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Neuromuscular compensatory strategies at the trunk and lower limb are not resolved following an ACL reconstruction

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ABSTRACT

Background: Following anterior cruciate ligament reconstruction (ACLR), patients present with greater trunk ipsilateral lean, which may affect knee kinetics and increase re-injury risk. However, there has been little research into neuromuscular factors controlling the trunk and their relation to the knee between healthy and ACLR subjects. This is critical to establish in order to develop more directed and effective interventions.

Hypothesis: As compared to healthy control subjects, ACLR subjects will demonstrate increased erector spinae and rectus abdominis co-contraction, greater rectus abdominis force and greater hamstring force that is correlated to increased forward trunk lean.

Study design: Cross-sectional study, Level of Evidence: 3.

Methods: Eleven healthy and eleven ACLR subjects were matched for age, mass and height. Subjects were asked to run at a self-selected speed while instrumented gait analysis was performed. An anthropometrically scaled OpenSim model was created for each subject. Trunk and hamstring muscle forces from Static Optimization were analyzed at impact peak. Additionally, directed co-contraction ratios were calculated for the erector spinae and erector spinae/rectus abdominis combinations.

Results: ACLR subjects showed more balanced erector spinae co-contraction \( p < 0.01 \), and greater hamstring force \( \text{biceps femoris long head} (p = 0.02), \text{semimembranosus} (0.01), \text{semitendinosus} (0.01) \). There was no statistical difference for any other muscle group.

Conclusion: Despite release to return to sport, ACLR subjects are continuing to increase the stiffness of their trunk as well increase their hamstring force to potentially reduce anterior tibial translation. Clinical relevance: Clinicians may anticipate ACLR subjects using their erector spinae and hamstrings to maintain a sense of stability in their trunk and at their knee.

1. Introduction

Over 130,000 individuals will undergo an anterior cruciate ligament (ACL) reconstruction annually in the United States [1]. The initial injury leaves individuals nearly six times more likely to suffer another ACL tear [2]. Additionally, only 65% of individuals who have had ACL reconstruction (ACLR) will return to the same activity level as prior to the injury [3]. Furthermore, individuals who go on to tear their ACL have been shown to have reduced neuromuscular control of their trunk and exhibit greater lean over their injured limb during stance [4,5]. Determining the differences in neuromuscular trunk control between healthy subjects and those who have had an ACLR could help construct more specific and effective interventions by improving our understanding of the underlying differences in muscle force production.

A growing body of evidence shows that even after completing rehabilitation, patients who have had an ACLR continued to have altered trunk and knee kinematics and kinetics [6–8]. For example, we recently found decreased contralateral and increased forward trunk lean in subjects who have had ACLR [9]. We and others have also shown that following ACLR, subjects have reduced peak knee extensor moments [9,10]. Reduced knee extensor moments decrease the demand on the quadriceps and consequently reduce anterior translation of the tibia.

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and strain of the ACL [11]. It has been suggested that reduced knee extensor moments may be the result of increased forward trunk lean, linking the trunk to the knee in both sagittal and frontal planes [7,9,10]. However, this previous work was unable to delineate whether the decreased knee extensor moment was due to increased forward trunk lean or increased hamstring forces. Despite differences in mechanics being well documented after ACLR, there has been little research into the differences in neuromuscular control responsible for the altered mechanics [7,9,12].

