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Restrictions in ankle dorsiflexion range of motion alter landing kinematics but not movement strategy when fatigued

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ABSTRACT

Context: Ankle dorsiflexion range of motion (DF ROM) has been associated with a number of kinematic and kinetic variables associated with landing performance that increase injury risk. However, whether exercise-induced fatigue exacerbates compensatory strategies has not yet been established.

Objectives: i) explore differences in landing performance between individuals with restricted and normal ankle DF ROM, and ii) identify the effect of fatigue on compensations in landing strategies for individuals with restricted and normal ankle DF ROM.

Design: Cross-sectional.

Setting: University research laboratory.

Patients or Other Participants: 12 recreational athletes with restricted ankle DF ROM (restricted group) and 12 recreational athletes with normal ankle DF ROM (normal group).

Main Outcome Measure(s): Participants performed five bilateral drop-landings, before and following a fatiguing protocol. Normalized peak vertical ground reaction force (vGRF), time to peak vGRF and loading rate were calculated, alongside sagittal plane initial contact angles, peak angles and joint displacement for the ankle, knee and hip. Frontal plane projection angles were also calculated.

Results: At baseline, the restricted group landed with significantly less knee flexion (*P* = 0.005, effect size [ES] = 1.27) at initial contact and reduced peak ankle dorsiflexion (*P* < 0.001, ES = 1.67), knee flexion (*P* < 0.001, ES = 2.18) and hip flexion (*P* = 0.033, ES = 0.93) angles. Sagittal plane joint displacement was also significantly less for the restricted group for the ankle ($P < 0.001$, ES = 1.78), knee ($P < 0.001$, ES = 1.78) and hip ($P = 0.028$, ES = 0.96) joints.

Conclusions: These findings suggest individuals with restricted ankle DF ROM adopt different landing strategies than those with normal ankle DF ROM. This is exacerbated when fatigued, although the functional consequences of fatigue on landing mechanics in individuals with ankle DF ROM restriction are unclear.

Keywords: joint mechanics, ankle restriction, drop-landings

INTRODUCTION

Peak vertical ground reaction forces (vGRF) > 8 times bodyweight have been reported during bilateral landings,[1] which has been identified as a causal factor for lower limb injuries.[2] To support dissipation of vGRF during landings, simultaneous flexion at the ankle, knee and hip joints following ground contact must occur.[3,4] Thus, movement strategies that assist in attenuating vGRF and enhancing sufficient load sharing across joint segments are advantageous for reducing injury risk. For example, sagittal plane ankle, knee and hip joint alignment at initial contact[5-7] and at peak knee flexion[4] influence the magnitude of peak vGRF during landings, while greater angular joint displacement for the ankle, knee and hip joint supports the load sharing of peak vGRF across each joint segment.[8] Adopting a movement strategy which keeps peak vGRF below an injury-provoking threshold reduces acute[9] and chronic[10] injury risk in the lower extremity.

The knee and hip joints have been identified as primary segments for shock absorption during bilateral drop-landings.[3] However, restrictions in ankle dorsiflexion range of motion (DF ROM) can negatively influence the coordination of the proximal segments during landings by imposing a mechanical organismic constraint that can limit an individual’s capacity to adopt effective movement strategies.[11-14] It is therefore possible that reduced ankle DF ROM contributes to the development of compensatory strategies throughout the lower extremity in an attempt to maintain peak vGRF below an intolerable threshold.[12] Consistent with this suggestion, several studies have reported no relationship between ankle mobility and landing forces.[12-14] However, ankle DF ROM measured using the weight-bearing lunge test (WBLT), is related to ankle dorsiflexion (r = -0.31 to -0.34) and knee flexion (r = -0.37 to -0.41) angles at initial contact during bilateral drop-landings from drop heights equating to 100% and

150% of countermovement jump (CMJ) height in recreational athletes.[12] In the same investigation, significant relationships were also found between ankle DF ROM and peak ankle dorsiflexion (*r* = -0.43 to -0.44), knee flexion (*r* = -0.42 to -0.52) and frontal plane projections angles (FPPA) (*r* = 0.37) at the moment of peak knee flexion during bilateral drop-landings. These findings suggest restrictions in ankle DF ROM cause a stiffer landing strategy through limiting knee flexion, necessitating compensations at initial ground contact and the moment of peak knee flexion to prevent excessive peak vGRF.

The kinematic and kinetic variables associated with landing performance can also be affected by exercise-induced fatigue (defined as the inability for the neuromuscular system to maintain mechanical work for a given task[15]), as it has been shown to increase injury risk.[16] This may occur during prolonged activities, such as repetitive jumping, which results in exercise-induced fatigue that reduces lower extremity force production.[17] To attenuate peak vGRF, altered movement strategies are required to compensate for diminished muscular force production. As such, ankle plantar flexion has acutely increased (mean difference = 10.6°) under fatigue whilst knee flexion angles have decreased (mean difference = 7.0°) at initial contact during bilateral drop-landings.[18] These alterations in coordination strategies help to prevent fatigue-induced increases in peak vGRF by increasing angular joint displacement for the ankle and knee joint.[8] Interestingly, such compensations are similar to those demonstrated at initial contact by individuals with restrictions in ankle DF ROM.[12] It may be that when in a fatigued state, individuals with limited ankle DF ROM are unable to alter joint alignment at initial contact as a strategy to manage peak vGRF due to the mobility restriction already requiring this compensation.

It is also feasible that reduced DF ROM limits degrees of movement freedom across key lower-limb segments at peak knee flexion during landings, which may control peak vGRF in a fatigued state. Madigan and Pidcoe[19] found that when participants were acutely fatigued, peak ankle dorsiflexion (mean difference = 4.5°) and knee flexion angles increased (mean difference = 6.7°) resulting in a 0.45 $N\cdot kg^{-1}$ reduction in peak vGRF during landings. Similarly, James, Scheuermann and Smith[20] detected increased angular joint displacement for the knee (mean difference = 7.9°) and a 22% decrease in peak vGRF during bilateral drop-landings after fatiguing exercise. Collectively, these studies show that when individuals are fatigued, attenuation of peak vGRF is achieved by increasing the vertical displacement of centre of mass. For individuals whose movement is constrained by a restriction in ankle DF ROM, this compensatory strategy may not be fully available and their ability to cope with the addition of fatigue may be compromised.

Therefore, the aims of this study were: i) to examine differences in landing performance between individuals with restricted and normal ankle DF ROM and ii) identify the effect of fatigue on the compensations in landing strategies for individuals with restrictions in ankle DF ROM. We hypothesized that: i) individuals with limitations in ankle DF ROM will present with detectable differences in landing mechanics, and ii) individuals with restricted ankle DF ROM would fail to adopt vGRF attenuation-related strategies demonstrated by individuals with sufficient ankle DF ROM, during landing in a fatigued state.

