A four-experiment examination of ankle kinetics, kinematics and lateral ligament strains during different conditions; an examination using musculoskeletal simulation.

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Abstract

PURPOSE: Firstly, to provide insight into the differences in ankle kinetics, kinematics and lateral ligament strain between males and females during functional sports movements. In addition, this study explores the prophylactic efficacy of different footwear and bracing modalities.

METHODS: Experiment 1 examined male and female athletes performing run, 45° cut and one-legged hop movements, experiment 2 observed court, energy return and trainer footwear conditions during a change of direction task, experiment 3 examined high-cut, low-cut and trainer footwear conditions in change of direction, run, 45° cut and vertical jump movements and experiment 4 explored an ankle sleeve and an ankle brace during a change of direction movement. In each experiment ankle kinetics and ligament strain were measured using a musculoskeletal simulation approach.

RESULTS: Experiment 1 indicates that males exhibited increased inversion velocity (male=260.39 & female=219.18°/s) in the cut movement as well as enhanced peak posterior force (male=2.24 & female=1.35BW), anterior talofibular ligament (ATFL) strain rate (male=266.77 & female=133.16%/s). Experiment 2 showed that both calcaneofibular ligament (CFL) and posterior talofibular ligament (PTFL) strain velocities were greater in the court footwear (CFL=90.86 & PTFL=151.45%/s) compared to the trainer (CFL=69.07 & PTFL=119.57%/s). Experiment 3 showed in the run movement anterior talofibular ligament (ATFL) strain was enhanced in the trainer (7.86%) compared to the high (3.61%) and low (5.87%) conditions and the trainer (8.14%) compared to the low footwear (5.39%) for the cut movement. PTFL strain velocity was greater in the high footwear (188.01%/s) compared to the trainer (175.60%/s) during the run movement and in both high (221.55%/s) and low (220.29%/s) footwear compared to the trainer (202.05%/s) during the cut. Experiment 4 revealed that PTFL strain was greater in the sleeve condition (17.05%) compared to the ankle brace (15.42%).

CONCLUSION: This study provides insight into the potentially increased incidence of lateral ankle ligament sprain injuries in males, whilst also highlighting the prophylactic efficacy of ankle braces in attenuating the ankle strain mechanisms linked to the aetiology of lateral ankle ligament injuries.

Introduction

The benefits of physical activity/ sport are universally recognized; however, sports/ physical activity is associated with a high incidence of injuries (1). Ankle sprains are an extremely common complaint among physically active individuals, particularly in court based athletic disciplines (2-3). Lateral ankle sprains have been shown to account for 14% of all orthopedic emergency cases (4), with an estimated 2.15 ankle sprains occurring per 1000-person years (5). The most frequently injured lateral ankle ligaments are the anterior talofibular ligament (ATFL), calcaneofibular ligament (CFL) and posterior talofibular ligament (PTFL) (6). Importantly, many patients develop long lasting problems after experiencing an ankle sprain; a condition broadly termed as chronic ankle instability (CAI) (7), and concerningly, 13% of individuals with a history of CAI go on to experience post-traumatic ankle osteoarthritis (8).

The main function of skeletal ligaments is to maintain passive joint congruency (9), however ligaments are also able respond to mechanical stimuli and provide proprioception and kinesthesia (10). Ligaments themselves are composed primarily of collagen fibers that encompass approximately 75% of the dry ligamentous mass with proteoglycans, elastin, glycoproteins and other proteins making up the remaining 25% (11). Sprain injuries experienced by the lateral ankle ligaments are mediated when excessive tension is applied to the ligaments during athletic movements, this causes the ligaments collagen fibers to elongate i.e. experience strain (9). Ligament injury can happen with a single acute episode which exceeds the ligaments maximum strain capacity resulting in a ligament rupture, or from cumulative overload with insufficient recovery time so that these chronic insults render the ligaments unable to properly support the joint, leading to instability and pain (12).

Because of the high incidence of lateral ankle sprain injuries (4) and the poor-long term prognosis following injury (7), prophylactic/ preventative strategies are a key priority for clinical sports research (13). The ankle joint can theoretically be supported by external equipment (14), therefore devices such as ankle braces and sleeves as well as athletic footwear with different collar heights and traction characteristics have received considerable attention. To this end court-based footwear are designed with high-cut ankle supports designed primarily to limit excessive ankle movement (15). The biomechanical analyses examining the effects of high cut footwear have provided contradictory results. In 45° cutting movements Commons & Low (16) showed that high-cut footwear increased the peak ankle inversion angle whereas Lam et al., (15) and Liu et al., (17) found that the ankle inversion angle and peak inversion velocity were reduced in high-cut footwear. In addition, Klem et al., (18) showed during a 45° cutting movement that a hinged ankle brace reduced peak inversion, dorsiflexion and compressive loading compared to no-brace. Similarly, Graydon et al., (14) found that ankle braces significantly reduced both sagittal and coronal plane movement of the ankle during running and Ubell et al., (19) revealed that ankle braces significantly enhanced participants ability restrict inversion below a threshold of 24° during a one-legged jump task. Similarly, high friction at the shoe-surface interface is a well-acknowledged risk factor for lateral ankle sprain injuries (20), although to our knowledge there has yet to be any investigation which has examined the effects of footwear with different frictional characteristics on the factors linked to the aetiology of lateral ankle strains. However, despite the wealth of biomechanical analyses investigating effects of external prophylactic devices on ankle sprain injury risk, there have not been any investigations examining their effects on the lateral ankle ligaments themselves during functional sports movements commonly associated with sprain injuries.

The high incidence of lateral ankle ligament sprain injuries indicate that prophylactic modalities have had limited success. However, accurate assessment of ankle ligament strain behavior is crucial in optimizing prophylactic treatment modalities (21). Therefore, it is clear that further investigation of these prophylactic modalities is required, which may provide important clinical information for the prevention of lateral ankle ligament sprains across different athletic activities. Recent developments in musculoskeletal simulation modelling techniques now allow indices of ankle ligament strain to be obtained during sports movement commonly associated with ankle sprain injury (21). Therefore, the effects of different prophylactic on the specific physiological parameters that cause strain parameters can now be explored, which will be of both practical and clinical relevance.

