Physical Therapy in Sport 69 (2024) 22–32

Contents lists available at ScienceDirect

Physical Therapy in Sport

journal homepage: www.elsevier.com/ptsp

The effect of surface on knee landing mechanics and muscle activity during a single-leg landing task in recreationally active females☆

Mark C. Richardson [a,*], Paul Chesterton [a], Abigail Taylor [b], William Evans [b]

[a] Allied Health Professions Department, School of Health and Life Sciences, Teesside University, Middlesbrough, United Kingdom
[b] Faculty of Health Sciences and Wellbeing, Sunderland University, Sunderland, United Kingdom

ARTICLE INFO

Handling Editor: Dr L Herrington

Keywords:
Sand
Knee abduction moment
Frontal plane
Muscle activity

ABSTRACT

Objective: Investigate the effect of surface on frontal plane knee angle, knee moment and muscle activity.
Design: Randomised cross over.
Setting: University Laboratory.
Methods: Twenty females performed single-leg hop-landings onto sand, grass and firm surfaces. Kinematic, kinetic and muscle activity data were obtained. Compatibility curves were used to visualise parameter estimates alongside *P*- values, and S-value transforms.
Results: Knee angle for firm-sand (mean difference $\overline{(d)} = -2.2°$; 95% compatibility interval (CI): −4.6 to 0.28, p = 0.083, s = 3.6) and firm-grass ($\overline{d} = -1.9$; 95% CI: −4.3 to 0.5, p = 0.125, S = 3) yielded <4 bits of reputational information against the null hypothesis (H). 5 bits (p = 0.025) of information against H were observed for knee moment between firm-sand ($\overline{d} = 0.17$ N m/kg-1. m-1; 95% CI: 0.02 to 0.31) with similar effects for firm-grass ($\overline{d} = 0.14$ N m/kg-1. m-1; 95% CI: −0.02 to 0.29, p = 0.055, S = 4). Muscle activity across surfaces ranged from almost no (S = 1) reputational evidence against H (Quadriceps and Hamstrings) to 10–13 'bits' against H for lateral gastrocnemius (lower on sand).
Conclusions: Our study provides valuable information for practitioners of the observed effect sizes for lower-limb landing mechanics across surfaces in asymptomatic females.

1. Introduction

Anterior Cruciate Ligament (ACL) tear is a serious injury most prevalent in sports that include jump-landing (Agel et al., 2005). Approximately 70% of ACL tears are non-contact injuries (Boden et al., 2000), often involving a unilateral landing (Bisciotti et al., 2019) occurring ~ 50ms post-impact (Koga et al., 2010; Krosshaug et al., 2007). Female athletes appear to have increased susceptibility to ACL injury, with a 2–3 times increased risk compared to their male counterparts (Montalvo et al., 2019). Whilst anatomical differences and hormonal changes have been cited as potential causative mechanisms (Wojtys et al., 2002), an excessive valgus position of the knee upon landing is frequently proposed (Decker et al., 2003; Ford et al., 2003; Koga et al., 2010; Olsen et al., 2004). The increased (mal)alignment of the lower extremity on landing is associated with increased knee abduction moment (KAbM) (Miyamoto et al., 2023; Sigurðsson et al., 2021), predicting ACL injury risk with 73% specificity and 78% sensitivity (Hewett et al., 2005).

Effective muscle control and subsequent increases in co-ordination and stability are crucial for preventing excessive KAbM and protecting the ACL (Donnell-Fink et al., 2015; Hewett et al., 2005; Morgan et al., 2014). Females have been shown to activate the quadriceps more than males during the landing phase (Hughes & Dally, 2015), which may cause significant anterior tibial translation and subsequent ACL strain (DeMorat et al., 2004). Preferential activation of the lateral over medial quadriceps, hamstrings and gastrocnemius has also been noted in females (Palmieri-Smith et al., 2009; Landry et al., 2007) and may limit their ability to resist abduction loads, increasing KAbM, knee valgus and subsequent ACL injury risk (Letafatkar et al., 2015).

Interventions aiming to improve muscle control and reduce landing

☆ The research was conducted at Teesside University, Middlesbrough, UK.

* Corresponding author.
E-mail addresses: m.c.richardson@tees.ac.uk (M.C. Richardson), p.chesterton@tees.ac.uk (P. Chesterton), abbie.taylor@sunderland.ac.uk (A. Taylor), will.evans@sunderland.ac.uk (W. Evans).

https://doi.org/10.1016/j.ptsp.2024.07.001
Received 9 October 2023; Received in revised form 4 July 2024; Accepted 5 July 2024
Available online 8 July 2024

knee valgus and KAbM are integral to ACL injury prevention and rehabilitation programmes. Jump training programmes in isolation have been shown to be as effective at reducing landing knee valgus, and potential ACL injury risk, as those with additional balance and strength training components (Herrington, 2010; Petushek et al., 2019). However, these programmes are frequently performed on firm surfaces which exacerbate musculoskeletal loading, potentially increasing injury risk (Pereira et al., 2021). The use of sand as an alternative training surface has been advocated (Binnie et al., 2014; Rago et al., 2018), with jumping and plyometric tasks on sand during injury prevention and rehabilitation programmes highlighted as one of the main uses in professional sport (Richardson et al., 2022). The absorption qualities of sand reduce peak deceleration forces encountered upon impact with the training surface (Gaudino et al. 2013). A significant reduction in peak vertical ground reaction force (vGRF) has also been demonstrated during jumping tasks on a sand compared to firm surface (Giatsis et al., 2022) highlighting the potential for the use of sand in jumping programs aimed at both injury prevention and rehabilitation in the lower limb.

Empirical evidence for sand-based training has revealed a plethora of advantages when compared to a firm surface including; reductions in muscle soreness, exercise induced muscle damage, recovery time (Impellizzeri et al., 2008; Miyama & Nosaka, 2004; Singh et al. 2014), improved muscle control strategies (Pinnington et al., 2005; Panebianco et al., 2021; Sebastia-Amat et al., 2020; Sharma & Chaubey, 2013) and improvements in a range of firm ground performance measures (Mirzaei et al., 2014; Arazi et al., 2014; Ozen et al., 2020; Hammami et al., 2020).