METHODS

Design

A mixed study design was employed in which participants were assigned to independent groups (based on ankle DF ROM) who all performed landing tasks in both a non-fatigued and fatigued state. Participants were classified as either having restricted ankle DF ROM (restricted group) or normal ankle DF ROM (normal group) according to performance on the overhead squat and forward arm squat tests.[21] This method was selected due to its ability to identify individuals with a functional restriction in ankle DF ROM, whilst producing a large disparity in ankle DF ROM values between groups.[21] Briefly, participants were required to complete the overhead squat test and forward arm squat test for six and three repetitions, respectively. Performance was graded in real-time by the lead investigator against the criteria rating outlined by Rabin and Kozol.[21] When participants were unable to perform a test using a movement strategy that corresponded with the criteria rating, participants were assigned a ‘fail’ for that test. Conversely, participants who performed a test with a movement strategy that matched the criteria rating, a ‘pass’ result was given for that test. Participants who passed the overhead squat and forward arm squat test, were invited to take part in a testing session and assigned to the normal group. Participants who failed both the overhead squat test and forward arm squat test were invited to participate in a testing session and assigned to the restricted group. Participants who failed the overhead squat test but passed the forward arm squat test were excluded from the investigation and did not attend a subsequent testing session.

After completing the tests for group allocation, participants attended a single-test session, where ankle DF ROM was measured for both limbs independently using the WBLT. Participants then performed three maximal CMJ to establish drop height for the bilateral drop-landings and the threshold for establishing the onset of fatigue. Five bilateral drop-landings were then completed from a drop height of 150% CMJ height, both before and after the performance of a fatiguing protocol. All participants were informed of the risks associated with the testing prior to completing a pre-exercise questionnaire and providing informed written consent. Ethical approval was provided by the Institutional Research Ethics Committee. All test sessions were conducted between 10:00-13:00 h to control for circadian variation.

Participants

Using the effect size of 0.47 presented by James, Scheuermann and Smith[20] for differences in knee joint displacement during landings following the performance of a fatigue protocol, we performed a representative analysis using G*power to determine the appropriate sample size. With an alpha of 0.05, calculations indicated that to achieve 80% statistical power, a minimum of eight participants per group were required to determine differences in landing mechanics following the fatigue protocol. All participants were required to meet the following inclusion criteria: (1) between the ages of 18-40 years; (2) no lower-extremity injury six-months prior to testing; (3) no history of lower-extremity surgery; (4) regularly compete/participate 1-3 times per week in sport events involving landings activities, such as court, racquet, or team sports.

Twenty-eight participants volunteered to take part in the experiment. Following the initial screening session using the criteria previously described, four participants were excluded from the analysis, with 12 participants assigned to the restricted group (6 males, 6 females; age = 21 ± 1 years, height = 173.4 ± 9.7 cm, body mass 72.4 ± 10.7 kg) and 12 participants to the normal group (6 males, 6 females; age = 23 ± 5 years, height = 170.0 ± 6.8 cm, body mass 63.7 ± 8.0 kg).

Procedures

Following the recording of height and body mass during the test session, participants performed the WBLT. Participants began the test by facing a bare wall, with the greater toe of the test leg positioned against the wall. The greater toe and the centre of the heel were aligned using a marked line on the ground, perpendicular to the wall. Participants were instructed to place the non-test foot behind them, with the heel raised and at a distance that they felt allowed them to maximise their performance on the test. In order to maintain balance, participants were asked to keep both hands firmly against the wall throughout. The participants were then instructed to slowly lunge forward by simultaneously flexing at the ankle, knee and hip on the test leg in an attempt to make contact between the centre of the patella and a vertical marked line on the wall, perpendicular to the line on the ground. Subtalar joint position was maintained by keeping the test foot in the standardized position and ensuring the patella accurately contacted the vertical line.[22] Any elevation of the heel during the test was regarded as a failed attempt and feedback was provided to the participants regarding their inability to prevent the heel from rising. Upon successful completion of an attempt, where contact between the patella and the wall was made with no change in heel position relative to the ground, participants were instructed to move the test foot further away from the wall by approximately 0.5 cm. No more than three attempts were allowed at any given distance. At the last successful attempt, the distance between the heel and the wall, and the distance between the base of the patella and the ground were recorded to the nearest 0.1 cm. To determine ankle DF ROM, the trigonometric calculation method (DF ROM = 90-arctan [knee-ground/heel-wall]) was employed for each attempt using the heel-wall and ground-knee distances.[23] This procedure was repeated three times for each limb. Intra-rater reliability for this procedure has previously been reported as excellent (intraclass coefficients

(ICC) = 0.98), with a standard error of measurement (SEM) as 0.6° being established.[23] To ascertain that inter-limb differences did not exist, an independent t-test was used to compare the mean of the three trials for left and right WBLT scores. Bland-Altman level of agreement analysis for inter-limb asymmetry were -0.2 ± 3.8° and 0.6 ± 4.7° for the restricted and normal group, respectively. Mean inter-limb differences were not significant ($P > 0.05$) and the right limb was used for data analysis.

Following a standardized warm-up, participants were then familiarized with the performance of a CMJ. For the CMJ, participants stood bare feet with a hip-width stance with their hands placed on their hips. Participants were then asked to rapidly descend prior to explosively jumping as high as possible, with no control being placed on the depth or duration of the countermovement. Jump height was measured using photoelectric cells (Optojump System, Microgate, Bolzano, Italy). Three maximal effort CMJs were performed, with 60 s recovery between attempts. The maximum value of the three attempts was used to calculate drop height for the bilateral drop-landings as well as to establish the onset of fatigue during the fatigue protocol.

