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Anterior tibiofemoral intersegmental forces during landing are predicted by passive restraint measures in women

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ABSTRACT

Background: Passive restraint capabilities may influence sagittal plane knee joint mechanics during activity. This study aimed to determine if measures associated with passive restraint of anterior translation of the tibia are predictive of peak anterior knee shear force during landing.

Methods: Passive restraint measures were assessed via joint arthrometry and during 40% body weight simulated weight acceptance using recreationally active students (73 F, 42 M; 21.8 ± 2.9 yr, 1.69 ± 0.1 m, 68.9 ± 14.1 kg). Anterior knee laxity (mm) at 133 N and initial (0–20 N) and terminal (100–130 N) anterior stiffnesses (N/mm) were calculated from arthrometer data. Peak anterior tibial acceleration ($m \cdot s^{-2}$) relative to the femur was assessed via electromagnetic position sensors during 40% body weight acceptance trials. Peak knee shear force was assessed during double-leg drop jumps.

Results: Sex specific linear stepwise regressions revealed that in females, increasing peak tibial acceleration ($5.1 \pm 1.8\ m \cdot s^{-2}$) ($R^2\Delta = 7.3\%$, $P\Delta = 0.021$), increasing initial anterior stiffness (31.0 ± 14.0 N/mm) ($R^2\Delta = 5.9\%$, $P\Delta = 0.032$), and decreasing terminal anterior stiffness (43.4 ± 17.4 N/mm) ($R^2\Delta = 4.9\%$, $P\Delta = 0.046$) collectively predicted greater peak knee shear forces (66.6 ± 12.03% BW) (multiple $R^2 = 18.1\%$). No male regressions were significant.

Conclusions: Sagittal laxity measures are associated with anterior knee shear loads during landing in females. Greater tibial acceleration during early axial load along with greater initial and lesser terminal anterior stiffnesses predicted increasing anterior knee shear forces. Future work should investigate the combined contribution of passive and active restraints to high-risk ACL biomechanics.

1. Introduction

Sagittal plane mechanisms are commonly thought to contribute in part to high risk biomechanics associated with ACL injury [1]. Specifically, upright landing styles associated with decreased knee [2,3] and hip flexion [4] angles are thought to increase sagittal plane knee loading [4], likely through increased reliance on passive restraints (ligamentous structures) for knee stability. This upright posture is thought to result in increased landing forces [5] and subsequently increase anterior knee shear forces and ACL loading [6]. Proximal anterior tibial shear force is considered to be a major contributor to the loading of the ACL [7,8] with the ACL acting as the primary restraint system to anterior displacement of the tibia with respect to the femur [9]. Previous work has demonstrated that a combination of biomechanical factors including posterior ground reaction force, external knee flexion moment, knee flexion angle, integrated EMG activity of the vastus lateralis, and sex predicted a large amount of the variance (~86%) in anterior tibial shear force during a deceleration task [10]. While it is understood that internal and external loads may influence anterior shear forces, it is not understood how passive restraints responsible for controlling anterior tibial motion (which is dominated by ACL function [9]) may affect sagittal knee plane joint biomechanics during functional activity. As greater anterior knee laxity has been associated with greater anterior tibial translation [11], and greater translations are ultimately a function of anteriorly directed forces, it can be theorized that anterior knee laxity may be related to intersegmental shear forces.

It has been suggested that risk of ACL injury may increase in the absence of sufficiently taut passive restraints [12]. While multiple factors likely contribute to greater ACL injury risk in females, both retrospective [13–17] and prospective [12,18] studies identify a relationship between ACL injury and increased sagittal knee joint laxity. Although it is acknowledged that ACL injury is likely multiplanar in nature and that non-sagittal plane laxity may also contribute to high-risk mechanics [19], other work has focused on the relation of anterior knee laxity to total anterior tibial translation during weight acceptance, demonstrating a positive association between increased anterior knee laxity and greater anterior tibial translation upon weight acceptance [11]. Further, a combination of sagittal plane laxity measures has been associated with a landing strategy that resulted in greater workload about the knee [20]. This was postulated to be an attempt to stabilize the knee

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and reduce the loads applied to the ligamentous tissue; however also potentially rendering the knee less able to resist injurious forces [20]. Collectively, these studies suggest that passive restraints may be an important contributor to dynamic knee stability and impact ACL injury risk [12].

A deconstruction of previously reported sagittal plane measures of anterior knee laxity and anterior tibial translation upon weight acceptance [21] may allow even greater insight into passive restraint mechanics of anterior knee loading during functional activity (e.g. landing from a jump). Characteristics of the load–displacement curve (stiffness) of anterior knee laxity are thought to be important with respect to the clinical functioning of the knee joint [22] with alterations in loading range specific anterior knee stiffness postulated to be a factor in ACL injury risk [23]. The terminal phase of arthrometer loading is thought to be where the ACL fully engages in restraint [24]. Decreases in incremental stiffness at higher arthrometer loadings may be associated with ligamentous restraint behavior that may potentially result in altered arthrokinematics during functional activity [25]. The initial loading phase is thought to largely represent the resistance provided by the weight of the limb; however, the impact of early loading range stiffness on functional biomechanics is unknown.

