Influence of trunk flexion on hip and knee joint kinematics during a controlled drop landing

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Abstract

Background. An erect posture and greater knee valgus during landing have been implicated as anterior cruciate ligament injury risk factors. While previous research suggests coupling of knee and hip kinematics, the influence of trunk positioning on lower extremity kinematics has yet to be determined. We hypothesized that greater trunk flexion during landing would result in greater knee and hip flexion and lesser knee valgus. Identification of a modifiable factor (e.g. trunk flexion) which positively influences kinematics of multiple lower extremity joints would be invaluable for anterior cruciate ligament injury prevention efforts.

Methods. Forty healthy individuals completed two drop landing tasks while knee, hip, and trunk kinematics were sampled. The first task constituted the natural/preferred landing strategy (Preferred), while in the second task, subjects actively flexed the trunk upon landing (Flexed).

Findings. Peak trunk flexion angle was 47° greater for Flexed compared to Preferred (P < 0.001), and was associated with increases in peak hip flexion angle of 31° (P < 0.001) and peak knee flexion angle of 22° (P < 0.001).

Interpretation. Active trunk flexion during landing produces concomitant increases in knee and hip flexion angles. A more flexed/less erect posture during landing is associated with a reduced anterior cruciate ligament injury risk. As such, incorporating greater trunk flexion as an integral component of anterior cruciate ligament injury prevention programs may be warranted.

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1. Introduction

Recent research on anterior cruciate ligament (ACL) injury mechanisms has begun to evaluate influences of kinematic factors proximal to the knee joint. Given the closed-kinematic-chain nature of the lower extremity following ground contact during landing and gait activities, it has been suggested that segmental motion of the mass superincumbent to the knee directly influences knee joint motion and loading. Specifically, it has been suggested that hip internal rotation and adduction contribute to knee valgus (McLean et al., 2004b; Zeller et al., 2003), a kinematic factor which has been linked to ACL injury risk (Hewett et al., 2005). This notion coincides with the “position of no return” during which non-contact ACL injury is hypothesized to occur, characterized by hip adduction and internal rotation, knee valgus and external tibial rotation, and subtalar pronation (Ireland, 1999). Additionally, females, a population which is at heightened risk for ACL injury (Arendt et al., 1999; Gwinn et al., 2000), simultaneously display lesser knee, hip, and trunk flexion during gait and landing tasks compared to males (McLean et al., 2004b; DiStefano et al., 2005; Salei et al., 2004; Yu et al., 2006; Decker et al., 2003), suggesting sagittal-plane
coupling of these joints. Furthermore, females also demonstrate greater knee valgus during these tasks compared to males (Russell et al., 2006; Ford et al., 2003), suggesting the coupling of lower extremity kinematics has a multiaxial influence on ACL injury risk. As excessive knee valgus and a more erect landing posture (evidenced by a more extended knee, hip, and trunk) have been postulated as ACL injury risk factors (Griffin et al., 2000), continued research regarding the coupling of these joints is warranted.

Previous research supports a link between landing forces and knee injury (Dufek and Bates, 1991), and numerous investigations have demonstrated relationships between landing forces and frontal and sagittal plane kinetics and kinematics. In a prospective study, Hewett et al. (2005) reported a significant correlation between peak knee valgus angle and peak vertical ground reaction force in individuals who sustained ACL injury, and that the vertical ground reaction force was 20% greater in individuals who sustained ACL injury compared to those who did not. Furthermore, those who sustained ACL injury displayed lesser knee flexion compared to those who did not, and knee flexion angle was correlated with the vertical ground reaction force in these individuals. McLean et al. (2005) demonstrated the dependency of peak knee valgus moments during cutting tasks on hip and knee joint kinematics. Yu et al. (2006) demonstrated that hip and knee joint angular velocities were correlated with posterior and vertical ground reaction forces. Lastly, videotape feedback has been demonstrated to increase knee flexion displacement and decrease vertical ground reaction forces simultaneously (Onate et al., 2005), supporting the link between landing kinematics and kinetics. These data suggest that lower extremity landing kinetics and kinematics and the subsequent load placed on the ACL are highly interlinked in a multiaxial manner, a notion which is supported by investigations of the effects of various knee joint kinematic configurations on ACL stress and strain in cadaveric specimens (Durselen et al., 1995; Fukuda et al., 2003; Kanamori et al., 2002).

