross-cutting, and side-jump movements (Malinzak et al., 2001; McLean et al., 2004; McLean et al., 2005). Given that anterior shear force at the tibia was not directly measured in the current study, it is unknown whether ACL loading changed or not due to sagittal plan mechanisms. Therefore, our results can only suggest that the more extended knee movement pattern demonstrated by athletes after the postseason break may present an increased risk factor for ACL injuries. However, the relationship between knee flexion angle and ACL strain is a theoretical tenet without considering other loading mechanisms of the ACL.

Pre-landing VL EMG and BF EMG were reduced at post-break as compared to pre-break. Nagano et al. (2007) found a higher hamstrings/quadriceps EMG ratio for the 50 ms pre-landing phase in males than in females during single limb drop landings. Chappell et al. (2007) observed that quadriceps EMG increased about 50 ms before landing for both males and females. They also found increased quadriceps and hamstring EMG during the pre-landing phase in females as compared to males. Post-break reductions in pre-landing BF EMG could be associated with reduced knee flexion angles at initial foot contact with the ground. Although decreased pre-landing VL EMG was not consistent with a decreased knee flexion angle, a decrease in muscle coactivation on the lateral side of the knee may cause a decrease in joint stiffness (Louie & Mote, 1987; Olmstead et al., 1986).

Besides sagittal plane mechanisms, abnormal frontal plane movements have also been proposed as risk factors for non-contact ACL injuries. Markolf et al.(1995) found that the addition of valgus and varus moments to anterior tibial shear force significantly increased ACL strain as compared to only anterior tibial shear force loading. Hewett et al. (2005) found larger knee abduction angles and moments during jump landing tasks in ACL injured athletes than in uninjured athletes. In contrast to sagittal plane movements, which are mainly determined by quadriceps and hamstring muscles with small constraints from passive tissues, the control of frontal plane motion is more complex. Knee adduction/abduction angles are associated with knee flexion angle (Pollard et al., 2010), while coactivation of quadriceps and hamstring muscles support knee adduction-abduction moments by increasing joint stiffness (Louie & Mote, 1987; Olmstead et al., 1986). Although not assessed in this study, hip abduction and external rotation strength also play an important role in control of frontal plane knee motion (Jacobs et al., 2007; Lawrence et al., 2008). In addition, knee joint anatomy including the ratio of tibial plateau width to intercondylar distance and the ratio of medial tibial slope to lateral tibial slope correlate with knee abduction angle (McLean et al., 2010). A larger frontal plane motion was expected when knee flexion angles, quadriceps EMG, and hamstring EMG decreased after the postseason break. However, it is unlikely that significant changes occurred in the knee joint anatomy and passive tissues during the break, which may explain the unchanged frontal plane biomechanics in the current study.

Changes in jump height and movement patterns could provide information in developing training programs after a postseason break. Six weeks of neuromuscular training have resulted in significant increases in jump height as well as knee flexion-extension range of motion (Myer et al., 2005). On the other hand, poor conditioning was suggested to be associated with increased knee injuries (Hutchinson & Ireland, 1995). Implementing training programs after a postseason break might help athletes recover their jumping performance and movement patterns to their pre-break level.

The current study served as a preliminary work in hope of providing information for future research. The use of collegiate volleyball players resulted in a small sample size, which decreased the statistical power of the results and is a limitation for generalizing the findings to other sports. The type I error rate was not adjusted when multiple statistical tests were conducted, and instead an “exploratory” alpha rate was utilized to test for significance. Only 5 out of 21 variables were statistically significant, with relatively small magnitudes of change that could have approached the range of measurement error. Future studies with a larger sample size, a stronger statistical power, and a diversity of athletes are needed. It should also be noted that the interpretations of our results were limited by the fact that strength data were not collected. Because we did not examine injury causing events, the links between the findings of the current study and ACL injury risks remain speculative. Kinematic and kinetic variables are associated with ACL loading, but a complete description of the three-dimensional loading mechanism involves structural geometry, joint positions, passive tissue deformation, and muscle forces. Future studies with musculoskeletal models are needed to estimate the effect of a postseason break on ACL loading.