Reflective markers were then placed directly onto the participants' skin by the same investigator using the anatomical locations for sagittal plane lower-extremity joint movements and FPPA, as outlined by Dingenen et al.[24] and Munro, Herrington and Carolan.[25] For sagittal plane views, reflective markers were placed on the right acromioclavicular joint, greater trochanter, lateral femoral condyle, lateral malleolus and 5th metatarsal head.[24] To establish FPPA for the knee joints, reflective markers were placed at the centre of the right knee joint (midpoint between the femoral condyles), centre of the right ankle joint (midpoint between the malleoli) and on the proximal thigh (midpoint between the anterior superior iliac spine and the knee marker). Midpoints for the knee and ankle were measured with a standard tape measure (Seca 201, Seca, United Kingdom), as described by Munro, Herrington and Carolan.[25]

Participants were then familiarized with the bilateral drop-landings from a drop height of 150% of maximum CMJ height. Bilateral drop-landings were performed with participants standing bare foot with their arms folded across their chest on a height-adjustable platform (to the nearest 1 cm). Participants were then instructed to step off the platform, leading with the right leg, before immediately bringing the left leg off and alongside the right leg prior to impact with the ground. During this manoeuvre, participants were instructed to ensure that they did not modify the height of the centre of mass prior to dropping from the platform.[4] For a landing to be deemed successful, participants were required to ensure they landed with each foot simultaneously and in complete contact with the respective portable force platform, which was positioned 15 cm away from the elevated platform. Each foot landed on a separate portable force platform recording at 1000 Hz (Pasco, Roseville, CA, USA), positioned side-by-side, 5 cm apart and embedded in custom-built wooden mounts that were level with the force platforms and did not allow any extraneous movement. Full contact with the force platform was visually monitored during landings throughout by the lead investigator, with landings being disregarded where participants failed to either make full contact with the platform or maintain balance (e.g. either taking a step or placing a hand on the ground to prevent falling) upon landing. To ensure participants displayed their natural landing strategy, no instructions were provided regarding heel contact with the ground during the landing phase of the movement and no feedback on landing performance was provided at any point during testing. All landings were performed barefoot so as to prevent any heel elevation associated with footwear from altering landing mechanics and weakening internal validity.[26] For each condition (baseline and post fatigue protocol), participants performed five bilateral-drop landings for data collection. Baseline testing allowed for 60 s recovery between landings, while post fatigue protocol no recovery was provided between landings beyond the time it took to ascend the height-adjustable platform.

For 2D video analysis, sagittal- and frontal plane joint movements were recorded using three standard digital video cameras sampling at 60 Hz (Panasonic HX-WA30) using the procedures outlined by Payton.[27] For sagittal plane joint movements, a camera was positioned 3.5 m from the centre of either force platform.[28] To record frontal plane kinematics, a camera was placed 3.5 m in front of the centre of the force platforms.[28] All cameras were placed on a tripod at a height of 0.6 m from the ground.

The fatiguing protocol consisted of participants performing 30 successive CMJs, while maintaining the same technique as described above. Participants were instructed to keep their hands on their hips and repeatedly jump as high as possible for 30 repetitions, while spending minimal time on the ground between repetitions. Verbal encouragement was provided to ensure participants demonstrated maximal effort throughout. Following the 30th repetition, participants rested 30 s before performing a maximal CMJ for testing purposes. Participants then repeated the protocol until a > 20% decline in CMJ jump height was demonstrated.[18] Once participants were unable to reach > 80% of their maximum CMJ height, five bilateral drop-landings were immediately performed using the procedures previously described, with no recovery between landings so as to maintain a fatigued state. The last maximal CMJs were recorded for data analysis, with the percentage of fatigue calculated as CMJ height post fatigue protocol divided by CMJ height pre fatigue protocol, multiplied by 100.[18]

Data analysis

Raw vGRF data were low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 50 Hz.[29] Peak vGRF data were calculated for each leg and normalized to body mass ($N\cdot kg^{-1}$). An independent *t*-test was performed between mean values of peak vGRF for the right and left leg for each participant, which revealed no difference between limbs ($t_{(46)} = 0.657$, $P = 0.515$). As such, peak vGRF, time to peak vGRF and loading rate were independently calculated for the right leg and used for data collection. Peak vGRF data were normalized to body mass and initial contact velocity ($N\cdot kg^{-1}\cdot m\cdot s^{-1}$). To normalize peak vGRF to drop height, initial contact velocity was calculated using the following equation[12]:

$$\text{Initial contact velocity } (m\cdot s^{-1}) = \sqrt{2g \cdot DH}$$

where g is the gravitational acceleration and DH is drop height. For time to peak vGRF to be determined, initial contact was identified as the point that vGRF exceeded 10 N.[30] Time to peak vGRF was then calculated as the time difference between initial contact and the time point where peak vGRF occurred. Loading rate was calculated as peak vGRF normalized to body mass divided by time to peak vGRF. Within-session reliability for kinetic measures associated with bilateral drop-landing performance from a drop height equating 150% of CMJ height has previously been reported as excellent (ICC ranging between 0.91 to 0.94), with normalized peak force, time to peak force and loading rate possessing SEM values of 0.23 $N \cdot kg^{-1}$, 0.004 s and 6.7 $N \cdot s^{-1}$, respectively.[31]

All video recordings were analysed with free downloadable software (Kinovea for Windows, Version 0.8.15). For sagittal plane joint movements, hip flexion, knee flexion and ankle dorsiflexion angles were calculated at initial contact and the point of peak knee flexion for the right limb. These angles were then used to calculate joint displacement for each joint by subtracting the peak flexion angle from the initial contact angle. Initial contact was defined as the frame prior to visual impact between the foot and the ground that led to visual deformation of the foot complex. Peak flexion was identified visually and defined as the frame where no more downward motion occurred at the hip, knee or ankle joints.[24] Hip flexion angle was calculated as the angle between the line formed between the acromioclavular joint and the greater trochanter and the line between the greater trochanter and the lateral femoral condyle. Knee flexion angle was calculated as the angle between the line formed between the greater trochanter and the lateral femoral condyle and the line between the lateral femoral condyle and the lateral malleolus. Ankle dorsiflexion angle was calculated as the angle between the line formed between the lateral femoral condyle and the lateral malleolus and the line between the lateral malleolus and the 5th metatarsal head. FPPA was determined for both sides at the deepest landing position, defined as the frame corresponding to peak knee flexion.[25] FPPA was calculated as the angle between the line formed between the proximal thigh marker and the knee joint marker and the line between the knee joint marker and the ankle joint marker.[25] For hip flexion, knee flexion and ankle dorsiflexion, smaller values represented greater flexion and ankle dorsiflexion. For FPPA, values < 180° represented knee valgus and values > 180° representing knee varus. Within-session reliability for kinematic measures of bilateral-drop landings from a drop height equating to 150% of CMJ height have been previously reported as very large to nearly perfect (ICC ranging between 0.87 to 0.94). SEM for lower extremity joint angles at initial contact and at peak flexion have been reported as ranging between 1.1° to 1.3° and 2.3° to 6.6°, respectively.[28]