The acute effects of surface on landing mechanics during jump-landing tasks hold the potential to advance our understanding, particularly in the context of applying this knowledge to jump-based training programs on sand surfaces. However, it is important to acknowledge that the existing evidence in this area is limited and requires further replication. Reductions in KAbM have been reported previously (Richardson, Murphy, et al., 2020) during single-leg jump tasks on sand compared to firm surfaces. Furthermore, decreased knee valgus in females specifically during a single-leg landing task on sand compared to firm has been demonstrated (Richardson, Wilkinson, et al., 2020). However, neither study included a comparison with an additional control surface (e.g., artificial grass) which might also be used during jump training in (p)rehabilitation scenarios (Impellizzeri et al., 2008) and could offer equally effective alternatives to sand. Further, the use of a detailed 6 degrees of freedom lower body model, to assess kinematic and kinetic responses on the different surfaces may overcome limitations noted for two-dimensional (2D) approaches (Richardson, Wilkinson, et al., 2020) and the limited (Plug-in-gait) marker-set used in the earlier study. Finally, identifying the magnitude of change in muscle activation of lower limb muscles would provide insight into the muscle activation strategies evoked when the lower limbs are challenged by various surfaces. Taking the limitations into consideration and sparsity of evidence; Further information is required to ascertain the magnitude and direction of effect of surface on lower body landing mechanics.

Therefore, the purpose of our study was to investigate the influence of landing surface on frontal plane knee angle (FPKA), KAbM, peak vGRF and muscle activity (hamstrings, quadriceps, gastrocnemius) during a single-leg hop (SLH) onto sand, pliable grass and firm ground. Our research hypothesis was that KAbM, FPKA (valgus) and vGRF would reduce, and muscle activation would increase when landing in sand compared to the other surfaces.

2. Materials and methods

2.1. Participants

Twenty-three females who participated in a minimum of 3 h of sporting activity per week and were involved in jump-related sports (e.g., basketball, soccer, volleyball, rugby) were recruited from a university population. Three females were excluded, two for previous ACL injury and one for a lower limb injury within the last six months. Subsequently, twenty participants (age: 23.4 ± 4.8 years; body mass: 65.3 ± 12.7 kg; height: 1.63 ± 0.08m) undertook testing on one occasion repeating all conditions in a randomised order. All participants had no history of ACL injury or other knee pathology, previous significant lower limb fracture or surgery, and were injury-free for six months prior to data collection. All participants provided written informed consent, with the study approved by the University's ethics committee (No. 035/19), in accordance with the Declaration of Helsinki.

2.2. Procedures

Prior to data collection, the participant's age (years), height (m), and mass (kg), were collected. To standardise jump-landings, participants were fitted with standard plimsoll shoes to control shoe-surface interface and were instructed to refrain from caffeine use at least 24h prior, and strenuous muscular exercise 48h prior to testing. A sub-maximal warm-up was performed which included 5-min on a stationary bike at a self-selected pace of 25% of the participants perceived maximum at a moderate resistance (5/10) (Walsh et al., 2012) followed by a supervised and standardised dynamic stretching protocol, consisting of the following exercises (light jog across 10 m, leg cross-overs, high knee-pull, high lunge-pull, high knees-to-chest, quadriceps pull, hip cradle, lunge with twist, reverse kick, high kicks/reach, spiderman, skip-hop, back pedal and high kicks), each performed for approximately 30 s (Avedesian et al., 2018).

Subsequently, participants performed five familiarisation trials of the SLH on each of the surfaces, followed by a series of five SLH's (from a 30 cm height) on firm, sand and pliable grass surfaces performed in a randomised order (generated via a computer randomiser). The SLH landing was chosen due to its test-retest reliability during 3D measurement of knee kinematics and kinetics (Myer, Bates, et al., 2015). Five successful trials were recorded for each participant on each surface, with the requirement for success being; 1) the landing was controlled whereby the participant kept their balance when landing on the dominant leg (defined as the leg used to kick a ball for maximum distance), ensuring the contralateral leg made no contact with the ground on landing, 2) the impact phase of the movement occurred on a precisely located force platform, and, 3) participants landed with a 'stop and hold' for several seconds. No instructions were given for their arm or hand position to mimic a natural SLH. Participants were instructed to hop forward and down onto the floor, sand, or grass from a 30 cm height. A predetermined floor marker 30 cm from the participants' starting position was used to standardise the landing position (Fig. 1). Trials where the subjects hopped in an upward direction prior landing were discarded and subsequently repeated.

The sand (particle size 0.02–0.2 mm) (Building Sand; Wickes, United Kingdom) was placed in a purpose-built pit allowing for lateral displacement of the sand and transmission of forces onto and from the force plate. Previous testing within our laboratory indicated that the participants mass recorded by the force plate was the same with and without the sandpit, and the centre of pressure also remained accurate with a sand covering, so inverse dynamics could be performed (Richardson et al., 2020a). The sand was at a depth of 10 cm and placed directly on top of the force plate (Kistler, Model 9281CA, Kistler Group, Winterthur, Switzerland). The stiffness of each surface was assessed via ten repetitions of a 4.5 kg clegg-hammer (SD instrumentation, Wiltshire, England) dropped from a height of 0.457m in line with previously used methods (Pinnington & Dawson, 2001; Binnie et al., 2013). This device measures the peak impact deceleration force exerted by the surface. The stiffnesses were; sand (368.5 ± 41.3N), grass (1182.5 ± 151.8N) and firm (1587 ± 191.4N). The sand was stored in the lab at a controlled ambient temperature of 20.5 °C.

When hopping onto the sand from a 30 cm height, a 40 cm box was used to account for the change in height. The pliable grass surface (Sherwood 30 mm, Artificial Grass Direct Ltd, UK) which was 3 cm in

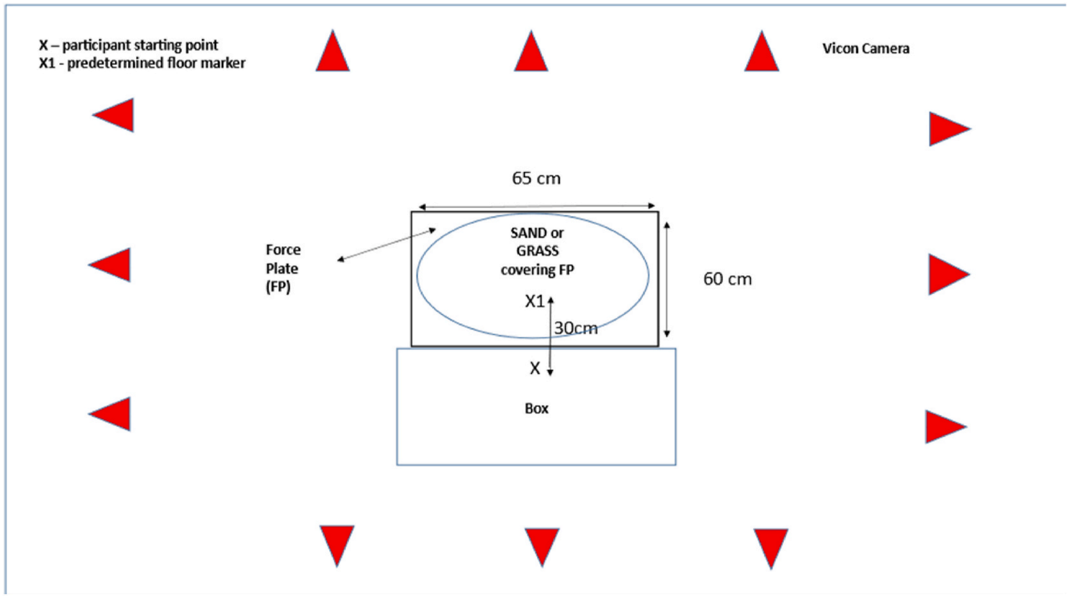

Fig. 1. An illustration of the experimental set up.

