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Combined effects of fatigue and decision making on female lower limb landing postures: Central and peripheral contributions to ACL injury risk

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Abstract

Background. In spite of ongoing prevention developments, anterior cruciate ligament injury rates and the associated sex-disparity have remained, suggesting an incomplete understanding of the injury mechanism. While both fatigue and decision making are known in isolation to directly impact anterior cruciate ligament injury risk, their combined manifestations remain unknown. We thus examined the combined effects of fatigue and decision making on lower limb kinematics during sports relevant landings.

Methods. Twenty five female National College Athletic Association athletes had initial contact and peak stance phase 3D lower limb joint kinematics quantified during anticipated and unanticipated single (left and right) leg landings, both before and during the accumulation of fatigue. Jump direction was governed by light stimuli activated prior to and during the pre-land phase of respective anticipated and unanticipated trials. To induce fatigue, subjects performed repetitive squat ($n = 5$) and randomly ordered jump sequences, until squats were no longer possible. Subject-based measures of each dependent factor were then calculated across pre-fatigue trials, and for those denoting 100% and 50% fatigue, and submitted to a 3-way mixed design analysis of covariance to test for the main effects of fatigue time, decision and leg.

Findings. Fatigue caused significant increases in initial contact hip extension and internal rotation, and in peak stance knee abduction and internal rotation and ankle supination angles. Fatigue-induced increases in initial contact hip rotations and in peak knee abduction angle were also significantly more pronounced during unanticipated compared to anticipated landings.

Interpretation. The integrative effects of fatigue and decision making may represent a worst case scenario in terms of anterior cruciate ligament injury risk during dynamic single leg landings, by perpetuating substantial degradation and overload of central control mechanisms.

Keywords: Anterior cruciate ligament; Neuromuscular fatigue; Decision making; Kinematics

1. Introduction

Non-contact anterior cruciate ligament (ACL) injuries continue to present in epidemic proportions, exposing individuals, particularly females, to significant and well documented short- and long-term debilitations (Griffin et al., 2006; Lohmander et al., 2004). A vast amount of research continues to focus on the likely role of altered or abnormal neuromuscular control within the underlying injury mechanism (Kernozek et al., 2005; McLean et al., 2005; Sigward and Powers, 2007), since these factors are in essence modifiable, and hence amenable to targeted intervention (Bahr and Krosshaug, 2005). Neuromuscular training programs have continued to evolve out of this research, with reported early successes in clinical trials (Mandelbaum et al., 2005; Myklebust et al., 2003). Despite the ever-increasing number and complexity of these programs however, injury rates

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and the sex-based disparity have persisted outside of such studies (Agel et al., 2005). It appears therefore, that the success of current ACL injury prevention strategies may be limited by an incomplete understanding of the true mechanism of injury.

Excluding more recent and progressive prevention modalities, which indeed integrate random perturbations within the training protocols (Myer et al., 2004; Myklebust et al., 2003), current prevention methods typically base their success on the promotion of safe neuromuscular control strategies via controlled and largely systematic lab-based interventions. Sports in which ACL injuries are common however, typically present as a random and often complex series of dynamic events, requiring an equally complex coordination of central and peripheral responses (Bonato et al., 2003; Ericsson and Lehmann, 1996). Thus, while there is no question current prevention methods afford significant domain-specific practice, they may not promote the level of perception-action coupling necessary to successfully and safely adapt to the inherently random and demanding sports environment. The integration of more sports-relevant factors within the *in vivo* experimental testing environment may provide further crucial insights into the causal factors of non-contact ACL injury, and hence, facilitate the formulation of more effective and adaptable prevention methods.

Recent studies have shown that fatigue (Chappell et al., 2005; Madigan and Pidcoe, 2003; McLean et al., 2007) and decision making (Besier et al., 2001; Houck et al., 2006; Pollard et al., 2004), both synonymous with sports participation, contribute directly in isolation to ACL injury risk via the promotion of high-risk joint neuromechanical strategies. Altered knee joint biomechanics, and in particular increased out of plane hip and knee motions and loads and sagittal plane ankle motions, are common postural outcomes when individuals are exposed to either factor during dynamic sports landings (Besier et al., 2001; Chappell et al., 2005; McLean et al., 2007). Within the true game environment however, fatigue and decision making effects rarely exist independent of one another, and it may well be that their combined manifestation presents as a worst case scenario in terms of ACL injury risk. Considering both central and peripheral processing mechanisms are compromised in the presence of fatigue (Gandevia, 2001; Lorist et al., 2005; Miura et al., 2004), poor perceptions, decisions, reactions and resultant movement strategies may be more likely when in a fatigued state.

To date, research evaluating potential interactions between neuromuscular fatigue and decision making within the sporting context has focused primarily on tasks inducing extreme exhaustive states, such as long distance cycling and running (Collardeau et al., 2001; Grego et al., 2005). Further, the few studies examining these combined effects in sports where ACL injuries are prevalent have centered on performance outcomes rather than injury risk (McMorris et al., 1999). Elucidating the integrative impact of fatigue and decision making on resultant lower limb joint postures during high risk dynamic sports maneuvers would provide an important step forward in understanding and hence combating the causality of sports related non-contact ACL injuries. With these facts in mind therefore, the current study examined the combined effects of neuromuscular fatigue and decision making on three-dimensional lower limb (hip, knee and ankle) kinematics during the stance phase of single leg landing tasks. To achieve these aims, we tested the following tenets, being based primarily on previous observations:

1. Neuromuscular fatigue will induce significant increases in initial contact (at landing) hip and knee extension, hip internal rotation and ankle plantar flexion, and peak stance (0–50% stance) phase knee abduction and internal rotation positions during single leg landings.
2. Fatigue induced changes in the above kinematic parameters will be more pronounced during unanticipated compared to anticipated landings.

