Effects of tennis shoes and running footwear on anterior cruciate ligament loading during a maximal change of direction manoeuvre.

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Key words: Biomechanics, anterior cruciate ligament, sports, footwear, injury.

Abstract

The aim of the current investigation was to examine the effects of tennis shoes and running footwear on the loads experienced by the ACL during a maximal change of direction task. Thirteen male participants performed maximal change of direction movements in tennis shoes and running footwear. Lower limb kinematics were collected using an 8 camera motion capture system and ground reaction forces were quantified using an embedded force platform. Anterior cruciate ligament (ACL) loading was examined via a musculoskeletal modelling approach and the frictional properties of the footwear were examined using ground reaction force information. Differences in ACL loading parameters between footwear were examined using one-way repeated measures ANOVA and multiple regression analyses were used to determine frictional predictors of ACL loading. Peak ACL force was significantly larger in the tennis shoes (2308.35 N) in relation to running footwear (1859.21 N) conditions. In addition, it was shown that the peak rotational moment was a significant predictor of peak ACL force in the tennis shoes (Adjusted R^2 = 0.68) and running footwear (Adjusted R^2 = 0.61) conditions. The findings from the current investigation indicate that the specific tennis shoes examined in the current investigation may place athletes at increased risk from ACL pathology during maximal change of direction movements. However, further exploration

using a more ecologically valid research design is required before this notion can be truly substantiated.

Introduction

Racquet court sports such as tennis are associated with repeated high-intensity, intermittent movement bouts, with rallies of between 5–20 seconds (Fernandez et al., 2006). Whilst biomechanical literature has predominantly focussed on linear running, this mode of locomotion is not ecologically relevant to the majority of sporting movements, particularly in court sports (Lees, 2003). Court sports such as tennis require players to perform an array of different movements including jumping, and rapid changes of direction/ cutting manoeuvres (Hewit et al., 2013). The ability to quickly change direction is important for effective performance in racquet court sports, allowing players more time to execute their strokes and providing a mechanism to gain positional advantage on the court (Baker & Newton, 2008).

Tennis is associated with a high rate of knee pathologies in relation to other athletic disciplines, with the anterior cruciate ligament (ACL) accounting for 11% of all knee injuries (Majewski et al., 2006). ACL injuries are extremely serious and can lead to long term absence from competitive sport (Olsen et al., 2004). ACL injuries typically lead to long term discomfort at the knee, which forces many athletes to permanently withdraw from training/ competition. Indeed, Roos et al., (1995) demonstrated that only 30 % of competitive footballers remained active 3 years after suffering an ACL injury. Even after full physiological recovery from ACL injury, athletes typically fail to return to their previous levels of function, as statistically significant performance decrements have been observed in relation to non-injured controls (Carey et al., 2006).

In addition, athletes who experience an ACL pathology are statistically more likely to experience degenerative knee osteoarthritis in relation to non-injured controls (Øiestad et al., 2009). Thus experiencing an ACL injury serves to reduce engagement with sport/ physical activity but also leads to chronic pain and disability in later life (Ajuied et al., 2014). In the US alone over 175,000 ACL reconstruction surgeries are conducted annually, with directly associated costs in excess of over $2 billion and total allocated costs of $3.4 billion (Gottlob et al., 1999).

Injuries to the ACL are predominantly non-contact in nature, in that the ligament is damaged without any physical interaction between athletes (Boden et al., 2009). ACL pathologies occur mechanically when excessive loading is experienced by the ligament itself (Smith et al., 2012). Non-contact ACL pathologies typically involve decelerations, cutting movements, sudden changes of direction, or landings from a jump (Olsen et al., 2004). Athletes are particularly at risk when the foot is in an everted closed-chain position at footstrike, the tibia is rotated internally, and the knee is minimally flexed (Shimokochi & Shultz, 2008).

Like all footwear, tennis shoes are designed in order to improve performance and to attenuate injury. The mechanical characteristics of tennis shoes are traditionally designed specifically in order to attenuate axial impact loading and promote lateral stability. In addition to this, the rapid changes of direction that are commonplace during tennis, means that friction at the outsole-surface interface is important to reduce undesirable levels of movement of the shoe relative to the surface (Carre et al., 2014). The frictional properties of athletic footwear are typically investigated in biomechanical analyses using both the peak translational coefficient of friction and rotational friction moment (Frederick, 1993). In tennis in particular, the frictional characteristics of sports footwear can affect both performance and the risk of injury (Frederick, 1993). Excessive friction can lead to injury due to overloading of the soft tissues in the lower extremities (Thomson et al., 2015), whereas insufficient friction can cause excessive foot motion relative to the surface, which causes decrements in performance (Frederick, 1993).

