Gender specific ACL loading patterns during the fencing lunge: Implications for ACL injury risk

Sinclair J[1] & Bottoms L[2]

1. *Centre for Applied Sport and Exercise Sciences, School of Sport and Wellbeing, Faculty of Health & Wellbeing, University of Central Lancashire, Lancashire, UK.*
2. *Department of Psychology and Sports Sciences, Centre for Research in Psychology and Sport Sciences, University of Hertfordshire, Hertfordshire, UK.*

Correspondence Address:

Dr. Jonathan Sinclair

Darwin Building

Centre for Applied Sport Exercise and Nutritional Sciences

School of Sport and Wellbeing

Faculty of Health & Wellbeing

University of Central Lancashire

Preston

Lancashire

PR1 2HE.

Abstract

Introduction

Fencing is an Olympic sport which requires the fencer to strike the opponent with their sword to score a hit (Turner et al., 2013). Fencing represents a high intensity and intermittent discipline that necessitates short bouts of high intensity exercise and periods of relatively low intensity activity. Bounces, steps and lunges occur frequently during the competition for the purposes of defence and attack, which place high demands the musculoskeletal system (Bottoms et al., 2011).

Epidemiological analyses have documented that injuries and pain associated with fencing training/ competition were apparent in 92.8% of fencers, with the majority of these injuries being experienced in the lower extremities (Harmer, 2008). Harmer (2008) showed that the knee was the most commonly injured musculoskeletal site in fencers, accounting for 19.6% of all pathologies with particular concern relating to the anterior cruciate ligament (ACL). The data of Mountcastle et al., (2007) support this notion indicating that the ACL was a common injury location in military recruits involved in fencing training/ competition.

The ACL one of the 4 predominant ligaments that are effective in providing stability to the knee joint. The primary function of the ACL is to resist anterior tibial translation (ATT), providing 87% of the total restraining force at 30° of knee flexion (Butler et al., 1980). The ACL also prevents excessive knee extension, knee adduction and abduction movements, and resists internal rotation of the tibia (Liu-Ambrose, 2003). Injuries to the ACL are debilitating, cause long term cessations from training/ competition and may ultimately be career threatening as current treatment modalities do always successfully return athletes to their previous levels of functionality (Ardern et al., 2011). ACL injuries are also associated with long term health implications, with athletes being up to 10 times more likely to develop early-onset degenerative knee osteoarthritis in relation to non-injured controls (Øiestad et al., 2009), leading not only to a reduction in sports activity but also chronic incapacity in later life (Ajuied et al., 2014). ACL injuries traditionally necessitate surgical intervention in order to restore function, followed by a significant and aggressive period of rehabilitation. Gottlob et al., (1999) determined that over 175,000 ACL surgeries are performed each year in the US with directly associated costs of over $2 billion.

The majority of ACL injuries (72%) are non-contact in nature, in that injury occurs without physical contact between athletes (Boden et al., 2009). Mechanically, ACL injuries manifest when an excessive loading is experienced by the ACL (Smith et al., 2012). Non-contact ACL injuries habitually occur at the point of foot strike with the knee close to full extension in athletic disciplines where sudden decelerations, landing and pivoting manoeuvres are repeatedly performed (Olsen et al., 2004). It has been demonstrated that most non-contact ACL injuries occur in activities that involve single-limb decelerations (Boden et al., 2009). The lunge is the most frequently used attack in fencing (Sinclair & Bottoms, 2013). However, the front leg must produce a rapid deceleration action on landing to stabilize the fencer (Sinclair et al., 2010), thus it appears that the lunge movement may be the movement that imposes that placers fencers at greatest risk from ACL pathologies.

Whilst male and female fencers often train concurrently fencing competitions are gender specific. Importantly, Harmer, (2008) showed that female fencers had a 35% greater risk for time-loss injuries in relation to males. Furthermore, ACL injuries are renowned for being prevalent in female athletes, with an incidence rate in the region of 4-10 times that noted in males (Arendt et al., 1999). The enhanced risk for ACL injury in female athletes has led to a significant amount of research attention focussed on the mechanical factors responsible for the gender disparity in the rate of ACL injuries. Gender differences in lower body mechanics in fencing have received only limited attention in biomechanical literature. Sinclair & Bottoms, (2013) examined gender differences in lower extremity kinematics during the fencing lunge. Their findings showed that females produced significantly greater knee abduction and hip adduction of the lead limb during the lunge. Furthermore, Sinclair et al., (2014) investigated gender specific loading of the Achilles tendon during the lunge movement. They demonstrated that males exhibited significantly greater Achilles tendon loading in comparison to females. However, gender differences in ACL loading during the fencing lunge have yet to be explored, thus gender specific risk for ACL injury in fencers is currently unknown.