In conclusion, collegiate female volleyball players demonstrated significantly reduced post-break jump heights, reduced initial knee flexion angles, reduced knee flexion angles at PATRF, reduced pre-landing VL EMG, and reduced pre-landing BF EMG as compared to pre-break. No differences were observed for frontal plane biomechanics, quadriceps landing EMG, and hamstring landing EMG between pre-break and post-break. The more extended knee movement pattern demonstrated by athletes after a one-month postseason break may present a risk factor for ACL injuries. However, this link between knee flexion angle and ACL injury risk remains speculative due to the three dimensional loading mechanism of ACL. The results may provide some insight into developing ACL prevention programs at different phases of a competitive season.

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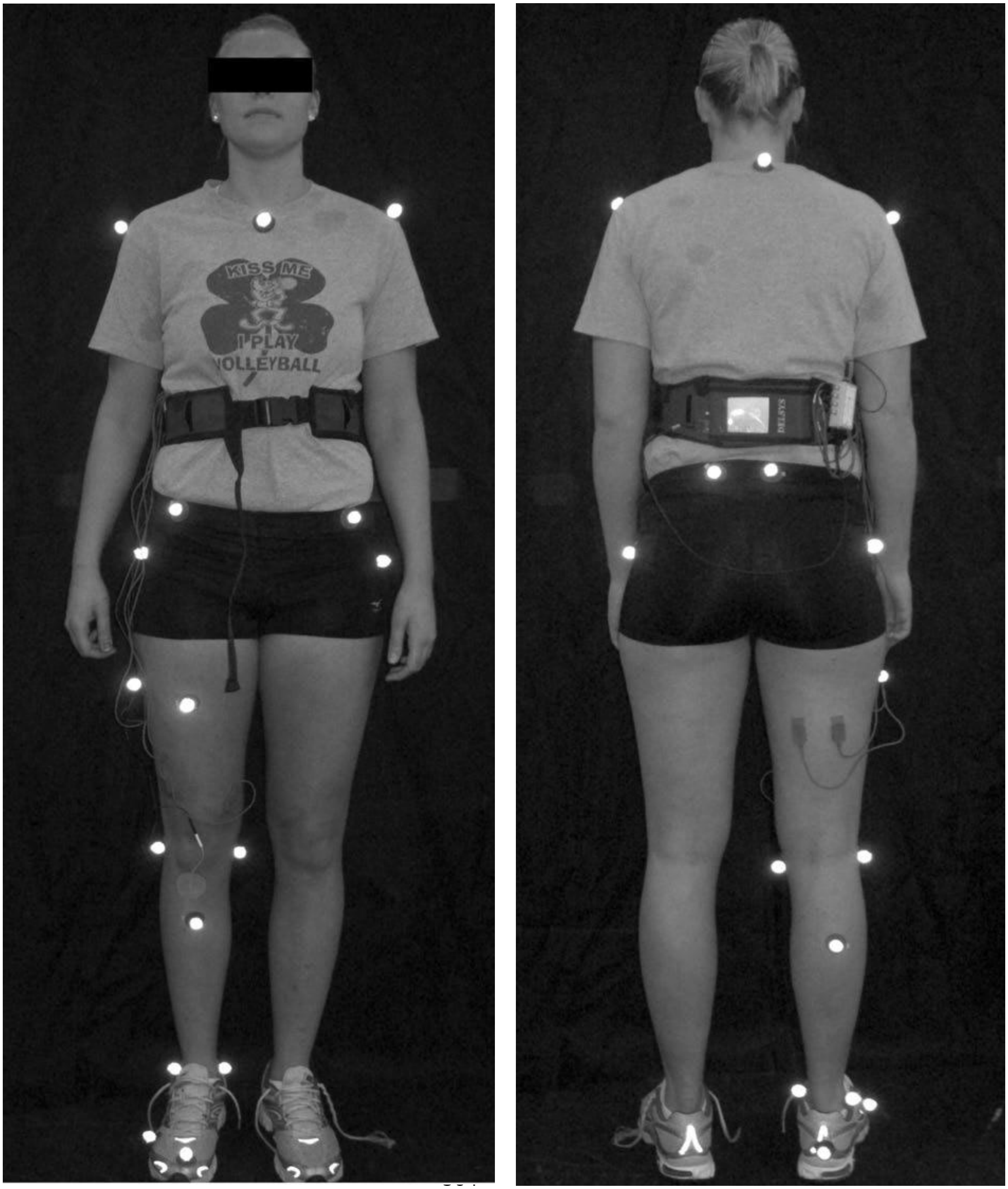