The aforementioned relationships between frontal and sagittal plane kinematics and landing forces suggest that greater hip and knee flexion and a smaller knee valgus angle are associated with lesser ground reaction forces, potentially shielding the ACL from excessive loading. While the coupling of hip and knee kinematics and the link between lower extremity kinematics, landing forces, and ACL injury are supported in the literature, it is unclear how trunk motion influences these factors. Specifically, it is unclear how the sagittal-plane positioning of the trunk influences hip and knee joint kinematics. Identification of a single modifiable factor (e.g., trunk flexion angle) which has the potential to alter lower extremity kinematics in a multiaxial manner that is associated with a lesser ACL injury risk would be invaluable to injury prevention efforts. Therefore, the purpose of this investigation was to evaluate the influence of trunk motion in the sagittal plane on hip and knee joint kinematics. It was hypothesized that trunk flexion during landing would result in greater knee and hip flexion, lesser knee valgus, and lesser hip adduction and internal rotation.

2. Methods

2.1. Subjects

Forty healthy volunteers (20 males, 20 females: age = 21.5 (SD 1.9) years, height = 1.73 m (SD 0.10), mass = 74.6 kg (SD 17.47) constituted the sample for this investigation. All subjects were physically active, participating in physical activity a minimum of 20 min 3 times per week, and had no history of (1) ACL injury, (2) lower extremity surgery, (3) neurological disorder, (4) chronic lower extremity injury, or (5) lower extremity injury within the six months prior to data collection. Subjects read and signed an approved informed consent document prior to data collection. All data were sampled from the subject’s right lower extremity, which corresponded with the dominant leg in 37 of the 40 subjects (93%).

2.2. Instrumentation

Electromagnetic tracking sensors (Flock of Birds, Ascension Technology Corp., Burlington, VT, USA) were positioned on the thorax, sacrum, thigh, and shank using double-sided tape. Global and segment axis systems were established with the X-axis designated as positive forward/anteriorly, the Y-axis positive leftward/medially, and the Z-axis positive upward/superiorly. The trunk and right lower extremity were modeled by digitizing the C7/T1, T12/L1, hip, knee, and ankle joint centers. The hip joint center was estimated using a least squares method (Leardini et al., 1999), while knee and ankle joint centers were defined as the midpoints between the digitized medial and lateral femoral condyles and medial lateral malleoli, respectively. Vertebral joints were defined as the digitized space between the corresponding spinous processes. The Motion Monitor motion capture system (Innovative Sports Training, Chicago, IL, USA) was used for model generation/calibration and data acquisition.

2.3. Experimental procedures

Subjects performed two controlled vertical drop landing tasks utilizing a repeated-measures design during which trunk, hip, and knee kinematics were assessed. For the first task, subjects stepped off a platform 60 cm in height with the right knee extended (McNitt-Gray, 1993) and performed a double-leg landing onto a non-conductive force plate (Bertec Corp., Columbus, OH, USA), with only the right foot making contact with the force plate. This task represented the subject’s natural/preferred landing strategy (Preferred). For the second task, subjects stepped off the platform in a manner identical to the first, but were
instructed to flex the trunk during landing (Flexed). The primary investigator demonstrated each task as well as the motion of trunk flexion. Vertical displacement of the sacrum during the takeoff from the platform was assessed following each trial to ensure that subjects stepped off the platform rather than jumping upward, thus standardizing the drop height and downward velocity. Trials during which upward sacral displacement exceeded 10 cm were discarded and repeated. Five acceptable trials were performed for each task. Due to the fact that subject knowledge of the intent of the second task (i.e. trunk flexion during landing) may have biased the natural/preferred landing strategy, Preferred was always performed prior to Flexed.