2. Methods

2.1. Subjects

Submitting data from previous studies investigating the isolated effects of neuromusclar fatigue (Chappell et al., 2005; McLean et al., 2007) or decision making (Besier et al., 2001; Pollard et al., 2004) on lower limb neuromechanics revealed that to currently achieve 90% statistical power with an exploratory alpha level of 0.05, a minimum of 19 (same-sex) subjects would be required. We subsequently recruited 24 female (age 21.2 (2.5) years) NCAA division 1 (basketball, soccer and volleyball) athletes (height = 176.3 (5.1) cm; mass = 67.5 (7.4) kg) to participate in the study. Prior to testing, research approval through the Cleveland Clinic Foundation Institutional Review Board and written informed consent was obtained for all subjects. Subject exclusion criteria was based on (1) a history of previous knee injury and/or surgery, (2) pain in lower extremity immediately prior to testing, (3) any injury to the lower extremity in last 6 months, (4) undertaking any exercise within 24 h of the test session, and, (5) a current pregnancy. All subjects were required to wear spandex bike shorts, sports shoes and sports brassier during testing. Subjects also had limb dominance ascertained prior to data collection, with the dominant leg being that which could kick a ball the farthest (McLean et al., 2007).

2.2. Experimental design

Subjects had bilateral three-dimensional (3D) lower limb (hip, knee and ankle) joint kinematic and 3D ground reaction force data recorded during execution of a series of jump landing tasks, both before and during exposure to a generalized neuromuscular fatigue protocol. A successful jump trial was based on the subject making a complete foot contact with one (or both) of two force plates (AMTI OR6-5 #4048, Advanced Mechanical Technology Inc., Watertown, USA) located 10 cm apart, within the field of view of an eight camera high-speed (240 fps) motion analysis system (Motion Analysis Corporation, Santa Rosa, USA). Prior to testing, subjects were provided with adequate time to warm up and to familiarize them with the required landing tasks.

Subjects performed an initial sequence of landing tasks from which baseline (pre-fatigue) data were obtained. Specifically, they were required to execute one of three randomly ordered landings, governed by activation of an explicit light stimuli (L1, L2 or L3) prior to the landing phase (Fig. 1). Light positions and the required movement response were chosen to loosely reflect different anticipatory demands typically evident during game play (McLean et al., 2007). Activation of L1 required subjects to land on their left foot only and immediately and aggressively jump laterally to the right. Conversely, activation of L2 necessitated a rapid jump off the right foot, laterally to the left. If L3 was activated, subjects were required to land on both feet, one foot on each force plate, and jump vertically as high as possible.

Each single leg landing trial was further discretized to incorporate either an anticipated or unanticipated movement response, with this order again being randomized. For anticipated trials, the specific light stimuli was activated prior to (approx. 5 s) the subject initiating the jump-phase of the movement. For unanticipated trials however, the stimuli was automatically triggered via a light beam switch (42RLU–4000B, Allen Bradley, Milwaukie, USA), which the subjects broke following take-off, such that the light did not come on until approximately 350 ms prior to ground contact. Vertical (two-legged) jump trials were utilized to compare relative subject fatigue levels only (see Statistical Treatment), and were hence all anticipated. Each subject was thus required to perform approximately 30 pre-fatigue landing trials in order to ensure adequate data were available for each (leg (2) × decision (2), and anticipated vertical) pre-fatigue jump condition. A single experimenter (Dr McLean) with experience in these methods delineated between successful and unsuccessful trials (e.g., landing on wrong foot), with the latter being repeated elsewhere along the randomly ordered trial sequence.

Following the pre-fatigue trials, subjects again performed a random series of anticipated and unanticipated landings while being simultaneously exposed to a general fatigue protocol. Specifically, subjects performed continuous sets of five double leg squats between jump trials, with the jump sequence and decision type again being randomized. Intra and inter subject variations in the squat tasks were minimized by having subjects squat at a consistent frequency (1 Hz) and enforcing that the thighs finished parallel with the ground at the end of each squat. Subjects continued to alternate between squat sequence and jump landing task until maximal fatigue was attained, being defined as the point where they could no longer complete three complete squats unassisted. The potential for subject-based variations in maximum fatigue level was also considered within the ensuing statistical treatment of the data (see below). Verbal cues were used to motivate subjects as they progressed through to maximal fatigue. These cues were provided by a single experimenter (McLean) throughout the fatigue protocol, and were consistent between subjects to minimize the potential for any confounding effects.

3. Kinematic analyses

Lower limb joint rotations were quantified for each landing trial based on the 3D coordinates of twenty-eight (12.7 mm diameter) precisely attached reflective skin markers (McLean et al., 2007) (Fig. 2). Markers were secured to pre-determined anatomical landmarks via hypoallergenic, air-permeable cross elastic tape (Cover-Roll Stretch, BSN Medical GmbH, Hamburg, Germany). All attachment sites were initially shaved and attachment over areas of large

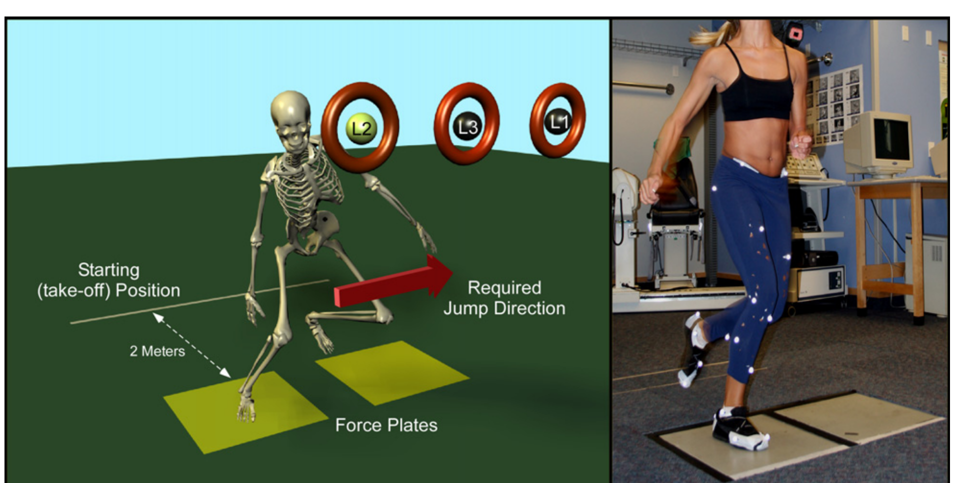

Fig. 1. Subjects reacted to a random light stimulus and moved in the appropriate direction upon landing from a forward jump. Three lights were used, corresponding to a rapid jump to the right (L1) or left (L2), or vertical (L3).