Figure 1a, b: Retroreflective marker placement

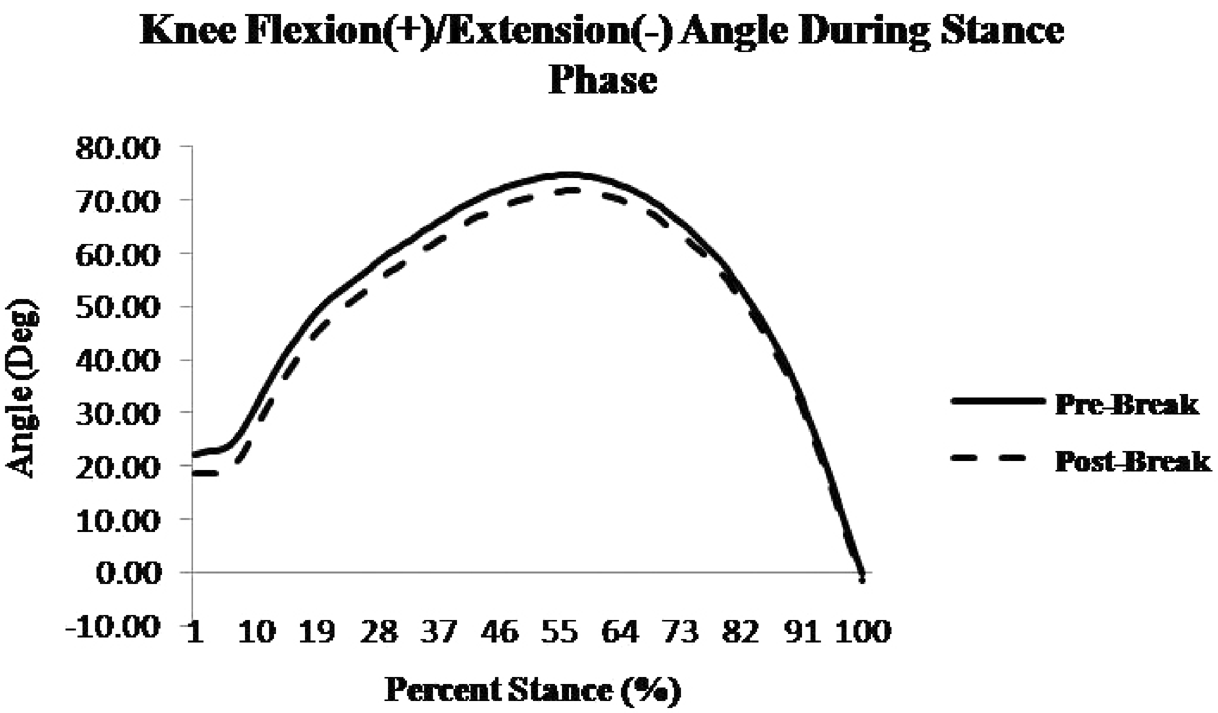

Figure 2: Knee flexion/extension angle during stance phase

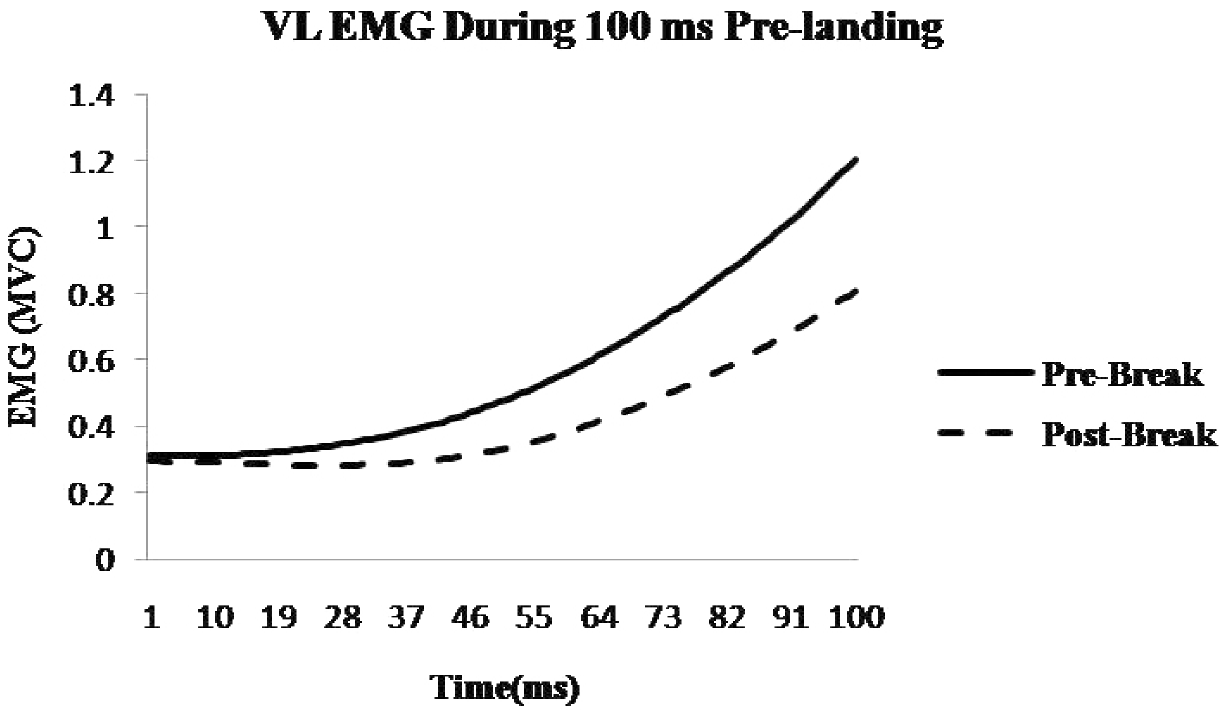

Figure 3: VL EMG during 100 ms pre-landing

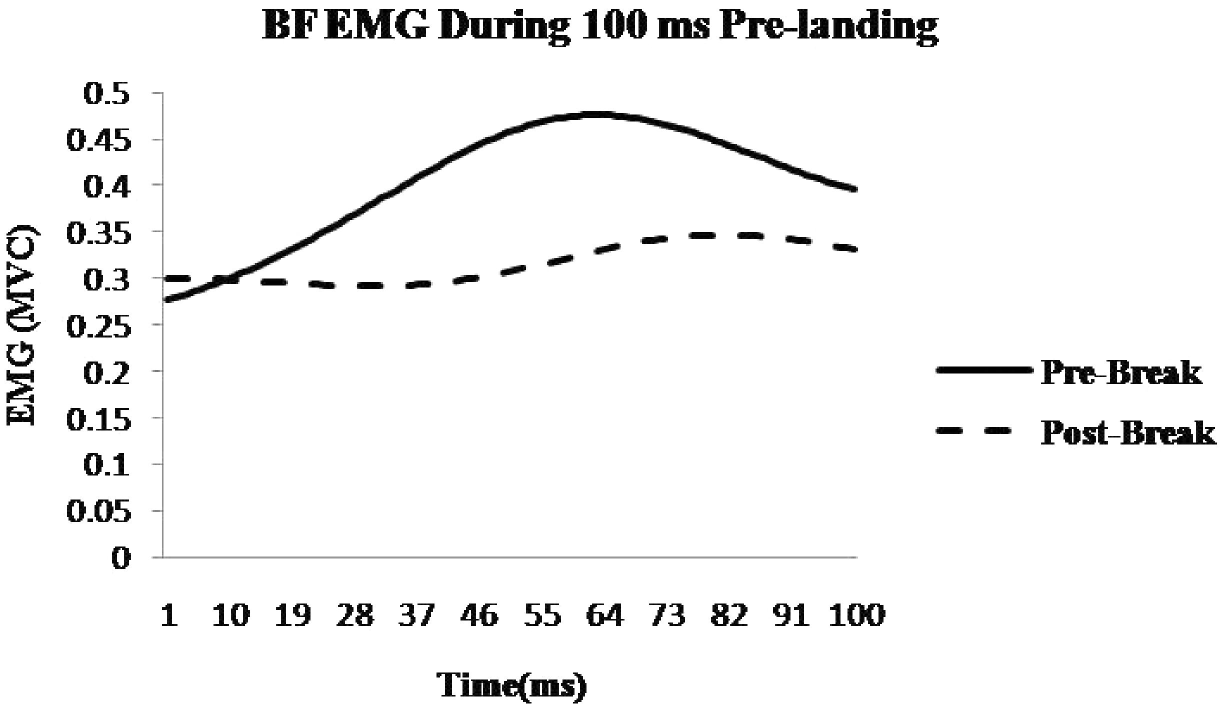

Figure 4: BF EMG during 100 ms pre-landing

Table 1 Subject characteristics

	Mean ± *SD*
Age (years)	19.25 ± 1.22
Experience (years)	8.75 ± 2.22
Prebreak height (cm)	178.49 ± 6.96
Postbreak height (cm)	178.37 ± 6.40
Prebreak mass (kg)	71.17 ± 5.65
Postbreak mass (kg)	70.88 ± 5.86

Table 2 Kinematics/kinetics and EMG comparisons between prebreak and postbreak (mean ± *SD*, $*p \leq 0.05$, $**p \leq 0.01$)