Studies examining trunk muscle function during other injuries, such as low back pain, have shown that individuals stiffen their trunk by increasing the co-contraction of the erector spinae, rectus abdominis and oblique muscles [13,14]. In low back pain, there is no significance to the direction of the deviation of the trunk. However, in the case of an ACL tear, directionality is important because further ipsilateral lean puts the individual at increased risk for an injury. Therefore, our current study will utilize a modified form of the co-contraction index, the directional co-contraction ratio (DCCR), taking into account the direction of the net force acting on the segment. This measure has been previously used in studies of compartmental osteoarthritis, cervical spine motion, and knee motion during cutting [15–17]. Muscle force estimates will be calculated using OpenSim, a biomechanical musculoskeletal modeling software tool whose accuracy against Electromyography (EMG) during walking and running has been demonstrated [18,19]. Although EMG is useful for assessing muscle activation, it presents challenges in modeling segments like the trunk where it is not practical to elicit muscle activations from deep tissue muscles during athletic movements. Unlike EMG, OpenSim is able to estimate the numerical force produced by all muscles with no invasive measurements.

The purpose of this study is to investigate differences in trunk and hamstring muscle function between healthy and ACLR individuals. We hypothesized that individuals who have had an ACLR would utilize increased co-contraction in their erector spinae and rectus abdominis muscles at impact peak to stiffen their trunks compared to healthy individuals. Additionally, we hypothesized that ACLR individuals would produce more rectus abdominis force and thus greater forward trunk lean than their healthy counterparts. Finally, we hypothesized that the ACLR individuals would produce greater hamstring force compared to the healthy individuals at impact peak, which would be correlated with increased forward trunk lean.

2. Methods

2.1. Subject demographics

Informed consent documents approved by the university Institutional Review Board were reviewed and signed by each individual prior to participation. Eleven individuals with ACLR (age 22.7 ± 3.5 yrs, mass 60.6 ± 8.5 kg, height 1.68 ± 0.08 m) and eleven healthy control subjects (age 19.7 ± 3.7 yrs, mass 64.2 ± 12.0 kg, height 1.67 ± 0.06 m) were matched for age, mass and height. ACLR individuals had undergone an ACL reconstruction 237 ± 65 days prior to gait analysis. ACLR individuals had undergone rehabilitation and been cleared by their physician to return to sport. One patient had a hamstring graft, all other patients had bone-patellar bone grafts. Control subjects were required to have no previous lower extremity surgeries. This study was a subsequent analysis of a data set previously published on in July 2014 [9]. A subject and their matched counterpart were excluded from this subsequent analysis if a trunk reflective marker was not present for over half the trial (6), inverse kinematics gave too large an error (1), the ground reaction force vector appeared out of alignment with the foot (1) or the static optimization failed (1).

<table>
<thead>
<tr>
<th>Joint</th>
<th>Angle Agreement</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pelvis Tilt</td>
<td>1.2° ± 0.6°</td>
</tr>
<tr>
<td>Pelvis List</td>
<td>1.8° ± 0.8°</td>
</tr>
<tr>
<td>Pelvis Rotation</td>
<td>1.8° ± 0.9°</td>
</tr>
<tr>
<td>Hip Flexion (stance)</td>
<td>1.6° ± 0.9°</td>
</tr>
<tr>
<td>Hip Adduction (stance)</td>
<td>1.2° ± 0.7°</td>
</tr>
<tr>
<td>Hip Rotation (stance)</td>
<td>1.6° ± 0.7°</td>
</tr>
<tr>
<td>Knee (stance)</td>
<td>2.1° ± 1.0°</td>
</tr>
<tr>
<td>Ankle (stance)</td>
<td>3.7° ± 2.5°</td>
</tr>
<tr>
<td>Hip Flexion (non-stance)</td>
<td>1.6° ± 0.6°</td>
</tr>
<tr>
<td>Hip Adduction (non-stance)</td>
<td>1.3° ± 0.5°</td>
</tr>
<tr>
<td>Hip Rotation (non-stance)</td>
<td>2.8° ± 1.1°</td>
</tr>
<tr>
<td>Knee (non-stance)</td>
<td>1.7° ± 0.8°</td>
</tr>
<tr>
<td>Ankle (non-stance)</td>
<td>4.3° ± 3.7°</td>
</tr>
<tr>
<td>Lumbar Extension</td>
<td>1.2° ± 0.4°</td>
</tr>
<tr>
<td>Ipsilateral Lumbar Bending</td>
<td>0.9° ± 0.4°</td>
</tr>
<tr>
<td>Lumbar Rotation</td>
<td>0.9° ± 0.5°</td>
</tr>
</tbody>
</table>