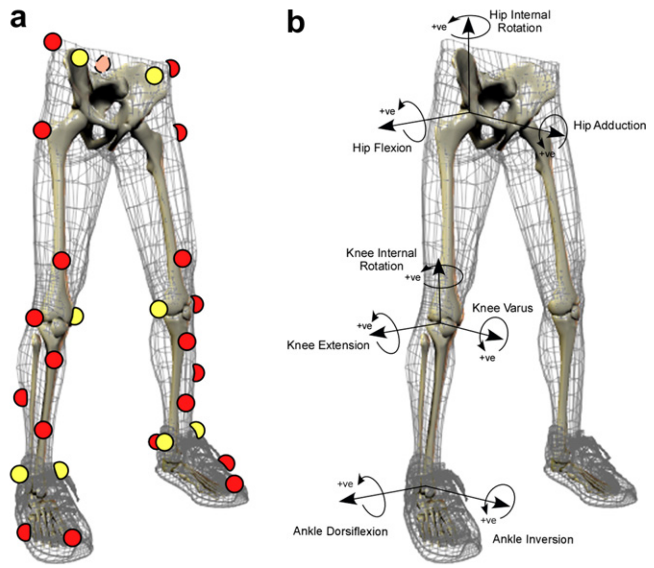

Fig. 2. Marker locations used to define a kinematic model comprised of nine skeletal segments (a). The left and right ASIS and bilateral medial femoral condyle and lateral and medial malleoli markers (yellow) were removed prior to the recording of movement trials. Pelvis (body) motion was described with respect to the Global (lab) coordinate system via three translational and three rotational degrees of freedom (b). The hip, knee and ankle joints were defined locally and each assigned three respective rotational DoFs. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

muscle mass was avoided to minimize excessive marker movement during ground contact. Ankle length spandex tights were subsequently pulled-up over the markers, ensuring their positions were maintained. Small incisions were then made in the tights and the makers were subsequently drawn through. These steps were taken to maximize the contrast disparity between the markers and background, and further, to minimize the potential for marker movement or loss, particularly when subjects began to sweat as they approached maximal fatigue.

Following marker placement, a high-speed video recording was first obtained with the subject standing in a stationary (neutral) position (McLean et al., 2007). A kinematic model was then defined based on these data, consisting of nine skeletal segments (foot, talus, shank and thigh of each limb, and the pelvis) and 24 degrees of freedom using Mocap Solver 6.17 software (Motion Analysis Corp., Santa Rosa, USA). We have used these methods extensively to successfully quantify lower limb joint rotations for these and similar movements (McLean et al., 2004, 2005, 2007). Specifically, the pelvis was assigned six DOF relative to the global (laboratory) coordinate system, with the hip, knee and ankle joints of each limb defined locally and assigned three rotational DOF respectively (McLean et al., 2003). Hip, knee and ankle joint centers were also defined in accordance with our previous work (McLean et al., 2003). The 3D marker trajectories recorded during each landing trial were processed by the Mocap Solver software, to solve for the 3D lower limb joint rotations at each time frame. Joint rotations were expressed relative to each subject's standing (neutral) position data (McLean et al., 2005, 2007). These data were subsequently time normalized to 100% of stance, being resampled at 1% increments ($N = 101$), with heel strike and toe-off defined as the instant when the vertical GRF first exceeded and went below 10 N respectively (McLean et al., 2007).

4. Statistical treatment

Planned statistical comparisons of key kinematic parameters were undertaken based on variables linked previously to an increased risk of ACL injury (Ford et al., 2003; Hewett et al., 2005; Kernozek et al., 2005; McLean et al., 2007). Specifically, initial contact (at landing) hip and knee flexion–extension, hip internal–external rotation, and ankle plantar–dorsi flexion angles, and peak stance (0–50%) phase knee abduction and internal rotation, and ankle supination angles were obtained from each single leg landing trial. Peak measures were defined between 0% and 50% of stance since there is increased suggestion that injury occurs within this time-frame (Griffin et al., 2006).

Mean subject-based values for each of the seven dependent factors defined above were first calculated across leg (2) and decision making (2) conditions from the pre-fatigue jump trial data. From the remaining (fatigue) trials, dependent measure data obtained from the last successful trial for each of the same four landing conditions was recorded, and denoted as the maximum (100%) fatigue value in each case. Data corresponding to 50% fatigue were also recorded, being the median trial of each condition.

Mean pre-fatigue maximum vertical (two-legged) jump heights were calculated for each subject based on the maximum height of the pelvis center of mass (McLean et al., 2007). Maximum jump heights were similarly calculated for the 0% (first), 50% (median) and 100% (last) fatigue vertical jump trial and represented as a percentage of the pre-fatigue baseline value. The potential for between-subject variations in initial, mid and maximum fatigue levels to confound outcomes were subsequently examined by considering these percentages as covariates within the ensuing statistical treatment. Kinematic data were subsequently submitted to a three-way mixed design ANCOVA, testing for the main effects of leg (dominant and non-dominant), fatigue (pre, 50% and 100%) and decision (anticipated and unanticipated), and interactions between these factors, with decision treated as a repeated measure within each subject. An alpha level 0.05 was used to denote statistical significance for all comparisons. In instances where a significant effect of fatigue or interactions between main effects was observed, a Tukey post-hoc analysis was used to determine precisely where they occured.

