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Authors(s)	Doherty, Cailbhe, Bleakley, Chris J., Hertel, Jay, Caulfield, Brian, Ryan, John, Delahunt, Eamonn
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Title: Single-leg drop landing motor control strategies, 6-months following an acute lateral ankle sprain injury.

Running title: Ankle sprain and landing performance.

Authors:

Cailbhe Doherty[1]

Chris Bleakley[3]

Jay Hertel[4]

Brian Caulfield[1]

John Ryan[5]

Eamonn Delahunt[1,2]

1. School of Public Health, Physiotherapy and Population Science, University College Dublin, Dublin, Ireland.
2. Institute for Sport and Health, University College Dublin, Dublin, Ireland.
3. Sport and Exercise Sciences Research Institute, Ulster Sports Academy, University of Ulster, Newtownabbey, Co. Antrim, Northern Ireland.
4. Department of Kinesiology, University of Virginia, Charlottesville, VA, United States.
5. St. Vincent's University Hospital, Dublin, Ireland.

Address for Correspondence:

Cailbhe Doherty

A101

School of Public Health, Physiotherapy and Population Science

University College Dublin

Health Sciences Centre

Belfield

Dublin 4

Ireland

Email: cailbhe.doherty@ucdconnect.ie

Telephone: 00 353 1 7166671

Fax: 00 353 1 716 6501

Abstract

No research exists predicating a link between acute ankle sprain injury-affiliated movement patterns and those of chronic ankle instability (CAI) populations. The aim of the current study was to perform a biomechanical analysis of participants, 6-months after they sustained an acute lateral ankle sprain (LAS) injury to establish this link.

Fifty-seven participants with a six-month history of LAS and twenty non-injured participants completed a single-leg drop landing task (DL) on both limbs. 3-dimensional kinematic (angular displacement) and sagittal plane kinetic (moment of force) data were acquired for the joints of the lower extremity, from 200ms pre-initial contact (IC) to 200ms post IC. Individual joint stiffnesses and the peak magnitude of the vertical component of the ground reaction force (GRF) were also computed. Injured participants displayed increases in hip flexion and ankle inversion on their injured limb ($p < 0.05$); this coincided with a reduction in the net flexion-extension moment at the hip-joint, with an increase in its stiffness ($p < 0.05$). There was no difference in the magnitude of the peak vertical GRF for either limb compared to controls. These results demonstrate that altered movement strategies persist in participants, 6-months following acute LAS which may precipitate the onset of CAI.

Key terms: ankle joint [MEsH]; biomechanics [MEsH]; kinematics [MEsH]; kinetics [MEsH]; task performance and analysis [MEsH].

INTRODUCTION

The biomechanics literature is replete with investigations which have performed laboratory analyses of dynamic movement tasks. Typically the periods of interest for researchers during such tasks include those in which movement is terminal and the final constraint of the task is relative stasis (such as in drop landings), or those in which acceleration is manipulated by the participant, and movement is non-terminal (such is in drop jumps and cutting manoeuvres) (Schmitz et al., 2010). A number of recently published laboratory analyses have revealed the dichotomy of demands dictated by the terminal and non-terminal components of a given dynamic movement task (Bates et al., 2013, 2013). Typically, these types of analyses quantify the organisation and control of the motor apparatus by the sensorimotor system [otherwise known as coordination (Bernstein, 1967)] using kinematic and kinetic profiling. It has been argued that these studies, which quantify the energetics of coordination, are the most informative of all biomechanical analyses because they conceptualise the force responsible for producing observed movement patterns (Norcross et al., 2010).

Terminal and non-terminal jumping movement tasks are utilised in laboratory analyses as they are seen to mimic the demands of activities which typically lead to injurious events; for example, participants of sports such as volleyball and basketball are at a significantly greater risk for ankle sprain injury compared to participants of sports such as soccer or field-hockey (Doherty et al., 2014), likely due to the greater jumping and landing requirements of the former sports. Thus, researchers have sought to evaluate the movement patterns of participants with lateral ankle sprain (LAS) using dynamic movement tasks across the spectrum of this injury: those who have recovered fully 1-year following their acute LAS (known as ‘copers’) (Brown et al., 2012; Brown et al., 2008; Brown et al., 2009), those who are suffering the chronic sequelae associated with LAS [collectively described by the umbrella term of ‘chronic ankle instability’ (CAI)] for a minimum of 1-year following the first acute episode (Brown et al., 2010; Brown et al., 2012; Brown et al., 2006; Brown et al., 2008; Brown et al., 2011; Delahunt et al., 2006; Gribble et al., 2009; Terada et al., 2013), and those with a current acute lateral ankle sprain injury (Doherty et al., 2014; Doherty et al., In press). For example, a recent energetic analysis completed in our laboratory during a terminal movement landing task elucidated that participants with an acute LAS injury constraint rely on their hip-joint complex to absorb the forces of landing more so than non-injured participants. They do so by adopting a position of greater hip flexion during the landing descent with reduced extension moment during floor contact, potentially in the aim of reducing the impact of landing (Doherty et al., 2014). The bilateral nature of the observations made in this analysis lends to the hypothesis that acute ankle sprain may cause impairment of centrally mediated motor control pathways (Khin-Myo-Hla et al., 1999). There is a gap in the biomechanics literature however, as the movement patterns which characterise individuals after they have sustained an acute LAS injury, but have yet to develop CAI or proceed to full recovery, have not been characterized or established to date. It is plausible that participants with a six-month history of a LAS injury constraint might display coordination strategies evolved from those adopted in the acute phase of their injury (Doherty et al., 2014) and potentially akin to their coper and CAI counterparts.

