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The influence of internal and external tibial rotation offsets on knee joint and ligament biomechanics during simulated athletic tasks

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ABSTRACT

Background: Following anterior cruciate ligament injury and subsequent reconstruction transverse plane tibiofemoral rotation becomes underconstrained and overconstrained, respectively. Conflicting reports exist on how rotations influence loading at the knee. This investigation aimed to determine the mechanical effects of internal and external tibial rotation offsets on knee kinematics and ligament strains during *in vitro* simulations of *in vivo* recorded kinematics.

Method: A 6-degree-of-freedom robotic manipulator arm was used to articulate 11 cadaveric tibiofemoral joint specimens through simulations of four athletic tasks produced from *in vivo* recorded kinematics. These simulations were then repeated with 4° tibial rotation offsets applied to the baseline joint orientation.

Findings: Rotational offsets had a significant effect on peak posterior force for female motion simulations ($P < 0.01$), peak lateral force for most simulated tasks ($P < 0.01$), and peak anterior force, internal torque, and flexion torque for sidestep cutting tasks ($P \leq 0.01$). Rotational offsets did not exhibit statistically significant effects on peak anterior cruciate ligament strain ($P > 0.05$) or medial collateral ligament strain ($P > 0.05$) for any task.

Interpretation: Transverse plane rotational offsets comparable to those observed in anterior cruciate ligament deficient and reconstructed patients alter knee kinetics without significantly altering anterior cruciate ligament strain. As knee degeneration is attributed to abnormal knee loading profiles, altered transverse plane kinematics may contribute to this. However, altered transverse plane rotations likely play a limited role in anterior cruciate ligament injury risk as physiologic offsets failed to significantly influence anterior cruciate ligament strain during athletic tasks.

1. Introduction

Approximately 250,000 anterior cruciate ligament (ACL) injuries occur each year in the United States. (Johnson and Warner, 1993) While a multitude of risk factors have been associated with increased risk of ACL injury, (Boden et al., 2000; Ford et al., 2003; Ford et al., 2010; Griffin et al., 2000; Hewett et al., 2005; Kaeding et al., 2015; Myer et al., 2015) poor neuromuscular control leading to out-of-plane kinematics and kinetics at the knee during rapid deceleration and change of direction tasks are the predominant physical presentation at time of injury. (Krosshaug et al., 2007) These injuries are catastrophic and debilitating to knee health as ACL deficiency has been shown to alter kinematics and tibiofemoral contact areas within the knee. (Andriacchi et al., 2006; Andriacchi and Dyrby, 2005; Tashman, 2004) These conditions lead to abnormal loading of articular cartilage which likely contributes to rapid knee degeneration and degradation in knee quality of life following injury. (Andriacchi et al., 2006; Lohmander et al., 2004) Specifically, literature has demonstrated that, during gait, the internal tibial rotation at the knee increases by 3–12° following ACL rupture. (Andriacchi and Dyrby, 2005; Georgoulis et al., 2003)

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Similarly, ACL reconstructed knees exhibit increased external tibial rotation by 4° compared to healthy controls. (Tashman, 2004) These alterations caused by ligament deficiency and repair would suggest that internal/external tibial rotation plays a critical role in ligament loading and injury. Notwithstanding, internal/external tibial rotation has not been associated with ACL injury risk (Hewett et al., 2005), has not been factored into clinical nomograms that predict potential for ACL injury (Myer et al., 2010), and has not been identified as a primary or secondary source of ACL mechanical resistance during gait. (Nesbitt et al., 2013)

Internal and external tibial rotations alter transverse plane torques at the knee. (Bates et al., 2017b) The effect of isolated internal tibial torques on knee loading has been evaluated *via* biomechanical testing and the current literature is conflicted. During impact testing, it was found that the incorporation of internal tibial torques during simulated landing conditions increased both internal tibial rotation and ACL strain. (Oh et al., 2011; Oh et al., 2012) Despite this, an impact-driven ACL injury simulator found that isolated internal tibial torque offsets only produced ACL rupture in 7% of specimens during a simulated landings. (Levine et al., 2013) Further, isolated internal tibial torques applied *via* robotic manipulators to cadaveric joints at fixed knee flexion angles found that no significant internal tibial rotation differences existed between ACL-intact, ACL-deficient, and ACL-reconstructed specimens. (Keklikci et al., 2013) Additionally, the mechanical response to increased internal/external tibial rotation at the knee has not been quantified relative to dynamic *in vivo* kinematics. With the recent advent of new methods in robotic simulation technology, (Bates et al., 2015b) investigators can now simulate athletic tasks that are directly derived from *in vivo* recorded kinematics on cadaveric joints. Specifically, investigators can precisely rotate the tibia in the transverse plane in order to offset the natural tibiofemoral alignment of the knee joint and directly examine how these malalignments would impact intra-articular mechanics during motion tasks.

The objective of this investigation was to determine the mechanical effects of internal and external tibial rotation offsets on knee kinematics and ligament strains during *in vitro* simulations of *in vivo* recorded kinematics. It was hypothesized that the rotational offsets would alter the kinetic loading profiles without significant influence on ACL strain. The results of this study will help synthesize the ACL contributions to the resistance of internal/external rotation at the knee under physiologic conditions and, consequently, the importance of those rotations to ligament loading.