Tennis players typically wear either court specific footwear or running shoes, however tennis footwear has received relatively little attention in biomechanical literature. Luethi et al., (1986) investigated the effects of tennis shoes with flexible and stiff midsoles, during a lateral hopping task. Their results indicated that the flexible footwear condition was associated with significantly larger peak vertical impact forces and peak angles of foot inversion. Strauss et al., (2009) explored the effects of multi-court, hard, grass and clay court specific tennis footwear during a running forehand drive, on hard, grass and clay surfaces. Their findings showed that on a hard court the specific footwear reduced the vertical load rate in comparison to the multi-court footwear. Conversely on the grass court, the specific footwear increased the vertical load rate in comparison to the multi-court footwear. Herbaut et al., (2015) identified the effects tennis shoe drop on the kinetics and kinematics of junior tennis players during an open-stance forehand. Their results indicated that the lower drop footwear condition was associated with a reduced vertical impact peak and also a less dorsiflexed ankle angle at the instance of foot contact. Finally, Sinclair, (2017) examined the effects of court specific, minimalist and running trainers during a change of direction task. The findings showed that the instantaneous load rate and peak tibial accelerations were significantly larger in the minimalist and court specific footwear compared with the running trainers. In addition, the peak angle of inversion was revealed to be significantly larger in the minimalist compared to the court footwear and running trainers. However, there is currently no quantitative information relating to the effects of tennis footwear on the loads experienced by the ACL during change of direction movements.

Therefore, the aim of the current investigation was to examine the effects of tennis shoes and running footwear on the loads experienced by the ACL during a maximal change of direction task. Research of this nature may provide important new information to athletes regarding the selection of appropriate footwear for the prevention of ACL injuries during tennis based activities.

Methods

Participants

Thirteen male court athletes volunteered to take part in this study. The mean characteristics of the participants were: age 23.15 ± 2.66 years, height 177.91 ± 4.55 cm and body mass 75.11 ± 5.74 kg. All were free from lower extremity pathology at the time of data collection and provided written informed consent. The procedure was approved by a University ethics committee STEMH 512.

Experimental footwear

The footwear used during this study consisted of, running footwear (New Balance 1260 v2), and tennis shoes (Hi-Tec Indoor Lite) (shoe size 8–10 in UK men's sizes) (Figure 1). The running footwear had an average mass of 0.285 kg, heel thickness of 25 mm and a heel drop of 14 mm. The running footwear tread pattern was a mixture of circular and elliptical grooves with a discontinuity between the rear and forefoot components. Whereas the tennis shoes had an average mass of 0.368 kg, heel thickness of 28 mm and a heel drop of 10 mm. The tennis shoes tread pattern was predominantly a curved herringbone configuration and also had discontinuity between the rear and forefoot components.

@@@ FIGURE 1 NEAR HERE @@@

Procedure

Participants were instructed to perform maximal 180° cutting manoeuvres whilst striking an embedded force platform (Kistler, Kistler Instruments Ltd., Alton, Hampshire; length, width, height = 0.6 x 0.4 x 0 m) with their right (dominant foot) foot. The force platform sampled at 1000 Hz. Participants commenced their trials from 6 m away from the force platform. This distance was selected as being approximately half the width of a tennis court and was deemed to be typical of the distances that tennis players may be expected to run and then change direction (Sinclair, 2017). Participants ran straight ahead for 6 m then planted their dominant foot on the force plate, and then changed direction to move 180° to their initial direction of motion. The stance phase was delineated as the duration over which > 20 N of vertical force was applied to the force platform (Sinclair et al, 2011).

Participants were given time to familiarize themselves with the experimental setup, this was conducted until they were able to confidently achieve the required foot position on the force platform. Five successful trials were obtained in each footwear condition. A successful trial was defined as one in which the foot made full contact with the force platform and there was no evidence of gait modifications due to the experimental conditions. The order in which participants performed in each footwear condition was counterbalanced. To ensure that participants utilized a similar approach velocity in each of the experimental footwear; the linear velocity of the pelvic segment was quantified. The approach velocity during the first trial was calculated and a maximum deviation of 5 % from this velocity was allowed throughout data collection for each participant.