5. Results

Subjects completed an average of 78.8 (9.2) squat-landing sequences prior to reaching maximal volitional fatigue, equating to approximately 20 min. Percentage maximum jump heights at 0% fatigue (98.2 (3.2) %) were similar ($P = 0.767$, observed power = 0.023) to mean baseline values (Fig. 3). Percentage height values at both 50% (61.3 (9.4) %) and 100% fatigue (52.8 (10.7) %) however, were statistically significantly ($P < 0.001$) lower than the 0% and pre fatigue values. Furthermore, percentage maximum jump height values at 50% and 100% fatigue were found to be similar (observed power = 0.064). Inter-subject variations in maximum percentage jump heights were also not found to influence the remaining statistical outcomes. Finally, fatigue (observed power = 0.069), decision making (observed power = 0.054) or leg conditions (observed power = 0.053) did not influence mean contact time data (Table 1). Considering the above results therefore, it was felt that the remaining statistical evaluations could be made with confidence.

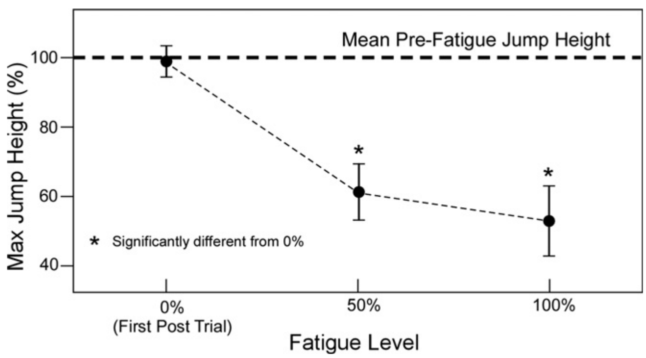

Fig. 3. Comparison of mean (±SD) jump heights as a function of fatigue level. Specifically, maximum jump heights were significantly lower than initial (0% fatigue) heights at 50% and 100% fatigue. Jump heights at 100% fatigue were also significantly lower than 50% fatigue values.

Both fatigue and decision making had a substantial impact on initial contact hip, but not on initial contact knee and ankle postures during the single leg landing tasks (Fig. 4). Fatigue evoked statistically significant decreases in initial contact hip flexion ($P < 0.001$) and increases in hip internal rotation ($P < 0.001$) positions compared to pre-fatigue baseline measures (Table 2). Specifically, significant changes were observed for each of these initial contact rotations at both the 50% and 100% fatigue levels compared to baseline. Differences between 50% and 100% fatigue levels however, were not found in either case. Significant decreases in initial contact hip flexion ($P < 0.001$) and increases in hip internal rotation ($P < 0.001$) positions also occurred during unanticipated compared to anticipated landings. Furthermore, a statistically significant ($P < 0.05$) interaction between the main effects of fatigue and decision making was observed in both cases, with fatigue induced changes in initial contact hip

Table 1
Combined effects of movement type, leg and level of fatigue on resultant mean (SD) ground contact times during execution of single leg landings

	Anticipated			Unanticipated		
	Pre	50%	100%	Pre	50%	100%
Dominant	0.342 (0.061)	0.341 (0.091)	0.348 (0.085)	0.339 (0.105)	0.345 (0.099)	0.340 (0.093)
Non-dominant	0.344 (0.078)	0.341 (0.101)	0.346 (0.084)	0.344 (0.065)	0.339 (0.103)	0.347 (0.096)

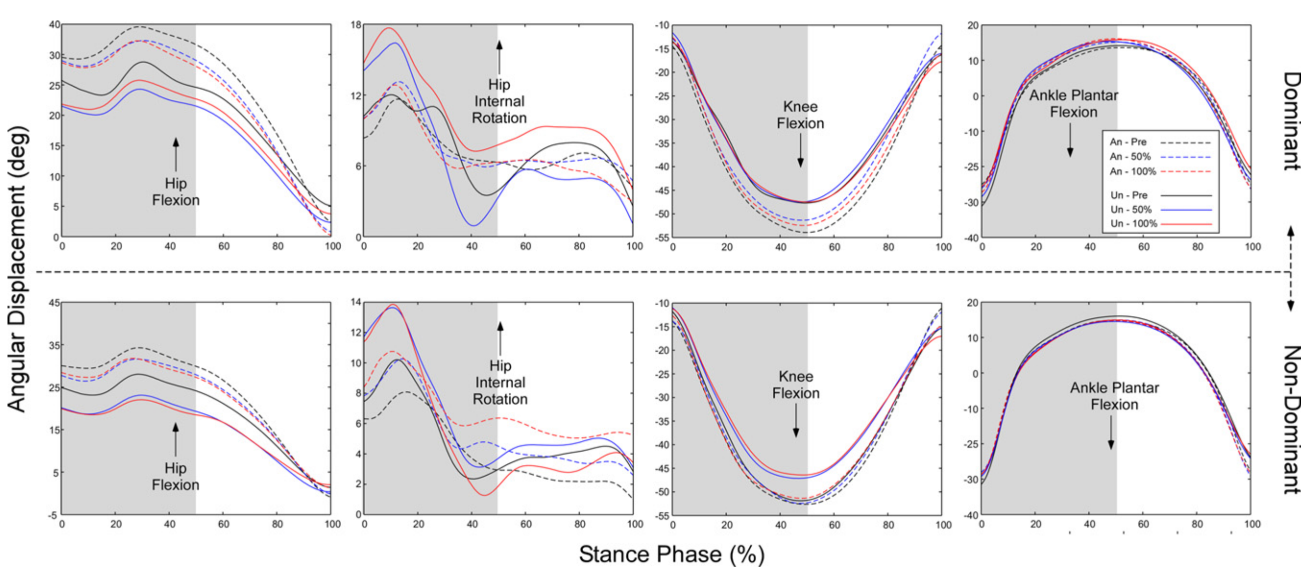

Fig. 4. Comparison of initial contact (at landing) 3D lower limb joint angular displacements across movement (anticipated and unanticipated), leg (dominant and non-dominant) and fatigue (pre, 50% and 100%) conditions.