Therefore, the aim of the current investigation was to determine whether gender differences in ACL loading linked to the aetiology of injures are evident during the fencing lunge. Research of this nature may provide important clinical information regarding potential ACL injury risk in fencers.

Methods

Participants

Ten male participants and ten female participants volunteered to take part in this investigation (all were right hand dominant). All were injury free at the time of data collection and provided written informed consent in accordance to guidelines outlined in the declaration of Helsinki. Participants were active competitive fencers who engaged in training a minimum of 3 training sessions per week. The mean characteristics of the participants were males; age 29.18 ± 4.30

years, height 1.79 ± 0.05 m and mass 75.33 ± 6.28 kg and females; age 23.04 ± 5.57 years, height 1.67 ± 0.06 m and mass 63.57 ± 3.66 kg. The procedure was approved by the University of Central Lancashire ethics committee.

Procedure

Participants were required to complete 5 lunges hitting a dummy with their weapon whilst returning to a starting point (pre-determined by each participant prior to the commencement of data capture) following each trial to control lunge distance. In addition to striking the dummy with their weapon participants also made contact with a force platform (Kistler, Kistler Instruments Ltd., Alton, Hampshire) embedded in the floor (Altrosports 6mm, Altro Ltd,) of a biomechanics laboratory with their right (lead) foot. The starting point for the movement was adjusted and maintained for each participant. Kinematics and ground reaction force data were synchronized using an analogue to digital interface board. The lunge movement was delineated as the period from foot contact (defined as > 20 N of vertical force applied to the force platform) to the instance of maximum knee flexion.

An eight camera motion analysis system (Qualisys^{TM} Medical AB, Gothenburg, Sweden) captured kinematic data. Calibration of the motion analysis system was performed before each data collection session. Only calibrations which produced average residuals of less than 0.85 mm for each camera for a 750.5 mm wand length and points above 4000 were accepted prior to data collection.

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 (Graydon et al., 2015; Sinclair et al., 2015), whereas the hip joint centre was obtained using the positions of the ASIS markers (Sinclair et al., 2014). Static calibration trials (not normalized to static trial posture) were obtained for the anatomical markers to be referenced in relation to the tracking markers/ clusters. 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.

Processing

Dynamic trials were processed using Qualisys Track Manager and then exported as C3D files. GRF and marker data were filtered at 50 Hz and 15 Hz respectively using a low-pass Butterworth 4th order filter and processed using Visual 3-D (C-Motion, Germantown, MD, USA). Joint moments were computed using Newton-Euler inverse-dynamics, allowing net knee joint moments to be calculated. Angular kinematics were calculated using an XYZ (sagittal, coronal and transverse) sequence of rotations (Sinclair et al., 2014). To quantify knee joint moments segment mass, segment length, ground reaction force and angular kinematics were utilized.

A musculoskeletal modelling approach was utilized to quantify ACL loading during the lunge movement. To accomplish this we firstly had to quantify the tibia-anterior shear force (TASF), 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 as the sum of ACL forces caused by the TASF, transverse plane knee moment, and transverse plane knee moment in accordance with EQ[1].

EQ[1] - **ACL load =** (*F100* / 100 * TASF) + (*F10TV* / 10 * transverse plane knee moment) + (*F10CR* / 10 * transverse plane knee moment)

The components of EQ[1] were obtained using the data described by Markolf et al., (1995), who examined ACL forces in vitro when a 100 N TASF (*F100*) was applied to cadaver knees from 0-90˚ of knee flexion. ACL forces were also measured when additional torques of 10 Nm in the coronal (*F10CR*) and transverse (*F10TV*) planes were combined with the 100 N TASF from 0-90˚ of knee flexion.

All force parameters were normalized by dividing the net values by body mass (N/kg). From the musculoskeletal models indices of peak ACL and TASF forces were extracted. In addition ACL and TASF instantaneous load rates (N/kg/s) were quantified as the peak increase in force between adjacent data points. In addition we also calculated the ACL impulse N/kg·s) during the lunge movement by multiplying the ACL load by the duration over which the movement occurred.

Analyses

Descriptive statistics of means, standard deviations (SD) and 95% confidence intervals (95% CI) were calculated. Gender differences in ACL loading parameters were examined using independent samples t-tests with significance accepted at the $P \leq 0.05$ level (Sinclair et al., 2013). Effect sizes were quantified using partial eta squared ($p\eta^2$). Shapiro-Wilk tests confirmed that the data were normally distributed in all cases. All statistical procedures were conducted using SPSS v23 (SPSS Inc., Chicago, IL, USA).

Results

Table 1 and figure 1 present the gender differences in ACL loading during the fencing lunge movement. The results indicate that ACL loading parameters were significantly influenced by gender.