Kinematics and ground reaction force information were synchronously collected. Kinematic data were captured at 250 Hz via an eight camera motion analysis system (Qualisys Medical AB, Goteburg, Sweden). Lower extremity segments were modelled in 6 degrees of freedom using the calibrated anatomical systems technique (Cappozzo et al., 1995). To define the segment co-ordinate axes of the right foot, shank and thigh, retroreflective markers were placed unilaterally onto the 1st metatarsal, 5th metatarsal, calcaneus, medial and lateral malleoli, medial and lateral epicondyles of the femur. To define the pelvis segment further markers were positioned onto the anterior (ASIS) and posterior (PSIS) superior iliac spines. Carbon fiber tracking clusters were positioned onto the shank and thigh segments. The foot was tracked using the 1st metatarsal, 5th metatarsal and calcaneus markers and the pelvis using the ASIS and PSIS markers. The centers of the ankle and knee joints were delineated as the mid-point between the malleoli and femoral epicondyle markers, whereas the hip joint centre was obtained using the positions of the ASIS markers. This method placed the hip joint centre 14% of the ASIS breadth medially, 19% posteriorly, and 30% distally from the ipsilateral (Right) ASIS (Bell et al., 1999). Static calibration trials were obtained in each footwear allowing for the anatomical markers to be referenced in relation to the tracking markers/ clusters. The Z (transverse) plane was oriented vertically from the distal segment end to the proximal segment end. The Y (coronal) plane was oriented in the segment from posterior to anterior. Finally, the X (sagittal) plane orientation was determined using the right hand rule and was oriented from medial to lateral.

Processing

Dynamic trials were digitized using Qualisys Track Manager in order to identify anatomical and tracking markers then exported as C3D files to Visual 3D (C-Motion, Germantown, MD, USA). Ground reaction force and kinematic data were smoothed using cut-off frequencies of 25 and 12 Hz with a low-pass Butterworth 4th order zero lag filter. Euler knee joint angles were calculated using an XYZ sequence of rotations and knee joint moments were calculated using Newton-euler inverse dynamics within Visual 3D.

A musculoskeletal modelling approach was utilized to quantify ACL loading, as described and validated by Dai & Yu, (2012). This approach has been shown to be sufficiently sensitive to resolve differences in ACL force during different movements (Dai & Yu, 2012) and also as a function of different prophylactic mechanisms (Sinclair & Taylor, 2017). The face validity of the current model has been evaluated from three key aspects in the literature. Firstly, Dai & Yu, (2012) showed that the model exhibited a high level of consistency with the values provided from in vivo ACL loading investigations (Cerulli et al., 2003; Taylor et al., 2011). Secondly, the timing of ACL injuries in dynamic tasks occurs within the first 50 ms after the initial foot contact (Krosshaug et al., 2007). The timing of the peak ACL force estimated using this model by both Dai & Yu, (2012) and Sinclair et al., (2017) was shown to be < 50

ms, which is consistent with this data and supports the face validity of the model. Thirdly Brown et al., (2012) demonstrated that landing with increased knee flexion reduced in vivo peak ACL loading. The data provided by Dai & Yu, (2012) supported this notion as they showed that peak ACL force was greater when landing with reduced knee flexion.

Firstly, the tibia-anterior shear force (TASF) was calculated, which was undertaken using a modified version of the model described in detail by Devita & Hortobagyi, (2001). Our model differed only in that gender specific estimates of posterior tibial plateau slope (Hohmann et al., 2011), hamstring-tibia shaft angle (Lin et al., 2009) and patellar tendon-tibia shaft angle (Nunley et al., 2003) were utilized.

ACL loading was determined in accordance with the below equation. Key input parameters into this model where TASF, transverse plane knee moment, coronal plane knee moment and also in vitro information based on the data of Markolf et al., (1995), which were extrapolated as a function of the knee flexion angle measured during the current study. The first component (*F100*) of the above equation was mediated via by the TASF. ACL forces caused by a 100 N TASF at different knee angles were obtained by digitizing and fitting a polynomial curve to the data described by Markolf et al., (1995), who examined ACL forces in vitro when a 100 N TASF was applied to cadaver knees from 0-90° of knee flexion. *F100* was extrapolated using the knee flexion data from the current investigation. The second component (*F10TV*) was caused by the knee transverse plane moment. The ACL forces caused by a 10 Nm transverse plane knee moment, across the different knee angles were obtained by digitizing and fitting a polynomial curve to the data of Markolf et al. (1995). *F10TV* was similarly extrapolated as a function of the knee flexion data from the current investigation. The final aspect (*F10CR*) was caused by the knee coronal plane moment. The ACL forces caused by a 10 Nm coronal plane knee moment, across the different knee angles were again obtained by digitizing the data reported by Markolf et al. (1995). *F10CR* was extrapolated using the knee flexion data from the current investigation.

