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

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

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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.

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

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. 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).

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


Table 1
Anthropometric, gait velocity and surgical data.

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

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

Fig. 1. Marker set used for gait analysis.
and trigonometry are used to process marker positions within a Cartesian frame into rotation angles. Hip, knee and ankle joint angular displacements were calculated in the sagittal, frontal and transverse planes. Kinematic data were analyzed using the CODA-Motion software (Charnwood Dynamics Ltd, Leicestershire, UK), with the following axis conventions; x axis correlating to frontal plane motion; y axis correlating to sagittal plane motion and z axis correlating to transverse plane motion. A link-segmental analysis method was used to calculate kinetic data using a seven segment rigid body model based on the principles of inverse dynamics. To allow for the calculation of moments in respect of each joint the limb segments are treated mechanically as free bodies to which we can apply three-dimensional Newtonian mechanics relating motions and forces. The analysis is simplified by modeling each limb segment as a fixed, uniform distribution of mass around the longitudinal axis connecting the joint centers. Joint moments were calculated for the hip, knee and ankle joints in the sagittal, frontal and transverse planes.

Knee angular kinematic data from each trial were accumulated over the entire stride and converted to Microsoft Excel file format by converting the number of output samples to 100 + 1 in the data export option of the CODA software, which represented the complete stride as 100%, for averaging and further analysis. The ten normalized trials for each subject were combined to create an average ensemble curve for each participant, with group profiles then being calculated. This specific analysis technique has previously been used in our laboratory (Delahunt et al., 2006; Monaghan et al., 2006, 2007). Independent two-sided t-tests were used to test for significant differences between magnitudes of the group means time averaged profiles for each variable recorded during the stride. This technique has been used previously (Delahunt et al., 2006; Monaghan et al., 2006). The level of significance was set at \( P < 0.05 \). All moments reported are external moments.

Peak knee adduction moment during early stance and normalized peak knee adduction moment (%BW–HT) (Mossio et al., 2003) during early stance were compared using an independent samples one sided t-test (PASSW Statistics, 24 Version 18.0, IBM Corporation, Armonk, NY, USA). A one sided test was used because we hypothesized that the ACL-R group would have higher knee adduction moment. The level of significance was set at \( P < 0.05 \). A Levene’s test was used to check the assumption that the variances of both groups were similar for each variable. Associated effect sizes (eta squared) were calculated using the formula described in Pallant (2010): \( \eta^2 = \frac{t^2}{t^2 + (N_1 + N_2 - 2)} \) and quantified according to Cohen (1988): 0.01 = small effect size, 0.06 = medium effect size and 0.14 = large effect size. The adduction moment was normalized as a percentage of body weight and height to allow for future cross study comparisons.

3. Results

Levene’s test for equality of variance revealed that the variance was equal for the peak knee adduction moment as well as for the normalized peak knee adduction moment. Peak knee adduction moment was lower for the ACL-R 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, Fig. 2). There was a large effect size (eta squared = 0.19). Normalized knee adduction moment was also lower for the ACL-R group (\( N = 17, M = 2.86, SD = 1.01 \)) than for the control group (\( N = 17, M = 3.89, SD = 1.13 \); \( t(32) = 2.770, p = 0.005 \), one tailed). There was a large effect size (eta squared = 0.19).

Comparison of time averaged profiles for knee angular displacement in coronal, frontal and sagittal planes showed that there were significant differences in knee joint kinematics between the groups during the swing phase yet not during the stance phase of the gait cycle (Fig. 3). The ACL-R subjects displayed a more extended and more adducted knee position during swing phase than the control subjects.

4. Discussion

Prior to carrying out this study we had hypothesized that ACL-R females would exhibit increased peak knee adduction moment during the stance phase of walking compared to an age and activity matched control group. However, the main finding was that the knee adduction moment during gait was actually significantly lower in a group of ACL-R female subjects compared to a group of healthy female controls.
The results of the present study appear to be in contrast to that of Butler et al. (2009), who showed that a group of ACL-R patients had elevated knee adduction moments during gait (Table 2). Butler et al. (2009) had only two ACL-R subjects who had knee adduction moments that were less than 1 SEM below the mean value for their controls, whereas in our study we had only two ACL-R subjects who had knee adduction moments that were greater than 1 SEM above the mean value for our control subjects (Fig. 4). However, Butler et al. (2009) did not separate males and females in their study so a direct comparison is not possible. Webster et al. (2012b) showed that a group of ACL-R females had peak knee adduction moments during gait that were similar to those values observed in the Butler et al. (2009) study. Our results are more in line with the Webster and Feller (2011) study that showed that a group of ACL-R males had a lower knee adduction moment than healthy controls.