There is a lack of clarity within epidemiological literature regarding the relative risk of lateral ankle sprain injuries in male and female athletes. Indeed, Waterman et al., (2010), Beynnon et al., (22) and Roos et al., (23) all demonstrated that males and females had similar ankle-sprain incidence ratios. However, Ristolainen et al., (24) and Hosea et al., (25) found that ankle injuries were most common in female athletes, yet conversely Tummala et al., (26) showed that the rate of ankle injuries was greater in males. From a biomechanical perspective, females have been shown to exhibit increased inversion-eversion laxity during a dynamic postural control task (27), reduced inversion and increased eversion during a 45° cutting movement (28) and increased inversion during side-step and jump landing tasks (29). However, owing to a lack of appropriate musculoskeletal modelling tools it is currently unknown whether there are sex differences in lateral ligament strain characteristics (27). Thus, with the advent of more advanced musculoskeletal simulation-based modeling approaches there is a clear need to further investigate the mechanics of the lateral ankle ligaments in both males and females in disciplines/movements commonly associated with ankle ligament strain injuries.

Therefore, the aims of the current investigation by using a four-experiment musculoskeletal simulation-based approach were (whilst measuring ankle kinetics, kinematics and lateral ankle ligament strain parameters to investigate): 1. sex differences during functional sports movements, 2. the effects of different court based (court, trainer and energy return) footwear during a change of direction task 3. the effects of high and low-cut court footwear during functional sports movements 4. the effects of an ankle brace and ankle sleeve during a change of direction task.

In relation to the aforementioned aims, the current investigation tests the following hypotheses; 1. no sex differences in ankle inversion, joint loading and lateral ankle strain characteristics will be evident; 2. court footwear will exhibit reduced ankle inversion, joint loading and lateral ankle strain characteristics; 3. high-cut footwear will reduce ankle inversion parameters, joint loading and lateral ankle strain characteristics and 4. ankle bracing will attenuate ankle inversion, joint loading and lateral ankle strain characteristics.

Methods

For each of the four investigations, participants provided written informed consent and ethical approval was obtained from the University of Central Lancashire, in accordance with the principles documented in the Declaration of Helsinki. All participants were free from lower extremity musculoskeletal pathology at the time of data collection and had not undergone surgical intervention at the ankle joint.

Experiment 1

Participants

Fifteen male (age 30.1 ± 5.2 years, height 1.75 ± 0.07 m and body mass 77.1 ± 10.8 kg) and fifteen female (age 29.6 ± 5.6 years, height 1.66 ± 0.06 m and body mass 65.8 ± 9.9 kg) recreational athletes volunteered to take part in the current investigation.

Procedure

Participants completed five trials of three sport-specific movements, (run, one legged hop and 45° cut) and the order in which participants performed each movement was counterbalanced. To ensure consistency, each participant wore the same footwear (Asics, Patriot 6). Kinematic information was obtained using an eight-camera motion capture system (Qualisys Medical AB, Goteburg, Sweden) with a capture frequency of 250 Hz. To measure ground reaction forces (GRF), an embedded piezoelectric force platform (Kistler National Instruments, Model 9281CA) operating at 1000 Hz was adopted. The GRF and kinematic information were synchronously obtained using an analogue board and interfaced using Qualisys track manager.

To define the anatomical frames of the thorax, pelvis, thighs, shanks and feet, passive retroreflective markers of 19mm diameter were placed at the C7, T12 and xiphoid process landmarks and also positioned bilaterally onto the acromion process, iliac crest, anterior superior iliac spine (ASIS), posterior super iliac spine (PSIS), medial and lateral malleoli, medial and lateral femoral epicondyles, greater trochanter, calcaneus, first metatarsal and fifth metatarsal. The hip, knee and ankle joint centre's were delineated according to previously established guidelines (30-32). Carbon-fibre tracking clusters comprising of four non-linear retroreflective markers were positioned onto the thigh and shank segments. The foot segments were tracked via the calcaneus, first and fifth metatarsal, the pelvic segment using the PSIS and ASIS markers and the thorax via the T12, C7 and xiphoid markers. Static calibration trials were obtained with the participant in the anatomical position in order for the positions of the anatomical markers to be referenced in relation to the tracking clusters/markers, following which those not required for dynamic data were removed. The Z (transverse) axis was oriented vertically from the distal segment end to the proximal segment end. The Y (coronal) axis was oriented in the segment from posterior to anterior. Finally, the X (sagittal) axis orientation was determined using the right-hand rule and was oriented from medial to lateral.

Data were collected during the cut and hop movements according to below procedures:

<u>Run</u>

Participants ran at 4.0 ± 0.2 m/s and struck the force platform with their right (dominant) limb. The average velocity of running was monitored using infra-red timing gates (SmartSpeed Ltd UK), and the stance phase of running was defined as the duration over > 20 N of vertical force was applied to the force platform.

<u>Cut</u>

Participants completed 45° sideways cut movements using an approach velocity of 4.0 ± 0.2 m/s striking the force platform with their right (dominant) limb. Cut angles were measured from the centre of the force plate and the corresponding line of movement was delineated using masking tape so that it was clearly evident to participants. The stance phase of the cut movement was defined as the duration over > 20 N of vertical force applied to the force platform.

Hop

Participants began standing by on their dominant limb, they were then requested to hop forward maximally, landing on the force platform with same leg without losing balance. The arms were held across the chest to remove arm-swing contribution. The landing phase of the this movement was analysed which was defined as the duration from foot contact (defined as > 20 N of vertical force applied to the force platform) to maximum knee flexion. The hop distance for each participant was established during practice trials, and the starting position was marked using masking tape.

Processing

Dynamic trials were digitized using Qualisys Track Manager (Qualisys Medical AB, Goteburg, Sweden) in order to identify anatomical and tracking markers then exported as C3D files to Visual 3D (C-Motion, Germantown, MD, USA). All data were linearly normalized to 100 % of the stance/ landing phases. GRF data and marker trajectories were smoothed with cut-off frequencies of 50 Hz at 12 Hz respectively, using a low-pass Butterworth 4th order zero lag filter. Within Visual 3D ankle were quantified using an XYZ cardan sequence (where X is dorsiflexion-plantarflexion; Y is inversion-eversion and is Z is internal-external rotation). Three-dimensional angular kinematic measures that were extracted in each of the aforementioned planes of rotation were peak angle, peak angular velocity and minimum angular velocity. Finally, within Visual 3D three dimensional joint moments were quantified using Newton-Euler inverse dynamics and the peak joint moment (Nm/kg) and the joint moment impulse (Nm/kg·s) (using a trapezoidal function) were extracted for analysis. Finally, the peak translation coefficient of friction (μ) of each footwear was determined from the ratio of horizontal and vertical force components during the initial period of shoe motion (33). The peak rotational moment of the ground reaction force (Nm/kg) was used to describe the rotational friction characteristics of the footwear (34).