depth was placed directly over the force plate. When hopping onto the pliable grass surface from a 30 cm height a 33 cm high box was used to account for the change in height. For the SLH, participants were instructed to stand on their dominant leg and position their toes as close as possible to a predetermined marker at the edge of the box. After each landing on the sand surface, the sand was raked before the next jump to ensure an evenly distributed surface and consistent 10 cm depth. Participants were given 1-min rest between each repetition and a standardised 3-min rest between surface conditions to avoid the effects of fatigue. Following completion of 5 successful jumps on each surface participants were asked to perform a cool down consisting of 5-min on a stationary bike at a self-directed pace and 3 × 60 s static stretches of the following muscles: gluteus maximus, quadriceps, hamstrings, and gastrocnemii.

2.3. 3D motion capture and electromyography

Three-dimensional kinematics were captured with twelve Vicon MX cameras sampling at 200 Hz (Vicon Nexus v12, Vicon Motion Systems Ltd, Oxford, UK). These were synchronised with a Kistler force plate and Delsys electromyography (EMG) system (Delsys Trigno, Delsys Inc, Boston, USA) both sampling at 2000 Hz for kinetic and muscle activity data respectively. For each trial, the Vicon software sent a digital pulse to the Delsys and Kistler systems, beginning the data acquisition across all systems at the same time. The cameras were calibrated the morning of data capture following procedures outlined as per the user manual. Forty-four, 14 mm retro-reflective markers were attached to the overlaying skin of each participant to create a lower body 6-degrees of freedom model including pelvis, thigh, shank, and foot segments. Markers were placed over the following landmarks: anterior superior iliac spines, posterior superior iliac spines, iliac crests, greater trochanters, medial and lateral femoral condyles, medial and lateral malleoli, posterior calcanei, and the head of the 1st, 2nd and 5th metatarsals (Alahmari et al., 2020). Tracking markers were mounted on technical clusters on the thigh and shank. The same individual (primary author) placed the markers on all participants. A static trial was taken for each participant in a neutral (standing) position, with each foot on the force plate, following marker placement. Following skin preparation, surface EMG electrodes were placed on the overlaying skin of the vastus lateralis (VL), vastus medialis (VM), lateral hamstring (LH), medial hamstring (MH) and lateral gastrocnemius (LG) and secured with double-sided tape and additional micropore tape. We followed SENIAM guidelines described by Hermens et al. (1999) to standardise EMG electrode placement.

2.4. Data processing and reduction

Marker trajectories were processed in Vicon Nexus (version 2.2) and exported (via.C3D file) alongside the GRF data to Visual 3D (Version 5, C-motion, MD, USA) for subsequent calculation of inverse dynamics. Trajectory and GRF data were filtered using a zero-lag fourth order Butterworth filter (10 Hz) (Kristianslund et al., 2012; Ford et al., 2003). A biomechanical model was defined within Visual 3D to determine joint angles and moments. For the knee joint, we created a virtual shank segment to align precisely with the segment coordinate system for the thigh to set the knee angle to zero degrees in the standing trial effectively normalising the knee angle to the static pose. The knee moment was resolved into local coordinate system of the thigh segment (Mizner et al., 2012). In Visual3D, positive frontal (y) plane knee angles and moments reflect knee adduction (varus) in the right knee, and negative values reflect knee abduction (valgus) for angles and moments, this is reversed in the left knee. Two out of the twenty participants landed on their left leg. Therefore, to compare changes in knee adduction/abduction in our cohort, we reversed the sign so right and left knee data were comparable. External knee moments are described in this article.

Raw GRF and EMG signals alongside the modelled outputs (from Visual3D) were imported in MATLAB (MathWorks, 2021, version 2019b) and processed via a custom-designed programme. The vGRF data were low-pass filtered using a zero-lag second order (50 Hz) Butterworth filter (Brown et al., 2014) and used to determine point of impact. Initial contact (IC) was determined as the first point above 20 N using a descending for-loop from the peak vGRF. Fifty milliseconds post-landing (i.e., IC) was used to extract our outcome measures of interest, with non-contact ACL injuries often occurring at this approximate time point (Koga et al., 2010; Krosshaug et al., 2007). The EMG signal was high-pass filtered (20 Hz) to remove the DC offset followed by a linear envelope which included full wave rectification and a 2nd order low-pass (3 Hz) Butterworth filter (Winter, 2009). EMG signals were normalised to the peak EMG amplitude observed during the landing phase (from initial contact to peak knee flexion) for each participant, from all trials, across all conditions. This normalisation approach allowed for the comparison of muscle activity across different tasks

(Cronin et al., 2015) and has been shown to reduce inter-individual variation (Cronin et al., 2015).

2.5. Statistical analyses

We used a frequentist framework to analyse our data. Our selected response variables of interest were differences in discrete local optima for the frontal plane knee angle, knee moment, peak vGRF and EMG activity of lower limb muscles on landing from a SLH task on firm, grass and sand surfaces (pre-specified in ClinicalTrials.gov). We did not perform formal sample size (n) calculations *a priori*, our sample size justification is based upon resource limitations (Lakens, 2022). To reflect on the power of our study to detect a range of effect sizes and improve the informational value of our work we performed sensitivity power analysis (based on fixed n, and α-level = 0.05). (Analysis outlined in Supplementary material 1; Section 1).

Data were analysed in RStudio (version 2022.12.0.353). To answer our research question, differences in outcome measures were analysed using a random-intercepts model (mixed approach) via the lme4 package (Bates et al., 2015) which accounts for individual differences in the data. For our model *Surface* was specified as a fixed factor (categorical) and participant ID as the random effect allowing each participant's intercept to vary. Satterthwaite's approximation method was used for the calculation of t-tests and subsequent p-value via the lmerTest package. The behaviour of the residuals were checked visually and were acceptable for the assumptions of normality (via QQ plots), constant variance (residuals vs. predicted), and independence (auto-correlation of residuals). The random variance of participant ID, expressed as an SD (0.95 CI via parametric bootstrapping), is reported as supplementary material (Supplementary material 1; Section B) for the main outcomes to demonstrate the between-participant variance in the response variable.