Dependent Variables	Prebreak	Postbreak
Approach speed (m/s)	2.17 ± 0 .45	2.17 ± 0.39
Jump height (m)	0.48 ± 0.05	0.44 ± 0.05**
PATSF during landing (BW)	0.92 ± 0.49	0.87 ± 0.31
Initial knee flexion angle (deg)	21.18 ± 5.76	18.55 ± 5.55*
Knee flexion angle at PATSF (deg)	30.51 ± 9.53	24.96 ± 8.68*
Maximum knee flexion angle (deg)	75.33 ± 5.07	72.19 ± 6.70
Knee extension moment at PATSF (BW × BH)	0.02 ± 0.04	0.03 ± 0.03
Maximum knee extension moment during landing (BW × BH)	0.07 ± 0.04	0.07± 0.04
Initial knee abduction angle (deg)	2.60 ± 2.35	2.53 ± 2.01
Knee abduction angle at PATSF (deg)	4.91 ± 3.26	4.83 ± 3.75
Maximum knee abduction angle (deg)	11.74 ± 6.74	12.71 ± 5.10
Knee adduction moment at PATSF (BW × BH)	0.02 ± 0.05	0.01 ± 0.04
Maximum knee adduction moment during landing (BW × BH)	0.07 ± 0.04	0.07 ± 0.04
Prelanding VM EMG (MVC)	0.73 ± 0.33	0.59 ± 0.27
Prelanding VL EMG (MVC)	0.79 ± 0.39	0.54 ± 0.37*
Prelanding ST EMG (MVC)	0.36 ± 0.19	0.36 ± 0.14
Prelanding BF EMG (MVC)	0.45 ± 0.17	0.33 ± 0.09**
Landing VM EMG (MVC)	1.74 ± 0.52	1.51 ± 0.61
Landing VL EMG (MVC)	2.22 ± 1.10	1.71 ± 0.75
Landing ST EMG (MVC)	0.32 ± 0.21	0.32 ± 0.15
Landing BF EMG (MVC)	0.44 ± 0.22	0.35 ± 0.13