2.2. Gait analysis

Fifty-four retroreflective markers were placed on each subject. This marker set has been previously reported and the placement accuracy of these markers has been previously verified [20,21]. Following a five-minute warm-up on the instrumented treadmill, subjects were asked to run at a self-selected speed for another two minutes. Ground reaction force data and marker trajectories were simultaneously collected. Marker trajectories were collected with a 15-camera motion capture system at 200 Hz (Motion Analysis, Santa Rosa, CA). Marker trajectories were filtered at 6 Hz with a 6th order low-pass Butterworth filter. Force data was collected with an instrumented Bertec treadmill at 1200 Hz (Bertec Corporation, Columbus, OH). Force motion files were produced from Visual3D files using Lichtwark’s Matlab Opensim Toolbox and the open-source Biomechanical ToolKit Package. Ground reaction forces were filtered at 35 Hz using a 6th order low-pass Butterworth filter.

2.3. Simulations

A 12-segment, 29′-of-freedom, full-body running model was used because it’s accuracy at running speeds had already been verified by Hamner et al. [19]. To provide a more detailed muscular anatomy, the internal and external oblique muscles in the Hamner model were replaced by the internal and external oblique muscles from a detailed lumbar spine model [22]. The maximum isometric muscle forces for the model were increased to three times their original value to permit the estimation of high muscle forces. This increase is within the range of previous studies that have had lighter subjects [23,24]. All joints were kept from the Hamner model. The subtalar joint was locked. The lumbar and lower extremity joints were actuated by 102 Hill-type muscle-tendon actuators. The arm joints were actuated by ideal torque actuators.

The model was anthropometrically scaled based on each subject’s height and mass. Inverse Kinematics was performed to determine joint kinematics from marker trajectories measured experimentally. Next, the Reduce Residual Algorithm (RRA) minimized the effect of non-physiological residual forces applied to the body to compensate for noise in marker data that causes segment accelerations to be slightly inconsistent with ground reaction forces. Using measured ground reaction forces and the results of Inverse Kinematics, RRA simulated actuator and residual forces at each time point. The model’s segment mass distribution was then marginally changed based on average residual values. Finally, a second simulation produced kinematics dynamically consistent with experimental ground reaction forces and the new segment mass distribution by imposing a higher cost on residual forces.
Kinematics from RRA were then used for Static Optimization. Static Optimization determines muscle activations and resulting muscle forces that produce the net joint moments at each instant in time consistent with the experimentally measured data.

Muscle forces were evaluated during the stance phase from initial contact to toe off for each subject. Trials in which the reconstructed limb was in contact with the ground were selected for ACLR subjects. The same side was selected for their matched, healthy counterpart. Muscle forces generated by static optimization were interpolated to 100 points.

2.4. Statistical analysis

The average force and DCCR at impact peak for the erector spinae, rectus abdominis, internal and external obliques, and hamstrings were analyzed for the ACLR and healthy cohorts. Impact peak is generally accepted to occur at 15% of stance. Reported stance lengths range from 240 ms to 270 ms for running speeds of our subjects around 2.8 m/s [25]. Therefore, impact peak is very close to the time to ACL rupture, 37 ± 6 ms, reported by Krosshaug et al. [26]. The DCCR was calculated for each trunk muscle according to Heiden et al. [17]. If agonist force is greater than antagonist force:

\[ DCCR = 1 - \frac{\text{antagonist force}}{\text{agonist force}} \]

If antagonist was greater than agonist force:

\[ DCCR = \frac{\text{agonist force}}{\text{antagonist force}} - 1 \]

A two-tailed t-test was performed to determine if any of the differences between the two cohorts were significant (α = 0.05). A Pearson product moment correlation were used to determine the relationship between trunk lean and hamstring and erector spinae force production at impact peak.