Following this, data during the stance/ landing phases were exported from Visual 3D into OpenSim 3.3 software (Simtk.org). The standard OpenSim Gait2392 musculoskeletal model with 12 segments, 19 degrees of freedom and 92 musculotendon actuators was adapted to include the three lateral ankle ligaments (ATFL, CFL and PTFL) (Figure 1). The model ligament insertion points were implemented in accordance with Golano et al., (6) and the resting ligament lengths (ATFL = 22.10mm, CFL = 31.71mm and PTFL = 21.39mm) compare well with the published in-vivo ligament data provided by Zhang et al., (21). The model firstly was scaled to account for the anthropometrics of each participant, then as muscle forces are the main determinant of joint forces (35), muscle kinetics were quantified using static optimization. From the static optimization procedure kinetics of the muscles within the model that cross the ankle joint (extensor digitorum longus, extensor hallucis longus, flexor digitorum longus, flexor hallucis longus, lateral gastrocnemius, medial gastrocnemius, peroneus brevis, peroneus longus, peroneus tertius, soleus, tibialis anterior, tibialis posterior) were analyzed. From these muscles the peak force (BW) and the impulse (BW·ms) (using a trapezoidal function) during the stance/ landing phases were obtained. Three-dimensional ankle joint forces were then calculated using the joint reaction analyses function using the muscle forces generated from the static optimization process as inputs. From the joint reaction process peak three-dimensional joint forces (BW) and impulses (BW·s) (using a trapezoidal function) during the stance/ landing phases were extracted.

@@@FIGURE 1 NEAR HERE@@@

ATFL, CFL and PTFL kinematics during each movement were calculated via the muscle analyses function within OpenSim. Peak ligament strain (%) was calculated and extracted by dividing the change in length of each ligament during movement by its resting length then multiplying by 100 to create a percentage. In addition, the peak strain velocity (%/s) was calculated and extracted as the maximum change in strain between adjacent data points using a first derivative function.

Following this, three-dimensional ankle joint kinematics, joint moments, joint forces, muscle forces, ligament strain and ligament strain velocity were extracted during the entire stance/ landing phase and time normalized to 101 data points using linear interpolation for each participant (36).

Statistical analyses

Differences across the stance/ landing phase of each movement were examined using 1-dimensional statistical parametric mapping (SPM) (MATLAB, MathWorks, Natick, USA) using the source code available at http://www.spm1d.org/. Differences between males and females were examined using independent t-tests (SPM t) and the alpha level for statistical significance was set at 0.05. For discrete parameters means and standard deviations were calculated, and differences examined using Bayesian independent samples t-tests with default prior scales using JASP software 0.10.2 (37). Bayesian factors (BF) were used to explore the extent to which the data supported the alternative (H_1) hypothesis. Bayes factors were interpreted in accordance with the recommendations of Jeffreys, (38), with values above 3 indicating sufficient evidence in support of H_1.

Experiment 2

Participants

Ten male recreational athletes volunteered to take part in the current investigation. The mean characteristics of the participants were: age 24.6 ± 2.8 years, height 1.77 ± 0.05 m and body mass 73.7 ± 7.1 kg.

Footwear

The footwear used during this study consisted of a conventional Trainer (New Balance 1260 v2), Court shoes (Hi-Tec Indoor Lite), and Energy return footwear (Adidas energy boost) (shoe size 8–10 in UK men's sizes) (Figure 2abc).

@@@FIGURE 2 NEAR HERE@@@

Procedure

Kinematic information was obtained using the procedure and biomechanical modelling approach outlined in experiment 1. For this experiment participants performed maximal 180° cutting maneuvers in each footwear condition (Court, Energy return and Trainer) whilst striking the force platform with their dominant foot. Participants commenced their trials 6 m away from the force platform, ran straight and planted their dominant foot on the force platform, and then changed direction to move 180° to their initial direction of motion. The order in which participants performed in each footwear condition was counterbalanced and to ensure that participants utilized a similar approach velocity, their approach velocity during the first trial was calculated and a maximum deviation of 5% from this was allowed (Sinclair & Stainton, 2017).

Processing

The same processing techniques as experiment 1 were adopted and peak ligament strain, peak ligaments strain velocities, peak angles, peak angular velocities, peak joint moments, joint moment impulses, coefficient of friction, peak muscle forces, muscle force impulses, peak joint forces and joint force impulses were extracted for each experimental condition.

Statistical analyses

SPM was implemented in a hierarchical manner, analogous to one-way repeated measures ANOVA with post-hoc paired t-tests (SPM t) in the event of a main effect to explore differences between footwear conditions. examined using (SPM t) and the alpha level for statistical significance was set at 0.05. Differences in discrete parameters were examined using Bayesian one-way repeated measures ANOVA, followed by Bayesian paired t-tests in the event of a main effect to explore differences between footwear.

<u>Experiment 3</u>

Participants

Ten male recreational athletes volunteered to take part in the current investigation. The mean characteristics of the participants were: age 24.3 ± 4.1 years, height 1.77 ± 0.07 m and body mass 78.7 ± 7.4 kg.

Footwear

The footwear used during this study consisted of a conventional Trainer (New Balance 1260 v2), high-cut (Nike Lebron XII High) and low-cut (Nike Lebron XII Low) (shoe size 8–10 in UK men's sizes) (Figure 3abc).

@@@FIGURE 3 NEAR HERE@@@

Procedure

Kinematic information was obtained using the procedure and biomechanical modelling approach outlined in experiment 1. For this experiment participants performed four different movements (run, cut, change of direction and vertical jump) in each of the aforementioned footwear conditions (High, Low and Trainer). The run, cut and change of direction movements were examined as described in experiments 1 and 2. For the vertical jump movement, participants completed counter movement vertical jumps in which they were required to use full arm swing and also to commence and land the jump on the force platform. The landing phase of the jump movement was quantified and was considered to have begun at foot contact (defined as > 20 N of vertical force applied to the force platform) and ended at the instance of maximum knee flexion.