The capacity of the sensorimotor system to reweight dependence on each of the lower extremity joints according to the demands of the task and the immediate capabilities of the individual is well established, and known as the energy absorption strategy (Norcross et al., 2010). For example, compared to males, females exhibit greater use of the ankle plantarflexors in absorbing the energy of a terminal landing task compared to males (Schmitz et al., 2007), and in instances where a task constraint of reducing landing ‘stiffness’ is prescribed individuals adopt a less erect body posture, adjusted primarily by moments of force produced at the hip (DeVita et al., 1992). In contrast to ankle sprain copers (Wikstrom et al., 2014), it is likely that the sensorimotor system of individuals suffering from CAI develop inappropriate coordination strategies which fail to effectively exploit the degrees of freedom available to their motor apparatus using suitable energetics (Brown et al., 2007). Evaluating participants with a 6-month history of LAS may elucidate certain coordination and energy absorption strategies which may persist following acute injury, and which may be precursors to those that belie CAI or coper status.

Therefore, the aim of this study was to perform a biomechanical energetic analysis of a group with a 6-month history of ankle sprain injury during a terminal-movement, drop land (DL) task. The current study is in continuation of one previously described (Doherty et al., 2014), which together form part of a larger study on the coordination strategy predictors of CAI or coper status during a terminal landing task. For the current study we hypothesised that participants with a 6-month history of LAS would display movement patterns similar to their acutely injured counterparts, when compared to a non-injured group: (i) They would adopt a position of increased hip flexion during landing descent which would persist into floor contact on both their previously injured, and non-injured limbs; (ii) This would manifest in a reduction in the extension moment of the hip moment of force profile, thus resulting in greater overall hip joint stiffness compared to controls; (iii) This hip-dominant strategy would be conducive to a reduction in the peak vertical ground reaction force (vGRF) of landing. Furthermore, in light of the evidence presented during a similar task in CAI populations, we hypothesised that the group with a 6-month history of LAS would display positions of increased knee flexion (Caulfield et al., 2002) and ankle inversion (Delahunt et al., 2006) during landing.

METHODS

Design

As part of the larger study conducted in our laboratory, fifty-seven participants (thirty-four males and twenty-one females, mean age of 22.6 years, mean height of 1.72 metres, mean body weight of 74.7 kg) were recruited from a University affiliated hospital Emergency Department (ED) within 2-weeks of sustaining a first-time acute LAS injury, and attended a single testing session 6-months after sustaining this injury in which all data for this study were acquired. Testing procedures for these participants in the acute phase of their injury has previously been reported (Doherty et al., 2014). An additional group of twenty participants (fifteen males and five females, mean age of 22.6 years, mean height of 1.73 metres, mean body weight of 71.4 kg) with no prior history of LAS were recruited from the hospital catchment area population using posters and flyers to act as a control group. Participants provided written informed consent, and the study was approved by the university's institutional review board.

Instrumentation

All participants were required to complete two questionnaires prior to the completion of testing protocol: the Cumberland Ankle Instability Tool (CAIT) was used to assess overall ankle joint function and symptoms (Hiller et al., 2006), and the activities of daily living and sports subscales of the Foot and Ankle Ability Measure (FAAMadl and FAAMsport) were used to quantify ankle-related function, patient reported symptoms and functional ability (Carcia et al., 2008).

Collection methods for this study have been previously documented (Doherty et al., 2014). Briefly, each participant was instrumented with the Codamotion bilateral lower limb gait set-up (Charnwood Dynamics Ltd, Leicestershire, UK) and asked to perform 3 DLs on both limbs, following a practice period. The DL task began with participants standing barefoot on a 0.4 m high platform with their test leg initially held in a non-weight bearing position and their knee flexed. Participants were then required to drop forward onto the test leg, landing on a force plate in front of the platform. Upon landing, participants were required to balance as quickly as possible on the test leg and hold this position for approximately 4–6 seconds. Kinematic data acquisition during trials of the DL was made at 200 Hz using 3 Codamotion cx1 units and kinetic data at 1000 Hz using 2 fully integrated AMTI (Watertown, MA) walkway embedded force plates. The Codamotion cx1 units were time synchronized with the force plates. Kinematic data were calculated by comparing the angular orientations of the coordinate systems of adjacent limb segments using the angular coupling set ''Euler Angles'' to represent clinical rotations in 3 dimensions (Winter, 2009). Marker positions within a Cartesian frame were processed into rotation angles using vector algebra and trigonometry (CODA mpx30 User Guide, Charnwood Dynamics Ltd, Leicestershire, UK). Kinematic and kinetic data were analysed using the Codamotion software and then converted to Microsoft Excel file format with the number of output samples per trial set at 100 + 1 in the data-export option of the Codamotion software, which represented the timeframe of interest during the DL trial as 100%, for averaging and further analysis. GRF data were passed through a third-order Butterworth low-pass digital filter with a 20-Hz cut-off frequency (Winter, 2009). The variables of interest were identified during the period from 200-ms pre-initial contact (IC) to 200-ms post-IC for the 3 successful DL trials for each subject on each limb. The vertical component of GRF (force plate registered vertical GRF greater than 10 N) was used to identify IC.

Time-averaged 3-dimensional angular displacement profiles for hip (flexion-extension; abduction-adduction; internal-external rotation), knee (flexion-extension; valgus-varus;

internal-external rotation), and ankle joints (plantarflexion-dorsiflexion; inversion-eversion; foot abduction-adduction) were calculated for each limb of all participants from 200-ms pre-IC to 200-ms post-IC. Total flexion displacement for the hip, knee, and ankle was calculated as the difference between the joint angle at ground contact and the peak joint angle.

Inverse dynamics were then used to calculate time averaged, sagittal plane hip, knee and ankle moments from the kinematic and force-plate data, with a net-supporting moment profile of all three joints from 200-ms pre IC to 200-ms post IC being identified for each limb of all participants during the DL task to identify the net-flexor/extensor pattern of all three joints (Winter, 1983). The supporting moment, Ms, during landing is defined as Ms = Mk – Ma - Mh, where Mk, Ma and Mh are the moments at the knee, ankle and hip respectively (Winter, 1980).