Table 2
Combined effects of movement type, leg and level of fatigue on resultant mean (±SD) lower limb initial contact (shaded gray) and peak (shaded white) stance phase (0–50%) joint kinematics during execution of single leg landing tasks

Rotation (degrees)	Dominant						Non-dominant					
	Anticipated			Unanticipated			Anticipated			Unanticipated		
	Pre	50%	100%	Pre	50%	100%	Pre	50%	100%	Pre	50%	100%
Hip fle[b,c,d]	31.0 (3.0)	28.9 (2.8)	27.2 (2.4)	25.8 (2.4)	21.8 (2.8)	21.9 (2.6)	31.2 (3.2)	28.8 (2.8)	29.1 (2.8)	25.0 (2.1)	20.1 (2.3)	200 (2.1)
Hip IR[b,c,d]	8.2 (2.1)	10.0 (1.9)	10.0 (2.2)	10.2 (1.5)	14.2 (1.5)	14.8 (2.7)	6.3 (1.6)	7.8 (1.8)	8.4 (2.4)	7.4 (1.4)	11.8 (1.9)	11.3 (2.3)
Knee fle	14.4 (4.3)	12.7 (3.8)	12.5 (3.6)	13.4 (3.2)	13.8 (3.5)	12.6 (4.8)	13.8 (4.8)	13.8 (3.5)	12.6 (3.6)	11.8 (2.2)	10.9 (3.8)	11.1 (4.5)
Ankle plan fle	25.5 (17.8)	26.5 (14.0)	27.7 (8.4)	28.9 (10.2)	28.3 (9.4)	28.9 (11.7)	31.6 (12.6)	28.9 (12.5)	26.1 (9.8)	31.7 (8.3)	29.5 (8.1)	28.9 (7.8)
Knee abd[b,c,d]	3.5 (3.2)	3.6 (3.3)	4.3 (3.1)	3.9 (2.8)	7.9 (3.3)	7.5 (3.8)	0.9 (2.0)	1.9 (3.5)	1.9 (3.7)	1.9 (3.2)	8.7 (3.2)	8.9 (4.1)
Knee IR[a,b,c]	13.1 (4.2)	14.1 (5.1)	14.4 (5.0)	15.2 (5.0)	18.7 (5.4)	19.0 (5.4)	12.7 (4.2)	14.5 (4.4)	13.5 (4.7)	13.9 (3.8)	18.8 (5.2)	18.8 (3.4)
Ankle sup[c]	9.1 (3.6)	13.0 (3.1)	13.0 (3.2)	9.5 (2.9)	12.5 (2.8)	13.1 (2.6)	8.6 (3.3)	12.4 (2.0)	12.1 (3.3)	9.3 (2.4)	13.3 (2.7)	12.3 (1.7)

[a] Denotes statistically significant effect of leg.
[b] Denotes statistically significant effect of movement (antic or unant).
[c] Denotes statistically significant effect of fatigue level (pre, 50% or 100%).
[d] Denotes statistically significant interaction between movement and fatigue level.

flexion and internal rotation positions being more pronounced during unanticipated compared to anticipated landings. No initial contact measures were found statistically to be influenced by the main effect of leg (see Table 1).

Several peak stance (0–50%) phase lower limb rotations were similarly influenced by the main effects of fatigue and/or decision making (Fig. 5). Specifically, fatigue promoted statistically significant increases in peak knee abduction ($P < 0.001$), knee internal rotation ($P < 0.001$) and ankle supination ($P < 0.01$) angles compared to baseline pre-fatigue values (see Table 1). Similar to initial contact hip rotation comparisons, peak values for each of these three rotations were significantly different from baseline at both 50% and 100% fatigue. Again however, significant differences were not explicitly observed between 50% and 100% fatigue levels.

Peak stance knee abduction and internal rotation postures were statistically significantly ($P < 0.001$) greater during unanticipated compared to anticipated landings. Furthermore, a statistically significant ($P < 0.001$) interaction between fatigue and decision effects was observed for peak knee abduction measures, with fatigue induced increases in this parameter being more evident during unanticipated compared to anticipated landings (see Fig. 5). While a similar trend was observed for peak stance knee internal rotation however (see Fig. 5), it was not substantiated statistically. Again, peak stance (0–50%) phase kinematic data were not found to be influenced statistically by the main effect of leg.

6. Discussion

Successful prevention of non-contact ACL injuries relies on a detailed understanding of the underlying injury mechanism. Previous studies have shown that both fatigue and decision making, factors synonymous with realistic sports participation, promote high-risk lower limb joint neuromechanical adaptations that may manifest within the resultant cause of injury. The outcomes of the current study strongly support these contentions and further suggest that the integrative impact of fatigue and decision making may present as a worst case scenario for high-risk dynamic landing strategies. If this is indeed the case, then combating the deleterious impact of these combined factors within the injury prevention model appears crucial.

In our previous work, we noted that variations in the underlying fatigue model may have precipitated the equally diverse theories on how fatigue implicates within the neuromuscular response and hence injury risk (Miura et al., 2004; McLean et al., 2007). Local fatigue models for instance, most often based on repetitive isokinetic loading of targeted muscle groups, precipitate substantial movement control adaptations (Nyland et al., 1997a,b) and mechanoreceptor dysfunction (Miura et al., 2004; Wojtys et al., 1996), which may culminate in an increased risk of ACL injury. Such models however, do not necessarily reflect how fatigue progresses and implicates during