Processing

The same processing techniques as experiment 1 were adopted and peak ligament strain, peak ligaments strain velocities, peak angles, peak angular velocities, peak joint moments, joint moment impulses, coefficient of friction, peak muscle forces, muscle force impulses, peak joint forces and joint force impulses were extracted for each experimental condition.

Statistical analyses

To examine differences between footwear the same statistical analyses as experiment 2 were adopted, with the same statistical principles and reporting as experiment 1 adhered to.

<u>Experiment 4</u>

Participants

Twelve male recreational athletes volunteered to take part in this study. The mean characteristics of the participants were: age 20.7 ± 1.6 years, height 1.81 ± 0.05 m and body mass 79.3 ± 8.2 kg.

Ankles braces

The ankle braces used during this study consisted of an ankle Sleeve (Compex, Trizone) and also an ankle Brace (Aircast A60 DJO), in sizes small, medium and large.

@@@FIGURE 4 NEAR HERE@@@

Procedure

Kinematic information was obtained using the procedure and biomechanical modelling approach outlined in experiment 1. For this experiment participants performed a 45° cutting maneuvers as described in experiment 1 in each ankle brace condition (ankle Sleeve, ankle Brace and no-brace).

Processing

The same processing techniques as experiment 1 were adopted and peak ligament strain, peak ligaments strain velocities, peak angles, peak angular velocities, peak joint moments, joint moment impulses, peak muscle forces, muscle force impulses, peak joint forces and joint force impulses were extracted for each experimental condition.

Statistical analyses

To examine differences between ankle brace conditions the same statistical analyses as experiment 2 were adopted, with the same statistical principles and reporting as experiment 1 adhered to.

Results

Tables 1-7 and supplemental figures 1-7 show comparisons between experimental conditions in discrete parameters using Bayesian analyses and SPM. In the interests of conciseness and clarity to the reader, only discrete values that exhibited a Bayes factor in excess of 3 and SPM comparisons that demonstrated statistical significance.

<u>Experiment 1</u>

Cut

Statistical parametric mapping

No significant (p>0.05) differences between males and females were detected using SPM.

Discrete parameters

Differences between males and females in discrete parameters and the associated BF's are presented in table 1.

@@@TABLE 1 NEAR HERE@@@

Run

Statistical parametric mapping

No significant (p>0.05) differences between males and females were detected using SPM.

Discrete parameters

None of the discrete parameters were associated with a BF greater than 3.

Hop

Statistical parametric mapping

Posterior force was shown to be significantly greater in males from 0-60% of the landing phase in comparison to females (Supplemental figure 1).

Discrete parameters

Differences in discrete parameters between males and females and the associated BF's are presented in table 1.

<u>Experiment 2</u>

Statistical parametric mapping

Inversion was shown to be significantly greater in the trainer from 15-40% and in the court footwear from 5-20% of the stance phase compared to energy return (Supplemental figure 2ab). Internal rotation was shown to be significantly greater in the energy return footwear from 20-40% and 5-50% of the stance phase compared to the trainer and court footwear and in the trainer compared to court footwear from 15-40% of the stance phase (Supplemental figure 2cde). Inversion velocity was greater in the court footwear from 5-10 and 90-95% but greater in the energy return condition from 20-25 and 40-45% of the stance phase (Supplemental figure 2f). In addition, anterior force was greater in the energy return from 5-40% and compressive force from 20-30% of the stance phase compared to court footwear (Supplemental figure2gh). Medial forces were shown to be larger in the court footwear from 20-50 and 40-50% of the stance phase compared to the energy return and trainer conditions (Supplemental figure2ij). Similarly, CFL strain velocity was shown to be larger in the court footwear from 10-20 and 15-20% of the stance phase compared to the energy return and trainer conditions (Supplemental figure2kl). Similarly, PTFL strain velocity was shown to be larger in the court footwear from

10-30 and 15-30% of the stance phase compared to the energy return and trainer conditions (Supplemental figure3ab).

Discrete parameters

Differences in discrete parameters between footwear and the associated BF's for the both main effect and post-hoc analyses are presented in table 2.

@@@TABLE 2 NEAR HERE@@@

Experiment 3

Change of direction

Statistical parametric mapping

Inversion shown to be larger in the high footwear from 20-80 and 10-80% of the stance phase compared to the low and trainer conditions (Supplemental figure 4ab). Internal rotation was shown to be greater in the trainer from 40-45% of the stance phase compared to the high footwear and in the low compared to the high footwear from 5-15 and 50-55% of the stance phase (Supplemental figure 4cd). Inversion velocity was shown to be larger in the high footwear from 5-20 and 5-10% of the stance phase compared to the low and trainer conditions (Supplemental figure 4ef). In addition, the transverse plane moment was shown to be larger in the high footwear from 15-20 and 15-20/ 25-30% of the stance phase compared to the low and trainer conditions (Supplemental figure 4ef). Similarly, anterior forces were shown to be larger in the high footwear from 80-85 and 70-80% of the stance phase compared to the low and trainer conditions (Supplemental figure 4ij).

Discrete parameters

Differences in discrete parameters between footwear and the associated BF's for the both main effect and post-hoc analyses are presented in table 3.

@@@TABLE 3 NEAR HERE@@@

Cut

Statistical parametric mapping

Inversion velocity shown to be larger in the high footwear from 10-20% of the stance phase compared to the low conditions (Supplemental figure 5a). External rotation velocity was also shown to be greater in the high footwear compared to low from 10-15% of the stance phase (Supplemental figure 5b). In addition, the transverse plane moment was shown to be larger in the high footwear from 5-10 of the stance phase compared to the low condition (Supplemental figure 5c). Similarly, the lateral gastrocnemius force was shown to be larger in the high footwear from 50-65 of the stance phase compared to the low condition (Supplemental figure 5d). Finally, ATFL strain was shown to be larger in the trainer compared to the low condition from footwear from 80-100% of the stance phase (Supplemental figure5e).

Discrete parameters

Differences in discrete parameters between footwear and the associated BF's for the both main effect and post-hoc analyses are presented in table 4.

