Published in final edited form as:

Clin Biomech (Bristol, Avon). 2018 February ; 52: 109–116. doi:10.1016/j.clinbiomech.2018.01.019.

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*

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CONFLICTS OF INTEREST: None

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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 ($P = 0.01$). 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.

Keywords

Anterior cruciate ligament; medial collateral ligament; jump landing; robotic manipulator; athletic task simulation

1. INTRODUCTION

Approximately 250,000 anterior cruciate ligament (ACL) injuries occur each year in the United States.(Johnson and Warner) While a multitude of risk factors have been associated with increased risk of ACL injury,(Boden et al.; Ford et al.; Ford et al.; Griffin et al.; Hewett et al.; Kaeding et al.; Myer et al.) 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.) 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) 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.) 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.) Similarly, ACL reconstructed knees exhibit increased external tibial rotation by 4° compared to healthy controls.(Tashman) 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.), has not been factored into clinical nomograms that predict potential for ACL injury(Myer et al.), and has not been identified as a primary or secondary source of ACL mechanical resistance during gait.(Nesbitt et al.)

Internal and external tibial rotations alter transverse plane torques at the knee.(Bates et al.) 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.) 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.) 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.) 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.) Specimens were frozen at -20°C and allowed to thaw 24 hours before testing. Prior to undergoing biomechanical simulation, each specimen was resected of all soft tissue outside of the knee joint capsule (Figure 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 kinematics were calculated through a Visual3D biomechanical model (version 4.0, C-Motion, Inc., Germantown, MD, USA) via custom MATLAB code (bversion 2012b, The MathWorks, Inc., Natick, MA, USA). (Ford et al.) 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. This 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) 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., Williston, VT, USA) parallel to fibril alignment on the ACL and medial collateral ligament (MCL). (Beynon et al.; Levine et al.)

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) As biplane radiography has indicated the knee experiences approximately 8° of transverse plane rotation during a jump-cut maneuver, (Miranda et al.) 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.)

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 forces and torques were reported as internal knee joint loads and analyzed in the tibial reference frame relative to the knee joint coordinate system.(Grood and Suntay) All forces and toques were low-pass filtered at 12 Hz through a Fourier transform. Translational forces were normalized to percent bodyweight and all data were time normalized to percent of landing phase which encompasses initial contact through minimum center of gravity.(Bates et al., 2013) All kinetic data were processed and calculated with custom MATLAB code.(Boguszewski et al.; Boguszewski et al.; Herfat et al.; Herfat et al.; Nesbitt et al.) The 8th and 9th cycles of each simulation were analyzed to eliminate cycle effects. A 3x1 ANOVA with Bonferroni post-hoc analysis was used to determine differences independently for each force (anterior/posterior, medial/lateral, compression/distraction), torque (adduction/abduction, flexion/extension, internal/external), and strains (ACL, MCL) between each of the three simulation offset conditions. Analyses were performed independently for each ligament and simulated task in JMP Pro (version 10.0.0, SAS Institute Inc., Cary, NC).

3. RESULTS

3.1 Kinetics

For both female simulated tasks, the rotational offsets had a significant effect on peak posterior force ($P < 0.01$). The 4° internal tibial rotation offset expressed greater peak posterior joint force than the 4° external tibial rotation offset ($P < 0.05$; Table 1). For three out of four simulated tasks, the rotational offsets also had a significant effect on peak lateral force ($P < 0.01$). Peak lateral force was lower in the 4° externally offset simulations than the 4° internally offset simulations ($P < 0.01$). Lateral force was also greater in the 4° internal offset than the baseline simulations for the female DVJ and male sidestep cut ($P < 0.04$). Within the sidestep cutting simulations, rotational offsets were found to have a significant influence on peak anterior force ($P < 0.01$), internal torque ($P < 0.01$), and flexion torque ($P = 0.01$). Peak anterior forces were greater during the 4° external offset than the 4° internal offset simulations ($P < 0.01$), while the 4° internal offset produced greater peak anterior force than the baseline condition during the female cut ($P = 0.04$). The 4° internal offset produced greater internal knee torque than the 4° external offset ($P < 0.01$). During the male sidestep cut task, the 4° external offset increased peak flexion torque relative to the 4° internal offset ($P < 0.01$).