### 2.4. Data sampling and processing

Data were sampled over a 4 s interval centered on initial ground contact (IGC), defined as the instant at which the vertical ground reaction force exceeded 10 N. Kinematic angles were evaluated at two points in the landing task: (1) IGC, and (2) peak value during the loading phase, defined as the interval between ICG and peak knee flexion (Yu et al., 2006). Respective values for trunk flexion/extension, knee flexion/extension and valgus/varus, and hip flexion/extension, abduction/adduction, and internal/external rotation were calculated for each landing phase.

Kinematic data were sampled at 100 Hz and lowpass filtered at 10 Hz (4th order, zero-phase-lag Butterworth) (Decker et al., 2003). The initial 500 ms were removed to eliminate filter effects. Kinematic angles were calculated using Euler angle sequences rotated in an order of flexion-extension (Y-axis), valgus-varus (X-axis), and internal-external rotation (Z-axis) (Yu et al., 2006). Trunk angles were calculated as the trunk reference frame relative to the thigh reference frame. Knee angles were calculated as the shank reference frame relative to the thigh reference frame, while hip angles were calculated as the sacrum reference frame relative to the thigh reference frame.

### 2.5. Statistical analyses

Mean values for each dependent variable were calculated across the five trials for each task. The effects of landing phase and task were evaluated by performing separate 2 (Task: Preferred and Flexed) × 2 (Phase: IGC and Loading) repeated measures ANOVA for each dependent variable. Significant task × Phase interaction effects were evaluated post hoc via two-tailed dependent-samples t-tests following a Bonferroni correction for Type I error rate. These planned pairwise comparisons were made between Preferred and Flexed conditions for each phase of the landing. Statistical significance was established a priori as \( \alpha = 0.05 \).

### 3. Results

The Task main effect and Task × Phase interaction effect were significant for trunk flexion, knee flexion, and hip flexion (\( P < 0.001 \) for each analysis). No other main effect or interaction effect reached statistical significance. With respect to post hoc analyses, the trunk, hip, and knee displayed significantly greater flexion for Flexed compared to Preferred on average (i.e. collapsed across landing phases). Additionally, this difference was significant at both landing phases (IGC and Loading) for the trunk and hip, but was only significant for the Loading Phase for the knee. Means and standard deviations for each dependent variable are presented in Tables 1–3.

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**Table 1**

Descriptive statistics for trunk kinematics: Mean (SD) and [lower, upper limits of 95% confidence interval]

<table>
<thead>
<tr>
<th></th>
<th>IGC</th>
<th>Loading (peak)</th>
<th>Collapsed across phases</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Flexion</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Preferred</td>
<td>14 (SD 11°) [11, 17]</td>
<td>49 (SD 21°) [43, 56]</td>
<td>32 (SD 24°) [24, 39]</td>
</tr>
<tr>
<td>Flexed</td>
<td>24 (SD 13°) [20, 28]°</td>
<td>96 (SD 20°) [89, 102]°</td>
<td>60 (SD 40°) [47, 73]°</td>
</tr>
</tbody>
</table>

\(^{a}\) Significantly different from preferred.