3. Results

There was no statistical difference between the two groups with regard to running speed (Healthy: 2.8 ± 0.3 m/s, ACLR: 2.8 ± 0.3 m/s; p = 0.94). Further, there was no significant difference
in age, height, or mass between the groups.

The greatest joint angle differences between the experimental kinematics and model data occurred at the ankle, where the mean deviation was 4.3 ± 3.7°. Deviations for the remaining joints were less than 2.1° (Table 1). Following RRA, the simulation was more dynamically consistent with the experimental GRF data as the RMS average residual forces and moments were less than 11 N and 25 N-m for all subjects, respectively. The moments generated about each joint from the summed contributions of the muscles agreed very well with net joint moments determined from RRA (Fig. 1).

There was no significant difference between groups for trunk-pelvis lateral angle (ACLR: 8.5° ± 2.6°, Healthy: 5.6° ± 5.6°; p = 0.178) or trunk-pelvis extension (ACLR: 0.5° ± 9.4°, Healthy: 0.1° ± 6.7°; p = 0.90) (Fig. 2). An analysis of the lower extremity joint kinematics...
found that the ACLR cohort exhibited significantly less knee flexion (28.0° ± 5.2°) at impact peak than their healthy counterparts (35.9° ± 6.0°) (p = 0.003).

At impact peak the ACLR individuals produced significantly greater co-contraction in their erector spinae muscles than their healthy counterparts, (ACLR: 0.04 ± 0.58, Healthy: 0.65 ± 0.11, p = 0.003). There was no significant difference in normalized erector spinae force between groups on the stance side (ACLR: 12.96 ± 9.12 N/kg, Healthy: 14.11 ± 5.10 N/kg, p = 0.69) or non-stance side (ACLR: 7.50 ± 5.59 N/kg, Healthy: 4.34 ± 3.49 N/kg, p = 0.18). There was a strong correlation between ipsilateral trunk-pelvis lateral lean and contralateral erector spinae force (r = 0.51, p = 0.02). There was no relationship between ipsilateral trunk-pelvis lateral lean and ipsilateral erector spinae force. While the ACLR group exhibited greater rectus abdominis and external and internal oblique forces than the healthy individuals on both the contact and non-contact side; none of these were significantly different (Table 2).

All three hamstring muscles produced significantly larger forces in the ACLR group than the healthy group at impact peak (Table 2). There was no statistical difference between the groups for quadriceps force (Table 2).

4. Discussion

The purpose of this study was to determine if differences in trunk and hamstring muscle function (i.e., forces) are related to trunk positioning differences seen during gait at impact peak between healthy and ACLR individuals. The hypothesis that the ACLR individuals would have more balanced co-contraction of their erector spinae, internal and external obliques and rectus abdominis muscles was not supported as only the erector spinae muscles produced a more balanced co-contraction than the healthy individuals at impact peak. The hypothesis that the rectus abdominis would produce greater force in ACLR patients was not supported though the force on both sides was greater for ACLR individuals than healthy individuals, however not significantly. The hypothesis that ACLR subjects would have greater hamstring force at