At initial contact orientation, the rotational offsets had a significant effect on medial/lateral force ($P < 0.01$) and internal/external torque ($P < 0.04$) during all simulated tasks. The 4° internal offset increased medial joint force at initial contact relative to both the 4° external offset and baseline conditions ($P < 0.03$). The 4° internal offset also generated greater internal tibial torque than the 4° external offset condition at initial contact ($P < 0.04$). With

respect to frontal plane torque, the 4° internal rotational offset was found to incur greater adduction torque at initial contact during sidestep cutting tasks ($P < 0.04$). Additionally, in the sagittal plane, 4° internal offsets exhibited greater flexion torque at initial contact than the baseline condition ($P < 0.03$) and 4° external offsets ($P < 0.01$) for female DVJ simulations as well as the 4° external offsets ($P < 0.03$) during male sidestep cutting simulations.

3.2 Strains

Following a 4° internal rotational offset, 6 out of 8 subjects exhibited a magnitude increase of $> 1.0\%$ in peak ACL strain during the female DVJ simulation; however, these alterations were not statistically different from the peak strains recorded in baseline alignment (mean difference = 1.7% , [95% CI -2.5 to 5.8%], $P = 0.57$; Table 1, Figure 2 & 3). This trend was maintained across male DVJ (6 of 8 subjects $> 0.4\%$, mean difference = 1.0% , [95% CI -3.8 to 5.8%], $P = 0.86$), female sidestep cut (7 of 8 subjects $> 0.8\%$, mean difference = 0.5% , [95% CI -4.1 to 5.2%], $P = 0.96$), and male sidestep cut simulations (7 of 8 subjects $> 0.9\%$, mean difference = 1.4% , [95% CI -3.8 to 6.5%], $P = 0.78$). Similarly, following a 4° external rotational offset, 6 of 8 subjects during the female sidestep cut simulation and all 8 subjects during the male sidestep cut exhibited a $> 0.4\%$ decrease in the magnitude of peak ACL strain. Again, these magnitude alterations were not statistically different from baseline (mean difference = 0.5% , [95% CI -4.1 to 5.2%], $P = 0.96$ and mean difference = 2.4% , [95% CI -2.7 to 7.5%], $P = 0.48$, respectively). For the MCL, internal rotation offsets raised the peak ACL strain by 0.5% in all but 3 cases across all 32 simulations; however, these mean differences from baseline also lacked significance within each task (female DVJ mean difference = 1.4% , [95% CI -1.1 to 4.0%], $P = 0.34$; male DVJ mean difference = 0.8% , [95% CI -1.8 to 3.5%], $P = 0.72$; female sidestep cut mean difference = 0.5% , [95% CI -2.2 to 3.2%], $P = 0.89$; male sidestep cut mean difference = 1.5% , [95% CI -2.3 to 5.4%], $P = 0.59$).

4. DISCUSSION

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. The data presented was largely in support of the hypothesis that internal and external tibial rotation offsets would alter the loading profile at the knee joint without significantly altering strain in the ACL. That the present data supported this hypothesis is in agreement with previous study that found neither the ACL or MCL to be primary or secondary resistors to internal rotation at the knee.(Nesbitt et al.) This data is further in agreement with similar mechanical testing that found 4° internal and external rotational perturbations of the tibia at initial contact orientation did not affect significant changes on ACL or MCL strain.(Bates et al., 2017b)

ACL-deficiency and ACL-reconstruction have been shown to alter loading profiles within the knee.(Andriacchi et al., 2006; Andriacchi and Dyrby, 2005; Gardinier et al.; Georgoulis et al.; Nesbitt et al.; Tashman) Correspondingly, these conditions respectively induce slight offsets in internal and external tibial rotations during gait.(Andriacchi and Dyrby, 2005;

Tashman) These rotational changes imply that the ACL plays a significant role in the stabilization of internal tibial motion at the knee. However, the present results indicate that the role of the ACL in this regard is limited. The 4° internal offsets in the present study did consistently increase the magnitude of peak ACL strain by an average of 1.1% while external offsets decreased peak ACL strain by an average of 1.9% across all four simulated tasks. However, these differences were statistically insignificant and further lacked significance as the peak strains achieved for any condition were only approximately half of previously-reported failure strains for the ACL.(Levine et al.; Quatman et al.) The present data indicates that a greater than 4° rotational offset is necessary to induce a significant change on the ACL's contribution to the restraint of internal tibial rotation.

