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# Timing sequence of multi-planar knee kinematics revealed by physiologic cadaveric simulation of landing: Implications for ACL injury mechanism

Ata M. Kiapour [a,b], Carmen E. Quatman [c,d], Vijay K. Goel [b], Samuel C. Wordeman [c,e], Timothy E. Hewett [c,d,e,f], Constantine K. Demetropoulos [g,*]

[a] Sports Medicine Research Laboratory, Department of Orthopaedic Surgery, Boston Children's Hospital, Harvard Medical School, Boston, MA, United States
[b] Engineering Center for Orthopaedic Research Excellence (ECORE), Departments of Orthopaedics and Bioengineering, University of Toledo, Toledo, OH, United States
[c] Sports Health and Performance Institute, The Ohio State University, Columbus, OH, United States
[d] Department of Orthopaedic Surgery, The Ohio State University, Columbus, OH, United States
[e] Department of Biomedical Engineering, The Ohio State University, Columbus, OH, United States
[f] Departments of Physiology and Cell Biology, Family Medicine and the School of Health and Rehabilitation Sciences, The Ohio State University, Columbus, OH, United States
[g] Biomechanics & Injury Mitigation Systems, Research & Exploratory Development Department, The Johns Hopkins University Applied Physics Laboratory, Laurel, MD, United States

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ABSTRACT

*Background:* Challenges in accurate, *in vivo* quantification of multi-planar knee kinematics and relevant timing sequence during high-risk injurious tasks pose challenges in understanding the relative contributions of joint loads in non-contact injury mechanisms. Biomechanical testing on human cadaveric tissue, if properly designed, offers a practical means to evaluate joint biomechanics and injury mechanisms. This study seeks to investigate the detailed interactions between tibiofemoral joint multi-planar kinematics and anterior cruciate ligament strain in a cadaveric model of landing using a validated physiologic drop-stand apparatus.
*Methods:* Sixteen instrumented cadaveric legs, mean 45(SD 7) years (*8 female and 8 male*) were tested. Event timing sequence, change in tibiofemoral kinematics (position, angular velocity and linear acceleration) and change in anterior cruciate ligament strain were quantified.
*Findings:* The proposed cadaveric model demonstrated similar tibiofemoral kinematics/kinetics as reported measurements obtained from *in vivo* studies. While knee flexion, anterior tibial translation, knee abduction and increased anterior cruciate ligament strain initiated and reached maximum values almost simultaneously, internal tibial rotation initiated and peaked significantly later ($P < 0.015$ for all comparisons). Further, internal tibial rotation reached mean 1.8(SD 2.5)°, almost 63% of its maximum value, at the time that peak anterior cruciate ligament strain occurred, while both anterior tibial translation and knee abduction had already reached their peaks.
*Interpretation:* Together, these findings indicate that although internal tibial rotation contributes to increased anterior cruciate ligament strain, it is secondary to knee abduction and anterior tibial translation in its effect on anterior cruciate ligament strain and potential risk of injury.


## 1. Introduction

Over 125,000 anterior cruciate ligament (ACL) injuries occur annually in the United States (Kim et al., 2011), mainly affecting the young athletic population. Non-contact injuries are reported to be the predominant mechanism of ACL injury (>70% of ACL injuries) (Griffin et al., 2000; Henrichs, 2004). These injuries often occur during landing with high ground reaction forces, muscle forces and segmental inertia (Boden et al., 2000; Olsen et al., 2004). Injury prevention strategies are an appealing option to avoid long-term joint instability, pain, and early development of osteoarthritis associated with ACL injury (Agel et al., 2005; Arendt and Dick, 1995; Hewett et al., 1999; Malone et al., 1993), as well as potential loss of sports participation (Maquirriain and Megey, 2006; van Lent et al., 1994) and high costs associated with surgical reconstruction (de Loes et al., 2000).