Sagittal-plane hip, knee, and ankle torsional joint stiffnesses were calculated as the change in normalized net internal moment (Nm) divided by the change in angular position (degrees) from initial contact to peak flexion excursion ($\mathrm{NM} \cdot \mathrm{Kg}^{-1} \cdot \mathrm{degrees}^{-1}$) during the defined landing phase (Farley et al., 1999; Schmitz et al., 2010).

Finally, absolute peak magnitude of the vertical component of the GRF within the first 200ms post-IC was also calculated for all participants. Prior to data analysis all values of force were normalised with respect to each subject's body mass (BM).

Statistical analyses

For the injured group, the limb with the recently incurred LAS was labelled as ‘‘involved’’ and the non-injured limb as ‘‘uninvolved’’. In all cases the limbs in the control group were side matched to the injured group; for each control subject, one limb was assigned as ‘‘involved’’ and one as ‘‘uninvolved’’ so that an equal proportion of right and left limbs were classified as ‘‘involved’’ and ‘‘uninvolved’’ in both the LAS and control groups.

To determine whether the LAS group would have persistent disability and poorer self-reported function, scores on the CAIT and subscales of the FAAM were compared to the control group using a multivariate analysis of variance. The independent variable was group (injured *vs.* control). The dependent variables were CAIT score, FAAMadl score and FAAMsport score for the involved limb. The significance level for this analysis was set a priori with a bonferonni adjusted alpha level of $p < 0.017$.

Between-group differences in involved and uninvolved limb 3-dimensional, time-averaged angular displacement profiles, and sagittal plane time-averaged net supporting moment profiles with their hip, knee and ankle components, were tested for statistical significance using independent-samples t-tests for each data point. The significance level for these analyses was set a priori at $p < 0.05$. Effect sizes were not calculated for the temporal kinematic and kinetic profiles secondary to the number of separate comparisons for each variable.

Independent samples, two-sided t-tests were undertaken for each limb to test for significant differences in sagittal plane hip, knee and ankle torsional stiffness in the time interval from 0 to 200-ms post-IC, and differences in the magnitude of the peak vertical GRF in the time interval from 0 to 200-ms post-IC during the DL task. The significance level for this analysis was set a priori at $p < 0.025$ (2 x limb).

Associated effect sizes (η^2) were calculated for all discrete variables with 0.01 = small effect size, 0.06 = medium effect size and 0.14 = large effect size.(Cohen, 1988) All statistical analyses were performed with IBM SPSS Statistics 20 (IBM Ireland Ltd, Dublin, Ireland).

RESULTS

With regards to the questionnaires, there was a statistically significant main effect for the combined dependent variables, $F (3, 63) = 14.80$, $p < 0.01$, Wilks' Lambda = 0.59, partial eta squared = 0.41. The means and standard deviations of the CAIT and subscales of the FAAM for the injured and control groups are presented in Table 1.

Time-averaged 3-dimensional kinematic profiles revealed that the injured group displayed altered movement patterns compared to control participants in sagittal plane hip motion on the involved and uninvolved limbs, sagittal plane knee motion on the uninvolved limb and frontal plane ankle motion on the involved limb. Kinematic profiles for the hip, knee and ankle joints are detailed in Figures 1-3 respectively.

Time-averaged sagittal plane kinetic profiles revealed between-group differences for the hip moment profile on the involved limb only. Sagittal plane kinetic profiles for the hip, knee and ankle with a coinciding support moment profile are presented in Figure 4.

There was a significant reduction in sagittal plane hip joint stiffness on both the involved and uninvolved limbs in the injured group compared to the control group in the time period from IC to 200ms post-IC (Table 2). Stiffness values for the involved and uninvolved limbs are depicted in figures 5 and 6 respectively.

There was no significant difference in the magnitude of the peak vertical GRF in the 0-200ms post-IC time interval for either the involved or uninvolved limbs (Table 3).

DISCUSSION

The findings of the current study both confirm and rebuke our experimental hypotheses. The participants with a 6-month history of ankle sprain injury displayed movement patterns and energetics akin to their acutely injured [bilateral increases in hip flexion, reduced post-IC hip extension moment on the involved limb (Doherty et al., 2014)], and CAI [increased pre-IC ankle inversion on the involved limb (Delahunt et al., 2009) and knee flexion (Caulfield et al., 2002)] counterparts. However, a number of the patterns of energy absorption displayed by these injured participants have either not been reported during a terminal landing task (increased stiffness at the hip joint), or were contrary to previous findings by Doherty et al. 2014 [no reduction in the peak vGRF (Doherty et al., 2014)].

These coordination and energy absorption strategies can be interpreted as the continuation of coping strategies adopted in the acute phase of injury (Doherty et al., 2014), in light of the persistence of reduced function reported by the injured group (based on the CAIT and subscales of the FAAM). Whether the emergence of these coping strategies following injury is contingent with complete recovery or chronicity is unknown based on the current findings, however, this study is part of a larger longitudinal analysis designed to clarify this.

Importantly, in analysing the results of the current study and those previously reported in other publications of participants across the spectrum of LAS injury (CAI, coper and acutely injured participants) we consider it inappropriate to directly compare the coordination and energetics that characterise these groups if the prescribed task is different in nature (terminal vs non-terminal dynamic movement tasks).

Landing impact during the DL task is controlled by lower extremity coordination strategies designed to arrest the downward velocity of the body's centre of mass in a safe and efficient manner (DeVita et al., 1992). The nature of the joint-specific contributions to the absorption of this kinetic energy should ideally limit the transmission of excess force to capsuloligamentous structures such as the lateral ligament complex of the ankle joint (DeVita et al., 1992), and exploit the geometry of the musculoskeletal system with its controlling musculature (Farley et al., 1999). The increased injury rate in CAI populations has been attributed to an increase or improper mediation of landing forces, secondary to the emergence of inappropriate coordination strategies following injury (Brown et al., 2010). Thus CAI populations are theorized to be vulnerable during landing as the preparatory and/or reactive coordination strategies devised by their sensorimotor system are poorly executed (Griller, 1972; Lees, 1981; Wikstrom et al., 2006); the current investigation provides a link as to how acute coordination strategies may be contingent with long-term outcomes during a terminal landing task.