Computer modeling has demonstrated that a 4° transverse tibial rotation offset without any changes in loading resulted in a 44% increase in cartilage loss compared to the native knee. (Andriacchi et al.; Andriacchi and Dyrby) As the offsets in this literature were determined by the relative movement of tibial and femoral axes defined by each subject's bony geometry, they should be comparable to the transverse plane offsets in the present study due to similarities in magnitude and definition methodology. If a 4° offset drastically influences the progression of osteoarthritis following ACLR and has a limited effect on ACL strain response, as demonstrated in the present study; reconstruction grafts and additional passive structures alone may not provide sufficient restraint to constrain the knee to native mechanics in the transverse plane following injury. Clinically, a slightly noticeable physical offset that does not overload or threaten the integrity of the ACL can have grave consequences regarding long-term knee degeneration. For these reasons proper neuromuscular control of the joint is essential to reduce potential for future injury and degradation.(Hewett and Bates; Hewett et al.; Sugimoto et al.; Sugimoto et al.)

Previous literature is divided over the influence of internal and external rotations and torques at the knee. During previous robotic simulation of clinical tests on cadaveric knees at fixed flexion angles, internal rotation was similarly demonstrated to have little influence on ACL loading.(Keklikci et al., 2013) In these robotic simulations it was again found that isolated abduction torques had greater influence over ACL loads than isolated internal and external torques.(Battaglia et al.; Gadikota et al.) This concept is further supported in that knee abduction torque has been prospectively associated with ACL injury risk, whereas internal tibial torque has not.(Hewett et al.) Similarly, in testing that simulated ground impact during drop landings, it was found that additional internal torque applied to the tibia significantly increased strain on the ACL, but not to the level of applied abduction or combined abduction and internal rotation torques.(Levine et al.; Quatman et al.) However, a second method of ground impact simulation indicated that the incorporation of internal tibial torque was a secondary source of ACL loading as the ligament accounted for 13% of internal tibial torsion resistance during landing.(Oh et al., 2011) The present data was more supportive of the former studies described here. Though the internal offsets generally increased the peak ACL strain in all four simulated tasks, the magnitude of those increases were not statistically or clinically significant.

Literature has suggested that some ACL reconstructions may overconstrain the knee joint compared to the native ACL. This is exhibited in that ACL reconstruction patients

demonstrate about 4° greater external rotation at the knee during gait than healthy controls. (Tashman) This is further exemplified in that the magnitude of mean ligament force in ACL reconstructions has exceeded that of intact ACLs during simulated pivot-shift tests performed at low knee flexion angles. (Bates et al., 2015a) According to the present data, the external tibial rotation offset applied to simulate relative ACL reconstruction kinematics did decrease peak ACL strain in the specimens, but this change was not significant. With that considered, the peak ACL strain for the simulated male sidestep cut motion decreased by 44% between the baseline and 4° external offset conditions. Therefore, it is conceivable that this external offset induced by ACL reconstructions would stress shield native ACLs from natural loading, but the variability in change across specimens was too high to be statistically significant.

In the present study, internal rotation offsets increased the posterior force and lateral forces at the knee joint center relative to external rotational offsets. As all kinetics reported in the present data were calculated as internal to the knee joint, they must be inverted (with the exception of compression) when compared to the externally calculated knee kinetics commonly reported in motion capture analysis. Therefore, relative to the more typically reported convention of externally applied knee kinetics, internal rotation offsets generated greater anterior and medial knee forces. Both of these are associated with increased demand on the ACL, especially anterior knee force as the ACL stabilizes up to 87% of anterior tibial translation at the knee. (Butler et al.; Nesbitt et al.) However, the 4° external rotation offsets produced greater flexion moments at the knee joint center, which translate to larger external knee extension torques working externally on the knee. Such external knee extension moments are similarly associated with increased demand on the ACL. (Nesbitt et al.) As such, this counterbalance of increased ACL demand from both rotational offsets in various degrees of freedom may account for the lack of statistically significant increases in ACL strain.