Table 2
Mean normalized muscle force estimates of the trunk muscles, hamstring muscle group and quadriceps muscle group.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Healthy Force (N/kg)</th>
<th>ACL Reconstructed Force (N/kg)</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Trunk Muscles – Stance</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Erector Spinae</td>
<td>14.11 ± 5.10</td>
<td>12.96 ± 9.12</td>
<td>0.69</td>
</tr>
<tr>
<td>Rectus Abdominis</td>
<td>0.89 ± 1.24</td>
<td>1.38 ± 1.94</td>
<td>0.52</td>
</tr>
<tr>
<td>External Oblique</td>
<td>0.27 ± 0.28</td>
<td>0.34 ± 0.72</td>
<td>0.78</td>
</tr>
<tr>
<td>Internal Oblique</td>
<td>0.01 ± 0.01</td>
<td>0.12 ± 0.28</td>
<td>0.26</td>
</tr>
<tr>
<td><strong>Trunk Muscles – Non-Stance</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Erector Spinae</td>
<td>4.34 ± 3.49</td>
<td>7.50 ± 5.59</td>
<td>0.18</td>
</tr>
<tr>
<td>Rectus Abdominis</td>
<td>0.41 ± 0.54</td>
<td>0.92 ± 1.55</td>
<td>0.34</td>
</tr>
<tr>
<td>External Oblique</td>
<td>0.32 ± 0.50</td>
<td>0.43 ± 0.93</td>
<td>0.73</td>
</tr>
<tr>
<td>Internal Oblique</td>
<td>0.70 ± 0.40</td>
<td>0.70 ± 0.88</td>
<td>0.98</td>
</tr>
<tr>
<td><strong>Hamstrings</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Biceps Femoris Long</td>
<td>2.50 ± 1.12</td>
<td>3.93 ± 1.42</td>
<td>0.02</td>
</tr>
<tr>
<td>Head</td>
<td>2.10 ± 1.00</td>
<td>4.05 ± 1.78</td>
<td>&lt; 0.01</td>
</tr>
<tr>
<td>Semimembranosus</td>
<td>0.72 ± 0.35</td>
<td>1.23 ± 0.46</td>
<td>0.01</td>
</tr>
<tr>
<td>Semitendinosus</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Quadriceps</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Rectus Femoris</td>
<td>0.0 ± 0.0</td>
<td>0.0 ± 0.0</td>
<td>0.29</td>
</tr>
<tr>
<td>Vastus Medialis</td>
<td>3.34 ± 2.51</td>
<td>2.87 ± 1.52</td>
<td>0.61</td>
</tr>
<tr>
<td>Vastus Intermedius</td>
<td>4.13 ± 3.00</td>
<td>3.45 ± 1.79</td>
<td>0.53</td>
</tr>
<tr>
<td>Vastus Lateralis</td>
<td>7.25 ± 5.53</td>
<td>6.31 ± 3.45</td>
<td>0.64</td>
</tr>
</tbody>
</table>

Fig. 2. Healthy (—) and ACLR (−) ensemble curves with standard deviations for selected joint angles.
impact peak was confirmed as the ACLR individuals exhibited increased forces in all three muscles. These increased hamstring forces were weakly to moderately correlated to increased forward trunk lean.

Activation patterns reported in this study were consistent with previously reported results. The activation pattern of the erector spinae was consistent with patterns seen in previous studies using both Static Optimization and Computed Muscle Control that showed large activation values near initial contact and slow reduction through stance [19,27] (Fig. 3). It has been suggested that this large initial activation is to decelerate the trunk following flight phase and prevent trunk flexion [28]. This action would require bilateral contraction and matches with the co-contraction observed early in stance in this study. The obliques aligned well with previously published patterns during running [27].

Previously published EMG of the rectus abdominis fell within the pattern of the margin of error for the simulated forces, though differing activation patterns between subjects in this study caused the margin to be large [27]. The hamstrings showed peaks during the weight acceptance phase of stance and little to no activity after 80% of stance (Fig. 3). These results are consistent with published EMG and simulation data [19,29]. At impact peak, the sole subject with a hamstring tendon graft had typical semitendinosus force within one half standard deviation of the mean ACLR force.