The overall support moment profiles and their constituents in the current analysis give valuable insight into this integration of preparatory and reactive actions of the sensorimotor system (Winter, 1980). The sinusoidal trajectory of the hip joint moment following IC during this terminal landing task equates to a reduction in the net moment pattern at this joint when compared to that of the knee and ankle joints. As a result, smaller stiffness values were evident at the hip when compared to the knee and ankle, which did not fluctuate between extensor and flexor moments in the same manner as the hip. Thus, this flexor (≃ 50ms post-IC)-extensor (≃90ms post-IC)-flexor (≃160ms post-IC) pattern at the hip may be the central joint strategy to controlling the kinetic energy event associated with this terminal landing task, as the fundamental role of the knee and ankle, which displayed almost complete extensor dominance, was simply to prevent collapse of the lower extremity during descent via an eccentric extension and flexion moments respectively.

That injured individuals displayed a reduced extensor moment at the hip joint (80 to 96ms post-IC), which was soon followed by an increase in its flexor moment (144-152ms post-IC) was reflected in the stiffness parameter for these participants: because the net moment pattern at the hip was negative for injured participants (increased flexor pattern, reduced extensor pattern compared to controls), hip stiffness was reduced. In contrast, control participants had a negative net moment pattern at the hip (reduced flexor pattern, increased extensor pattern) with a resultant positive hip stiffness value or greater hip stiffness.

The results of the investigation pertaining to this injured groups' coordination strategies in the acute phase of injury elucidated contrasting hip-joint moment patterns to those displayed in the current study (when compared to control participants): a sequential pattern of reduced extension moment (≃80ms post-IC) followed by increased flexor moment at the hip (≃150ms post-IC) compared to controls in the current study was different, as participants in the acute phase of their injury actually displayed a reduced flexion moment (≃50ms post-IC) followed by the same pattern of reduced extension moment (≃80ms post-IC) (Doherty et al., 2014). It is likely that this reduction in flexor moment at the hip in the immediate impact phase of the landing (Lees, 1981) was an important contributor to the decrease in peak post-IC vGRF which was evident in the latter study (Doherty et al., 2014), and not in the current one. Whether the pattern consistent between the acute study and the current study (reduced hip extension moment ≃80ms post-IC) is appropriate to complete recovery following the initial LAS , or contributes to instability associated with CAI, is unknown based on the current findings; this can only be clarified at the 1-year follow-up.

The apparent importance of the hip in controlling the landing event based on the current observations is in agreement with the findings of other research groups (DeVita et al., 1992; Dufek et al., 1990). Furthermore, in light of the findings of a number of studies during other movement tasks, it is plausible that individuals who have sustained a LAS injury increase weighted dominance on hip-joint movement strategies to maintain static balance (Doherty et al., 2014), to fulfil both slow (Doherty et al., In press) and fast dynamic movement tasks

(Doherty et al., 2014) and in reacting to significant balance perturbation (Beckman et al., 1995), which is further evidenced by altered hip muscle activation onsets and patterns (Bullock-Saxton, 1994). It is possible that the reduction in hip stiffness in injured participants observed in the current study may be strategic in shifting dependence to a joint ideally suited to the transmission of larger impact forces (Alexander et al., 1990), and away from the ankle joint, which is more suited to subtle alterations in balance (Nashner et al., 1985).

Inspection of the kinematic data for the hip reveals a preparatory (pre-IC) increase in hip flexion on the involved limb, a trait shared by these participants in the acute phase of their injury (Doherty et al., 2014). That injured participants landed in increased hip flexion is intuitively linked to the decrease in the observed hip stiffness: Farley and Morgenroth (1999) report that if the leg is more extended at the instant of touchdown, the vGRF vector will be more closely aligned with the joints, simultaneously decreasing joint moments but increasing leg stiffness. However, describing the relationship of torsional joint stiffness and the position of the motor apparatus using a discrete point in such a linear fashion may be an over-simplification, as landing in a position of extreme hip flexion could theoretically increase the tension on passive and active structures crossing the joint, thus reducing the available range through which this joint can move, and increasing joint stiffness. That the increase in hip flexion displayed by injured participants in the current study was linked with a decrease in hip stiffness compared to controls lends to the original theory but must be considered with caution, and may be part of strategy contingent with that of the acute phase data (Doherty et al., 2014). This strategy has since become redundant, in consideration of the absence of a between-groups difference for the vGRF. Furthermore, because the impact forces of landing occur less than 30-50ms post-IC (DeVita et al., 1992; Nigg et al., 1981), and because the quickest potential reaction response by the sensorimotor system is insufficient to modify these forces, any attempt to reduce external impact must include some activity prior to contacting the landing surface (DeVita et al., 1992; Lees, 1981); apart from the brief period of significant increase in hip flexion, the injured group displayed no other preparatory coordination strategies in the sagittal plane.

The final between groups difference in involved limb kinematics was an increase in inversion at the ankle joint exhibited by the injured group (starting 96ms pre-IC). This may have occurred as a result of damage to the calcaneofibular ligament (CFL) when the injury was first sustained; the CFL is responsible for limiting inversion and frontal plane motion at the ankle, and is often injured in lateral ankle sprains (Stormont et al., 1985). This ligament was likely stretched or torn in the injured group, thus increasing frontal plane motion during the DL task. This increased in frontal plane motion is potentially anomalous in its capacity to increase the risk of sustaining further injury; increased inversion about the ankle joint axis equates to an increase in the ground reaction force moments about the sub-talar joint with significant potential for re-sprain of the injured ankle (Tropp, 2002; Wright et al., 2000).