A lack of significant change in MCL strain relative to internal and external rotational offsets was expected. The MCL does not contribute as a primary or secondary resistor to transverse plane torques at the knee. (Boguszewski, 2012; Nesbitt et al.) Additionally, the MCL has previously demonstrated a lack of response to 4° transverse plane rotational stimuli applied to the tibia at initial contact limb orientations. (Bates et al., 2017b) As such physiologically relevant magnitudes of knee rotation in this plane would not be expected to impart drastic change on the ligament.

As with all simulation studies, the present investigation faced limitations. One limitation of the current study is that the internal and external offsets applied were consistent throughout the duration of each simulation. Though offsets were recorded on *in vivo* subjects, as mentioned, these values likely varied throughout tasks execution. As offset data was not explicitly provided for time points throughout task performance, it would be impossible to estimate fluctuations; thus, a constant value was selected for simulation. Secondly, it would have been ideal to incorporate abduction and adduction offsets into the presented simulation in order to directly compare their effects on knee loading. However, it was discovered in pilot testing that specimens could often not sustain the loads generated by abduction and adduction offsets throughout the duration of a simulated task. As such frontal plane offsets

were not incorporated. A third limitation relates to the implantation of the DVRTs. DVRTs are optimized to function with rigid material; however, in the present study they were implanted in elastic ligaments. Though the custom barbs on our DVRTs help secure them in place, they still experience a small degree of play at the insertion site that may increase variability in reported strain and, thus, likelihood of statistically insignificant results. Further limitation exists in the inherent biologic variability that exists in tissue mechanical properties between specimens. Though robotic manipulators are highly-precise systems they do not compensate for variability between specimens.(Fujie et al., 1995; Fujie et al., 1993) This variability is observed in that ACL failure during simulated impact occurs at a mean of 18.7% strain, but exhibits a standard deviation of 10.0% and a range of 7% to 36%.(Levine et al.) This influence of this significant biologic variability in loading outcomes has been documented by the presence of substantial standard deviations in previous robotic-manipulator-driven investigations.(Woo et al.) Therefore, the standard deviations in the present study may be due variability in ACL mechanical properties that preclude statistical significance, in spite of robotic precision. Larger magnitudes of rotational offsets would likely have introduced significant changes in ACL strain, but also could have potentially lead to specimen damage during simulation.

5. CONCLUSIONS

Four degree transverse plane rotational offsets applied to the tibia prior to simulated DVJ and sidestep cutting tasks did not significantly alter the peak strain produced in the ACL or MCL during simulation. However, these offsets did significantly influence intra-articular kinetics in several DOFs. A 4° offset from neutral is comparable to the malalignment that has been reported following ACLR. While the 4° offset may not be detrimental to ACL strain during task performance, the altered intra-articular mechanics may contribute to the early onset osteoarthritis that is frequently reported within 20 years post ACL injury.

Acknowledgments

This work was supported by the National Institutes of Health/NIAMS Grants #R01-AR049735, #R01-AR05563, #R01-AR056660 and #R01-AR056259. The authors would also like to acknowledge the support of the staff at the Orthopedic Biomechanics Laboratory at Mayo Clinic and the Sports Medicine Biodynamics Laboratory at Cincinnati Children's Hospital.

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HIGHLIGHTS

- Robotic manipulator executed athletic task simulation on lower extremity specimens.
- 4° tibial rotation offsets applied to knee alignment prior to simulation.
- Offsets mimicked potential malalignment following ligament reconstruction.
- Offsets did not significantly alter knee ligament strains from baseline alignment.
- Offsets did alter intra-articular knee kinetics from baseline alignment.

Figure 1.

Stages of the testing methodology presented. A) Resect soft tissue outside of the major knee ligaments, B) resect the patella and patellar tendon, C) mount prepared specimen to 6DOF load cell on the end effector of a robotic manipulator using custom fixtures, D) implant DVRT on distal aspect of the anteromedial bundle of the ACL, E) implant DVRT sensors on the midsubstance of the MCL, F) use the robotic manipulator to articulating the tibia about the stationary femur and simulate *in vivo* recorded kinematic pathways. Step F) is performed with the tibia in a neutral orientation as well as offset by a 4° internal rotation and offset by 4° external rotation. Simulation is initiated with the specimen positioned to match the orientation of initial ground contact recorded for each task.

Figure 2.

Depiction of peak ACL and MCL strain values and standard deviations for all three rotational offset conditions for each motion task simulated. For both ligaments, there were no statistically significant differences in peak strains between offset conditions.