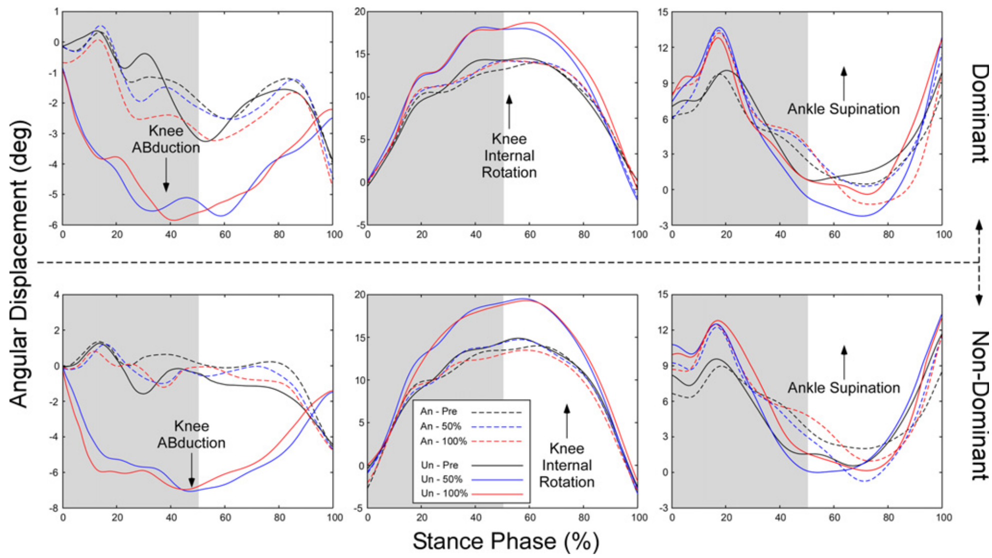

Fig. 5. Statistically significant ($P < 0.006$) interactions between the main effects of movement and fatigue conditions were observed for peak stance phase (0–50%) hip internal rotation and knee Abduction measures. Specifically in both cases, fatigue effects were more pronounced during unanticipated compared to anticipated landings.

realistic sports participation, suggesting a more general induction strategy is warranted. General fatigue models appear to more practically simulate sports relevant movement tasks and may induce both local and central fatigue effects along the entire proprioceptional pathway (Lattanzio and Petrella, 1998; Miura et al., 2004). Similar to more recent investigations therefore (Chappell et al., 2005; McLean et al., 2007), we currently chose to examine fatigue effects within a more general induction model.

A major concern in adopting a general fatigue approach is that it is extremely difficult to quantify and hence compare fatigue induced strength and/or power adaptations (McLean et al., 2007). To address this issue, recent studies have defined maximal fatigue levels based on volitional exhaustion, being the point that the subject can no longer perform a specific component of the fatiguing task/s (Chappell et al., 2005; Madigan and Pidcoe, 2003). Assuming subjects indeed progress through to a realistic exhaustive end point, and that muscle groups being fatigued are actually utilized within the comparative performance task, this approach likely minimizes the potential for confounding between-subject variations in maximal neuromuscular fatigue levels. We thus chose to induce fatigue via repetitive squatting tasks, since they represent a closed kinetic chain exercise involving hip, knee and ankle agonistic and antagonistic musculature, which are equally prominent in the control and coordination of jump landing tasks similar to that tested herein (Chappell et al., 2005; Besier et al., 2003; Decker et al., 2003). While it is difficult to determine whether comparable levels of fatigue were actually reached in each subject, visual assessment suggests this was the case with all subjects being physically unable to complete any more squatting tasks. Maximum (100%) fatigue jump height percentages, which we used to denote maximal fatigue (McLean et al., 2007) were also consistent between subjects, being well below those recorded prior to fatigue (see Fig. 3).

As clearly stated in the title of this paper, we wished to examine the potential for fatigue to impact lower limb neuromechanics during landing. In simple terms, muscle fatigue represents a reduction in the ability to generate force or power (Gandevia, 2001). This concept is distinctly different however from exhaustion, being an inability to sustain exercise at predefined target intensities (Vollestad, 1997). Considering the nature of our cumulative fatiguing task, it is thus plausible that while subjects currently experienced maximal exhaustion, they may not have progressed through to maximal fatigue. Regardless of whether control and execution pathways were maximally affected through this process however, we feel we have reproduced a fatigued/exhaustive state that is more typical of that experienced with the true sporting context. Hence, while we recommend future studies adopt similar protocols to best examine movement strategies adopted under sports related physical and/or mental duress, the term exhaustion rather than fatigue may be a more appropriate governing term.

The majority of previous studies investigating relationships between fatigue and joint neuromechanics have adopted a pre-test–fatigue – post-test model, with resultant injury based on disparities between pre and post fatigue measures (Chappell et al., 2005; McLean et al., 2007; Nyland et al., 1997a, Wojtys et al., 1996). Recent research however, suggests that a rapid deterioration in fatigue effects may present within such a design (McLean et al., 2007), compromising outcome and conclusion efficacies. We thus quantified and compared lower limb joint kinematics in parallel with the progression towards maximum fatigue (Madigan and Pidcoe, 2003). This approach not only negates the above concern, but also enables specific time points along the fatigue continuum to be identified and compared. To achieve the purposes of the current study, and for statistical integrity, we chose only to compare data at 50% and 100% fatigue levels. We intend however, to examine complete neuromechanical time histories in more detail in our future work, affording precise determination of when deleterious fatigue effects initiate.

Lower limb stance phase kinematic patterns and the associated initial contact and peak (0–50%) rotation magnitudes and timings were reasonably consistent with those reported previously for similar landing tasks (Decker et al., 2003; Kernozek et al., 2005; McLean et al., 2007). Subtle differences among these data likely stem from concomitant variations in the movement tasks (McLean et al., 2005), subject skill levels (McLean et al., 2004; Ford et al., 2003), and the way in which the underlying kinematic model was defined (Cappozzo et al., 2005). We have used the current model, and the underlying processing methods extensively (McLean et al., 2003, 2004, 2005, 2007) and are confident it provides reliable within-subject comparisons of lower limb joint kinematic data for dynamic sports postures, within the known constraints of an external marker-based analysis method (Cappozzo et al., 2005; Leardini et al., 2005).