The hypothesis that ACLR subjects would produce more balanced co-contraction in their erector spinae was confirmed. ACLR subjects also showed greater trunk-pelvis lateral lean than healthy subjects. This ipsilateral lean moves the center of mass of the trunk away from the pelvis-trunk joint and lengthens the contralateral erector spinae, both of which will lead to an increase in muscle force. This was supported by the strong correlation between ipsilateral lean and contralateral erector spinae force. However, there was no correlation between trunk-pelvis lean and ipsilateral erector spinae force. Therefore, as the subject further displaces their trunk, the co-contraction around the pelvis-trunk joint in the frontal plane will become more balanced. More balanced co-contraction has been repeatedly shown to increase the stiffness of the joint it is acting on [13,14]. Static Optimization solves the inverse dynamics problem by minimizing muscular activation so this co-contraction is not a reflection of a conscious decision by ACLR subjects. Rather, the increased co-contraction is a result of kinematics that put a demand on the erector spinae group for both lumbar flexion and lumbar bending moments. ACLR subjects may be relying on this induced stiffness for a sense of stability. It is not clear whether a stiffer trunk protects the individual by transferring energy absorption to the trunk or whether the stiffer trunk may harm the individual by locking the trunk into a more at-risk position of ipsilateral lean. Further research should focus on the effects of distally applied loads to trunks with varying degrees of stiffness.

We have previously shown decreased peak knee extensor moments in ACLR subjects but were not able to isolate whether this was an effect of increased forward trunk lean or increased hamstring forces [9]. By taking advantage of additional analyses in OpenSim, we were able to show that ACL subjects did produce significantly greater force than healthy subjects in all three hamstring muscles at impact peak. ACL subjects’ increased hamstring force may be a mechanism to reduce anterior translation of the tibia and therefore reduce ACL strain [11]. However, there was no correlation between hamstring force and forward trunk lean indicating that the effects of the hamstrings may not be propagating proximally.

We also found that the hypothesis that greater pelvis-trunk flexion would be associated with greater rectus abdominis force was not supported. Our prior work showed greater forward trunk lean in ACL patients when measured against a fixed laboratory coordinate system [9]. However, this study showed no statistical difference for trunk flexion, defined as the relative angle between the trunk and pelvis. This suggests that the observed difference in global trunk lean does not originate because of a difference at the lumbar joint. Instead, if the lumbar angle between two subjects is kept similar, a difference in the trunk global angle can still be manifested if the subjects’ hip, knee or ankle angles are significantly different. This suggests that further work to identify muscular differences responsible for forward trunk lean should focus on hip and knee musculature.

This study was limited by several factors. Static optimization solves inverse dynamics by minimizing muscle activation, which is not necessarily minimizing co-contraction. However, static optimization has been shown to accurately reflect activities that require co-contraction.
like running and hopping [19,30]. The only change to the Hamner model was to replace the single muscle representing each of the internal and external obliques with 6 fascicles from the Christophy model. The abdomen segment of the Christophy model was added for attachment points of several fascicles. However, z-axis rotation of the abdomen/pelvis joint was locked because no tracking data was available. This may have altered the force needed for several internal and external oblique muscle fascicles attached to this abdomen. Other work with a detailed trunk model has shown the obliques to not be significantly active around impact peak, indicating this may not have significantly altered muscle force estimates.

In conclusion, ACLR subjects demonstrated significantly increased co-contraction of their erector spinae muscles at impact peak compared to matched, healthy controls who showed a directed co-contraction ratio favoring their stance side erector spinae. The ACLR subjects’ increased co-contraction may be allowing these subjects to maintain a sense of stability, however, further work is needed to determine whether increased stiffness from co-contraction is a protective or harmful strategy in this population. ACLR subjects had significantly greater hamstring forces at impact peak. This may serve as a protective mechanism to reduce anterior translation of the tibia in their reconstructed limb as well as explain decreased knee extensor moments previously seen in this data set. ACLR subjects additionally showed significantly decreased knee flexion. Results of this study indicate that even in straight line running, ACLR subjects are utilizing protective strategies to prevent trunk motion and anterior translation of the tibia.

Conflict of interest statement
The authors have no conflicts of interest to disclose.

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References