$$\text{ACL load} = (F100 / 100 * \text{TASF}) + (F10TV / 10 * \text{transverse plane knee moment}) + (F10CR / 10 * \text{coronal plane knee moment})$$

From the musculoskeletal model, peak ACL force (N) was extracted. In addition, ACL average (N/s) and instantaneous load rates (N/s) were quantified. Average load rate was obtained by dividing the peak ACL force by the duration over which the peak force occurred and instantaneous load rate was quantified as the peak increase in force between adjacent data points. Finally, ACL impulse (N·s) during the stance phase was quantified using a trapezoidal function.

In addition, 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 (Stiles & Dixon, 2006). The peak rotational moment of the ground reaction force (Nm) was used to describe the rotational friction characteristics of the footwear (Holden & Cavanagh, 1991).

Statistical analyses

Means, standard deviations (SD) and 95 % confidence intervals (95% CI) were calculated for each outcome measure for both footwear conditions. Differences in ACL loading parameters between footwear were examined using one-way repeated measures ANOVAs. Effect sizes were calculated using partial eta^2 ($p\eta^2$). The data was screened for normality using a Shapiro-Wilk which confirmed that the normality assumption was met. In addition, multiple regression analyses with peak ACL force as criterion and peak translation coefficient of friction and peak rotational moment as predictor variables were conducted for each footwear condition using a forward stepwise procedure. An alpha level of $P \leq 0.05$ was used throughout as the criterion for statistical significance (Sinclair et al., 2013), and statistical actions were conducted using SPSS v23.0 (SPSS Inc., Chicago, USA).

Results

Tables 1-2 and figure 2 present the ACL loading parameters that were obtained as a function of the different footwear conditions examined as part of this investigation.

@@@ TABLE 1 NEAR HERE @@@

@@@ FIGURE 2 NEAR HERE @@@

ACL loading parameters

It was revealed that peak ACL force was significantly ($P = 0.009$, $p\eta^2 = 0.55$) larger in the tennis shoes in relation to the running footwear. In addition, ACL average load rate was significantly ($P = 0.004$, $p\eta^2 = 0.63$) larger in the tennis shoes in relation to the running footwear. Finally, ACL instantaneous load rate was significantly ($P = 0.002$, $p\eta^2 = 0.69$) larger in the tennis shoes compared to the running footwear.

Frictional parameters

@@@ TABLE 2 NEAR HERE @@@

Peak rotational moment was significantly ($P = 0.003$, $p\eta^2 = 0.64$) larger in the tennis shoes in relation running footwear. In addition, peak translational coefficient of friction was significantly ($P = 0.003$, $p\eta^2 = 0.63$) greater in the running footwear in relation to the tennis shoes.

Regression analyses

The multiple regression analyses showed that for the tennis shoes (Adjusted $R^2 = 0.68$, $P < 0.05$), and running footwear (Adjusted $R^2 = 0.61$, $P < 0.05$) the peak rotational moment was a significant predictor of peak ACL force.

Discussion

The aim of the current investigation was to examine the effects of tennis shoes and running footwear on the loads experienced by the ACL during a maximal effort change of direction task. To the authors knowledge this represents the first comparative investigation to quantify the effects of different tennis footwear on ACL loading during a change of direction movement. Quantitatively investigating the parameters linked to the aetiology of ACL injury may provide tennis players with key clinical information regarding the selection of appropriate footwear for their training/ competition.

Importantly the current investigation showed that ACL loading parameters were significantly greater in the tennis shoes in relation to the running footwear. The mechanical aetiology of ACL injury in athletic populations is caused by excessive loading of the ACL itself (Smith et al., 2012). ACL injuries are considered to be extremely serious and habitually require reconstructive intervention leading to long term absences from competition (Myklebust & Bahr, 2004). Therefore, given the statistical increases in ACL loading in the tennis shoes, the results from the current observation may be clinically relevant for tennis based athletes. It can be conjectured based on the findings from this investigation that the specific tennis shoes examined in this investigation may increase the risk from ACL injury during sport specific change of direction movements.

