Clinical Biomechanics 29 (2014) 138–142

Contents lists available at ScienceDirect

Clinical Biomechanics

journal homepage: www.elsevier.com/locate/clinbiomech

Peak knee adduction moment during gait in anterior cruciate ligament reconstructed females

Matthew R. Patterson [a,c,*], Eamonn Delahunt [a,b], Brian Caulfield [a,c]

[a] School of Public Health, Physiotherapy and Population Science, University College Dublin, Dublin Ireland Health Sciences Centre, Belfield, Dublin 4, Ireland
[b] Institute for Sport and Health, University College Dublin, Dublin, Ireland
[c] Clarity Centre for Sensor Web Technologies, University College Dublin, Ireland

ARTICLE INFO

Article history:
Received 18 February 2013
Accepted 26 November 2013

Keywords:
Gait analysis
walking
ACL
knee
anterior cruciate ligament reconstruction

ABSTRACT

Background: Recent work has shown that anterior cruciate ligament reconstructed patients exhibit an increased peak knee adduction moment during walking gait compared to healthy controls. An increased peak knee adduction moment has been suggested to be a potential mechanism of degeneration for knee osteoarthritis. The few studies in this area have not considered an exclusively female anterior cruciate ligament reconstructed group. This study tested the hypothesis that female anterior cruciate ligament-reconstructed patients would have higher peak knee adduction moments than controls.

Methods: Peak knee adduction moment during walking was compared between a group of anterior cruciate ligament reconstructed females and a group of female activity matched controls over ten 15 m walking trials in a laboratory at a self-selected pace.

Findings: Peak knee adduction moment was lower for the anterior cruciate ligament reconstructed group ($N = 17$, $M = 0.31$ Nm/kg·m, $SD = 0.08$) than for the control group ($N = 17$, $M = 0.41$ Nm/kg·m, $SD = 0.12$; $t(32) = 2.483$, $p = 0.010$, one-tailed, eta squared effect size = 0.16).

Interpretation: A group of female anterior cruciate ligament reconstructed subjects did not exhibit a gait characteristic which has been suggested to be associated with knee osteoarthritis development and has been shown to be present in male and mixed sex anterior cruciate ligament reconstructed populations previously.

1. Introduction

The anterior cruciate ligament (ACL) is important for knee joint stability. It is commonly ruptured in sports that require fast changes in direction as well as single leg landing at awkward angles (Agel et al., 2005; Krosshaug et al., 2007). An ACL rupture is commonly fixed via reconstruction surgery in order to restore knee joint stability. Epidemiological research has shown that nearly half of those who rupture their ACL develop knee osteoarthritis (OA) within 15–20 years post-surgery (Lohmander et al., 2007) and it is estimated that the knee is aged 30 years by an ACL rupture (Lohmander et al., 2004). It has been theorized that this high incidence of knee OA in the ACL reconstructed (ACL-R) population could be a result of abnormal gait mechanics (Bulgheroni et al., 1997; Ferber et al., 2002; Knoll Et Al., 2004). Small deviations in movement patterns during walking cause slightly different loading at the knee and over days, weeks, months and years this continuous, abnormal loading may result in knee joint degeneration and OA.

Most of the research on ACL-R gait patterns has investigated knee joint angular kinematics. ACL-R subjects just after surgery have been shown to have a decreased level of peak knee extension during midstance (Gao and Zheng, 2010; Gokeler et al., 2003). This is to reduce the functional need for the ACL, since one of its major functions is to limit the anterior tibial translation when the knee is in extension. As time from surgery increases ACL-R patients have been shown to regain normal knee extension levels (Webster et al., 2012a). It has been proposed that a more adducted and internally rotated knee position is likely to cause abnormal loading at the knee which may lead to cartilage degeneration over time (Gao and Zheng, 2010; Webster et al., 2012a). However, to date the literature has shown a large level of heterogeneity in terms of ACL-R patients' frontal and coronal plane knee angular kinematics over the gait cycle (Bulgheroni et al., 1997; Butler et al., 2009; Gao and Zheng, 2010; Scanlan et al., 2010).

Knee adduction moment has recently been investigated as a potential factor that may be contributing to the high prevalence of knee OA in the ACL-R population. Knee OA patients have been shown to have higher knee adduction moments than the general population and an increased knee adduction moment has been suggested to be associated with knee OA progression (Amin et al., 2004; Kaufman et al., 2001). It is thought that frontal plane knee malalignment results in an increased distance between the ground reaction force and the knee joint center, causing increased frontal plane knee moments (Hurwitz et al., 2006). These larger frontal plane moments may result in degradation of the medial

* Corresponding author at: G26 Science North, University College Dublin, Belfield, Dublin 4, Ireland.
E-mail address: matt.patterson@ucd.ie (M.R. Patterson).