It is not clear why the female ACL-R group in our study had such lower peak knee adduction moments than what has been found in previous studies on female or predominately female ACL-R patients (Butler et al., 2009; Webster et al., 2012b). However, there are several factors that may contribute to this difference. In our study, data from 10 gait trials were averaged to derive kinematic and kinetic variables for each subject. We took this step to minimize potential errors that might be introduced by the measurement variability that is inherent in kinematic and kinetic analysis (Monaghan et al., 2007). Butler et al. (2009) did not report the number of trials that they used. Webster et al. (2012b) used a minimum of eight trials. It is possible that the use of fewer trials would make a study more susceptible to having outlier gait trials cause a large effect on the averaged data. Another explanation for the discrepancies between our findings and other authors is the fact that the surgical procedures, and subsequent rehabilitation programs, might have differed across studies. These discrepancies may also be due to the fact that different biomechanical models have been used to obtain kinetic data in these studies.

The higher peak knee adduction moment found in our study may have also been due to the activity levels of our subjects; our study consisted of subjects who were playing sport at a club or county level (equivalent to a Tegner activity score of 9 or higher), this may have resulted in our patient group having better movement patterns than patient groups from previous studies (Butler et al., 2009; Webster and Feller, 2011). However, our control group was activity matched, so they would also be expected to have better movement patterns than the less active control groups in previous studies. Our control group had an increased knee adduction moment compared to a group of predominately females of lower activity levels (Butler et al., 2009) and a similar knee adduction moment to a group of males of lower activity level (Webster and Feller, 2011). Recent preliminary work has shown that knee adduction moment may not be related to medial contact force during gait, suggesting that knee adduction moment might not be an appropriate measure to obtain when looking for gait patterns that may result in knee joint degeneration (Walter et al., 2010).

It is important to consider 3D knee joint angular kinematics in order to more fully understand the causes for the decreased knee adduction moment in the ACL-R group. In our study there was no difference in peak knee extension during stance. Longitudinal research has shown that over time, ACL-R patients do achieve greater extension during stance (Webster et al., 2012a). Knee joint kinematics were different between the ACL-R group and the control group over the swing phase of gait only. ACL-R knees were more extended and adducted during the swing phase. Such a position may imply an increased translational shear stress on the joint and may be a risk factor to knee joint degeneration over time.

Estimating knee osteoarthritis development is not simple, but a complex multifaceted problem that is beyond the scope of a single cross-sectional study. Data from the current work does show in what ways this group of ACL-R patients have different gait patterns compared to the closely matched control group. Such work is useful in order to guide future longitudinal research. It has been shown that ACL-R females have lower knee adduction moments than controls. However, we do not know if they have increased moments compared to their pre-surgery levels. This may be a factor for the increased incidence of knee OA in the ACL-R population. In order to control for activity levels we only included ACL-R subjects who had returned to playing sport. This may have created a self-selecting factor that worked in our patient group. Perhaps ACL-R patients who did not achieve sufficient quality of movement patterns after surgery, perhaps due to poor rehabilitation, were not able to return to high level sport and were not included in this study in the ACL-R group. Another limitation was that we did not control for graft type. Two graft types (hamstring and patellar autographs) were used in the ACL-R group. Recent research has shown that there is no significant difference in knee adduction moments between hamstring or patellar autographs in male subjects (Webster and Feller, 2011). Also, a literature review has suggested that graft type may not be a crucial factor in the outcome after ACL-R surgery (Spindler et al., 2004), however there may still be a minimal effect related to graft type.

In conclusion, the participants in this study who have had ACL-R surgery exhibited a decreased peak knee adduction moment compared to healthy controls. This finding is in contrast to previous research which has not considered an exclusively female patient group.

Table 2

<table>
<thead>
<tr>
<th></th>
<th>ACL-R group</th>
<th>Control group</th>
</tr>
</thead>
<tbody>
<tr>
<td>(Butler et al., 2009)</td>
<td>0.36 (13 F, 3 M)</td>
<td>0.30 (13 F, 3 M)</td>
</tr>
<tr>
<td>(Webster et al., 2012b)</td>
<td>0.30 (18 M) 0.38 (18 F)</td>
<td>0.28 (32 M)</td>
</tr>
<tr>
<td>(Webster and Feller, 2011)</td>
<td>0.28 (32 M)</td>
<td>0.40 (16 M)</td>
</tr>
<tr>
<td>Current Study</td>
<td>0.31 (17 F)</td>
<td>0.41 (17 F)</td>
</tr>
</tbody>
</table>

Fig. 4. Peak knee adduction moments for the ACL-R patient group. Black bars represent subjects who had knee moments less than 1 SEM below the mean value of the control group, white bars represent subjects who had knee moments within 1 SEM of the mean control value and hatched bars represent subjects who had knee moments greater than 1 SEM above the mean of control subjects.

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