Noncontact ACL injury mechanisms are multi-planar in nature, involving tibiofemoral joint articulation in all three anatomical planes (Kiapour, 2013; Koga et al., 2010; Quatman et al., 2010). Despite considerable efforts to characterize ACL injury mechanisms (Agel et al., 2005; Arendt and Dick, 1995; Boden et al., 2000; Chappell et al., 2002; Decker et al., 2003; Ford et al., 2003; Griffin et al., 2000; Hewett et al., 1999, 2005; Joseph et al., 2011; Kiapour et al., 2013a,b; Koga et al., 2010; Krosshaug et al., 2007; Malone et al., 1993; Moran and Marshall, 2006; Olsen et al., 2004), the relative contribution of each loading axis in the

* Corresponding author at: Biomechanics & Injury Mitigation Systems, Research & Exploratory Development Department, The Johns Hopkins University Applied Physics Laboratory, 11100 Johns Hopkins Rd, Mail Stop: MP2-N143, Laurel, MD 20723, United States.
*E-mail address:* constantine.demetropoulos@jhuapl.edu (C.K. Demetropoulos).

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multi-axial (multi-planar) injury mechanism during landing is unclear. Due to the high-rate dynamic environment of injurious events, precise *in vivo* measurements of tibiofemoral joint six-degrees of freedom kinematics, its interaction with ACL tension and the associated timing sequence remain a challenge.

While clinical studies ultimately represent the gold standard for the evaluation of ACL injuries, studies of cadaveric biomechanics (*ex vivo*) under controlled laboratory conditions complement and often precede such work. Biomechanical testing of human cadaveric tissue offers a practical means for the investigation of various disorders, and can evaluate associated conservative and non-conservative treatments. *Ex vivo* techniques serve to enhance our knowledge of joint biomechanics and ligament functions, and generate direct measurements of mechanical parameters (i.e. force and strain) that are challenging, if not impossible to obtain *in vivo*. Further, these techniques provide a standard framework in which to conduct robust parametric studies.

Over the past three decades, extensive efforts have been undertaken to study ACL biomechanics utilizing *ex vivo* approaches (Bach and Hull, 1998; Berns et al., 1992; Butler et al., 1980; Csintalan et al., 2006; DeMorat et al., 2004; Draganich and Vahey, 1990; Durselen et al., 1995; Fukubayashi et al., 1982; Gabriel et al., 2004; Hashemi et al., 2010; Kiapour et al., 2012a; Markolf et al., 2004; Mazzocca et al., 2003; Meyer and Haut, 2008; Oh et al., 2012; Renstrom et al., 1986; Romero et al., 2002; Wall et al., 2012; Wu, 2010; Yeow et al., 2009; Zantop et al., 2007). The majority of these studies simulate low-rate, sub-injurious tasks through the application of static and/or quasi-static loading conditions (Bach and Hull, 1998; Berns et al., 1992; Butler et al., 1980; Csintalan et al., 2006; Draganich and Vahey, 1990; Durselen et al., 1995; Fukubayashi et al., 1982; Gabriel et al., 2004; Kiapour et al., 2012a; Markolf et al., 2004; Mazzocca et al., 2003; Renstrom et al., 1986; Romero et al., 2002; Wu, 2010; Zantop et al., 2007). Reported findings from such studies help to understand ACL biomechanics and overall joint function. However, they are not strong representations of high-rate (dynamic) injurious conditions that occur during high-risk activities (i.e. landing and cutting maneuvers).

Experimental strategies have been developed to replicate high-risk, potentially injurious conditions and reproduce ACL injury (DeMorat et al., 2004; Hashemi et al., 2010; Meyer and Haut, 2008; Oh et al., 2012; Wall et al., 2012; Withrow et al., 2006; Yeow et al., 2009). Such experiments have focused on a variety of causative factors including muscle loading (DeMorat et al., 2004; Hashemi et al., 2010; Wall et al., 2012; Withrow et al., 2008), axial compression (Meyer and Haut, 2008; Wall et al., 2012; Yeow et al., 2009), and off-axis external loads (Meyer and Haut, 2008; Oh et al., 2012; Withrow et al., 2006) to simulate landing. Yet, such models are primarily limited by non-physiologic simulation of dynamic loading conditions (i.e. sharp impact peaks generated by a small mass, lack of muscle forces and insufficient magnitudes of off-axis external loads), unlike those experienced during actual ACL injuries.