**Table 2**

Descriptive statistics for knee kinematics: Mean (SD) and [lower, upper limits of 95% confidence interval]

<table>
<thead>
<tr>
<th></th>
<th>IGC</th>
<th>Loading (peak)</th>
<th>Collapsed across phases</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Flexion</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Preferred</td>
<td>6 (SD 7°) [3, 8]</td>
<td>69 (SD 16°) [64, 75]</td>
<td>38 (SD 34°) [27, 48]</td>
</tr>
<tr>
<td>Flexed</td>
<td>9 (SD 11°) [5, 12]</td>
<td>91 (SD 16°) [86, 96]°</td>
<td>50 (SD 43°) [36, 64]°</td>
</tr>
<tr>
<td><strong>Valgus</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Preferred</td>
<td>–6 (SD 7°) [–8, –3]</td>
<td>–15 (SD 8°) [–18, –12]</td>
<td>--10 (SD 9°) [–13, –8]</td>
</tr>
</tbody>
</table>

\(^{a}\) Significantly different from preferred. 
\(^{b}\) (–) value indicates valgus.
Table 3
Descriptive statistics for hip kinematics: Mean (SD) and [lower, upper limits of 95% confidence interval]

<table>
<thead>
<tr>
<th></th>
<th>IGC</th>
<th>Loading (peak)</th>
<th>Collapsed across phases</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Flexion</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Preferred</td>
<td>14 (SD 12°) [10, 17]</td>
<td>40 (SD 20°) [34, 47]</td>
<td>27 (SD 21°) [20, 34]</td>
</tr>
<tr>
<td>Flexed</td>
<td>20 (SD 12°) [16, 24]*</td>
<td>71 (SD 19°) [65, 77]*</td>
<td>46 (SD 31°) [36, 55]*</td>
</tr>
<tr>
<td><strong>Adduction</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Preferred</td>
<td>3 (SD 9°) [0, 6]</td>
<td>−5 (SD 7°) [−7, −2]</td>
<td>−1 (SD 9°) [−4, 2]</td>
</tr>
<tr>
<td>Flexed</td>
<td>2 (SD 12°) [−2, 6]</td>
<td>−6 (SD 12°) [−10, −2]</td>
<td>−2 (SD 12°) [−6, 2]</td>
</tr>
<tr>
<td><strong>Internal rotation</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Preferred</td>
<td>11 (SD 9°) [8, 14]</td>
<td>−11 (SD 18°) [−17, −5]</td>
<td>0 (SD 18°) [−6, 6]</td>
</tr>
<tr>
<td>Flexed</td>
<td>12 (SD 10°) [9, 15]</td>
<td>−7 (SD 21°) [−14, −1]</td>
<td>2 (SD 19°) [−4, 8]</td>
</tr>
</tbody>
</table>

*Significantly different from preferred.

b (+) value at IGC indicates abduction.

c (+) value at ICG indicates external rotation.

c (+) value at ICG indicates external rotation.

4. Discussion

The primary finding of this investigation was that active trunk flexion during landing produced concomitant increases in knee and hip flexion compared to a more erect/extended trunk posture. These results are in agreement with previous evidence of coupling of the knee and hip joints in the closed-kinematic-chain. Previous literature suggests that a more erect posture during landing, gait, and cutting activities as identified by more extended trunk, hip, and knee positions, may be a risk factor for ACL injury (Griffin et al., 2000). Additionally, females typically display a more erect posture during these activities relative to their male counterparts (Decker et al., 2003; Huston et al., 2001; Malinzak et al., 2001; Salci et al., 2004), potentially contributing to the higher female ACL injury rate. The results of the current investigation suggest that active trunk flexion during landing alters lower extremity kinematics in a manner which potentially reduces ACL loading and injury risk.

We are unaware of any previous investigations which have evaluated the effects of trunk flexion angle on lower extremity kinematics during landing tasks, thus comparison to previous literature is somewhat limited. However, peak knee flexion angle for the natural/preferred landing strategy in the current investigation is in agreement with Huston et al. (2001) and Dufek and Bates (1991), while peak knee valgus angle is in agreement with Ford et al. (2003) for similar drop landing tasks. With respect to hip kinematics, peak values for adduction and internal rotation are in agreement with Pollard et al. (2006), while peak hip flexion angle is in agreement with DiStefano et al. (2005).