In addition, it was also revealed that the tennis shoes were statistically associated with the highest values for the peak rotational moment and the lowest values for the peak translational coefficient of friction in relation to the running footwear. A likely explanation for this observation is based on the tread patterns of each shoe outsole which are distinct between the three footwear examined as part of the current study (Figure 1) (Valiant et al., 1985). This observation concurs with the observations of Severn, et al., (2011) and Wannop & Stefanyshyn, (2015) which indicates that manipulating the outsole patterns of different footwear can alter both rotational and translational friction characteristics.

It also appears based on the findings from the current analysis that the tennis shoes were effective in enhancing rotational friction but not optimal in promoting translational friction. The frictional properties between the shoe and surface are an important determinant of athletic performance, but high levels of friction at the outsole-surface interface may also be related to increased risk of soft tissue injury (Wannop et al., 2009). There is currently no agreement regarding the optimal frictional values that are required to provide sufficient traction, but also attenuate risk from injury during sports movements (Frederick, 1993). Importantly the current investigation showed that the rotational friction moment as opposed to the translational coefficient of friction was a significant predictor of the peak ACL force in all of the experimental footwear. This supports the proposition of Thomson et al., (2015) and indicates that during maximal change of direction tasks the peak rotational moment is the most clinically meaningful frictional parameter in relation to the development and prevention of ACL pathologies.

A potential limitation to the current analysis is that ACL loading parameters were quantified using a musculoskeletal modelling approach. This was a requirement of the current investigation given the impracticalities of obtaining in vivo measures of ligament loading during dynamic movements. Although the current model has been shown to exhibit good face validity (Dai & Yu, 2012), musculoskeletal models by definition are always subject to some mathematical assumptions that may compromise their efficacy across a range of participants. A further potential drawback is the laboratory based nature of the data collection protocol. Specifically, the stiffness and frictional properties of the laboratory surface are likely to be distinct from those experienced in field based testing scenarios in which participants perform tennis specific movements in realistic conditions. The current investigation utilized a repeated measures design and thereby the statistical comparison between footwear is sound, as participants performed in the same conditions in both footwear. However, the ecological validity of the procedure from a practical context was compromised as ACL loading may have differed had participants performed on a tennis specific surface. Therefore, it is strongly recommended that the current investigation be repeated using a field based data collection protocol.

In conclusion; although the biomechanical effects of tennis shoes have been examined previously; current knowledge regarding differences in ACL loading when performing change of direction tasks is limited. The current investigation thus adds to the current literature base by performing a comprehensive evaluation of ACL loading parameters when performing a change of direction task in tennis shoes and running footwear. Importantly, the current study showed ACL loading parameters were significantly greater in tennis shoes in relation to the running footwear. The findings from the current investigation indicate that the specific tennis shoes examined as part of this investigation may place athletes who undertake court based activities at increased risk from ACL pathology during maximal change of directions movements. However, further exploration using a more ecologically valid research design is required before this notion can be truly substantiated.

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Table 1: ACL loading parameters (mean, SD & 95% CI's) as a function of the experimental footwear conditions.

	Court			Running footwear		
	Mean	SD	95% CI	Mean	SD	95% CI
Peak ACL load (N)	2308.35	380.01	2036.51-2580.20	1859.21	395.80	1576.07-2142.34
Time to peak ACL force (ms)	48.20	14.74	37.70-58.78	49.80	13.81	39.88-59.65
ACL average load rate (N/s)	54295.37	12832.58	45115.49-63475.24	42930.23	10059.78	35733.89-50126.56
ACL instantaneous load rate (N/s)	147762.11	41376.27	118163.31-177360.91	103200.24	24934.95	85362.85-121037.63
ACL impulse (N·s)	330.14	87.71	267.40-392.89	312.25	65.62	265.32-359.19

Table 2: Frictional parameters (mean, SD & 95% CI's) as a function of the experimental footwear conditions.

	Court			Running footwear		
	Mean	SD	95% CI	Mean	SD	95% CI
Peak rotational moment (Nm)	24.63	7.25	17.39-29.71	19.56	6.52	14.49-23.91
Peak translational coefficient of friction (µ)	0.57	0.07	0.53-0.63	0.64	0.08	0.58-0.70