Figure 3.

Depiction of ACL and MCL strains values and standard deviations at initial contact for all three rotational offset conditions for each motion task simulated. For both ligaments, there were no statistically significant differences in initial contact strains between offset conditions.

Table 1

Displays force, torque, and ligament strain values generated during baseline, 4° internally offset, and 4° externally offset conditions for each simulated athletic task.

	Female DVJ			Female Sidestep Cut			Male DVJ			Male Sidestep Cut		
	Baseline	4° Internal	4° External	Baseline	4° Internal	4° External	Baseline	4° Internal	4° External	Base	4° Internal	4° External
Forces (BW)												
Posterior	0.11 (0.06)	0.18 (0.09)	0.06 (0.06) ϕ	0.07 (0.04)	0.11 (0.05)	0.03 (0.04) ϕ	0.12 (0.06)	0.16 (0.08)	0.08 (0.06)	0.05 (0.04)	0.08 (0.06)	0.02 (0.03)
Anterior	-0.15 (0.05)	-0.11 (0.06)	-0.17 (0.05)	-0.06 (0.02) ϕ	-0.02 (0.01) *	-0.09 (0.04) ϕ	-0.14 (0.04)	-0.11 (0.05)	-0.15 (0.05)	-0.14 (0.06)	-0.09 (0.06)	-0.20 (0.05) ϕ
Lateral	0.21 (0.05) ϕ	0.28 (0.06) *	0.11 (0.05) * ϕ	0.05 (0.03)	0.09 (0.05)	0.05 (0.03)	0.16 (0.07)	0.22 (0.05)	0.09 (0.05) ϕ	0.17 (0.06) ϕ	0.28 (0.09) *	0.1 (0.04) ϕ
Medial	-0.02 (0.03)	-0.03 (0.02)	-0.04 (0.06)	-0.03 (0.03)	-0.03 (0.02)	-0.05 (0.04)	-0.02 (0.04)	-0.02 (0.02)	-0.07 (0.08)	0.00 (0.01)	-0.02 (0.02)	-0.04 (0.05)
Distraction	-0.11 (0.34)	-0.06 (0.34)	-0.18 (0.30)	-0.33 (0.32)	-0.2 (0.25)	-0.42 (0.33)	-0.14 (0.44)	-0.07 (0.45)	-0.22 (0.47)	-0.07 (0.18)	-0.14 (0.16)	-0.09 (0.15)
Compression	-1.84 (0.19)	-1.95 (0.34)	-1.77 (0.17)	-1.67 (0.23)	-1.62 (0.25)	-1.61 (0.24)	-1.86 (0.34)	-1.87 (0.36)	-1.75 (0.39)	-1.68 (0.27)	-1.86 (0.37)	-1.56 (0.24)
A/P @ IC	-0.02 (0.02)	-0.03 (0.02)	-0.02 (0.03)	-0.02 (0.01)	0.00 (0.02)	-0.02 (0.03)	0.00 (0.03)	0.00 (0.02)	-0.01 (0.04)	-0.02 (0.02)	-0.03 (0.02)	-0.01 (0.02)
M/L @ IC	0.00 (0.01) ϕ	-0.03 (0.02) *	0.01 (0.01) ϕ	0.01 (0.01) ϕ	-0.02 (0.02) *	0.03 (0.02) ϕ	0.01 (0.01) ϕ	-0.02 (0.02) *	0.02 (0.01) ϕ	0.00 (0.00) ϕ	-0.02 (0.02) *	0.02 (0.01) ϕ
C/D @ IC	-0.51 (0.39)	-0.49 (0.39)	-0.50 (0.33)	-0.78 (0.31)	-0.71 (0.30)	-0.72 (0.23)	-0.60 (0.50)	-0.56 (0.40)	-0.56 (0.40)	-0.24 (0.33)	-0.27 (0.33)	-0.22 (0.29)
Torques (N°m/Kg)												
External	0.04 (0.02)	0.05 (0.03)	0.04 (0.03)	0.01 (0.01)	0.01 (0.02)	0.02 (0.01)	0.03 (0.03)	0.03 (0.03)	0.04 (0.03)	0.04 (0.02)	0.05 (0.03)	0.05 (0.02)