Statistical Analyses

Descriptive statistics (means ± standard deviation) were calculated for each kinetic and kinematic variable. Normality was confirmed for all dependent variables using the Shapiro-Wilk test. Independent *t*-tests were employed to determine between group differences for WBLT scores, maximum CMJ height and percentage of fatigue for CMJ height following the fatigue protocol. To test our first hypothesis, between-group differences at baseline for landing performance were examined using an independent *t*-test for kinetic and kinematic measures. Effect sizes (Cohen's *d*) were calculated as the difference between the means divided by the pooled standard deviation for all baseline measures and interpreted using the following criteria: < 0.2, a trivial difference; 0.21–0.5, a small difference; 0.51-0.8, a moderate difference; > 0.81, a large difference.[32]

A one-way analysis of covariance (ANCOVA) was performed to test our second hypothesis for between-group differences for landing performance following the fatigue protocol. This statistical analysis was chosen so as to provide greater statistical power and reduce variability, while accounting for between-group differences at baseline caused by the procedures for group allocation.[33,34] Values for kinetic and kinematic variables associated with landing performance following the fatigue protocol were used as the dependent variable, with baseline (pre) values used as the covariate. The *a-priori* level of statistical significance was set at $P < 0.05$, with a Bonferroni correction applied *post-hoc* in order to reduce the likelihood of Type I errors. As statistical significance is not a contextual factor and its use as the sole measure of significance has been contested[35], we also present 95% confidence intervals and effect sizes for a more complete, quantifiable description of the size of the effect. To that end, partial eta squared (η^2) values were calculated to indicate the magnitude of group differences in landing mechanics following the fatigue protocol using the following criteria: 0.02, a small difference; 0.13, a medium difference; 0.26, a large difference.[32] All statistical tests were performed using SPSS® statistical software package (v.24; SPSS Inc., Chicago, IL, USA).

RESULTS

Between-group differences at baseline

There were a between-group difference for WBLT scores, with the normal group demonstrating greater ankle DF ROM ($t_{(22)} = -10.19$, $P < 0.001$). However, there were no between-group differences at baseline in CMJ height ($t_{(22)} = -1.96$, $P = 0.062$). Table 1 presents both groups' landing performance scores at baseline for WBLT performance, CMJ height, kinetic and kinematic measures, including effect sizes and associated 95% confidence intervals. There were no between-group differences for any kinetic measures associated with landings between groups at baseline.

At initial contact, the restricted group landed with less knee flexion ($t_{(22)} = 3.12$, $P = 0.005$) and greater ankle plantarflexion ($t_{(22)} = 1.64$, $P = 0.116$). At the moment of peak knee flexion for all joints in the sagittal plane, the restricted group displayed less ankle dorsiflexion ($t_{(22)} = 4.10$, $P < 0.001$), knee flexion ($t_{(22)} = 5.34$, $P < 0.001$) and hip flexion ($t_{(22)} = 2.28$, $P =$

0.033). Joint displacement for the ankle ($t_{(22)} = -4.35$, $P < 0.001$), knee ($t_{(22)} = -4.35$, $P < 0.001$) and hip ($t_{(22)} = -2.35$, $P = 0.028$) were also significantly less for the restricted group. Other between-group differences were small to trivial.

INSERT TABLE 1 HERE

Effects of fatigue

Figure 1 presents between-group differences for post-test kinematic measures of bilateral drop-landing performance. All participants achieved a > 20% reduction in CMJ height following the performance of the fatigue protocol (restricted group = 68.2 ± 9.8%; normal group = 71.0 ± 6.9%), with no difference between groups for scores of percentage of fatigue ($t_{(22)} = -0.99$, $P = 0.333$). There were no main effects of group on post-test normalized peak vGRF ($F_{(1,21)} = 0.59$, $P = 0.451$, $\eta^2 = 0.03$), time to peak vGRF ($F_{(1,21)} = 1.17$, $P = 0.291$, $\eta^2 = 0.05$) and loading rate ($F_{(1,21)} = 0.42$, $P = 0.523$, $\eta^2 = 0.02$). Furthermore, the ANCOVA revealed no effect of group on post-test ankle plantar flexion angle ($F_{(1,21)} = 0.03$, $P = 0.868$, $\eta^2 = 0.00$), knee flexion angle ($F_{(1,21)} = 0.00$, $P = 0.965$, $\eta^2 = 0.00$) or hip flexion angle ($F_{(1,21)} = 2.12$, $P = 0.160$, $\eta^2 = 0.09$) at initial contact. There was a main effect of group on peak flexion for ankle dorsiflexion ($F_{(1,21)} = 5.80$, $P = 0.025$, $\eta^2 = 0.22$). Changes from baseline showed that the restricted group displayed less ankle dorsiflexion (mean difference = 0.3˚) than the normal group (mean difference = 2.7˚) following the fatiguing protocol. There were no main effects of group on peak knee flexion angle ($F_{(1,21)} = 0.60$, $P = 0.809$, $\eta^2 = 0.00$), peak hip flexion angle ($F_{(1,21)} = 0.20$, $P = 0.661$, $\eta^2 = 0.01$) and FPPA ($F_{(1,21)} = 1.92$, $P = 0.180$, $\eta^2 = 0.08$). There was a main effect of group on ankle joint displacement following the fatiguing protocol ($F_{(1,21)} = 7.88$, $P = 0.011$, $\eta^2 = 0.27$). Pairwise comparisons revealed greater ankle joint displacement for the normal group (mean difference = 2.4°) relative to the restricted group (mean difference = 0.1°). There was no main effect of group on knee joint displacement ($F_{(1,21)} = 0.66$, $P = 0.427$, $\eta^2 = 0.03$) and hip joint displacement ($F_{(1,21)} = 0.37$, $P = 0.557$, $\eta^2 = 0.02$) post-test.

INSERT FIGURE 1 HERE

DISCUSSION

This study had two main aims; first we examined the kinetic and kinematic characteristics of landing technique among recreational athletes with either functional restrictions or no restrictions in ankle DF ROM. Secondly, we assessed the effects of acute fatigue on landing technique between these two groups. We hypothesized that the restricted group would show different landing strategies to the normal group. Further, we hypothesized that this would affect their ability to compensate for reduced force production capability whilst fatigued, resulting in greater disparities in landing mechanics between groups. Consistent with our first hypothesis, the results revealed that individuals with limited ankle DF ROM land with less knee flexion at initial contact and reduced ankle, knee and hip flexion at the moment of knee peak knee flexion. This resulted in the restricted group displaying significantly less ankle, knee and hip joint displacement relative to the normal group. However, despite these disparities in kinematic patterns, there were no differences in kinetic variables during landing. Furthermore, our findings show that recreational athletes with limited ankle DF ROM were incapable of utilizing greater ankle joint motion when landing in an exercise induced fatigued state, which was in contrast to the normal group. However, this movement compensation did not result in differences between groups for any other kinematic or kinetic variable analysed, meaning that the functional relevance of this finding is uncertain.