We have approached our analysis cautiously aiming to provide a nuanced interpretation of our data and have attempted to replace '*confidence with compatibility and surprise*' (Rafi & Greenland, 2020). To avoid dichotomising our inferences, our observations are based on the magnitude and uncertainty of each outcome whether it is close to zero or not and reporting the observed p-value as a measure of the degree of compatibility between the target hypothesis and background model (Greenland, 2017) which encompasses all auxiliary assumptions. Thus, we make no reference to α-levels or decisions per se (Greenland, 2019). Our primary analysis involved using estimates from the model to obtain compatibility curves via the *concurve* package (Rafi & Vigotsky, 2020) for the estimated mean response between the reference surface (e.g., firm) and the other two surfaces. The compatibility curve contains horizontally stacked compatibility intervals (CI), estimated via the profile likelihood method, at every possible level (e.g., 95%, 90%, 75%) to visualise estimates consistent with model and all background assumptions used to compute the test statistic. Thus, we present relevant information for alternative hypotheses (non-nil effect sizes within the region of high compatibility for a given CI) as well as the nil-null (target) hypothesis (H). To preserve space, only curves for frontal plane knee angle and moment at 50ms are presented. We visualised the curves for all our outcomes to inform our inferences. The point estimate and 95% CI alongside the exact p-value (to 3 decimal places) and corresponding S-value transform is presented for all interactions. The S-values provide a measure of 'surprise' or amount of reputational information in the form of 'bits' or binary digits (e.g., p = 0.05 provides 4.3 bits of information) the P value supplies against our model, calculated by taking the negative base-2 logarithm of the P value (-log2(p)) (Rafi & Greenland, 2020).

3. Results

Descriptive statistics (mean and SD) for the dependent variables are displayed in Table 1, Ensembled average waveforms for knee angle (y), knee moment (y) and EMG between IC and 50 ms (loading phase) are presented to further aid interpretation (Fig. 2a and b). Model outcomes for comparisons of dependent variables between surfaces are displayed in Table 2.

Table 1
Mean ± SD for primary outcomes across the three surfaces (n = 20).

Surface	Firm	Sand	Grass
Outcome (at 50ms unless stated)			
Knee moment (y) (Nm·kg^{-1}.m^{-1})	−0.2 (0.2)	−0.1 (0.3)	−0.1 (0.3)
Knee angle (y) (deg)	3.4 (8.7)	1.5 (7.7)	1.2 (6.6)
Peak vGRF (N)	2791.1 (544.5)	1927.9 (270.0)	2866.7 (477.2)
vGRF(BW)	4.6 (1.2)	3.2 (0.7)	4.8 (1.0)
VM_EMG ratio (relative to peak during landing phase)	0.65 (0.22)	0.59 (0.25)	0.62 (0.22)
VL_EMG ratio (relative to peak during landing phase)	0.59 (0.22)	0.57 (0.27)	0.61 (0.23)
LG_EMG ratio (relative to peak during landing phase)	0.65 (0.20)	0.50 (0.13)	0.69 (0.21)
MH_EMG ratio (relative to peak during landing phase)	0.69 (0.19)	0.70 (0.19)	0.70 (0.17)
LH_EMG ratio (relative to peak during landing phase)	0.58 (0.23)	0.57 (0.20)	0.65 (0.26)

Abbreviations: vGRF, vertical ground reaction force; BW, body weights; EMG, electromyography; VM, vastus medialis; VL, vastus lateralis; LG, lateral gastrocnemius; MH, medial hamstrings; LH, lateral hamstring. For knee moment and angle negative values reflect abduction and positive values reflect adduction.

3.1. Frontal plane knee angle and moment

For FPKA, there was ≤4 bits of reputational information against the target hypothesis for firm vs. sand and firm vs grass. We observed almost no information (S = <1 bit) against the null hypothesis of no effect for grass vs. sand. Whilst zero (nil effect) was one of the possible effect sizes compatible with our background model data, the compatibility curves (Fig. 3a and b) shows negative effect sizes as large as −4.6° difference for sand, and −4.3° difference for grass are also compatible with the data, given the background model and contain the same information (4 bits) against the model as positive effect sizes of 0.5° and 0.3° which could be considered less practically important (Table 2). For knee abduction moment, we observed 5 bits of reputational information against the target hypothesis for firm vs. sand which would be as surprising as seeing five heads in a row conditioned on a fair coin toss. Similar effect sizes were observed in the region of high compatibility for firm-grass (Fig. 4a and b).

3.2. EMG

Almost no reputational information (S-value ≤1 for nearly all comparisons) against the target hypothesis was observed for VM, VL, MH and LH (Table 2) for firm vs. sand vs. grass. However mean muscle activity of the LG muscle was 15%–19% lower on sand compared to firm and grass surfaces which constituted 10–13 bits of reputational information against the target hypothesis (Table 2).

3.3. vGRF

The comparison of vGRF on firm vs sand (MD = 862.1N; 95% CI: 715.2 to 1011, p < 0.00001, S = 52) and grass vs. sand (MD = 938.7N; 95% CI: 790.8 to 1086.6, p < 0.00001, S = 49) yielded substantial reputational evidence against the target hypothesis whilst only 2 bits of information against H were observed for firm vs. grass (see Table 2 for body weight adjusted).

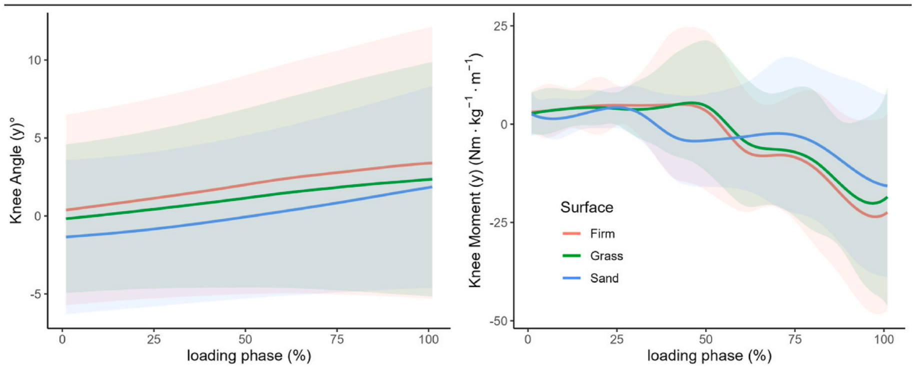

Fig. 2a. Ensembled average waveforms for knee angle (y), knee moment (y) between IC and 50 ms (loading phase).