Hip control was substantially altered during fatigue, with subjects tending to land with decreased hip flexion and increased hip internal rotation when in this state. With the hip joint being a primary support and stabilizing mechanism during the absorption phase of dynamic landing tasks (Decker et al., 2003; McNitt-Gray, 1993), a more extended initial hip position in the presence of fatigue seems intuitive. Specifically, having subjects perform repetitive squatting tasks likely compromised the ability of hip and knee extensors to eccentrically control the downward progression of the body center of mass immediately following contact. Landing in a more extended position when fatigued may thus prevent the excessive lower limb collapse that would otherwise occur. The fact that subjects elicited fatigue-induced increases in initial contact hip internal rotation positions may also stem from the underlying fatigue model. Apart from being primary hip extensors for example, the gluteals, which were likely fatigued as a result of the squatting tasks, also play an important secondary role in controlling axial hip rotations (Delp et al., 1999; Zeller et al., 2003). Of course, we did not quantify muscle activation data explicitly in this study and hence can only speculate on the precise interactions between fatigue-induced muscle control and resultant joint kinematics. Considering the intensity of our fatigue protocol however, we are confident that reasonable neuromuscular adaptation was evoked. Regardless, a detailed assessment of muscle activations and amplitudes in future work appears warranted.

The relationship between neuromuscular fatigue and resultant hip kinematics during jump landings may be somewhat dependent on the nature of the task itself. Fatigue-induced changes in sagittal plane hip control for example, have not been observed during two legged landings (Chappell et al., 2005; McLean et al., 2007), while they appear to be a more consistent in single leg tasks (Augustsson et al., 2006; Orishimo and Kremenic, 2006). Single leg landings present as a more functionally demanding task, since one leg is charged with decelerating the center of mass in both the vertical and horizontal directions over a relatively short time period (Orishimo and Kremenic, 2006). Ongoing evaluations of possible links between fatigue, hip neuromechanics and resultant non-contact ACL injury risk may thus be best served via a single leg landing model.

Fatigue-induced increases in peak stance phase knee abduction and internal rotation positions were consistent with previous observations (Chappell et al., 2005; McLean et al., 2007; Nyland et al., 1997b). Both internal rotation and abduction motions and loads are known to induce substantial increases in ACL loading (Kanamori et al., 2000; Markolf et al., 1995), with the latter recently shown prospectively to predict ACL injury risk in young female athletes (Hewett et al., 2005). Reasonable levels of neuromuscular fatigue may thus place the female knee joint in extreme, if not hazardous postures during dynamic single leg landing tasks. While the precise mechanisms for fatigued increases in out-of-plane knee motions remain largely unclear, there appear to be several possible explanations. Exercise at or near exhaustion for example, promotes increased knee joint laxity (Wojtys et al., 1996), possibly compromising ligament mechanoreceptor feedback and hence, muscle contributions to out-of-plane knee joint stability (Lattanzio and Petrella, 1998). Initial contact hip internal rotation postures elicited during dynamic landings have also been found to correlate directly with resultant peak stance phase knee abduction moments (McLean et al., 2005). Increases in knee abduction postures currently observed in the presence of fatigue may have thus stemmed from concomitant increases in the initial hip rotation position. Landing in this position likely promotes less than optimal bi-articular quadriceps and hamstring lengths (Delp et al., 1999), compromising their ability to successfully oppose external knee abduction loads (Besier et al., 2003; McLean et al., 2005). Considering further that the contractility and hence maximal force production of these muscles was likely drastically reduced as squatting trials progressed (Madigan and Pidcoe, 2003; Padua et al., 2006), substantial increases in peak knee abduction positions appear plausible. Fatigue-induced increases in peak tibial internal rotation postures may have stemmed from similar mechanisms, with the squatting task likely inhibiting soleus, gastrocnemius and deep posterior compartment calf muscle contractility (Manabe et al., 2007; Padua et al., 2006), which all act to limit internal tibial rotation motions (Nyland et al., 1997a,b).

We are somewhat unsure as to why peak ankle supination postures were increased in the presence of fatigue. We did not explicitly hypothesize a directional change in ankle pronation–supination, largely because of the lack of consistency in our own and other previous observations for this parameter (Kernozek et al., 2005; McLean et al., 2004, 2007). Extreme variations exist in human foot morphology (Isman and Inman, 1969), and the likely discrepancies in the ensuing talocrural axis definitions between studies may explain this phenomenon. Subtle differences in the movement patterns may similarly perpetuate vastly different stance phase ankle postures (McLean et al., 2007). With direct regard to current observations, it could simply be that our fatiguing task compromised peroneal contractility to the point where increased ankle supination was inevitable (Padua et al., 2006). We proposed above, that fatigue induced changes in the hip and knee positions may increase the potential for non-contact ACL injury risk. It may be however, that altered ankle strategies in the presence of fatigue may perpetuate to reduce rather than promote such an outcome. Increased ankle motions for example, may increase shock attenuation at the ankle joint (Self and Paine, 2001), minimizing the propagation of energy to the knee joint and ACL (Decker et al., 2003; Kernozek et al., 2005). Furthermore, with subjects required to aggressively push off in the opposite direction to the landing foot immediately following contact (see Fig. 1), increased supination may afford a more stable, laterally aligned base of support to successfully perform this movement phase when fatigued. Regardless, and especially considering the apparent inconsistencies reported for these postures, further investigations into the precise role of fatigue-induced ankle kinematics within resultant non-contact ACL injury risk appear warranted.

We originally hypothesized that fatigue would promote substantial modifications in lower limb kinematics during single leg landings and our results have largely supported this tenet. We did not however, expect that fatigue-induced kinematic adaptations would present with similar magnitudes earlier along the fatigue pathway. Specifically, fatigue induced changes in kinematic parameters did not differ between the 50% and 100% level. High-risk changes in lower limb joint postures may thus occur at much lower levels of fatigue than previously assumed (Chappell et al., 2005; McLean et al., 2007). Madigan and Pidcoe (2003) suggested that earlier changes may stem from a combination of learning effects, ongoing adaptive strategies to reduce impact forces, and neuromuscular protective mechanisms. Central fatigue effects may have also proliferated much earlier along the pathway than the associated peripheral adaptations (Knorr and Cafarelli, 2007; Kalmar and Cafarelli, 2006), acting to compromise resultant joint control even when muscle force was maintained (Gandevia, 2001). Considering our landing tasks incorporated reasonable processing and movement complexity, particularly during the unanticipated trials (Besier et al., 2001; Ulrich et al., 2006), early kinematic changes via such a phenomenon appear feasible. As noted earlier, we intend to explore the temporal evolution of fatigue during dynamic landing tasks more explicitly in our future work, which will have direct implications for current injury screening and prevention strategies.