A primary finding of the current study was that participants with ankle DF ROM restriction modified their landing mechanics at initial contact and at peak flexion. This occurred throughout the lower extremity, resulting in significant differences for angular joint displacement at the ankle, knee and hip joints. Specifically, at initial contact, participants with restricted ankle DF ROM landed with 5.5° less knee flexion. This is consistent with the findings of others,[12,36] where relationships between ankle DF ROM and knee flexion angles at initial contact during single-leg ($r = 0.33$) and double-leg landings ($r = -0.31$) were reported. Collectively, these results suggest that individuals compensate for restrictions in ankle DF ROM (as measured using the WBLT) by landing with greater knee extension prior to contacting the ground. It is likely that this movement strategy occurs in an attempt to maintain knee joint displacement, as peak knee flexion angles are significantly reduced by restrictions in ankle DF ROM.[12,36] The majority of acute non-contact knee injuries occur close to the point of initial contact during landings.[37] Landing with greater knee extension at initial contact has been associated with increased tibia anterior shear forces;[6] a known mechanism for anterior cruciate ligament injury.[38] Therefore, reduced ankle DF ROM may expose the knee to greater shear forces during landings, with the potential to increase injury risk.

Compensations at initial contact for restricted ankle DF ROM did not occur at the ankle joint itself. This was an unexpected finding, given that moderate negative relationships have been reported between ankle DF ROM and ankle plantar flexion angles at initial contact ($r = -0.34$)

during bilateral drop-landings from 100% of CMJ height.[12] Increasing ankle plantar flexion at initial contact provides a functional strategy for managing vGRF,[7] resulting in preservation of ankle joint displacement.[8] However, the relationship between ankle DF ROM and ankle plantar flexion angle at initial contact is not always consistent. Dowling, McPherson and Paci[36] found no such relationship during single-leg drop landings, while Howe et al.[12] reported a non-significant relationship during bilateral drop-landings from drop heights equalling 150% of CMJ height. As the present investigation found no difference in ankle plantar flexion angles at initial contact between groups, we suggest that the ankle does not provide a means of movement compensation at this stage of the landings for those with restrictions in ankle DF ROM.

In the current study, ankle DF ROM restriction significantly reduced baseline measures of peak flexion angles and joint displacement for the ankle, knee and hip joints, with large effect sizes found between groups. This is consistent with previous studies, where ankle dorsiflexion and knee flexion angles at peak flexion, along with joint displacement for these segments, have each been related to WBLT performance among both healthy[12,36] and injured[30] populations. The current finding is, therefore, in keeping with the sagittal plane coupling observed between the ankle and knee joints, whereby dorsiflexion at the ankle complex facilitates flexion at the knee joint during landings.[3] This coordination pattern allows for greater shock absorption,[3] supporting the management of vGRF when loading is greater due to task constraints. Manipulating the demand of a bilateral drop-landing by increasing drop height from 0.32 m to 1.03 m was reported to increase ankle and knee joint displacement by 4.2˚ and 11.6˚, respectively.[4] Reduced peak knee flexion has been shown to increase peak vGRF,[4] quadriceps muscle activity[5] and frontal plane knee abduction moments.[39] Each of these variables has been associated with increased anterior cruciate ligament injury risk.[40] Therefore, limitations in ankle DF ROM may cause individuals to adopt landing strategies that could potentially cause knee ligament injury.

This is the first investigation, to our knowledge, that has shown restrictions in ankle DF ROM significantly reduces hip flexion angles at peak flexion and hip flexion joint displacement during bilateral landings in a healthy and athletic population. During both unilateral[36] and bilateral landings,[12] ankle DF ROM has a small relationship with hip flexion angles at the moment of peak flexion (r = -0.23 to -0.28). In the current study, we found that the restricted group had lower peak hip flexion angles, with a mean difference of 16.3º compared to the normal group. Furthermore, mean hip joint displacement was 14.7˚ less for the restricted group. The hip joint has been shown to provide an important contribution to the dissipation of forces during landing tasks,[3] with a vital role for managing vGRF when landing from higher drop heights.[4] As a result, restrictions in ankle DF ROM potentially limits the hip joint's capacity to contribute to vGRF attenuation during landings, particularly from greater drop heights.

We found no difference for kinetic measures of landing performance between the restricted and normal group. Studies exploring the relationship between ankle DF ROM and kinetic variables have been inconclusive. A number of studies have found no significant relationship for ankle DF ROM and peak vGRF, time to peak vGRF and loading rate.[12-14] However, Fong et al.[11] did identify a moderate negative relationship between ankle DF ROM and peak vGRF during a jump-landing task. It has been proposed that the frontal plane compensations in the lower extremity reported by Whitting et al.[14] and Malloy et al.[13] may provide a strategy that assists in preserving the descent of the centre of mass to allow for vGRF attenuation.[12]

However, the data reported here challenges this suggestion, with FPPA for both groups showing no significant difference. The present findings indicate kinetic variables associated with landing performance are unlikely to be regulated exclusively by angular joint displacement or postures at specific time points (i.e. peak flexion) in lower extremity. Peak vGRF has been negatively correlated with angular velocity for the knee (r = -0.60) and hip joint (r = -0.45) at initial contact during a stop-jump task.[41] Similarly, increased eccentric work performed by the knee and hip extensors[4] and increased muscular activity prior to initial contact[42] also contributes to energy dissipation and aids in the attenuation of peak vGRF. Therefore, variables such as knee and hip angular velocity at initial contact and the eccentric work performed by the knee extensors may compensate for the reduced lower extremity joint displacement caused by restrictions in ankle DF ROM, resulting in the management of peak vGRF during landings. These findings indicate that ankle DF ROM may alter the requirements during landings for lower extremity strength qualities, due to a limited capacity to flex the knee and hip joints following ground contact. However, this suggestion is speculative, with research required to establish whether restricted ankle DF ROM demands greater rates of force development to effectively manage peak vGRF during landings.