4. Discussion

The aim of our study was to investigate the influence of different surfaces (sand, grass and firm) on frontal plane knee angle, KAbM, vGRF and muscle activity (hamstrings, quadriceps, gastrocnemius) upon landing from a SLH in females. The main findings were, 1) Comparisons between surfaces for knee angle yielded ≤4 bits of information against the target hypothesis. Though effects as large as −4.6 and −4.3 were observed in the 95% compatibility interval. 2) Effect sizes in line with a reduction in KAbM (~0.3Nm·kg^{-1}. m^{-1}) on sand and grass compared to firm were more compatible and less surprising as seeing no effect, with approximately 5 bits of reputational information against the target hypothesis observed. 3) Almost no reputational information against the target hypothesis in muscle activation across the surfaces for VM, VL, MH, LH was observed though CI were wide and more data is required. LG activation on the sand surface was reduced compared to grass and firm and yielded substantial information against the target hypothesis, and 4) Substantial reputational information against the target hypothesis for comparisons in peak vGRF for sand compared to grass and firm were observed. Effect sizes (alternative hypothesis) in favour of lower peak vGRF on sand constituted to ≤ 4.3 bits of reputational information against them.

To the authors' knowledge, this is the first study to provide detailed evidence for the magnitude of change and associated levels of compatibility in frontal plane angle, KAbM, vGRF, and lower limb muscle activation during a SLH landing on a firm, sand and grass surface for female (recreationally trained) athletes. As such, there is limited evidence with which to directly compare findings. Our cohort generally landed with healthy landing patterns and could be considered 'good landers' with 85% of participants landing below the knee abduction angle (KAA) thresholds (4.6°–6.3°) for ACL injury risk set previously for females (Bates et al., 2020; Hewett et al., 2005) (Fig. 2a), and the mean frontal plane knee angle across surfaces reflect knee varus. Thus, any observable adjustments in knee landing mechanics because of surface might be smaller with our current cohort. This is inconsistent with previous work which demonstrated reductions in knee valgus (~5°) when landing on a sand compared to firm surface, with females landing in ~12° of valgus on a firm surface, and thus could be considered an 'at risk' population (Richardson, Wilkinson, et al., 2020) with greater scope to improve. It should be noted that we used 3D motion capture, whereas the aforementioned authors calculated the frontal plane projection angle via a 2D approach which is susceptible to perspective and parallax errors. As such, the two approaches only show moderate correlation and are not interchangeable (Alahmari et al., 2020), making direct comparison difficult. The compatibility curve (Fig. 3b) shows the range of compatible effect sizes for firm vs. sand and thus effect sizes where $p > 0.05$. Whilst zero is one of the plausible effect sizes, an effect size of ~ −4.6 (counter-null) would have the same p value as no effect and shares the same bits of information against the statistical model. It is interesting to note that the range of effect sizes we observed are smaller than previous work (Richardson, Wilkinson, et al., 2020) and point in the opposite direction. It is not clear whether this means that sand (or grass) might have exacerbated the amount of valgus an individual landed in with our cohort, or allows individuals who land in adduction, to land closer to the midline when landing on these surfaces. It is feasible to suggest that the range of effect sizes we observed in our cohort may not be practically meaningful and therefore the landing patterns between surfaces, when the individuals are generally 'good landers', are practically equivalent to each other, though at this time there is not consensus on the smallest effect size of interest. We suggest researchers should now compare surfaces on participants who land consistently with KAA >5° (Bates et al., 2020; Hewett et al., 2005) on firm surfaces, or compare landing mechanics across surfaces between 'good' and 'poor' landers.

Knee abduction moments were observed (Table 1, Fig. 2a) at 50ms post impact across surfaces. Frontal plane moments during landing have been demonstrably associated with ACL injury/injury risk in athletes (Boden et al., 2000; Hewett et al., 2005; Krosshaug et al., 2016; Myer, Bates, et al., 2015). Higher loads are likely to have greater influence on ligament strain, which is supported by cadaveric impact simulations, with only the highest risk and rupture loading profiles demonstrating increases in ACL strain (Bates et al., 2019a; Bates, Schilaty, Nagelli, et al., 2019). Myer et al. (2011) reported that female athletes who exceeded a threshold of 25.3 N m during landing had a 6.8% risk for subsequent ACL injury compared with a 0.4% risk in athletes who were below this threshold. Richardson, Murphy, et al. (2020) previously reported KAbM values of 17.3 (5.9) N.m during the landing of a SLH task from a 30 cm height in a predominantly female cohort, which reduced to 14.8 (5.2) N.m (or by approximately 0.04 Nm· kg^{-1}. m^{-1}) when landing on a sand surface compared to firm. We also observed effect sizes in line with a reduction in KAbM (~0.3Nm·kg^{-1}. m^{-1}) on sand and additionally grass compared to firm, which were more compatible and less surprising as seeing no effect, with approximately 5 bits of reputational information against the target hypothesis observed (Table 2). It is possible that the observed changes in KAbM, despite landing in knee varus, were a result of differences at the foot/ankle (Dempsey et al., 2012; Tait et al., 2022; Teng et al., 2017), hip (Pollard et al., 2010), and trunk (Shimokochi et al., 2013; Chijimatsu et al., 2020; Taniguchi et al., 2022). Furthermore, when landing on a soft compared to firm surface, the compliance of the surface allows the body to sink into it resulting in lower vGRF (Gaudino et al., 2013; Giatsis et al., 2022), which was evident in our study for sand specifically (Tables 1 and 2) and might contribute to a decreased KAbM. Given the importance placed on reducing KAbM during landing to help prevent ACL injuries (Sugimoto et al., 2015), practitioners may wish to consider the use of sand or grass surfaces when planning ACL injury prevention or rehabilitation programs with females, which involve a single leg jump landing component. Whilst the reduced KabM on sand and grass may have the potential to reduce ACL injury risk, it may also enable an accelerated rehabilitation