Our findings lend to the hypothesis that participants with a history of LAS exhibit a seemingly re-weighted dominance on their hip joint in the completion of the movement tasks (Beckman et al., 1995; Doherty et al., 2014)(Bullock-Saxton, 1994; Doherty et al., 2014; Doherty et al., In press). This advances current understanding of the compensatory mechanisms which may lend to the onset of chronicity in groups with ankle sprain as, to the authors knowledge to date, no research has previously been conducted evaluating joint energetics and movement patterns in a group 6-months after they sustained their injury. In consideration of our findings clinicians must be cognizant of the potential persistence of injury-affiliated movement patterns with a coinciding decrease in self-reported functional capacity, even six-months following LAS injury. The inability to determine whether the movement patterns observed in the current study are predictors of chronicity or recovery may be a significant flaw, but as we have previously alluded to, this analysis is part of a larger study designed to tackle such a dilemma.

PERSPECTIVE

This analysis has elucidated a continuation of certain landing movement patterns previously exhibited by a group with acute LAS injury (Doherty et al., 2014), this time 6-months following the initial biomechanical evaluation. That many of these movement patterns are also exhibited by patients with CAI potentiates a link between the movement strategies adopted in the acute phase of injury and long-term outcome, although future longitudinal analyses of LAS participants are required to confirm this.

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Table 1. Participant self-reported function and disability questionnaire scores [mean ± SD] for the involved limb of injured and control groups.

Group	CAIT (/30)	FAAMadl (%)	FAAMsport (%)
Injured	21.44 ± 5.90[a]	95.78 ± 5.82[a]	83.72 ± 13.41[a]
Control	30 ± 0.00[a]	100 ± 0.00[a]	100 ± 0.00[a]

[a] significantly different from control group;

		Injured		Control				95% CI of the difference		
Limb		Mean	SD	Mean	SD	P	Mean difference	Lower	Upper	η^2
Involved	Hip	-0.01	0.06	0.06	0.09	0.01*	0.07	0.02	0.11	0.24
	Knee	0.07	0.02	0.08	0.03	0.20	-0.01	-0.02	0.01	0.03
	Ankle	-0.04	0.01	-0.04	0.02	0.97	0.00	-0.01	0.01	0.00
Uninvolved	Hip	-0.00	0.05	0.04	0.07	0.06	0.04	0.00	0.08	0.16
	Knee	0.07	0.02	0.08	0.02	0.58	0.01	-0.02	0.01	0.01
	Ankle	-0.03	0.01	-0.03	0.01	0.88	0.00	-0.01	0.01	0.00

Table 2. Mean ± SD values, p values, mean difference, 95% Confidence Interval of the difference (CI) and associated effect sizes for sagittal hip, knee and ankle torsional stiffnesses ($\mathrm{NM} \cdot \mathrm{kg}^{-1} \cdot \mathrm{degrees}^{-1}$) for the injured and control groups. Positive values indicate extensor dominance (greater stiffness); Negative values indicate flexor dominance (greater flexibility). *Indicates statistically significant difference.

	Injured		Control						
Limb	Mean (x BW)	SD	Mean (x BW)	SD	p	Mean difference	95% CI of the difference		η^2
Involved	2.44	.23	2.48	.24	.48	-.04	-.17	.08	0.01
Uninvolved	2.44	.20	2.51	.24	.20	-.07	-.19	.04	0.02

Table 3. Statistical output of peak post-IC vGRF analysis for injured and control participants on their involved and uninvolved limbs during the drop land task with associated effect sizes. Abbreviations: BW = bodyweight; IC = initial contact; vGRF = vertical ground reaction force.

Figure legends

Figure 1. Hip-joint adduction-abduction, flexion-extension and internal-external rotation during performance of the drop land task from 200ms pre-IC to 200ms post-IC for the involved and uninvolved limbs of injured and control groups. Adduction, flexion and internal rotation are positive; abduction, extension and external rotation are negative. Values are mean ± SEM. Black line with arrow = initial contact (IC). Shaded area = area of statistical significance.

Figure 2. Knee-joint varus-valgus, flexion-extension and internal-external rotation during performance of the drop land task from 200ms pre-IC to 200ms post-IC for the involved and uninvolved limbs of injured and control groups. Varus, flexion and internal rotation are positive; valgus, extension and external rotation are negative. Values are mean ± SEM. Black line with arrow = initial contact (IC). Shaded area = area of statistical significance.

Figure 3. Ankle-joint inversion-eversion, dorsiflexion-plantarflexion and foot abduction-adduction during performance of the drop land task from 200ms pre-IC to 200ms post-IC for the involved and uninvolved limbs of injured and control groups. Inversion, dorsiflexion and adduction are positive; eversion, plantarflexion and abduction are negative. Values are mean ± SEM. Black line with arrow = initial contact (IC). Shaded area = area of statistical significance.

Figure 4. Sagittal plane joint moment-of-force profiles for the hip, knee and ankle during performance of the drop land task from 200ms pre-IC to 200ms post-IC for the involved limb of injured and control groups. Extension moments are positive for Ms and Mk; flexor

moments are positive for Mh and Ma . Values are mean ± SEM. Black line with arrow=initial contact. Shaded area = area of statistical significance. Abbreviations: Mh = Hip moment; Mk = Knee Moment; Ma = Ankle moment; Ms = Support moment (Mk-Mh-Ma).

Figure 5. Injured and control relative joint stiffness on the involved limb during the drop land task. Positive values indicate extensor dominance (greater stiffness); Negative values indicate flexor dominance (greater flexibility).

[a] Indicates statistically significant difference from injured participants.

Figure 6. Injured and control relative joint stiffness on the uninvolved limb during the drop land task. Positive values indicate extensor dominance (greater stiffness); Negative values indicate flexor dominance (greater flexibility).

[a] Indicates statistically significant difference from injured participants.

Figure 1.3-dimensional hip kinematics during the landing task.