@@@TABLE 4 NEAR HERE@@@

Run

Statistical parametric mapping

Plantarflexion velocity was shown to be larger in the trainer from 70-100 and 70-95% of the stance phase compared to the high and low footwear conditions (Supplemental figure 6ab). Eversion velocity was also shown to be greater in the high footwear compared to the trainer from 0-10% of the stance phase (Supplemental figure 6c). In addition, compressive forces were larger in the high footwear from 30-35 and 5-10/ 20-25/ 60-65% of the stance phase compared to the trainer and low footwear conditions (Supplemental figure 6de). ATFL strain was shown to be larger in the trainer from 95-100 and 95-100% of the stance phase compared to the high and low footwear conditions (Supplemental figure 6fg). Similarly, ATFL strain velocity was larger in the trainer from 75-100 and 75-100% of the stance phase compared to the high and low footwear conditions (Supplemental figure 6hi). Finally, PTFL strain velocity was shown to be greater in the high compared to the low footwear from 10-30/ 85-100% of the stance phase (Supplemental figure 6j).

Discrete parameters

Differences in discrete parameters between footwear and the associated BF's for the both main effect and post-hoc analyses are presented in table 5.

@@@TABLE 5 NEAR HERE@@@

Vertical jump

Statistical parametric mapping

Flexor digitorum longus force was shown to be larger in the high footwear from 30-40% of the stance phase compared to the trainer (Supplemental figure 7a). Similarly, flexor hallucis longus force was greater in the high footwear from 35-40% of the stance phase compared to the trainer (Supplemental figure 7b).

Discrete parameters

Differences in discrete parameters between footwear and the associated BF's for the both main effect and post-hoc analyses are presented in table 6.

@@@TABLE 6 NEAR HERE@@@

<u>Experiment 4</u>

Statistical parametric mapping

Inversion was shown to be larger in the sleeve compared to the brace from 35-85% of the stance phase and in the no-brace condition compared to the sleeve from 15-15% of the stance phase (Supplemental figure 8ab). Medial forces were also larger in the no-brace condition compared to brace from 15-20% of the stance phase (Supplemental figure 8c).

Discrete parameters

Differences in discrete parameters between footwear and the associated BF's for the both main effect and post-hoc analyses are presented in table 7.

@@@TABLE 7 NEAR HERE@@@

Discussion

The current investigation using a four-experiment approach represents the first study to explore differences in ankle kinetics, kinematics and lateral ankle ligament strain parameters between males and females in addition to examining the influence of different footwear and ankle brace conditions. A study of this nature provides further insight into potentially distinct incidence rates of lateral ankle ligament injures in female/males athletes, in addition to the potential efficacy of different prophylactic modalities for the prevention of ankle ligament pathologies in different sports movements.

The findings from experiment 1 do not support hypothesis 1, as differences in ankle inversion, joint loading and lateral ankle strain characteristics were evident during both the cut and hop movements. Specifically, both discrete parameters and SPM showed that males were associated with enhanced inversion velocity, peak posterior force and ATFL strain rate compared to females. Increased ligament strain magnitudes are linked to the aetiology of lateral ankle ligament pathology, either as an acute occurrence or through repeated manifestations/ exposures (12). Similarly, enhanced ankle joint loading is associated, through repeated exposure with the initiation and progression joint osteoarthritic degeneration (40). Therefore, experiment 1 collectively indicates that males are most susceptible to the mechanisms linked to the aetiology of lateral ankle ligament and joint pathologies compared to females.

Importantly, in partial support of hypothesis 2 the findings from experiment 2 showed via both discrete parameters and SPM that the energy return footwear were associated with enhanced ankle joint loading in all three planes in comparison primarily to the court footwear, although medially directed forces were also greater compared to the trainer during early and midstance. Because of the proposed association between contact loading and the initiation/ progression of joint degeneration (40), this observation may be clinically meaningful. It can be conjectured based on the findings from this investigation that the specific energy return footwear examined in this investigation may increase the risk from degenerative ankle joint injury during sport specific change of direction movements.

Further to this however, in contradiction of hypothesis 2, both discrete parameters and SPM showed that the court footwear were found to be enhance both CFL and PTFL velocity compared to both the energy return and trainer conditions. Taking into account the concurrent increases in inversion and inversion velocity in the court footwear this observation was to be expected as excessive inversion itself is recognized as a primary kinematic mechanism associated with ankle sprain injuries (12). One of the main functions of the lateral ligaments is to resist inversion (9) and importantly the in-vivo data of Zhang et al., (21) showed that both the CFL and PTFL are lengthened by inversion. In addition, it has been ventured that friction at the shoe surface interface influences the risk for lateral ankle sprain injuries (41). However, the findings from experiment 2 indicate that whilst the coefficient of friction was reduced in the court footwear, the rotational moment was enhanced. This observation concurs with those of Sinclair & Stainton, (36) in that the rotational moment rather than the coefficient of friction were enhanced in court footwear and that it is this parameter that most strongly influences soft tissue injury risk during change of direction tasks. Regardless, the enhanced ligament strain observation in the court footwear may be clinically meaningful, as the aetiology of lateral ankle ligament injury is considered to be mediated through enhanced ligamentous strain characteristics (12). Experiment 2 therefore indicates that the biomechanical mechanisms responsible for lateral ankle ligament strain injuries were enhanced in the court specific footwear; a concerning observation taking into account that change of direction movements are fundamental to court-based activities (36).

In contradiction to hypothesis 3, collective consideration of the observations from experiment 3 in relation to joint loading showed using both SPM and discrete parameters that the high cut footwear enhanced joint loading in all three planes during the change of direction, cut and run movements in comparison primarily to both the low and trainer conditions. As such, taking into account the aforementioned association between joint loading and the aetiology of joint degeneration (40), the findings from experiment 3 indicate that high cut footwear may enhance the risk from degenerative ankle joint injury during sport specific movements. Furthermore, in partial support of hypothesis 3 ATFL strain parameters were shown to be enhanced in the trainer compared to the high and low conditions in the run movement and the high footwear for the cut movement. However, conversely PTFL strain characteristics were greater in the high-cut footwear for the run and vertical jumps movement and both high and low-cut conditions during the cut. The aforementioned results can also be contextualized taking into account the in-vivo observations of Zhang et al., (21) as both the discrete and SPM based findings from experiment 3 showed that inversion/ eversion parameters were enhanced in the high-cut footwear and plantarflexion variables larger in the trainer condition. Zhang et al., (21) showed that the ATFL is lengthened during plantarflexion and eversion and that the PTFL exhibits lengthening during dorsiflexion and inversion. As lateral ankle ligament sprain injuries are linked to the magnitude of the strain experienced by the ligaments themselves (12), experiment 3 provides interesting observations in that the experimental footwear affect ligament strain characteristics differently. This indicates that the trainer may enhance the risk for ATFL pathologies and the high-cut footwear condition appears to increase the risk from PTFL sprain injuries.