0268-0033/$ – see front matter © 2013 Elsevier Ltd. All rights reserved.
http://dx.doi.org/10.1016/j.clinbiomech.2013.11.021

tibio-femoral compartment of the knee. The link between knee OA and increased peak knee adduction moment has been only suggested, not proven. Butler et al. (2009) have shown that a mixed group of male and female (3m, 13f) ACL-R patients demonstrated increased peak knee adduction moment during stance compared to age and activity matched healthy controls. However, Webster and Feller (2011) demonstrated that male ACL-R patients actually have lower peak knee adduction moments during stance compared to controls. In a separate study Webster et al. (2012b) compared knee adduction moments during stance in groups of male and female ACL-R patients and found that females had higher adduction moments than the males. This suggests there may be a gender effect and that ACL-R females have an elevated peak knee adduction moment. To date, no studies have directly compared knee adduction moments during gait in a group of female ACL-R patients and age and activity matched healthy controls. Thus, further work is required.

The purpose of this study is to determine if peak knee adduction moment is altered in a group of ACL-R female patients compared to a group of controls. In addition, because moments at the knee may be affected by knee angular kinematics in all three planes, we want to examine if knee rotations were altered in the patient group. It is our hypothesis that peak adduction moment will be increased in the ACL-R group.

2. Methods

2.1. Participants

Seventeen lower limbs of fourteen females aged 20.8 (SD 1.17) years, constituted the ACL-R group (Table 1). Of these athletes, 3 participants had previously ruptured both right and left ACL, thus both lower limbs were included for the analysis in these participants. Of these involved lower limbs, eight were reconstructed via a hamstring auto-graft surgical procedure, with the remaining being a bone-patellar tendon-bone autograft. At the time of testing all athletes were fully engaged in field or court based sports (e.g. Gaelic football, soccer, hockey, basketball) at a club or county level and no athlete was undergoing any form of formal rehabilitation. Seventeen females aged 23.7 (SD 3.12) years, with no previous history of knee joint injury constituted the control group. All participants played field or court based sports (e.g. Gaelic football, soccer, hockey, basketball) at a club or county level. All participants had a Tegner activity level of 9 or 10. The ACL-R group and control group were weight and height matched. The ACL-R group was 2.9 years on average older than the control group (Table 1). Ethical approval for the study was approved by the Universities ethics committee. Before each subject began the study, they were told the risks of participation and then they read and signed an informed consent form.

2.2. Laboratory protocol

Gait analysis data was obtained using an active marker CODA Motion Analysis System (Charnwood Dynamics, Ltd, Leicestershire, UK) that consisted of three MPX30 cameras sampling at 200 Hz. The system was integrated with two AMTI force plates sampling at 1000 Hz (Watertown, Massachusetts). Subjects were familiarized with the test procedure and the equipment prior to testing. Kinematic and kinetic data was obtained using the marker set-up and data analysis methods described in Monaghan et al. (2007). Anthropometric data was obtained for the calculation of internal joint centers at the hip, knee and ankle joints, this included pelvic width (left ASIS to right ASIS), pelvic depth (ASIS to PSIS on right side), knee width and ankle width. Limb lengths of the thigh, shank and foot were measured with a measuring tape. The markers and marker wands were applied according to the manufacturer's guidelines by the same investigator on all subjects. Markers were positioned on the lateral aspect of the knee joint line, lateral malleolus, heel and fifth metatarsal head. Wands with anterior and posterior markers were positioned on the pelvis, sacrum, thigh and shank (Fig. 1). The markers were fixed to the skin with double sided tape. After the markers were in place, a physiotherapist put each subject into a subtalar neutral standing position prior to collection of a neutral stance trial that was used to align the participant with the laboratory coordinate system and to function as a reference position for subsequent kinematic and kinetic analysis. Subjects then completed several practice walking trials through the laboratory walkway. This allowed the subjects to get accustomed to the markers as well as allowed a starting point to be identified so that the subjects would contact the force plates in normal stride. Subjects walked barefoot across the 15 m walkway at their self-selected *normal* walking speed. Subjects were not aware of the presence of the force plates until the data collection was completed. Any trials in which there were not two consecutive foot strikes onto the force plates were discarded. 10 *clean* gait cycles, in which two full foot strikes onto the force plates were detected, were saved. The measurement system has been shown to be reliable when using the same marker set up and ten gait trials (Monaghan et al., 2007).

Table 1
Anthropometric, gait velocity and surgical data.

		Age (yrs)	Height (m)	Weight (kg)	Gait velocity (m/s)	Time since surgery (yrs)
Control	Mean	20.8	1.65	64.7	1.42	
	SD	1.17	0.06	7.06	0.131	
ACL-R	Mean	23.7	1.64	64.9	1.370	3.50
	SD	3.12	0.05	9.02	0.130	3.25

Differences between group means are non-significant ($P > 0.05$) except age.

2.3. Data analysis

Kinematic data was calculated by comparing the angular orientations of the co-ordinate systems of adjacent limb segments using Euler angles to represent clinical rotations in three dimensions. Vector algebra

Fig. 1. Marker set used for gait analysis.