The second major aim of this study was to investigate the effect of exercise-induced fatigue on landing mechanics in individuals with restricted ankle DF ROM. In this regard, another primary finding was the difference found between groups in ankle joint coordination during landings after an acute bout of exercise-induced fatigue. We found moderate and large effects for post-intervention ankle joint angle at peak flexion and ankle joint displacement respectively. These findings suggest that the restricted group was unable to access additional ankle dorsiflexion when performing landings in a fatigued state (Figure 1). This was in contrast to the normal group, who increased peak ankle dorsiflexion by 2.7˚ and ankle joint displacement by 2.4˚ when acutely fatigued. However, no differences were found when comparing groups and the effect of fatigue for the knee or hip joints for any kinematic measure associated with landing performance. Furthermore, no differences between groups were identified for any kinetic variable analysed following the fatigue protocol. Whether such small differences in peak flexion angles and joint displacement at the ankle are functionally relevant is unknown. As both groups were still able to access greater joint displacement at the knee and hip during landings it seems that the additional ankle DF ROM used by the normal group played no role in facilitating motion at the proximal segments.

Another consideration is whether 2D video analysis is able to detect such differences in landing strategy. Howe et al.[28] investigated the reliability of using 2D video analysis for bilateral drop-landings from drop heights equating to 150% of maximum CMJ height and reported minimal detectable change values for ankle dorsiflexion angle at peak flexion and ankle joint displacement were 6.8˚ and 6.0˚, respectively. As differences for the normal group following fatigue protocol did not exceed these thresholds it may be that the change in joint kinematics for this group can be defined as ‘real’. Therefore, individuals with restrictions in ankle DF ROM are no more constrained in their ability to adjust their landing strategy when fatigued, than individuals with normal ankle mobility. These findings suggest the presence of ankle DF ROM hypomobility does not exponentially increase injury risk when performing landings in a fatigued state.

This study is not without potential limitations. Firstly, this investigation used 2D video analysis to measure kinematic variables at distinct time points during bilateral-landings. While three-dimensional motion capture is considered the gold standard, many practitioners do not have access to such equipment in practical environments. The technologies used in this study are readily accessible in clinical settings and, consequently, provide clear practical application. Additionally, all kinematic measures presented in this investigation have shown acceptable within-session reliability, with CV% ranging between 1.1–11.4%.[28] Intra-rater reliability has also been reported, with typical error values <1.5° for all measures.[12] Another limitation was that our investigation did not control for menstrual cycle status for female participants, which has been shown to affect joint laxity[43] and landing mechanics.[44] As a result, it is possible that the differences found in our investigation may have been influenced by the menstrual cycle, which should be controlled for in future research.

CONCLUSION

Individuals who have restricted ankle DF ROM based on their performance of closed-chain activities adopt different landing strategies compared to non-restricted controls. In particular, individuals with functional limitations in ankle DF ROM use less ankle motion relative to controls during bilateral drop-landing landings. This is further exaggerated with the addition of fatigue, although these differences must be interpreted with caution due to the sensitivity of 2D video analysis for detecting changes in landing kinematics. At the knee, individuals compensate for reduced peak knee flexion angles by landing in a more extended posture at initial contact, in an attempt to maintain knee angular joint displacement and limit peak vGRF to a manageable level. This is also the first investigation to demonstrate that restrictions in ankle DF ROM affect sagittal plane hip kinematics during bilateral landings, with reduced peak flexion angles and angular joint displacement at the hip. As restrictions in ankle DF ROM appear to promote landing strategies that are more extended and stiffer in nature, injury risk may be increased during landing tasks for individuals with limited ankle DF ROM.

REFERENCES

1. Yanci J, Camara J. Bilateral and unilateral vertical ground reaction forces and leg asymmetries in soccer players. *Biol Sport.* 2016;33:179-183.
2. Hewett TE, Myer GD, Ford KR, Heidt Jr RS, Colosimo AJ, McLean SG, Van den Bogert AJ, Paterno MV, Succop P. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *Am J Sports Med*. 2005;33:492-501.
3. Yeow C, Lee P, Goh J. Non-linear flexion relationships of the knee with the hip and ankle, and their relative postures during landing. *Knee*. 2011;18:323-328.
4. Zhang S, Bates B, Dufek J. Contributions of lower extremity joints to energy dissipation during landings. *Med Sci Sports Exerc.* 2000;32:812-819.
5. Blackburn J, Padua D. Sagittal-plane trunk position, landing forces, and quadriceps electromyographic activity. *J Athl Train*. 2009;44:174-179.
6. Chappell JD, Yu B, Kirkendall DT, Garrett WE. A comparison of knee kinetics between male and female recreational athletes in stop-jump tasks. *Am J Sports Med*. 2002;30:261-267.
7. Rowley M, Richards J. Increasing plantar flexion angle during landing reduces vertical ground reaction forces, loading rates and the hip’s contribution to support moment within participants. *J Sports Sci.* 2015;33:1922-1931.
8. Begalle R, Walsh M, McGrath M, Boling M, Blackburn J, Padua, D. Ankle dorsiflexion displacement during landing is associated with initial contact kinematics but not joint displacement. *J Appl Biomech*. 2015;31:205-210.
9. Hewett T, Myer G, Ford K. Anterior cruciate ligament injuries in female athletes: Part 1, mechanisms and risk factors. *Am J Sports Med.* 2006;34:299-311.

10. Dierks TA, Manal KT, Hamill J, Davis I. Lower extremity kinematics in runners with patellofemoral pain during a prolonged run. *Med Sci Sports Exerc.* 2011;43:693–700.
11. Fong C, Blackburn J, Norcross M, McGrath M, Padua D. Ankle-dorsiflexion range of motion and landing biomechanics. *J Athl Train*. 2011;46:5-10.
12. Howe LP, Bampouras TM, North J, Waldron M. Ankle dorsiflexion range of motion is associated with kinematic but not kinetic variables related to bilateral drop-landing performance at various drop heights. *Hum Mov Sci.* 2019;64:320-328.
13. Malloy P, Morgan A, Meinerz C, Geiser C, Kipp K. The association of dorsiflexion flexibility on knee kinematics and kinetics during a drop vertical jump in healthy female athletes. *Knee Surg Sports Traumatol Arthrosc*. 2015;23:3550-3555.
14. Whitting JW, Steele JR, McGhee DE, Munro BJ. Dorsiflexion capacity affects Achilles tendon loading during drop-landings. *Med Sci Sports Exerc.* 2011;43:706–713.
15. Fousekis K, Tsepis E, Vagenas, G. Intrinsic risk factors of noncontact ankle sprains in soccer: a prospective study on 100 professional players. *Am J Sports Med*. 2012;40:1842-1850.
16. Borotikar BS, Newcomer R, Koppes R, McLean SG. Combined effects of fatigue and decision making on female lower limb landing postures: central and peripheral contributions to ACL injury risk. *Clin Biomech*. 2008;23:81-92.
17. Zadpoor AA, Nikooyan AA. The effects of lower-extremity muscle fatigue on the vertical ground reaction force: a meta-analysis. *Proc Inst Mech Eng H.* 2012;226:579-588.
18. Weinhandl JT, Smith JD, Dugan EL. The effects of repetitive drop jumps on impact phase joint kinematics and kinetics. *J Appl Biomech.* 2011;27:108-115.