In support of hypothesis 4, the observations from experiment 4 showed using both SPM and discrete parameters that the no-brace condition was associated with enhanced medially directed ankle joint loading compared to both the sleeve and brace. Importantly, excessive joint loading is linked to the aetiology of joint degeneration, and thus the findings from experiment 4 indicate that the ankle sleeve and brace conditions are able to attenuate the biomechanical mechanisms associated with joint pathology (40). In addition, further supporting hypothesis 4 it was also shown that PTFL strain was shown to be larger in the sleeve in comparison to the brace condition. As the PTFL exhibits lengthening during and inversion, which was shown to be concurrently enhanced in the sleeve condition this observation was to be expected (21). Therefore, as lateral ankle ligament pathologies are associated with excessive ligamentous strain magnitudes (12), experiment 4 indicates that the brace may attenuate the biomechanical risk factors associated with PTFL sprain injuries.

Limitations

A potential limitation is that musculoskeletal simulation modelling approach adopted in order to quantify ligament strain mechanics was not able to account for the inter-variability in the ligamentous construction and insertion points (42). Whilst should be noted that direct measures are not possible in human participants and that the resting lengths of the modelled ligaments as well as the strain magnitudes are consistent with those presented in the scientific literature and within the physiological range (21). There is nonetheless, considerable scope for future development of simulation-based models to address and improve upon these limitations; to provide more accurate and individually adapted musculoskeletal simulations of lateral ankle ligament mechanics linked to the aetiology of sprain injuries.

Conclusion

The findings from the current four-experiment investigation provide further insight into differences in lateral ankle ligament parameters between male and female athletes the mechanisms responsible for the potentially increased incidence of lateral ankle ligament sprain injuries in males, whilst also highlighting the prophylactic efficacy of ankle braces in attenuating the ankle strain mechanisms linked to the aetiology of lateral ankle ligament injuries.

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<u>Figure labels</u>

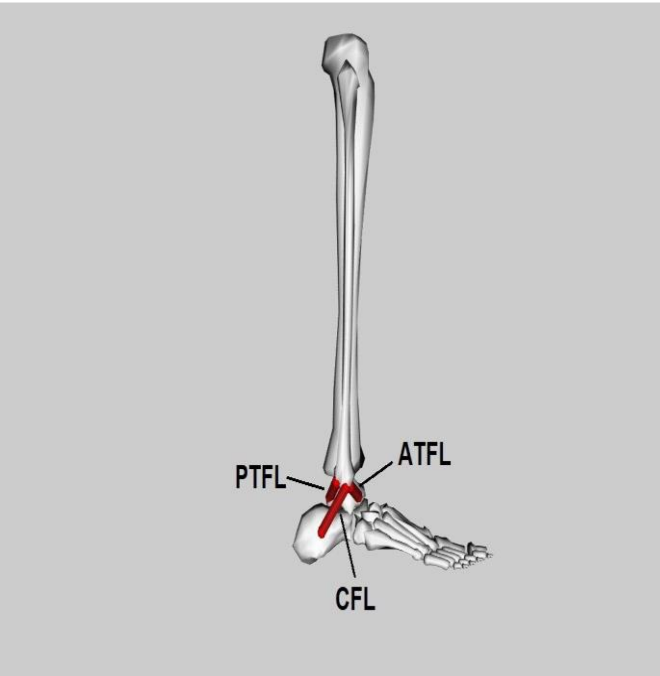

Figure 1: Lateral ankle ligaments (ATFL, CFL and PTFL) in the OpenSim based model.

Figure 2: Footwear from experiment 2 (a. = court, b. = energy return & c. = trainer).

Figure 3: Footwear from experiment 3 (a. = high-cut, b. = low-cut & c. = trainer).

Figure 4: Ankle braces from experiment 4 (a. = brace & b. = sleeve)

Table 1: Mean, standard deviation and Bayes factors (BF) for experiment 1.

	Male		Female		BF
	Mean	*SD*	**Mean**	*SD*	
	Cut				
Peak transverse plane angle (°)	8.80	5.98	3.84	4.24	***3.83***
Peak inversion velocity (°/s)	260.39	133.18	219.18	105.32	***6.33***
Extensor digitorum longus impulse (BW·s)	0.06	0.02	0.03	0.02	***5.74***
Peak peroneus longus force (BW)	1.23	0.17	1.48	0.23	***12.80***
	Hop				
Peak posterior force (BW)	2.24	0.48	1.35	0.63	***166.40***
Peak ATFL strain rate (%/s)	266.77	124.08	133.16	109.39	***4.40***

Notes: Bold Bayes factors indicate that they exceed 3.

Table 2: Mean, standard deviation and Bayes factors (BF) for experiment 2.