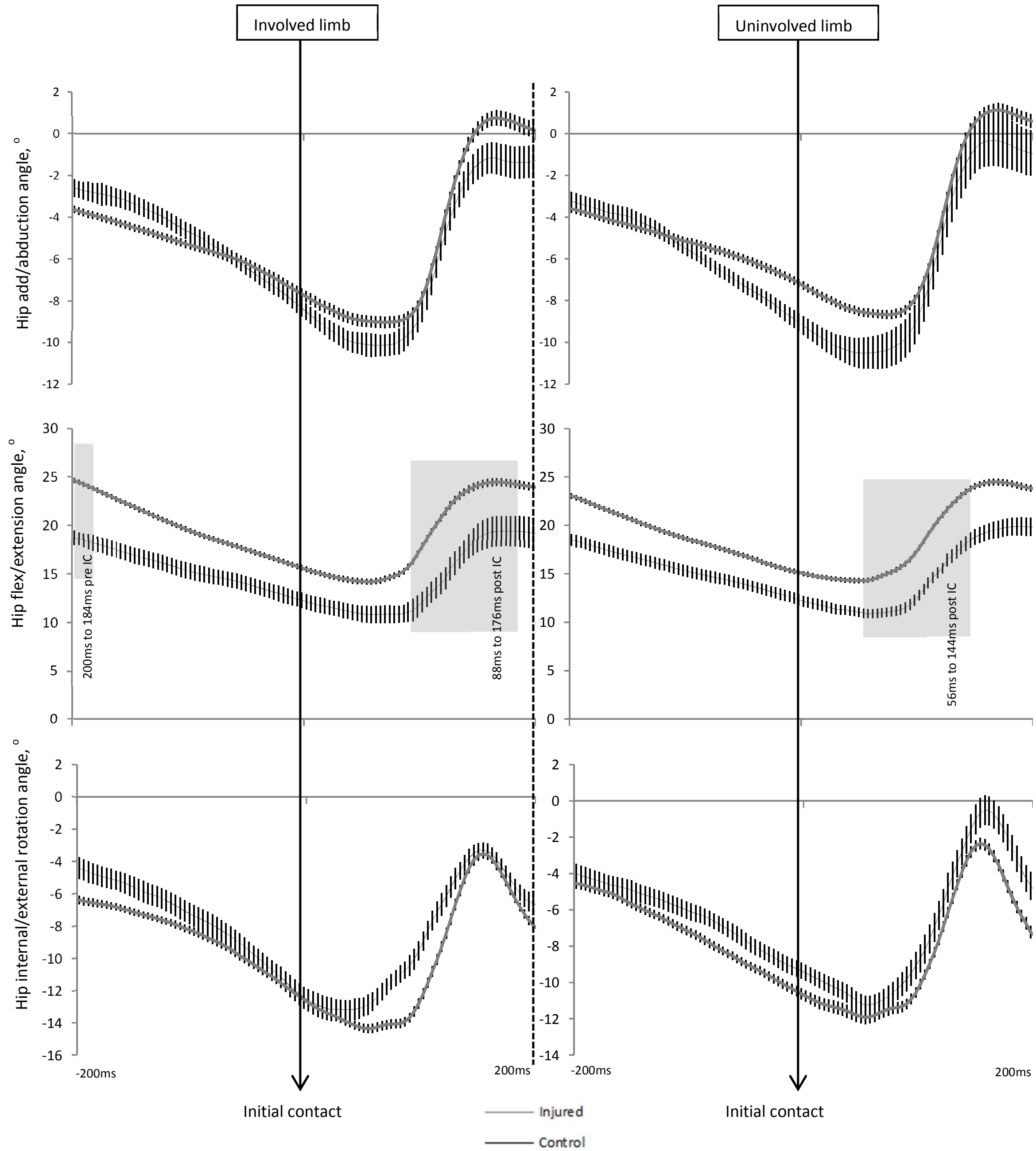

Figure 2.3-dimensional knee kinematics during the landing task.