While the load imposed on the ACL in vivo is determined by multi-directional influences, under controlled experimental conditions, this force is proportional to the force generated by the knee extension mechanism (Li et al., 1999; Durselen et al., 1995; Beynnon et al., 1995; Withrow et al., 2006b). Additionally, isolated quadriceps loading is capable of inducing ACL injury and rupture (DeMorat et al., 2004). The influence that quadriceps force has on ACL loading is mediated by two factors related to knee flexion angle: patellar tendon angle of insertion and ACL elevation angle.

As the knee progresses into flexion, the patellar tendon insertion angle with respect to the tibial longitudinal axis decreases (Zheng et al., 1998). This change in patellar tendon orientation has a profound influence on tibial shear force, as the anteriorly-directed component of the quadriceps/patellar tendon force is derived as a multiple of the sine of the insertion angle (Fig. 1a and b). Numerous investigators have demonstrated the effect of patellar tendon insertion angle on ACL loading, reporting decreases in anterior tibial shear force (Nisell et al., 1989), in situ ACL force (Li et al., 1999), anterior tibial translation (Li et al., 1999), and ACL strain (Durselen et al., 1995; Beynnon et al., 1995; Withrow et al., 2006b) as the knee moves into flexion. Therefore, greater knee flexion during the loading phase of landing would reduce anterior tibial shear force for a given quadriceps/patellar tendon force, thus reducing the force placed on the ACL. Due to the fact that actively flexing the trunk during landing simultaneously increased the peak knee flexion angle during the loading phase, trunk flexion likely exerts an indirect influence on ACL loading.

The elevation angle of the ACL with respect to the tibial plateau is also influenced by knee flexion angle, decreasing with knee flexion such that the ACL is essentially parallel to the tibial plateau with knee flexion past 90° (Li et al., 2006; Zheng et al., 1998). This change in orientation influences the load placed on the ACL and its ability to sustain elastic deformation without injury. The structural properties of the ACL are maximized under tensile (longitudinal) loading conditions and minimized under non-axial (shear) loading conditions (Lyon et al., 1989; Woo et al., 1991, 1987). As the knee progresses into extension, the ACL elevation angle is maximized. Under this configuration, the anterior tibial shear force generated by the quadriceps/patellar tendon and imparted to ACL is increasingly shear in nature. Conversely, as the ACL elevation angle decreases with knee flexion, the shear component of the resultant ACL force decreases while the tensile component
increases reciprocally (Fig. 1a and b). The greater peak knee flexion angle observed during the Flexed landing strategy in the current investigation suggests that active trunk flexion during landing potentially reduces ACL loading by simultaneously reducing the anterior tibial shear force induced by quadriceps contraction as well as the shear component of the resultant force imparted to the ACL.

A third manner in which active trunk flexion during landing may reduce ACL loading is via the change in insertion angles of the hamstrings muscles as functions of knee flexion angle. The hamstrings are suggested as being agonistic/supplemental to the function of the ACL, providing posterior tibial shear force to counter anterior tibial shear force introduced by the quadriceps. As the knee progresses into flexion, the angle of insertion of the hamstrings with respect to the tibial longitudinal axis increases (Fig. 1a and b) such that at knee flexion angles greater than 100°, the resultant hamstring force is directed parallel to the tibial plateau (Zheng et al., 1998). Li et al. (1999) demonstrated in the cadaveric knee that quadriceps/hamstring co-activation progressively reduced anterior tibial translation and in situ ACL force when compared to isolated quadriceps activation with knee flexion of 15–60°. Similarly, Withrow et al. (2006a) demonstrated that a simulated lengthening hamstring contraction reduced ACL strain in a cadaveric model. Similar results have been reported in humans using model simulation techniques (Pandy and Shelburne, 1997) and in vivo ACL strain measurement (Beynnon et al., 1995) during isometric knee extension exercises. The biarticulate hamstrings are subjected to a unique form of loading during landing, as they are shortened via knee flexion, yet lengthened via hip flexion. Trunk flexion during landing also corresponded with an increase in peak hip flexion angle in the current investigation. The hamstring lengthening and subsequent increase in passive tension imposed by greater hip flexion may account for the decreased tension induced by knee flexion. The combination of increased passive tension and angle of insertion potentially provides the hamstrings with a mechanical advantage for resisting anterior tibial translation when the knee is in a flexed position following trunk flexion. These data suggest that the greater knee and hip flexion angle associated with active trunk flexion during landing in the current investigation may be an effective approach to limiting ACL injury risk.