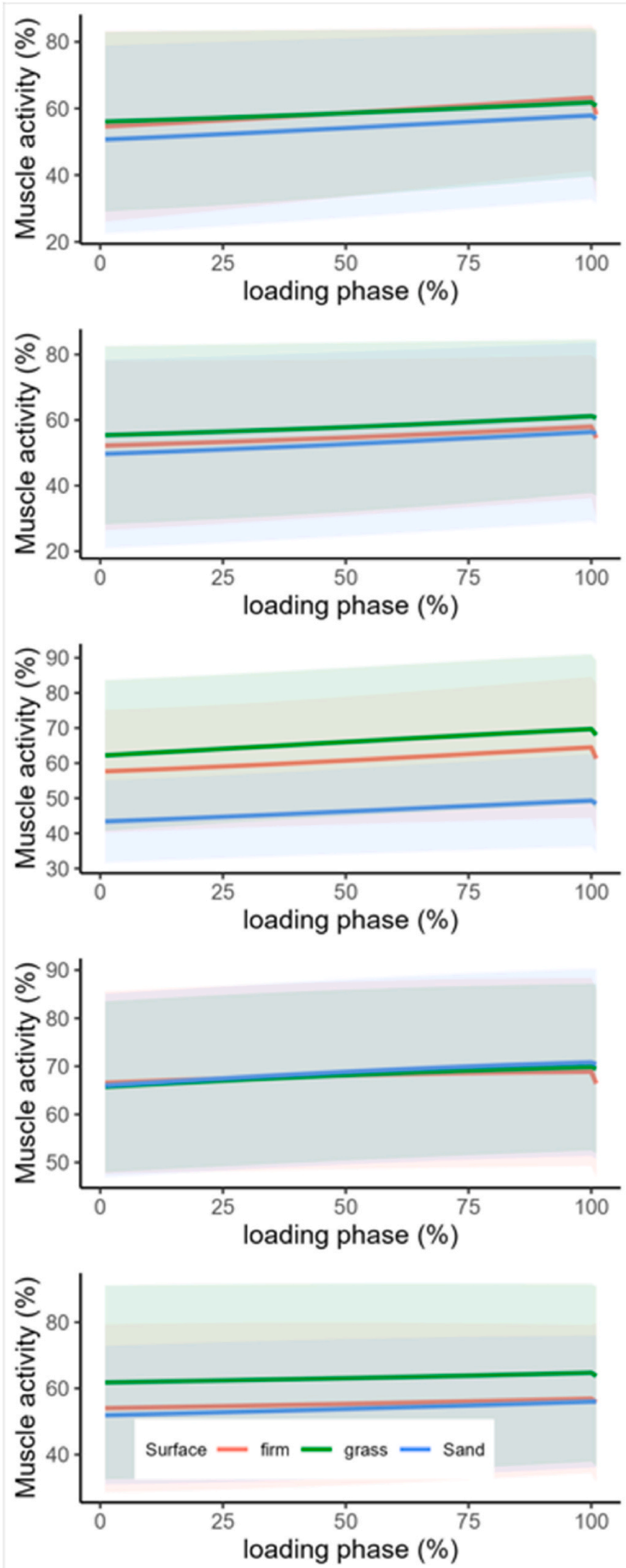

Fig. 2b. Ensembled average waveforms for EMG (VM, VL, LG, MH, LH – top to bottom) between IC and 50 ms (loading phase).

Table 2
Between surface differences for selected response variables.

	Firm-Sand	Firm-Grass	Grass-Sand
Outcome (at 50ms unless stated)			
Knee angle (y) (deg)	−2.2 (−4.6 to 0.28)	−1.9; (−4.3 to 0.5)	−0.26; (−2.17 to 2.70)
	P = 0.083	P = 0.125	P = 0.829
	S = 4	S = 3	S = 0.3
Knee moment (y) (N.m/ $kg^{-1}.m^{-1}$)	0.17 (0.02–0.31),	0.14 (−0.02 to 0.29)	−0.03; (−0.17 to 0.12)
	P = 0.025	P = 0.055	P = 0.724
	S = 5	S = 4	S = 0.5
Peak vGRF (N)	862.1 (715.2–1011.0)	−75.6 (−223.5 to 72.3)	938.7 (790.8–1086.6)
	P = <0.00001	P = 0.311	P = <0.00001
	S = 52	S = 2	S = 49
vGRF (BWs)	1.4 (1.7–1.2)	−0.1 (-0.4 to 0.1)	1.6 (1.3–1.8)
	P = <0.00001	P = 0.378	P = <0.00001
	S = 42	S = 2	S = 45
VM_EMG ratio (relative to peak during landing phase)	0.06 (−0.02 to 0.14)	0.03 (−0.05 to 0.10)	0.03 (−0.04 to 0.11)
	P = 0.118	P = 0.487	P = 0.375
	S = 3	S = 1	S = 1
VL_EMG ratio (relative to peak during landing phase)	0.02 (−0.08 to 0.12)	−0.02 (−0.11 to 0.08)	0.03 (−0.06 to 0.13)
	P = 0.609	P = 0.755	P = 0.478
	S = 1	S = 0.4	S = 1
LG_EMG ratio (relative to peak during landing phase)	0.15 (0.07–0.24)	−0.03 (−0.12 to 0.05)	0.19 (0.10–0.27)
	P = 0.001	P = 0.437	P = 0.0001
	S = 10	S = 1	S = 13
MH_EMG ratio (relative to peak during landing phase)	−0.01 (−0.08 to 0.06)	−0.01 (-0.08 to 0.07)	1.0 (−0.07 to 0.07)
	P = 0.796	P = 0.853	P = 0.941
	S = 0.4	S = 0.1	S = 0.1
LH_EMG ratio (relative to peak during landing phase)	0.02 (−0.07 to 0.10)	−0.07 (−0.15 to 0.01)	0.09 (0.00–0.20)
	P = 0.697	P = 0.097	P = 0.043
	S = 0.5	S = 3	S = 5

Abbreviations: vGRF, vertical ground reaction force; BW, body weights; EMG, electromyography; VM, vastus medialis; VL, vastus lateralis; LG, lateral gastrocnemius; MH, medial hamstrings; LH, lateral hamstring. For knee moment and angle negative values reflect abduction and positive values reflect adduction.

program because jumping activities could potentially be implemented more safely at an earlier stage in the process. However, this is speculative, and the magnitude of reduction in KAbM that is practically meaningful is currently unknown and thus our data should be interpreted cautiously. To develop our preliminary findings, future research should look to establish whether jump training on these surfaces provides the stimulus needed to reduce KAbM on landing during firm ground performance, and whether the differences exceed minimally detectable or clinically important differences.