Due to the complex, multi-factorial dynamic nature of knee injuries, validated experimental models that simulate realistic inciting events leading to consistent physiologic injuries are essential. Such models can be utilized to study the overall interaction between knee joint kinematics/kinetics with ACL tension and further investigate the relative contribution of each loading axis in the overall risk of ACL injury. Hence, this study aims to develop a novel, physiologic cadaveric model of landing (as a well-established high-risk task in non-contact ACL injury (Olsen et al., 2004; Boden et al., 2000)) in order to investigate detailed interactions between tibiofemoral joint multi-planar kinematics and ACL strain. We hypothesized that there are significant differences in temporal knee joint kinematics in different planes such that the peak knee sagittal and frontal plane motions coincide with peak ACL strain, while knee axial rotation peaks significantly later. Detailed understanding of knee joint dynamic motion during high-risk activities can lead to improved knowledge of ACL injury mechanisms and associated risk factors. This may in turn help clinicians to optimize current prevention and rehabilitation strategies in an effort to minimize the high incidence of ACL injury and early-onset post-traumatic osteoarthritis.

## 2. Methods

### 2.1. Specimen preparation

Sixteen unembalmed fresh frozen cadaveric lower limbs, mean 45(SD 7) years (*8 female and 8 male*), were acquired. Each specimen was inspected visually, and by computed tomography (CT) and magnetic resonance imaging (MRI) for signs of soft or hard tissue pathology including indications of prior surgery, mal-alignment deformities and ACL disruption. Specimens were stored at −20 °C. Specimens were slowly thawed to room temperature 24 h prior to testing. Thawed specimens were sectioned at the mid-femoral shaft (30 cm above the joint line) and all soft tissue up to 15 cm proximal to the joint line were dissected. Subsequently, the remaining segment of the proximal femur of each specimen was potted in a 3.8 cm (1.5 in.) diameter polyvinyl chloride (PVC) tube with polyester resin for rigid attachment to the testing frame.

The quadriceps (rectus femoris) and hamstring (semitendinosus, biceps femoris and semimembranosus) tendons were then isolated and clamped inside metal tendon grips to allow for the application of simulated muscle loads. The remaining musculatures along with the skin were maintained intact. The foot and ankle were also maintained intact to provide a realistic load transfer interface. The exposed tissue around the knee joint was kept moist with 0.9% buffered saline solution at all times.

### 2.2. Testing apparatus

A novel testing apparatus was designed to maintain specimens in an orientation that simulates lower extremity posture during ground strike while landing from a jump (Fig. 1) (Kiapour et al., 2013c; Levine et al., 2013). The unconstrained nature of this experimental setup allows for a broad range of loading conditions (i.e. anterior shear force, knee abduction and tibial axial rotation) to be applied during simulated landing (Levine et al., 2013; Quatman et al., 2013). Each specimen was rigidly fixed at the proximal femur to a fixture with an embedded custom-designed six-axis load cell (B9401, Denton, Rochester Hills, MI, USA). Specimens were positioned inverted with the tibia orientated vertically and the foot positioned above the tibia. The knee was positioned at 25° of flexion to simulate the orientation of this joint during injurious events, as reported by video analyses of ACL injuries (Koga et al., 2010). The femoral fixture was able to rotate and translate about five axes (no translation in the Z-direction) in order to orient the tibia in line with the axis of the impactor, while maintaining 25° of knee flexion.

As shown in Fig. 1, the drop stand apparatus is comprised of two independent platforms (floor and impactor). The lower platform (floor platform) acts to simulate floor contact, while the upper platform (impactor platform) imparts a simulated ground reaction force (GRF) during landing. Six vertically aligned linear bearings (three on each platform) were used to maintain platform alignment and guide the motion of each platform during the simulated landing. A second six-axis load cell (2586, Denton, Rochester Hills, MI, USA) incorporated into the floor platform captured all forces and moments applied to the specimen during simulated landing representing the GRF.

Muscle forces were simulated by multiple cable-pulley systems along with static weights that served to apply constant forces to the quadriceps and hamstring tendons. Adjustable pulley systems were used to maintain the physiologic line of action of each muscle group (Fig. 2). In order to simulate different postures during landing, an external fixation frame with an integrated pulley system was rigidly attached to the tibia. Additional cable-pulley systems along with static weights were designed to produce forces to generate anterior tibial shear, and force couples to generate pure abduction/adduction and internal/