It is important to note that frontal and transverse plane mechanics play important roles in determining the resultant force placed on the ACL. In our opinion, it is unlikely that sagittal plane mechanisms such as the anterior tibial shear force introduced by the quadriceps act in isolation to induce ACL injury. While the greater hip and knee flexion induced by trunk flexion during landing appear to align the lower extremity in a manner consistent with reduced ACL loading and injury risk, these benefits could potentially be counteracted by greater knee valgus, and hip adduction and internal rotation. However, frontal and transverse plane kinematics did not differ significantly between Preferred and Flexed landings. These results suggest that trunk flexion during landing appears to alter sagittal plane kinematics in a positive manner relative to ACL injury risk while having a limited influence on frontal and transverse plane kinematics.

5. Clinical implications

Considerable debate exists regarding the precise mechanisms of ACL injury and the greater incidence of ACL injury in females. While some investigators have implicated sagittal plane mechanisms as the primary contributors with minimal influence from the frontal plane (Chappell et al.,
2002), others suggest that frontal plane factors are the culprits with sagittal plane mechanisms playing insignificant roles (Hewett et al., 2005; McLean et al., 2004a). Given the predisposition for females to perform gait, landing, and cutting tasks in a more erect position compared to males, and the “quadriceps-dominant” activation strategy noted in females during these tasks (Van Doren et al., 2007; Malinzak et al., 2001; Sigward and Powers, 2006; Yu et al., 2006), we suggest that the load imparted to the ACL via quadriceps force is the primary mechanism for ACL injury. However, frontal and transverse plane mechanisms also load the ACL (Durselen et al., 1995; Kanamori et al., 2002; Fukuda et al., 2003), and greater knee valgus has been associated with increased ACL injury risk (Hewett et al., 2005). Rather than acting in isolation, we suggest that frontal and transverse plane motions at the knee and hip “preload” the ACL such that when coupled with a high velocity, large magnitude quadriceps contraction, ACL stress can exceed the elastic limit.

This investigation provides an initial evaluation of the influence that a single modifiable kinematic factor has on kinematic variables suggested as ACL risk factors. Due to the fact that active trunk flexion during landing alters kinematics of the lower extremity in manners thought to be consistent with a reduced ACL injury risk, incorporating greater trunk flexion as an integral part of ACL injury prevention programs may be warranted. However, the results of this investigation should be evaluated with caution, as a number of limitations potentially influence their applicability. Specifically, the direct implications of altering trunk flexion angle during landing has for the load borne on the ACL and ACL injury risk are unclear, and future research is necessary to determine the influence of trunk flexion on these factors. Additionally, our operational definition of trunk flexion (i.e. angle between the trunk and thigh segments) may not accurately reflect the intended motion, as the trunk and pelvis do not truly function as a single rigid segment. However, this measure is more readily assessable in the clinical setting in the absence of a motion capture system compared to more conventional measures (e.g. trunk relative to pelvis or world vertical axis), and is likely more appropriate for future injury prevention efforts. Lastly, it is unclear if the large magnitude change in trunk flexion observed in this investigation is feasible for implementation in an injury intervention program. In our opinion, it is highly unlikely that such a large change is feasible, particularly in athletic participation, as this positioning would likely limit function and performance. Future research is necessary to determine if lesser changes in trunk position can induce similar kinematic changes in the lower extremity, and if a critical value for trunk flexion exists beyond which no additional gains are achieved.

References