Minimal differences were noted in the normalised sEMG activity of the VM, VL, MH, LH muscles at 50 ms post-landing across the firm, grass and sand surfaces, with almost no reputational information (less than one heads) against the target hypothesis (Table 2). Larger differences (and more precision) in the EMG activity of the LG muscle (~15–19%) were found and were lower on the sand compared to both the firm and grass surface (Table 2). The gastrocnemius may play an elevated role in supporting the knee during single-leg landing (Nyland et al., 2010) working in conjunction with or replacing the forces required by the hamstrings to counter quadriceps force and stabilise the knee through joint compression, reducing subsequent ACL strain and valgus loading (Morgan et al., 2014). The substantial decrease in vGRF noted on sand compared to firm and grass, may indicate a lower requirement to absorb

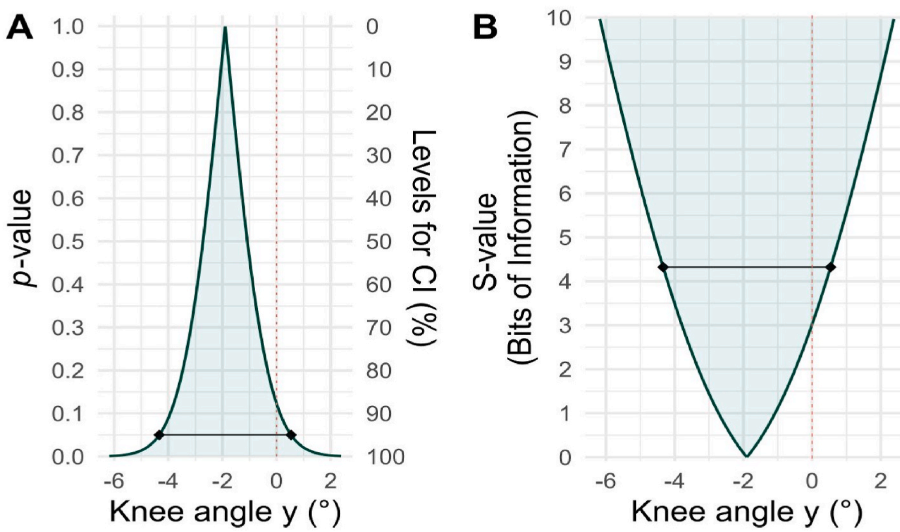

Fig. 3a. (A, B), shows the p value function, and corresponding S- value for frontal plane angle (firm-grass). The red line denotes the null (zero) hypothesis. The horizontal black line provides the effect sizes for p = 0.05 (95% CI). (For interpretation of the references to colour in this figure legend, the reader is referred to the Web version of this article.)

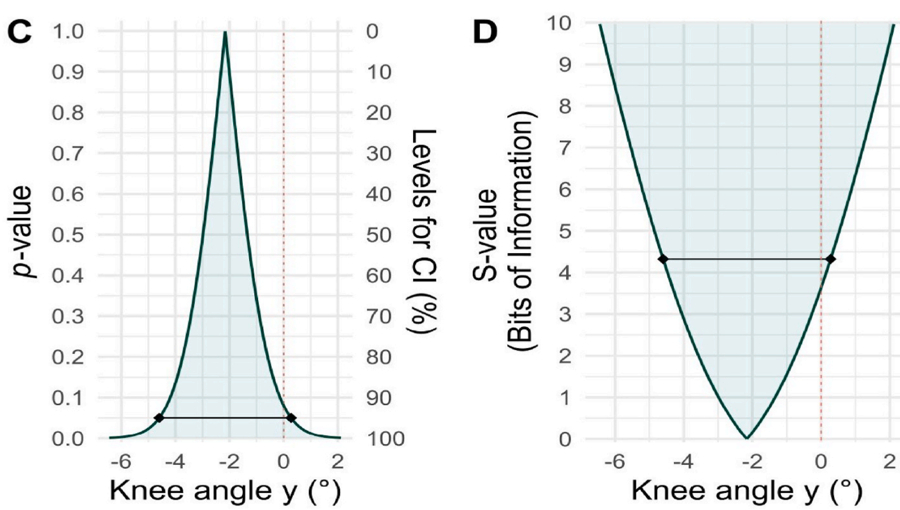

Fig. 3b. (C, D) shows the p value function, and corresponding S-value for frontal plane angle (firm-sand). The red line denotes the null (zero) hypothesis. The horizontal black line provides the effect sizes for p = 0.05 (95% CI). (For interpretation of the references to colour in this figure legend, the reader is referred to the Web version of this article.)

these forces and stabilise the knee on this surface. However, this alone does not explain the observed range of effect sizes for other lower limb muscles when landing on sand compared to firm and grass surfaces. Kinematic changes at the ankle in addition to the knee on landing may also be responsible for the reductions noted in LG on the sand surface and warrants future investigation.

Previous authors (Panebianco et al. 2021; Sebastia-Amat et al., 2020; Smith, 2006) have suggested that the deformation of sand increases the requirements for dynamic stability upon contact with the surface, compared to firm ground, and has been demonstrated in walking, running and balance tasks. During drop jump tasks, Peng et al. (2023) demonstrated increases in lower limb muscle activation (quadriceps, hamstrings and gastrocnemius) 30–60ms post landing on sand compared to firm surfaces, across a range of drop heights (30–60 cm). It is proposed that the increased muscle activation patterns noted on sand help an individual cope with the unstable nature of the surface and could improve stability if repeatedly exposed to the surface (Pinnington et al., 2005; Rafols Parellada et al., 2020). However, the nature of the SLH task differs to running and change of direction tasks, where a backward movement of the foot due to the deformation of sand may require further stabilisation to prevent slippage (Gaudino et al., 2013) compared to a 'land and hold' action during the SLH. Further, a lower degree of muscle activation has also been reported for drop-landing compared to countermovement jump and drop-jump tasks (Ambegaonkar et al., 2011; Arianasab et al., 2017) where stability is likely required in the eccentric phase of the task in preparation for an immediate explosive forward (and upward) movement. Therefore, the 'land and hold' nature of the SLH with the addition of the shock absorption qualities of sand (reduced vGRF) may reduce the perturbations to the postural system, and any observable difference in EMG between surfaces may be trivial, as we observed. The reduced mechanical demand as surface stiffness decreases has been noted previously in single-leg landing tasks (Hollville et al., 2020). From a practical standpoint, if the mechanical demands on the musculoskeletal system are reduced in jump-landing compared to countermovement jump and drop-jump tasks on sand, this may limit muscular adaptations following training interventions that involve only jump-land activities on this surface. Furthermore, the greatest differences in terms of muscle activation during running and change-of-

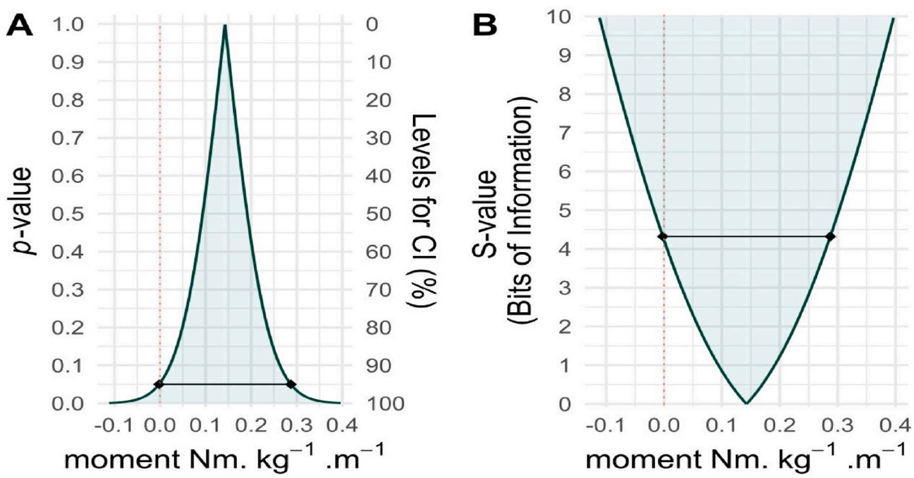

Fig. 4a. (A, B) shows the p value function, and corresponding S- value for Knee Abduction Moment (firm-grass). The red line denotes the null (zero) hypothesis. The horizontal black line provides the effect sizes for p = 0.05 (95% CI). (For interpretation of the references to colour in this figure legend, the reader is referred to the Web version of this article.)