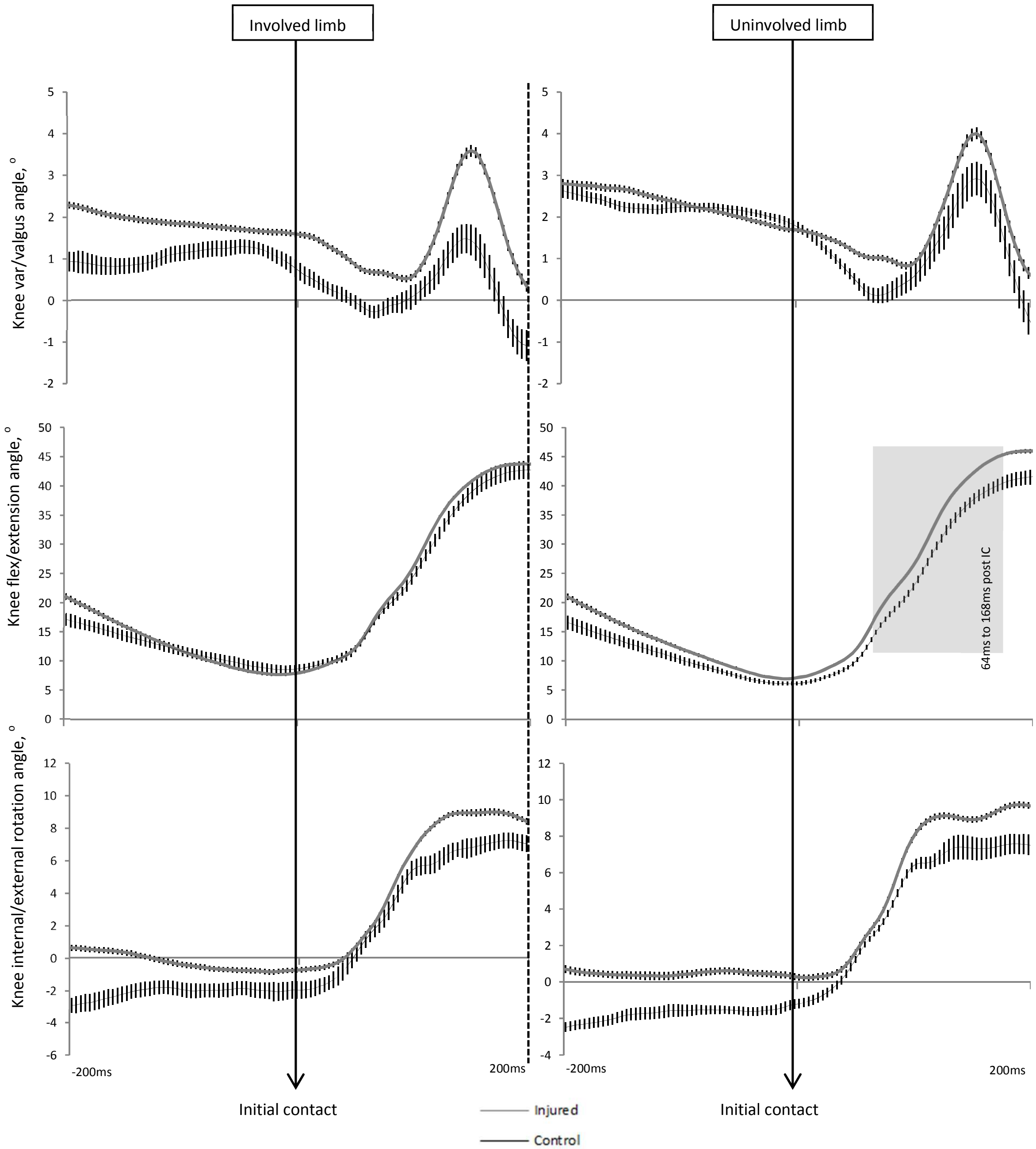

Figure 3.3-dimensional ankle kinematics during the landing task for the injured and control groups.

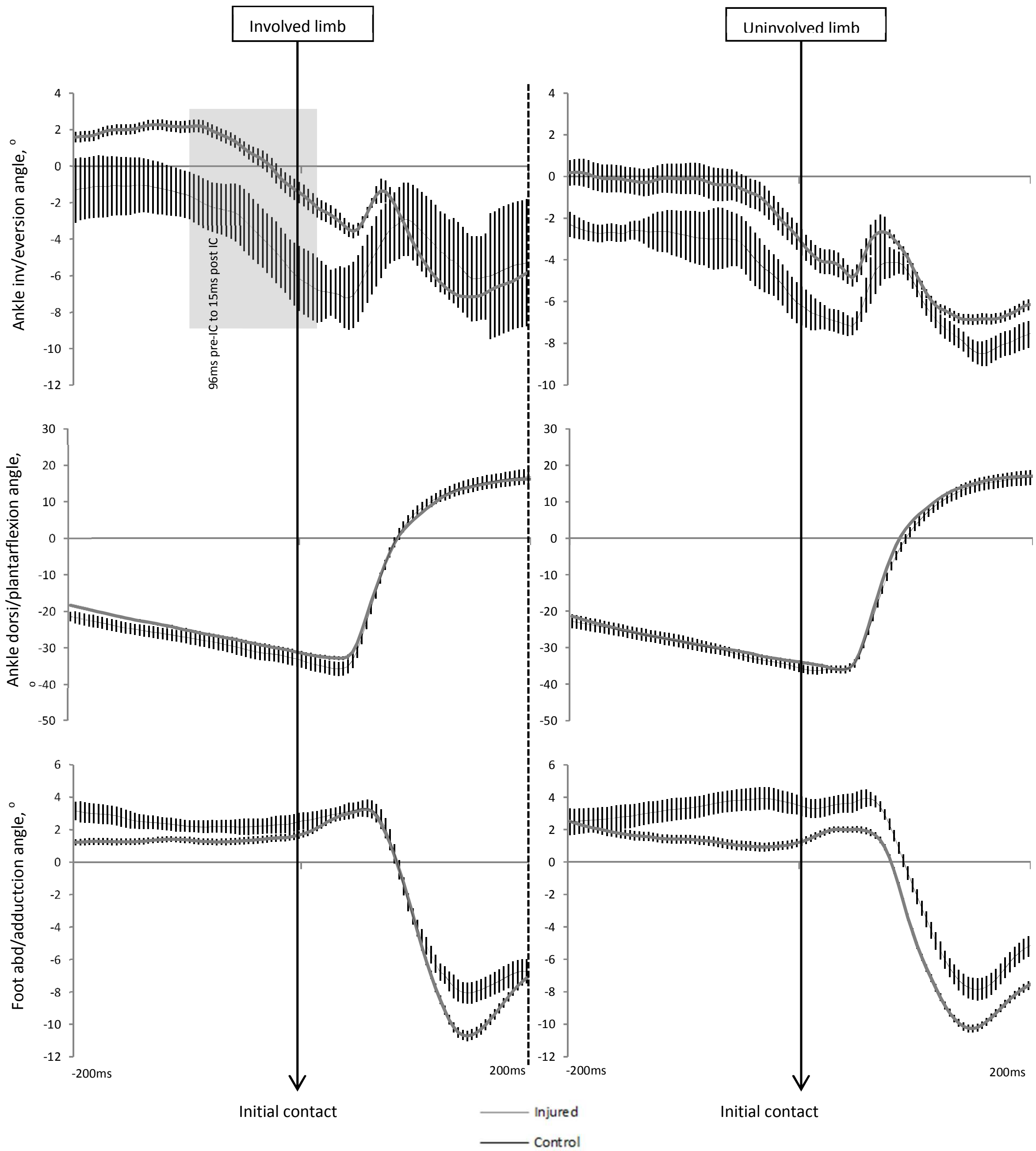

Figure 4.Sagittal plane lower extremity joint kinetics during the landing task.