	Trainer		Energy return		Court		BF Main effect	BF post-hoc		
	Mean	*SD*	**Mean**	*SD*	**Mean**	*SD*		**Trainer vs Energy return**	**Trainer vs Court**	**Energy return vs Court**
Compressive force impulse (BW·s)	2.63	1.04	2.91	1.16	2.71	1.09	11.18	2.33	0.32	***47.17***
Peak medial force (BW)	0.97	0.56	1.05	0.70	0.88	0.58	23.03	0.67	1.12	***13.55***
Medial/lateral force impulse (BW·s)	0.18	0.10	0.21	0.13	0.15	0.12	49.92	0.69	1.04	***128.10***
Peak coronal plane moment (Nm/kg·s)	0.68	0.19	0.63	0.23	0.61	0.22	7.81	1.40	***3.13***	0.47
Peak transverse plane moment (Nm/kg·s)	0.28	0.07	0.23	0.09	0.25	0.09	27.92	***31.56***	0.57	1.63
Peak inversion (°)	20.93	4.92	18.01	3.83	20.52	4.23	43.60	2.64	0.32	***8.98***
Peak internal rotation (°)	-6.04	3.93	-5.87	3.46	-8.35	2.98	209.01	0.22	***5.30***	***10157.19***
Peak inversion velocity (°/s)	410.07	90.83	338.49	93.50	421.06	66.23	4.55	0.62	0.23	***34.82***
Peak internal rotation velocity (°/s)	256.81	82.32	219.11	87.98	299.27	85.55	28.13	0.49	1.46	***99.93***
Peak extensor digitorum longus force (BW)	0.44	0.32	0.50	0.32	0.61	0.22	108.95	0.81	***8.91***	***4.90***
Extensor digitorum longus impulse (BW·s)	67.55	69.24	83.39	81.64	86.02	72.40	573.57	***4.90***	***235.80***	0.28
Peak extensor hallucis longus force (BW)	0.13	0.10	0.17	0.12	0.19	0.10	191.84	1.53	***781.61***	0.65
Extensor hallucis longus impulse (BW·s)	18.61	20.21	24.67	25.26	25.18	24.62	96.84	2.87	***67.01***	0.23
Peak CFL strain rate (%/s)	69.07	42.26	78.81	45.45	90.86	36.19	5.34	***204.79***	***9.41***	0.49
Peak PTFL strain rate (%/s)	119.57	38.62	134.47	48.76	151.45	40.67	7.68	0.65	***15.67***	0.53
Coefficient of friction (µ)	0.64	0.08	0.60	0.07	0.57	0.08	665.37	***3.85***	***15.63***	***5.18***
Peak rotational moment (Nm/kg)	0.27	0.09	0.25	0.07	0.36	0.07	6062.27	0.42	***16.20***	***70.53***

Notes: Bold post-hoc Bayes factors indicate that they exceed 3.

Table 3: Mean, standard deviation and Bayes factors (BF) for the ***change of direction*** movement from experiment 3.

	High		**Low**		**Trainer**		**BF Main effect**	**BF post-hoc**		
	Mean	*SD*	**Mean**	*SD*	**Mean**	*SD*		**High vs Low**	**High vs Trainer**	**Low vs Trainer**
Peak anterior force (BW)	-1.56	1.09	-1.68	0.84	-1.05	0.96	5.64	0.27	1.74	***3.02***
Peak medial force (BW)	0.99	0.63	0.71	0.35	0.88	0.39	5.61	***3.92***	0.36	1.76
Medial/lateral force impulse (BW·s)	0.16	0.08	0.11	0.07	0.16	0.08	5.23	***11.11***	0.24	1.30
Peak dorsiflexion moment (Nm/kg)	2.16	0.28	2.09	0.26	1.91	0.23	608.38	0.68	***10.01***	***62.71***
Peak internal rotation moment (Nm/kg)	0.36	0.11	0.30	0.08	0.28	0.11	2422.01	***19.48***	***23.57***	0.79
Transverse plane moment impulse (Nm/kg·s)	0.08	0.02	0.06	0.01	0.06	0.03	14.82	2.84	***10.93***	0.23
Peak inversion (°/s)	25.25	1.86	22.54	1.96	22.30	2.77	31.08	***7485.33***	***5.13***	0.24
Peak inversion velocity (°/s)	351.79	80.72	311.14	61.56	325.95	94.50	3.96	***90.91***	0.76	0.31
Peak internal rotation velocity (°/s)	-205.32	33.82	-187.38	34.61	-160.77	51.10	28.12	1.51	***19.23***	0.73
Peak rotational moment (Nm/kg)	0.36	0.07	0.33	0.09	0.32	0.07	3.50	***3.40***	1.93	0.26

Notes: Bold post-hoc Bayes factors indicate that they exceed 3.

Table 4: Mean, standard deviation and Bayes factors (BF) for the ***cut*** movement from experiment 3.

	High		Low		Trainer		BF Main effect	BF post-hoc		
	Mean	*SD*	**Mean**	*SD*	**Mean**	*SD*		**High vs Low**	**High vs Trainer**	**Low vs Trainer**
Compressive force impulse (BW·s)	1.56	0.17	1.52	0.16	1.43	0.20	140.30	0.65	***11.36***	***4.41***
Peak internal rotation moment (Nm/kg)	0.33	0.05	0.27	0.07	0.32	0.06	9.49	***4.95***	0.27	***3.70***
Sagittal plane moment impulse (Nm/kg·s)	0.40	0.04	0.36	0.07	0.33	0.06	211.74	2.06	***37.04***	0.96
Peak inversion (°)	15.11	2.16	12.50	2.65	11.90	2.35	115.27	***13.51***	***333.33***	0.27
Peak dorsiflexion velocity (°/s)	396.85	105.55	409.55	121.62	358.22	94.37	229.57	0.56	***18.18***	***6.71***
Peak inversion velocity (°/s)	189.72	109.82	146.82	76.76	171.35	100.24	31.64	***4.63***	1.58	1.32
Peak plantarflexion velocity (°/s)	-553.44	80.56	-580.91	66.22	-600.79	78.28	9.95	2.40	***14.49***	0.37
Peak flexor digitorum longus force (BW)	0.19	0.06	0.18	0.10	0.14	0.06	5.45	0.26	***9.01***	1.64
Flexor digitorum longus impulse (BW·ms)	7.49	3.36	6.42	3.15	5.88	2.38	18.17	***3.92***	***3.28***	0.46
Peak flexor hallucis longus force (BW)	0.19	0.06	0.17	0.09	0.13	0.06	41.47	0.39	***45.45***	2.09
Soleus impulse (BW·ms)	588.54	114.66	561.75	121.01	480.03	153.49	3.14	0.29	1.39	***3.06***
Peak tibialis posterior force (BW)	0.61	0.25	0.51	0.14	0.35	0.10	1047.63	0.67	***32.26***	***250.00***
Tibialis posterior impulse (BW·ms)	41.94	30.74	27.95	15.51	14.21	4.47	42.50	1.06	***4.42***	***8.62***
ATFL peak strain (%)	6.67	3.39	5.39	4.44	8.14	3.79	16.03	0.58	1.32	***7.46***
Peak PTFL strain velocity (%/s)	221.55	71.61	220.29	71.42	202.05	60.73	874.96	0.24	***43.48***	***23.26***

Notes: Bold post-hoc Bayes factors indicate that they exceed 3.

Table 5: Mean, standard deviation and Bayes factors (BF) for the ***run*** movement from experiment 3.