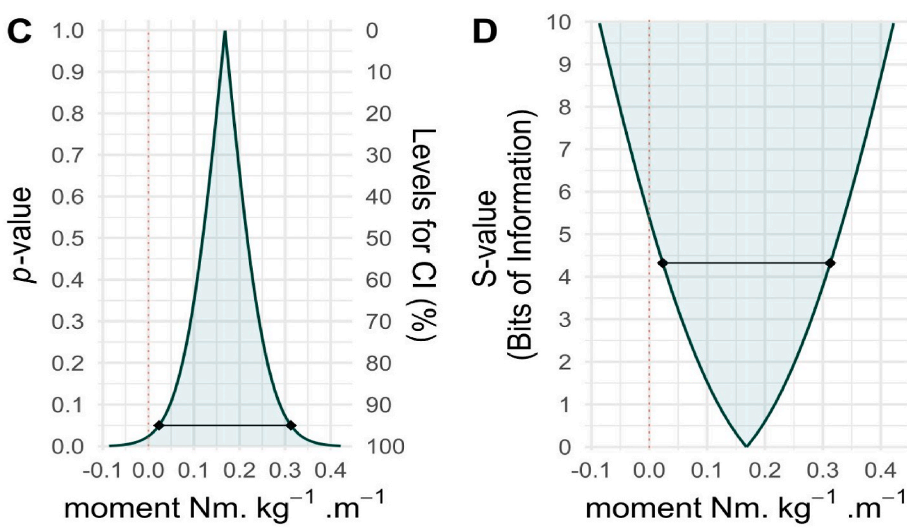

Fig. 4b. (C, D) shows the p value function, and corresponding S-value for Knee Abduction Moment (firm-sand). The red line denotes the null (zero) hypothesis. The horizontal black line provides the effect sizes for p = 0.05 (95% CI). (For interpretation of the references to colour in this figure legend, the reader is referred to the Web version of this article.)

direction tasks on sand compared to firm surfaces were found in the glutei muscles and tensor fascia latae (Rafols Parellada et al., 2020; Pinnington et al., 2005), which were not investigated in our study. Investigating muscle activation in these muscles when landing on different surfaces, would be beneficial, given the proposed contribution of gluteal muscle activity to reduced frontal plane landing angle and moments during single leg landing tasks (Llurda-Almuzara et al., 2021; Dix et al., 2019; Maniar, Schache, Pizzolato, & Opar, 2022; Neamatallah et al., 2020).

Our study is not without its limitations. We acknowledge that our observations are specific to a healthy recreationally active female cohort. This research paves the way for comparisons in poorer landers and across pathologic populations to allow wider generalisation. We appreciate that using a discrete time point for our outcomes may limit the ability to capture nuanced variations throughout the entire landing phase, which might be seen with other analysis techniques (e.g., statistical parametric mapping) which assess the continuous waveform. Additionally, EMG is prone to large between-participant variation, does not directly measure muscle force and evidence of a practically meaningful difference is lacking in the literature which would be useful to inform our inferences. It is possible that the characteristics of the sand surface such as granulation, moisture content, and depth contributed to different levels of stiffness between testing days which may have affected results (Pinnington & Dawson, 2001). Although we quantified surface stiffness and ensured a consistent depth of 10 cm for each SLH on the surface, we only used one type of sand, under single laboratory-controlled conditions. Future work should quantify the effects of different sand conditions (i.e., depth and stiffness) on lower limb kinematics, kinetics and muscle activity during jump landing tasks. Finally, whilst we have made every attempt to standardise our procedures and address background assumptions, we accept that there are uncertainties not captured in the statistical models which could explain our findings. We also acknowledge that values just outside the 95% CI also have limited information against them and could be considered compatible. As such, advice on the role of sand for (p)rehabilitation should acknowledge and quantify these uncontrolled biases with consideration of cost to benefit ratios.

5. Conclusion

Our study provides valuable information on the magnitude of changes in KAbM when landing on sand compared with a firm surface, and additionally a pliable grass surface. Observed differences in FPKA on sand compared to other surfaces were smaller in magnitude, less precise and pointed in the opposite direction to previous literature and our research hypothesis. This could be partially explained by the healthy landing strategies adopted by our participants across surfaces. It is not clear if the magnitude of differences we observed in KAbM and FPKA were of practical importance, and further work is required to establish these thresholds for both practitioners and researchers alike. Although larger effects for reduced LG on sand compared to grass and firm surfaces were observed, we found almost no reputational information against 'no effect' hypothesis in muscle activity for VM, VL, LH, and MH. Estimates of changes in vGRF were lower when landing on sand compared to firm and grass surfaces. Our findings are of interest to practitioners who consider the use of different surfaces in training, injury prevention or rehabilitation programmes with female athletes, which involve a jump landing component.

Data availability statement

The data that support the findings of this study are available from the corresponding author, [MR], upon reasonable request.

Funding

None received.

Ethical approval

All participants provided written informed consent, with the study approved by the University's ethics committee (No. 035/19), in accordance with the Declaration of Helsinki.

Clinical trials registry

The trial was registered with clinicaltrials.gov prior to study recruitment (NCT04502615).

CRediT authorship contribution statement

Mark C. Richardson: Conceptualization, Data curation, Formal analysis, Investigation, Methodology, Project administration, Resources, Software, Validation, Visualization, Writing – original draft, Writing – review & editing. **Paul Chesterton:** Conceptualization, Formal analysis, Investigation, Methodology, Supervision, Writing – original draft, Writing – review & editing. **Abigail Taylor:** Data curation, Formal analysis, Methodology, Writing – original draft, Writing – review & editing. **William Evans:** Conceptualization, Data curation, Formal analysis, Investigation, Methodology, Resources, Supervision, Validation, Visualization, Writing – original draft, Writing – review & editing.

Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

Acknowledgements

We would like to thank all participants for being part of this project.

Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.ptsp.2024.07.001.

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