While anterior knee laxity is predictive of anterior tibial translation during weight acceptance [11], ACL strain has been reported to be proportional to anterior tibial acceleration during weight acceptance [26]. Thus, it would be of benefit to further understand how passive (ligamentous) knee joint behavior translates to functional behavior of the knee during activity commonly associated with the ACL injury mechanism of landing from a jump. Thus the purpose of our investigation was to determine if measures associated with passive restraint of anterior tibial translation at the knee joint are predictive of peak anterior knee shear force during a drop jump landing in females and males. Given the widely reported sex differences in landing biomechanics, we chose to include sex-specific analyses to account for these sex differences in landing mechanics. It was hypothesized that a combination of measures associated with passive restraint of anterior translation of the tibia (anterior knee laxity, initial and terminal knee stiffness during anterior loading, and tibial acceleration during simulated weight acceptance) would collectively predict greater proximal anterior knee shear forces during a landing maneuver.

2. Methods

For purposes of a larger study examining the effects of hormone mediated knee joint laxity on weight bearing knee joint neuromechanics [21], 73 females and 42 males (21.8 ± 2.9 yr, 1.69 ± 0.1 m, 68.9 ± 14.1 kg) between 18 and 30 years of age volunteered to participate. Eight fewer males than the original study [21] were included due to technical problems with continuous anterior load–displacement data acquisition The study was approved by the University Institutional Review Board and all participants gave written informed consent to participate. Participants were recreationally active (2.5–10 h/week) for the past 3 months and non-smokers, and had a body mass index (weight/height^2) ≤ 30 and no history of ligament or cartilage injury to the knee. Females were tested during the first 6 days of menses to control for any potential acute hormone effects on joint laxity [27] and subsequently, neuromuscular control [21]. To ensure inclusion of a broad range of knee laxity values, participants were prescreened on anterior knee laxity and needed to fall within a predetermined anterior knee laxity distribution matrix. All testing was performed on the dominant leg (preferred stance leg when kicking a ball). Subjects were familiarized to all study procedures approximately 2 weeks prior to testing, and were asked to refrain from any physical activity on the day of testing until all measurements were obtained. Subjects completed a 5-minute warm-up on a stationary bike before data collection.

To determine anterior knee laxity (AKL) and initial and terminal knee stiffness, instrumented joint arthrometer testing was performed. Procedures for obtaining anterior knee laxity (AKL) data and its measurement and consistency have been previously reported [27]. Following common clinical practice and previously established methods of instrumented knee laxity testing, AKL represented the anterior displacement (mm) of the tibia relative to the femur produced by an anterior load of 133 N applied to the posterior tibia with the knee flexed to 25 ± 5° using the KT-2000TM knee arthrometer (Medmetric Corp, San Diego, CA) [21,27] Real-time load and displacement data were collected from the three AKL trials and were exported to a spreadsheet for later calculation of incremental stiffness values [25]. Using methods previously described [25], we extracted the initial (0–20 N load) and terminal (100–130 N load) stiffnesses (N/mm). The average of the three trials was used for analysis. Measurement consistency and prediction were previously assessed on 38 males tested 2 weeks apart (Initial Stiffness − $ICC_{2,3}$ = 0.87, SEM = 4.2 N/mm & Terminal Stiffness − $ICC_{2,3}$ = 0.90, SEM = 5.3 N/mm) [25].

To measure anterior tibial acceleration during the initial phase of weight acceptance, tibiofemoral kinematics during 40% weight bearing acceptance was assessed with the Vermont Knee Laxity Device (VKLD) as described previously in detail [11]. The VKLD measures displacement of the tibia relative to the femur as the knee transitions from non-weight bearing to weight bearing, and characterizes the anterior–posterior load–displacement behavior of the knee [28]. Features of the VKLD include the capability to apply quantifiable loads to the tibiofemoral joint under the control of gravity, by first creating an absolute zero shear load condition across the tibiofemoral joint while it is un-weighted to establish a reproducible neutral initial position of the tibia relative to the femur, and then to apply standardized compressive loads through the ankle and hip axes of rotation of the limb to simulate weight-bearing [29].

Subjects were placed in the VKLD and the foot was strapped to the foot cradle connected to a calibrated six-degree force transducer. The second metatarsal was visually aligned to the anterior superior iliac spine (ASIS) and the greater trochanter and the lateral malleolus were aligned to the axes of the hip and ankle counter-weight systems respectively. These counter-weight systems were applied to the shank and thigh to eliminate gravity forces caused by the shank and thigh segments and created zero shear forces across the knee joint. Three electromagnetic position sensors (Mini Birds, Ascension Technologies, Colchester, VT USA) were attached on the midpoint of the lateral thigh, the center of the patellar and the midpoint of the shaft of the tibia. The centroid method estimated the center of rotation of the ankle, knee, and hip joints. After determination of joint centers, the ankle and knee were flexed to 90° and 20° respectively and subjects were asked to relax their leg muscles. Knee flexion angle (20°) was confirmed manually and with the electromagnetic position sensors. Once properly positioned in the VKLD, three anterior to posterior forces were applied to the tibia just below the knee joint line to standardize the neutral position of the knee joint at the beginning of every trial. An initial zero compressive load to the tibia was also confirmed prior to each trial with a six degree-of-freedom load transducer (Model MC3A, Advanced Medical Technology, Inc; Watertown, MA).

Prior to actual data collection, we performed 3–5 practice trials to further familiarize the subject with the weight acceptance trials. Once the zero compressive and shear load were obtained, compressive loads equal to 40% BW were applied by the release of the prescribed weight via a pulley system, which acted through the ankle and hip joint axes to simulate the transition from non-weight bearing to weight bearing (Fig. 1). The 40% BW load is consistent with what would be experienced during double-leg stance assuming 50% of BW applied to each leg and 10% of body weight distributed below the knee [11] and is intended to represent the early weight bearing phase [29]. Starting from a relaxed neuromuscular state, participants were instructed to respond to the axial force as quickly as possible