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Peak TASF was found to be significantly ($t_{(9)} = 2.65$, $P<0.05$, $p\eta^2 = 0.29$) larger in female fencers in relation to males (Table 1; Figure 1a). In addition peak ACL was found to be significantly ($t_{(9)} = 2.65$, $P<0.05$, $p\eta^2 = 0.35$) larger in females in comparison to males (Table 1; Figure 1b).

TASF instantaneous load rate was also found to be significantly ($t_{(9)} = 2.65$, $P<0.05$, $p\eta^2 = 0.24$) higher in female fencers in compared to males (Table 1). ACL instantaneous load rate was similarly shown to be significantly ($t_{(9)} = 2.65$, $P<0.05$, $p\eta^2 = 0.26$) larger in females in comparison to males (Table 1). Finally, it was demonstrated that ACL impulse was significantly ($t_{(9)} = 2.65$, $P<0.05$, $p\eta^2 = 0.38$) greater in females in relation to male fencers (Table 1).

Discussion

The aim of this investigation was to investigate gender differences in ACL loading during the fencing lunge. To the authors knowledge this study represents the first quantitative examination of the ACL loading during fencing specific manoeuvres. Research of this nature may provide important clinical information regarding potential ACL injury risk in fencers.

The primary observation from the current study is that ACL loading parameters were found to be significantly larger in female fencers. Females exhibit distinct knee mechanics during deceleration/ landing tasks, involving reduced knee flexion, increased hip rotation/ adduction and knee valgus (Shimokochi & Shultz, 2008). Female athletes are regarded as being over reliant on the anterior kinetic chain due to diminished neuromuscular control in the posterior chain (Hewett et al., 2010). The knee posterior kinetic chain musculature, in particular the hamstring group are considered a synergist with the ACL and serve to mediate ATSF by pulling the tibia posteriorly (Hewett et al., 2010). This may help clarify the mechanism by which increases in ACL loading were observed in female fencers as knee ligament forces are strongly influenced by the ATSF (Shelburne et al., 2004). The lunge is renowned as one of the primary attacking mechanisms in fencing (Sinclair & Bottoms, 2013), thus the observations from the current investigation may have potential clinical relevance regarding the aetiology of injury in female fencers. Mechanically, ACL during dynamic tasks occur when excessive loading is experienced by the ACL itself (Smith et al., 2012). This study therefore provides insight into the increased incidence of ACL injuries in female athletes and also shows that female fencers may be at increased risk from ACL pathologies when performing the lunge movement.

The current study represents the first to quantitatively evidence that female fencers exhibit great ACL loading in relation to males. ACL injuries are one of the most common pathologies in athletic populations (Kiapour & Murray, 2014) and female athletes are considered to be at much greater risk from this injury in relation to males (Arendt et al., 1999). Thus it is important that training/ conditioning adaptations be incorporated by fencing coaches which are designed to decrease the risk from ACL injuries in females. Neuromuscular deficiencies are regarded as a key modifiable risk factor for ACL injuries, and controlling the magnitude of ACL loading through preventive neuromuscular training has been demonstrated as an effective intervention for the modification of ACL injury risk (Mandelbaum et al., 2005). Therefore it is strongly recommended that specific neuromuscular training protocols focussed on the muscles of posterior kinetic chain be implemented for female fencers in order to attenuate their risk from ACL injury.

In conclusion, whilst gender differences in lower extremity biomechanics have received limited information within clinical literature, the effects of gender on ACL loading parameters linked to the aetiology of ACL injuries has not been explored. As such the current study adds to the current literature base in the field of clinical biomechanics by providing a comprehensive analysis of gender specific loading patterns experienced during the fencing lunge. The findings from this investigation showed that female fencers experienced significantly larger ACL loading parameters than males during the lunge movement. Given the association between ACL loading and ACL injury risk, this investigation firstly provides insight into the high incidence of ACL injuries in female athletes and secondly indicates that female fencers may be at increased risk from ACL pathologies. Future analyses should seek to investigate and implement strategies aimed at reducing ACL loading in female fencers.

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Table 1: ACL loading parameters as a function of gender.

	Male			**Female**		
	Mean	***SD***	**95% CI**	**Mean**	***SD***	**95% CI**
Peak ACL load (N/kg)	4.04	0.78	3.27-5.47	6.21	1.26	4.83-7.88
ACL Instantaneous rate of loading (N/kg/s)	378.77	45.12	330.04-427.50	511.18	145.91	376.04-646.31
ACL Impulse (N/kg·s)	1.46	0.21	1.15-2.01	2.22	0.35	1.55-3.08
Peak TASF (N/kg)	3.75	0.46	3.74-4.59	4.61	0.55	4.07-5.32
TASF Instantaneous rate of loading (N/kg/s)	175.61	35.24	122.86-219.37	220.66	51.13	159.48-276.22