This appears to be the first time that decision making and fatigue effects have been examined in combination with regard to ACL injury, and our results suggest that their integrative impact may be extremely problematic. With subjects currently required to perform ongoing sequential squatting and dynamic landing drills, extensive peripheral muscle fatigue was likely. It is equally plausible that substantial central fatigue was evoked with this progression, both at the spinal level (Lattanzio and Petrella, 1998; Melnyk and Gollhofer, 2007; Woods et al., 1987) via muscle spindle and tendon organ reflex inhibition, and the supraspinal level (Gandevia et al., 1996; Smith et al., 2007; Taylor et al., 2000) where reduced volitional drive of the descending motor pathways may have occurred due to focal adaptations in cortical inhibitability and excitability (Gandevia, 2001). For anticipated movements, positional adaptations and adjustments are orchestrated primarily through these same central control mechanisms (Benvenuti et al., 1997; Cuisinier et al., 2005), which may explain why some fatigue-induced kinematic increases were already evident during anticipated landings (see Figs. 4 and 5). If we thus extend this rationale to consider the combined fatigued unanticipated case, necessary additional temporal constraints may further compromise an already depleted central nervous system, promoting substantial and potentially catastrophic kinesthetic adjustments (Bayramoglu et al., 2007; Proske, 2006).

Apart from directly impacting spinal and supraspinal control mechanisms, integrated fatigue and decision making effects may have also precipitated adverse movement behaviors via cognitive deterioration. As muscle fatigue accumulates during a sub-maximal task sequence, such as that currently undertaken, force outputs can be maintained centrally for some time by increasing the number or discharge rate of activated motor neurons (Gandevia, 2001). This progressive increase in central control is often paralleled by a perceived need to exert increased "effort" (Lorist et al., 2002), in turn promoting extreme mental load and reduced time-on-task (Lorist et al., 2000; Zijdewind et al., 2006). In the current study, as central control systems were likely increasingly taxed to maintain movement success, mirrored increases placed on cognitive demands may have caused delayed and/or erroneous choice reaction responses. Considering therefore, that successful neuromuscular control strategies during dynamic landings are largely pre-planned (Besier et al., 2003; McLean et al., 2005), exaggerated kinematic postures elicited during the combined fatigued unanticipated trials may have been due in part to delayed initiation of an appropriate central response. The fact that subjects more frequently elicited erroneous movements as they continued to progress towards fatigue may further support this tenet.

We are hopeful that outcomes of the current study will provide the impetus for ongoing research aimed at a more precise understanding of the non-contact ACL injury mechanism, and the potential combined manifestations of fatigue and decision making within this mechanism. With this in mind, we thus feel it is important to present several inherent study limitations, outcome applications and future research considerations to assist in this ongoing development.

This study has highlighted the potentially important role of central fatigue in promoting hazardous lower limb joint postures during unanticipated landings. This outcome has important and immediate implications for ongoing prevention strategy developments, where trained adaptations to targeted central control mechanisms may provide substantial improvements in the resultant movement response, particularly within the inherently random sporting environment. The inclusion of more complex and challenging decision making tasks within the training regimen for example, may afford improved decisions, reactions and resultant movement control, particularly in a fatigued state. Integrating fatiguing drills within the training program may also enable individuals to hone newly learned and coordinated skills to the physically demanding nature of sports. Evaluating movement tasks incorporating combined fatigue and decision making components at the end of training may also enable program effectiveness and relative participant progressions to be more readily assessed. The addition of combined fatigue-decision making tasks to current prevention modalities thus appears to be a feasible progression.

Of course, before various facets of control can be successfully "trained", their specific contributions to the dynamic movement strategy, the relative contributions of fatigue within each, and the means through which it implicates must be clearly identified. While definitely challenging, methods currently exist that may allow these important initial steps to be undertaken and subsequently built upon (Taylor et al., 2000; van Duinen et al., 2007). Our own follow-up research for example, will necessarily include concomitant assessments of muscle activation timings and amplitudes (Besier et al., 2003) and associated reflex behaviors (Biro et al., 2007), enabling links between fatigue and resultant central (spinal) control to be more readily assessed. Exposing subjects to varying levels of cognitive and/or physical demand within the current assessment paradigm would also afford a simple, yet immediate means to better evaluate the relative and combined contributions of fatigue and decision making to the resultant movement strategy. Worth noting however, is that the dynamic, high impact nature of sports tasks linked to non-contact ACL injury does not automatically lend itself to the intricacies of state-of-the-art assessment modalities that may ultimately be necessary here, particularly if an assessment of supraspinal contributions is to be included (e.g., functional Magnetic Resonance Imaging and Transcranial Magnetic Stimulation). More innovative and controlled research models may thus be necessary, possibly migrating from the traditional lower limb model, so that the precise means through which fatigue and decision making implicate and ultimately integrate within the non-contact ACL mechanism can be explored.

7. Conclusions

Based on the research outcomes obtained for the population tested, the following conclusions can be drawn:

1. Neuromuscular fatigue promotes significant decreases in initial contact hip flexion and significant increases in initial contact hip internal rotation, and in peak stance (0–50%) phase knee abduction, knee internal rotation and ankle supination positions during the execution of dynamic single leg landings.
2. Fatigue-induced modifications in lower limb kinematics observed at maximum (100%) fatigue during single leg landings are already evident at the 50% fatigue level.
3. Fatigue-induced changes in initial contact hip flexion and internal rotation, and peak stance (0–50%) phase knee abduction positions are significantly more pronounced during unanticipated compared to anticipated single leg landing tasks, suggesting substantial degradation in both peripheral and central processing mechanisms.

Conflict of Interest Statement

No authors listed in conjunction with this manuscript submission demonstrate any form of conflict of interest, be it financial or otherwise.

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