Internal	-0.01 (0.01)	-0.02 (0.01)	0.00 (0.01)	-0.01 (0.01)	-0.03 (0.01)	0.00 (0.01) ϕ	-0.01 (0.01)	-0.02 (0.01)	0.00 (0.01)	-0.01 (0.01)	-0.03 (0.01)	0.00 (0.01) ϕ
Extension	0.28 (0.13)	0.34 (0.22)	0.19 (0.12)	0.17 (0.10)	0.22 (0.15)	0.12 (0.09)	0.27 (0.17)	0.32 (0.18)	0.22 (0.17)	0.17 (0.10)	0.20 (0.13)	0.12 (0.08)
Flexion	-0.34 (0.17)	-0.31 (0.15)	-0.37 (0.21)	-0.12 (0.09)	-0.09 (0.07)	-0.17 (0.11)	-0.32 (0.19)	-0.30 (0.16)	-0.35 (0.22)	-0.36 (0.16)	-0.25 (0.15)	-0.52 (0.19) ϕ
Abduction	0.03 (0.09)	0.01 (0.09)	0.07 (0.08)	0.02 (0.07)	-0.02 (0.07)	0.05 (0.06)	0.01 (0.06)	-0.02 (0.08)	0.07 (0.07)	0.04 (0.06)	0.01 (0.06)	0.05 (0.05)
Adduction	-0.18 (0.13)	-0.18 (0.11)	-0.17 (0.12)	-0.18 (0.13)	-0.2 (0.11)	-0.18 (0.11)	-0.24 (0.16)	-0.25 (0.15)	-0.22 (0.15)	-0.24 (0.13)	-0.24 (0.11)	-0.25 (0.14)
I/E @ IC	0.00 (0.01)	-0.01 (0.01)	0.00 (0.01) ϕ	-0.01 (0.01)	-0.02 (0.01)	0.01 (0.01) ϕ	-0.01 (0.01)	-0.02 (0.01)	0.00 (0.01) ϕ	-0.01 (0.01)	-0.02 (0.02)	0.00 (0.00) ϕ
F/E @ IC	0.02 (0.04) ϕ	-0.06 (0.08) *	0.06 (0.04) ϕ	0.02 (0.03)	0.01 (0.04)	0.04 (0.03)	0.06 (0.04)	0.03 (0.06)	0.06 (0.07)	0.01 (0.04)	-0.03 (0.06)	0.03 (0.04) ϕ
A/A @ IC	-0.02 (0.05)	-0.04 (0.06)	0.01 (0.05)	0.00 (0.05)	-0.04 (0.06)	0.03 (0.04) ϕ	-0.02 (0.04)	-0.04 (0.06)	0.03 (0.07)	0.00 (0.02)	-0.03 (0.04)	0.01 (0.02) ϕ
Strains (%)												
ACL peak	6.3 (3.2)	8.0 (3.7)	6.1 (3.0)	4.2 (3.8)	5.2 (3.8)	3.4 (3.8)	6.7 (3.6)	7.2 (3.6)	7.2 (3.8)	7.2 (3.8)	8.5 (4.6)	4.8 (3.7)
ACL @ IC	3.8 (3.4)	4.8 (3.3)	2.8 (3.6)	3.8 (3.8)	4.5 (3.4)	2.7 (4.1)	3.0 (3.8)	4.2 (3.2)	2.4 (3.8)	3.4 (3.0)	3.6 (3.1)	2.4 (3.3)
MCL peak	1.8 (1.9)	3.3 (2.3)	1.0 (1.8)	0.6 (2.1)	1.1 (2.1)	0.4 (2.3)	1.1 (2.1)	1.9 (2.0)	0.7 (2.3)	2.5 (3.1)	4.0 (3.2)	1.7 (2.9)
MCL @ IC	0.2 (2.3)	0.4 (2.2)	0.3 (2.0)	0.4 (2.2)	0.6 (2.2)	0.4 (2.3)	0.4 (2.3)	0.6 (2.4)	-0.1 (2.7)	0.4 (3.0)	0.4 (2.7)	0.6 (2.8)

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* indicates statistically significant difference from baseline;
 ϕ indicates statistically significant difference from 4° internal rotation offset;
indicates statistically significant difference from 4° external rotation offset
BW = bodyweights; IC = initial contact; I/E = internal/external; M/L = medial/lateral; A/A = abduction/adduction