19. Madigan M, Pidcoe P. Changes in landing biomechanics during a fatiguing landing activity. *J Electromyogr Kinesiol*. 2003;13:491-498.

20. James C, Scheuermann B, Smith M. Effects of two neuromuscular fatigue protocols on landing performance. *J Electromyogr Kinesiol*. 2010;20:667-675.

21. Rabin A, Kozol, Z. Utility of the overhead squat and forward arm squat in screening for limited ankle dorsiflexion. *J Strength Cond Res*. 2017;31:1251-1258.

22. Dill KE, Begalle RL, Frank BS, Zinder SM, Padua DA. Altered knee and ankle kinematics during squatting in those with limited weight-bearing-lunge ankle-dorsiflexion range of motion. *J Athl Train.* 2014;49:723–732.

23. Howe LP, Bampouras TM, North JM, Waldron M. Within-session reliability for inter-limb asymmetries in ankle dorsiflexion range of motion during the weight-bearing lunge test. *Int J Sports Phys Ther.* 2020;15:64-73.

24. Dingenen B, Malfait B, Vanrenterghem J, Robinson M, Verschueren S, Staes F. Can two-dimensional measured peak sagittal plane excursions during drop vertical jumps help identify three-dimensional measured joint moments?. *Knee.* 2015;22:73-79.

25. Munro A, Herrington L, Carolan M. Reliability of 2-dimensional video assessment of frontal-plane dynamic knee valgus during common athletic screening tasks. *J Sport Rehabil*. 2012;21:7-11.

26. Lindenberg KM, Carcia CR. The influence of heel height on vertical ground reaction force during landing tasks in recreationally active and athletic collegiate females. *Int J Sports Phys Ther*. 2013;8:1-8.

27. Payton CJ. Motion analysis using video. In: C. J. Payton CJ, Bartlett RM, ed. *Biomechanical Evaluation of Movement in Sport and Exercise*. New York: Routledge; 2007;8-32.

28. Howe LP, Bampouras TM, North J, Waldron M. Reliability of two-dimensional measures associated with bilateral drop-landing performance. *Mov Sport Sciences*. Epub ahead of print. 2019.
29. Roewer BD, Ford KR, Myer GD, Hewett TE. The ‘impact’ of force filtering cut-off frequency on the peak knee abduction moment during landing: artefact or ‘artifiction’?. *Br J Sports Med*. 2008;48:464–468.
30. Hoch M, Farwell K, Gaven S, Weinhandl J. Weight-bearing dorsiflexion range of motion and landing biomechanics in individuals with chronic ankle instability. *J Athl Train.* 2015;50:833-839.
31. Howe LP, North JS, Waldron M, Bampouras TM. Reliability of independent kinetic variables and measures of inter-limb asymmetry associated with bilateral drop-landing performance. *Int J Phys Educ Fitness Sports*. 2018;*7*:32-47.
32. Cohen J. *Statistical power analysis for the behavioural sciences*. 2nd ed. Hillsdale, NJ: Lawrence Erlbaum Associates, Inc;1988.
33. de Boer MR, Waterlander WE, Kuijper LD, Steenhuis IH, Twisk JW. Testing for baseline differences in randomized controlled trials: an unhealthy research behavior that is hard to eradicate. *Int J Behav Nutr Phys Act*. 2015;12:4.
34. Zhang S, Paul J, Nantha-Aree M, Buckley N, Shahzad U, Cheng J, DeBeer J, Winemaker M, Wismer D, Punthakee D, Avram V. Empirical comparison of four baseline covariate adjustment methods in analysis of continuous outcomes in randomized controlled trials. *Clin Epidemiol.* 2014;6:227.
35. Hurlbert SH, Levine RA, Utts J. Coup de grâce for a tough old bull: “Statistically significant” expires. *Am Statistician*. 2019;73:352-357.

36. Dowling B, McPherson AL, Paci JM. Weightbearing ankle dorsiflexion range of motion and sagittal plane kinematics during single leg drop jump landing in healthy male athletes. *J Sports Med Phys Fitness.* 2018;58:867-874.

37. Krosshaug T, Nakamae A, Boden BP, Engebretsen L, Smith G, Slauterbeck JR, Hewett TE, Bahr R. Mechanisms of anterior cruciate ligament injury in basketball: video analysis of 39 cases. *Am J Sports Med*. 2007;35:359-367.

38. Boden BP, Sheehan FT, Torg JS, Hewett, TE. Noncontact anterior cruciate ligament injuries: mechanisms and risk factors. *J Am Acad Orthop Surg*. 2010;18:520–527.

39. Pollard C, Sigward S, Powers C. Limited hip and knee flexion during landing is associated with increased frontal plane knee motion and moments. *Clin Biomech*. 2010;25:142-146.

40. Renstrom P, Ljungqvist A, Arendt E, Beynnon B, Fukubayashi T, Garrett W, Georgoulis T, Hewett TE, Johnson R, Krosshaug T, Mandelbaum B. Non-contact ACL injuries in female athletes: an International Olympic Committee current concepts statement. *Br J Sports Med.* 2008;42:394-412.

41. Yu B, Lin C, Garrett W. Lower extremity biomechanics during the landing of a stop-jump task. *Clin Biomech*. 2006;21:297-305.

42. Devita P, Skelly WA. Effect of landing stiffness on joint kinetics and energetics in the lower extremity. *Med Sci Sports Exerc.* 1992;24:108-115.

43. Shultz SJ, Sander TC, Kirk SE, Perrin DH. Sex differences in knee joint laxity change across the female menstrual cycle. *J Sports Med Phys Fitness*. 2005;45:594-603.

44. Cesar GM, Pereira VS, Santiago PR, Benze BG, da Costa PH, Amorim CF, Serrão FV. Variations in dynamic knee valgus and gluteus medius onset timing in non-athletic females related to hormonal changes during the menstrual cycle. *Knee*. 2011;18:224-30.