2. Methods

A total of 11 specimens from 9 unique donors were acquired for this investigation from an anatomical donation program (Anatomy Gifts Registry, Inc., Hanover, MD, USA). Two specimens from one unique donor were excluded as they exhibited non-functional ACLs during specimen preparation. This left 9 total specimens from 8 unique donors (age 46.1 (7.7) years; height 169 (12) cm; mass 87.8 (20.7) kg; BMI 30.5 (5.6)) for statistical analysis. Prior to statistical analysis, results for the contralateral specimens from the same donor were averaged into a single sample in order to avoid confounding the data. (Bates et al., 2017a) These adjustments left 8 total samples for the present analysis. Methods of specimen preparation have been previously documented. (Bates et al., 2015b; Boguszewski et al., 2011; Herfat et al., 2012b) Specimens were frozen at −20 °C and allowed to thaw 24 h before testing. Prior to undergoing biomechanical simulation, each specimen was resected of all soft tissue outside of the knee joint capsule (Fig. 1).

The method of robotically simulating knee joint motion adapted for use in the present study has previously been described in the literature. (Bates et al., 2015b) Briefly, three-dimensional motion analysis data was recorded from a male (age 24 years; height 175 cm; mass 675 N) and female (age 25 years; height 170 cm; mass 632 N) subject matched for age, height, mass, and athletic ability during drop vertical jump (DVJ) and sidestep cutting maneuvers. Positional data was collected using passive markers at 240 Hz with a 10-camera system (Eagle cameras, Motion Analysis Corp., Santa Rosa, CA, USA). Data were filtered through a fourth-order, low-pass, digital filter at 6 Hz and 3D joint kinematic were calculated through a Visual3D biomechanical model (version 4.0, C-Motion, Inc., Germantown, MD, USA) *via* custom MATLAB code (version 2012b, The MathWorks, Inc., Natick, MA, USA). (Ford et al., 2007) Mathematical factors were then applied to adjust the resultant kinematics and constrain skin skin-artifact errors as described in the literature. (Bates et al., 2015b) The adjusted kinematics were used as input to control robotically-driven simulations of knee motion.

Custom biomechanical fixtures were affixed to the tibia and used to define its mechanical axis. These tibial axis was then aligned and attached the specimen to the primary axis of a 6-axis load cell (Theta Model, ATI Industrial Automation, Apex, NC) mounted on the end effector of a 6-degree-of-freedom robotic arm manipulator (KR210; KUKA Robotics Corp., Clinton Township, MI). The femur was then secured to a fixed table, where a coordinate measuring machine (Faro Digitizer F04L2, FARO Technologies, Inc., Lake Mary, FL) was used to digitize anatomical landmarks across the specimen and define its joint coordinate system. (Grood and Suntay, 1983) This setup allowed the robotic manipulator to articulate the tibia around the femur according to the path defined by the recorded *in vivo* kinematics. Prior to simulation, each specimen was articulated to 45° of knee flexion and implanted with 3 mm microminiature differential variable resistance transducers (DVRT, LORD MicroStrain, Inc., Willinston, VT, USA) parallel to fibril alignment on the ACL and medial collateral ligament (MCL). (Beynnon et al., 1992; Levine et al., 2013)

Each specimen was simulated through four separate recorded motion tasks (male DVJ, female DVJ, male sidestep cut, female sidestep cut) in a randomized order. Prior to the simulation of each task, the specimen orientation was matched to within 0.5° of the *in vivo* limb orientation recorded at the point of initial contact with the ground. From this initial position, the specimens were incrementally loaded in compression until a peak force of 2.0–2.5 or 1.5–2.0 bodyweights was achieved for DVJ and sidestep cut simulations, respectively. (Bates et al., 2013; Bates et al., 2015a) All simulations were performed at room temperature while the specimens were consistently hydrated with saline. Following the baseline simulation for each task, a specimen would be offset through a 4° internal tibial rotation, then run through the same kinematic pattern. This step was then repeated for a 4° external tibial offset from the initial baseline orientation. The 4° offsets were selected relative to the shift in knee kinematics that has been observed following ACL-deficiency and ACL-reconstruction. (Andriacchi and Dyrby, 2005; Tashman, 2004) As biplane radiography has indicated the knee experiences approximately 8° of transverse plane rotation during a jump-cut maneuver, (Miranda et al., 2013) our selected offset relatively represents 50% of the knee's natural range of motion during an athletic task. Prior to each simulation, specimens were articulated through 10 preconditioning cycles that were followed by 10 cycles where data was collected. After all simulations were completed the knee was resected of all load-bearing structures save the ACL or MCL. In the isolated-ligament condition, the specimen was articulated back to initial contact orientation, compressed to an unloaded position, and then distracted until a constant force was registered on the force sensor. The resulting position was recorded as the neutral strain location for the ligament and used to calculate absolute strain throughout simulation. Once neutral strain was determined, the remaining ligament was resected and all simulations were performed in a tibia-only condition. Forces and torques generated in the tibia-only condition were subtracted from the previous simulations as they represent values generated from gravity and inertia. (Boguszewski et al., 2011; Herfat et al., 2012b)

During simulation, the force sensor recorded 6-DOF forces and torques in line with the tibial axis and extrapolated them to the knee joint center, while the implanted DVRTs recorded ligament strain. All