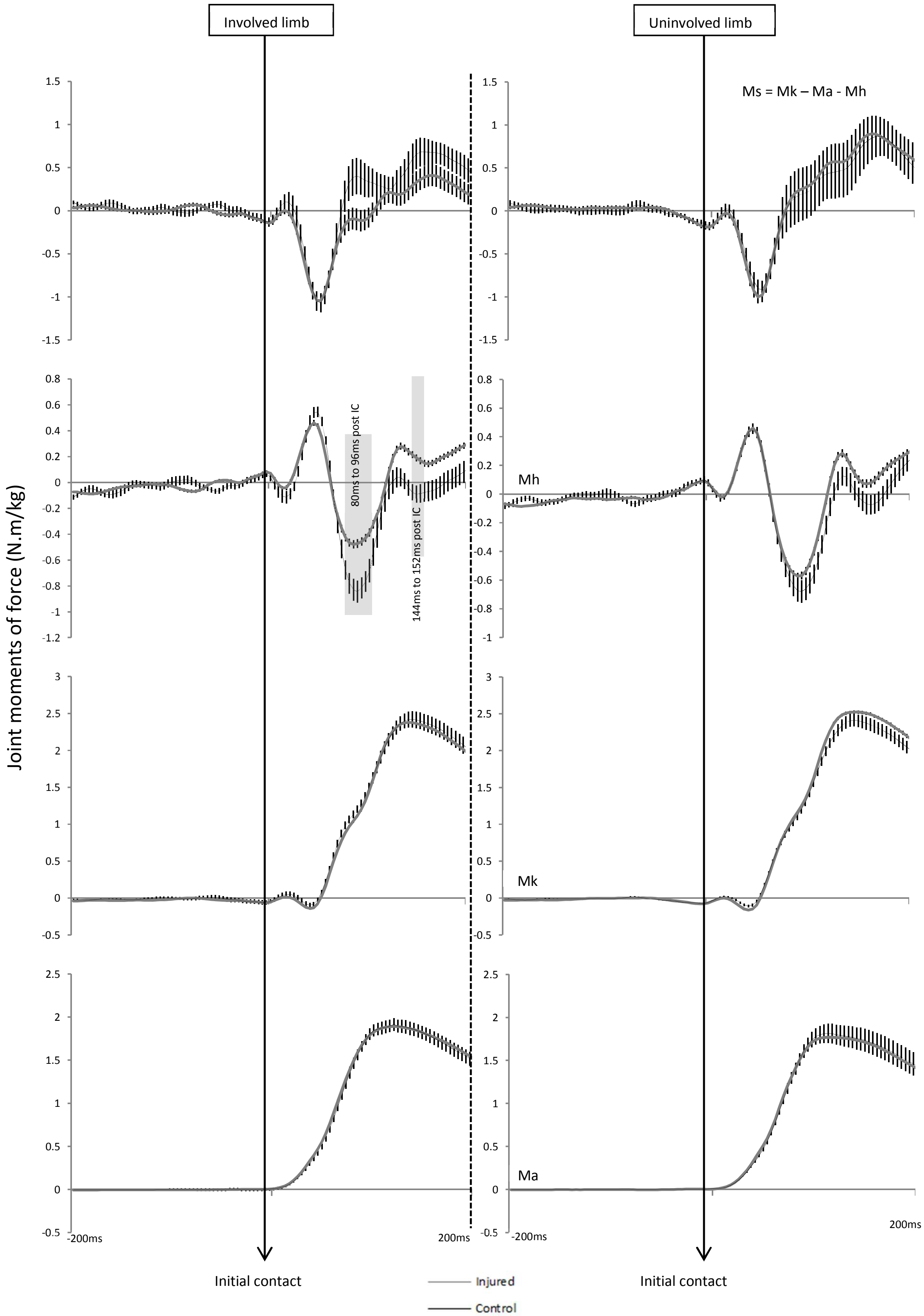

Figure 5. Joint stiffness values for the involved limb of the injured and control groups.

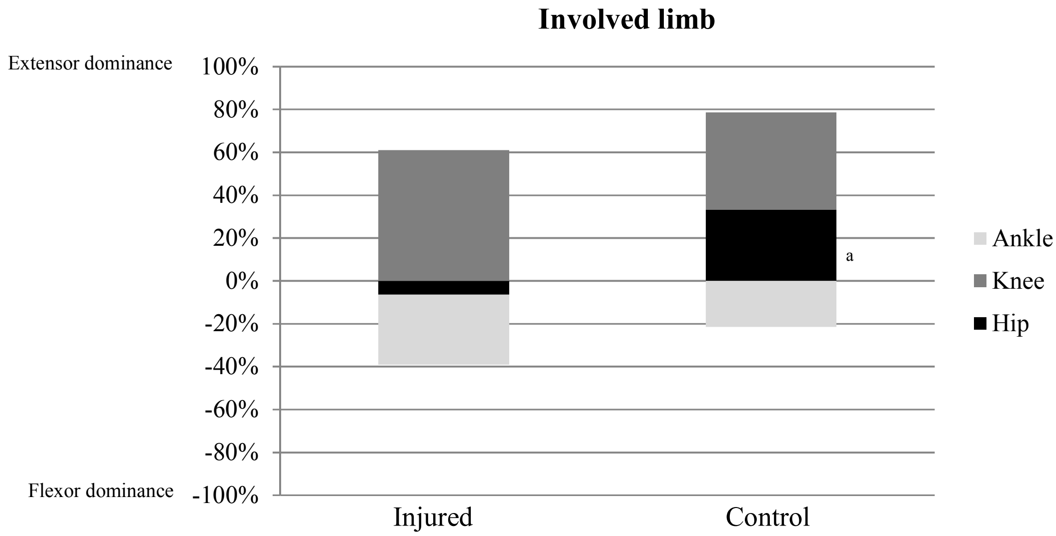

Figure 6. Joint stiffness values for the uninvolved limb of the injured and control groups.