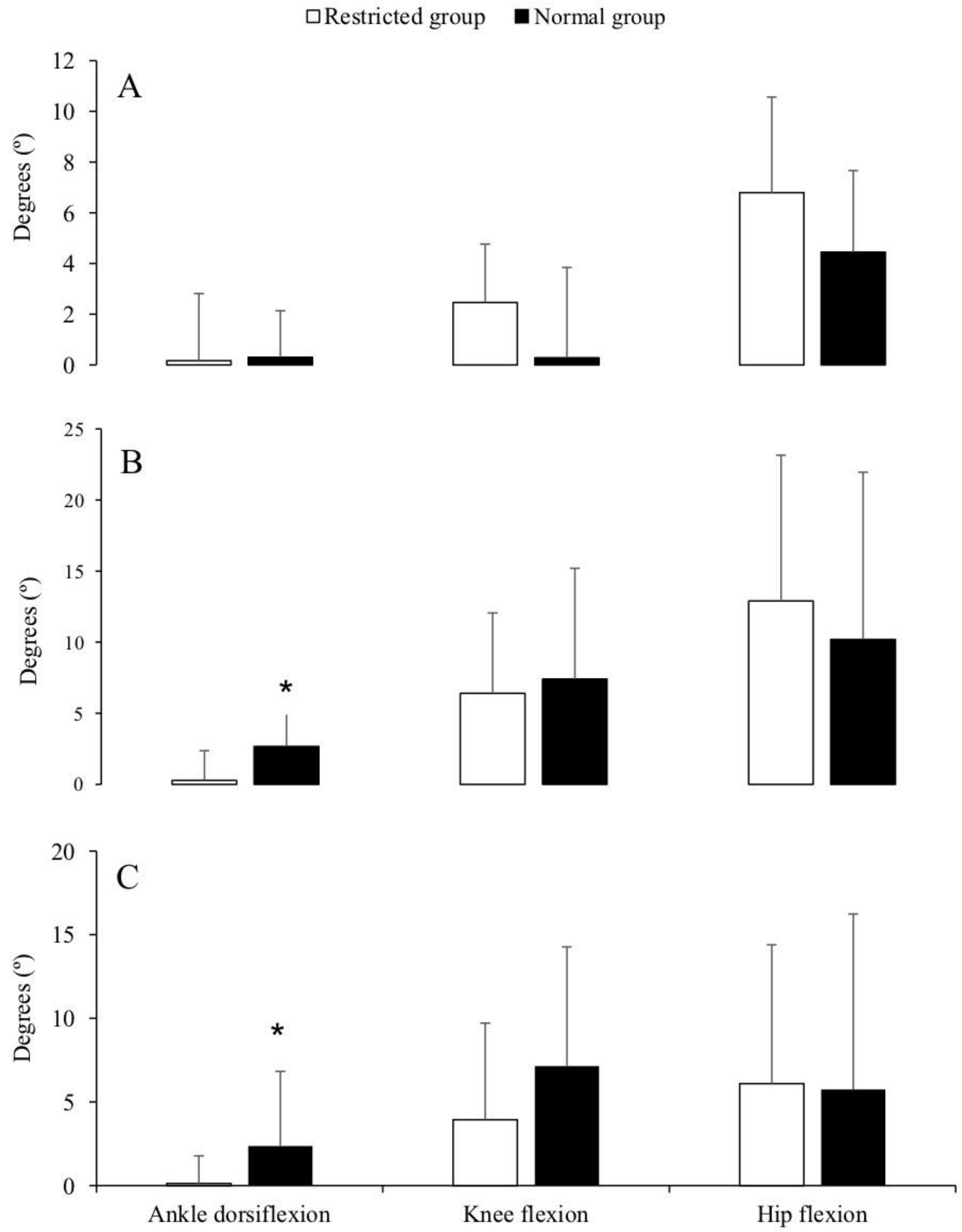

Figure 1. Group differences for kinematic measures of bilateral drop-landing performance following the fatigue protocol A) initial contact, B) peak flexion and C) sagittal plane joint displacement. Values represent differences from baseline testing. Means ± SD. * Between-group difference (*P* <0.05).

Table 1. Between-group differences at baseline for kinetic and kinematic measures associated with landing performance.

	Restricted (*n*=12)	Normal (*n*=12)	Mean difference (95% Confidence interval)	Effect size (95% Confidence interval)
	Mean ± SD	Mean ± SD		
Weight-bearing lunge test (°)	32.0 ± 3.3	44.6 ± 2.7	-12.6 (-15.1 – -10.0)*	4.2 (3.8 – 4.6)
Countermovement jump height (m)	0.30 ± 0.08	0.37 ± 0.10	-0.07 (-0.14 – 0.00)	0.8 (0.6 – 1.1)
Kinetic variables				
Peak force (N·kg^{-1}· m·s^{-1})	0.068 ± 0.021	0.064 ± 0.011	0.004 (-0.010 – 0.018)	0.2 (0.0 – 0.5)
Time to peak force (s)	0.058 ± 0.011	0.055 ± 0.010	0.003 (-0.005 – 0.012)	0.3 (0.1– 0.5)
Loading rate (N·s^{-1})	38.7 ± 21.3	38.0 ± 11.3	0.7 (-13.7 – 15.2)	0.0 (-0.2 – 0.4)
Initial contact angles				
Ankle (°)	153.1 ± 3.7	150.4 ± 4.8	2.9 (-0.8 – 6.5)	0.7 (0.4 – 0.9)
Knee (°)	170.2 ± 3.1	164.7 ± 5.3	5.5 (1.9 – 9.3)*	1.3 (1.0 – 1.5)
Hip (°)	161.8 ± 4.9	160.3 ± 5.8	1.6 (-3.0 – 6.1)	0.3 (0.1 – 0.5)
Peak flexion angles				
Ankle (°)	110.8 ± 7.6	96.8 ± 9.0	14.0 (6.9 – 21.1)*	1.7 (1.4 – 2.0)
Knee (°)	102.1 ± 6.4	79.2 ± 13.4	22.8 (13.8 – 31.9)*	2.2 (1.9 – 2.5)
Hip (°)	95.0 ± 17.1	78.7 ± 17.9	16.3 (1.5 – 31.1)*	0.9 (0.7 – 1.2)
Frontal plane projection angles (°)	200.0 ± 20.8	207.1 ± 19.2	-7.1 (-24.1 – 9.8)	0.4 (0.1 – 0.6)
Joint displacement				
Ankle dorsiflexion (°)	42.5 ± 5.9	53.6 ± 6.6	-11.1 (-16.4 – -5.8)*	1.8 (1.5 – 2.1)
Knee flexion (°)	68.2 ± 5.9	85.5 ± 12.8	-17.3 (-25.5 – -9.1)*	1.8 (1.5 – 2.1)
Hip flexion (°)	66.9 ± 14.0	81.6 ± 16.5	-14.7 (-27.7 – -1.7)*	1.0 (0.7 – 1.2)

* different between groups at the $P < 0.05$ level.