	High		**Low**		**Trainer**		**BF Main effect**	**BF post-hoc**		
	Mean	*SD*	**Mean**	*SD*	**Mean**	*SD*		**High vs Low**	**High vs Trainer**	**Low vs Trainer**
Peak anterior force (BW)	-2.49	0.51	-1.96	0.62	-1.95	0.32	187.85	***11.49***	***50.00***	0.23
Anterior/posterior force impulse (BW·s)	-0.12	0.10	-0.07	0.09	-0.07	0.06	13.39	***4.18***	***6.45***	0.23
Peak compressive force (BW)	9.40	1.48	8.48	1.00	8.28	1.30	28.94	***4.13***	***4.39***	0.30
Compressive force impulse (BW·s)	1.32	0.14	1.19	0.12	1.12	0.14	521.13	***33.33***	***23.81***	0.56
Sagittal plane moment impulse (Nm/kg·s)	0.36	0.07	0.32	0.06	0.29	0.07	442.99	***18.87***	***9.62***	1.05
Peak external rotation velocity (°/s)	154.11	20.81	141.41	18.13	164.13	30.27	3.40	0.95	0.33	***23.26***
Peak plantarflexion velocity (°/s)	-497.12	33.09	-530.79	16.75	-607.92	25.15	7928259	***5.03***	***250000***	***2040.82***
Peak eversion velocity (°/s)	-217.36	50.85	-208.96	62.79	-184.20	50.44	11.88	0.36	***3.83***	1.56
Peak lateral gastrocnemius force (BW)	0.43	0.16	0.31	0.07	0.30	0.05	49.27	***2.67***	***5.88***	0.24
Lateral gastrocnemius impulse (BW·ms)	37.20	10.90	26.30	8.60	27.30	6.74	624.50	***6.80***	***125.00***	0.26
Medial gastrocnemius impulse (BW·ms)	159.70	66.67	132.50	49.53	129.50	41.68	5.23	***6.49***	1.20	0.24
Peak soleus force (BW)	4.70	0.51	4.38	0.33	4.36	0.34	11.79	***3.53***	***6.54***	0.24
Soleus impulse (BW·ms)	569.40	71.12	504.10	81.00	475.70	81.89	44.60	***8.77***	***12.82***	0.36
ATFL peak strain (%)	3.61	4.02	5.87	1.99	7.86	2.89	149.87	1.31	***9.52***	***12.50***
Peak ATFL strain velocity (%/s)	345.61	25.29	362.75	16.19	392.86	11.39	4318067.79	2.04	***2331.00***	***6622.52***
Peak PTFL strain velocity (%/s)	188.01	41.03	182.46	35.72	175.60	37.13	4.99	0.39	***3.00***	0.63

Notes: Bold post-hoc Bayes factors indicate that they exceed 3.

Table 6: Mean, standard deviation and Bayes factors (BF) for the ***vertical jump*** movement from experiment 3.

	High		**Low**		**Trainer**		**BF Main effect**	**BF post-hoc**		
	Mean	*SD*	**Mean**	*SD*	**Mean**	*SD*		**High vs Low**	**High vs Trainer**	**Low vs Trainer**
Peak compressive force (BW)	5.00	1.08	5.59	0.95	5.25	1.27	6.37	***4.72***	0.63	0.63
Peak dorsiflexion (°)	22.96	7.17	24.62	7.11	25.72	8.29	3.56	1.07	***16.95***	0.32
Extensor digitorum longus impulse (BW·ms)	0.04	0.03	0.02	0.02	0.02	0.02	14.94	1.59	***9.52***	0.24
Extensor hallucis longus impulse (BW·ms)	0.01	0.01	0.01	0.00	0.01	0.01	7.24	1.37	***6.76***	0.25
Flexor digitorum longus impulse (BW·ms)	0.01	0.00	0.01	0.00	0.01	0.00	151.30	1.46	***250.00***	0.75
Peak flexor hallucis longus force (BW)	0.11	0.02	0.11	0.01	0.09	0.01	10.50	0.26	2.21	***250.00***
Flexor hallucis longus impulse (BW·ms)	0.01	0.00	0.01	0.00	0.01	0.01	115.76	1.07	***249.00***	0.90
Peroneus brevis impulse (BW·ms)	0.03	0.01	0.02	0.01	0.02	0.02	7.49	***3.88***	0.97	0.69
Peak tibialis posterior force (BW)	0.48	0.10	0.69	0.24	0.43	0.19	5.09	1.46	0.25	***6.10***
Peroneus tertius impulse (BW·ms)	0.01	0.00	0.01	0.00	0.01	0.00	81.36	***3.66***	***18.52***	0.26

Notes: Bold post-hoc Bayes factors indicate that they exceed 3.

Table 7: Mean, standard deviation and Bayes factors (BF) from experiment 4.

	No-brace		Sleeve		Brace		BF Main effect	BF post-hoc		
	Mean	*SD*	**Mean**	*SD*	**Mean**	*SD*		**No-brace vs Sleeve**	**No-brace vs Brace**	**Sleeve vs Brace**
Medial/lateral force impulse (BW·s)	0.16	0.06	0.14	0.06	0.12	0.09	10.20	***10.55***	***3.08***	0.55
Peak coronal plane moment (Nm/kg)	1.48	0.28	1.55	0.48	1.15	0.45	3.44	0.23	1.07	***28.47***
Coronal plane moment impulse (Nm/kg·s)	0.19	0.04	0.20	0.07	0.15	0.06	3.38	0.23	0.96	***34.86***
Peak inversion (°)	13.28	4.44	14.64	4.17	10.83	5.53	3.29	0.30	0.55	***5.83***
Extensor hallucis longus (BW·s)	2.98	1.74	2.79	1.45	6.59	4.65	44.09	0.24	***3.16***	***5.08***
Flexor hallucis longus impulse (BW·s)	5.31	2.27	4.83	1.64	10.46	7.14	9.82	0.35	1.86	***3.06***
Peroneus longus impulse (BW·s)	252.96	40.60	241.98	43.89	213.12	32.75	28.90	0.56	***5.18***	2.21
PTFL peak strain (%)	16.27	5.08	17.05	4.25	15.42	4.72	3.35	1.10	0.44	***3.53***

Notes: Bold post-hoc Bayes factors indicate that they exceed 3.