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New parameters describing how knee ligaments carry force *in situ* predict interspecimen variations in laxity during simulated clinical exams

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ABSTRACT

Knee laxity, defined as the net translation or rotation of the tibia relative to the femur in a given direction in response to an applied load, is highly variable from person to person. High levels of knee laxity as assessed during routine clinical exams are associated with first-time ligament injury and graft reinjury following reconstruction. During laxity exams, ligaments carry force to resist the applied load; however, relationships between intersubject variations in knee laxity and variations in how ligaments carry force as the knee moves through its passive envelope of motion, which we refer to as ligament engagement, are not well established. Thus, the objectives of this study were, first, to define parameters describing ligament engagement and, then, to link variations in ligament engagement and variations in laxity across a group of knees. We used a robotic manipulator in a cadaveric knee model (n = 20) to quantify how important knee stabilizers, namely the anterior and posterior cruciate ligaments (ACL and PCL, respectively), as well as the medial collateral ligament (MCL) engage during respective tests of anterior, posterior, and valgus laxity. Ligament engagement was quantified using three parameters: (1) *in situ* slack, defined as the relative tibiofemoral motion from the neutral position of the joint to the position where the ligament began to carry force; (2) *in situ* stiffness, defined as the slope of the linear portion of the ligament force–tibial motion response; and (3) ligament force at the peak applied load. Knee laxity was related to parameters of ligament engagement using univariate and multivariate regression models. Variations in the *in situ* slack of the ACL and PCL predicted anterior and posterior laxity, while variations in both *in situ* slack and *in situ* stiffness of the MCL predicted valgus laxity. Parameters of ligament engagement may be useful to further characterize the *in situ* biomechanical function of ligaments and ligament grafts.

1. Introduction

Injuries to musculoskeletal joints pose major clinical and public health burdens due to their high incidence across all ages and sexes leading to the frequent need for costly surgeries and extensive rehabilitation (Friel and Chu, 2013; Gage et al., 2012; Golan et al., 2016; Passanante et al., 2016; Roos et al., 2016; Simon et al., 2006; Tubbs et al., 2011; Yang et al., 2005). In the case of the tibiofemoral joint, over 6.5 million knee injuries were presented to U.S. Emergency Departments from 1999 to 2008, with nearly half (42%) of these injuries being sprains and strains of cruciate and collateral ligaments (Gage et al., 2012). Patients who experience such an injury are five times as likely to develop post-traumatic osteoarthritis (PTOA) within 15 years (Gelber et al., 2000; Roos, 2005), leading to loss of function and a decreased quality of life, often during their most productive working years (ages 25–50) (Friel and Chu, 2013; Lyman et al., 2009).

Wide variations exist in the native physiology of the knee joint, most notably the structural properties of the soft tissues and the geometries of the articular surfaces, both of which play major stabilizing roles (Butler et al., 1992; Hashemi et al., 2010). Variations in knee physiology manifest in highly heterogeneous laxity, defined as the net translation or rotation of the tibia relative to

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the femur in a given direction in response to an applied load, across the population, especially between males and females (Boguszewski et al., 2015; Hsu et al., 2006; Park et al., 2008; Roth et al., 2015; Shultz et al., 2007). These laxity variations are associated with the risk of ligament injury (Uhorchak et al., 2003; Vacek et al., 2016) and of graft reinjury following surgical ligament reconstruction (Magnussen et al., 2016); however, little work has been done to explain variability in tibiofemoral laxity across individuals.

One potential explanation for interpersonal variations in knee laxity lies in the variability in the biomechanical function of ligaments. Most studies describe the *in situ* biomechanical contribution of knee ligaments in cadaveric models by reporting the force carried by a given structure at the peak applied load, where *in situ* ligament force is measured using robotic technology or by instrumentation of the ligament (Kanamori et al., 2000a, 2000b; Markolf et al., 1990; Rudy et al., 1996; Sakane et al., 1997). Although this parameter is critical for ranking the contribution of ligamentous stabilizers, it is not known whether *in situ* ligament force is related to variations in knee laxity. Moreover, the way in which force builds in ligaments *in situ* as the knee moves through its passive envelope of motion, which we refer to as ligament engagement, has not been explored. To describe ligament engagement beyond the commonly-used ligament force at the peak applied load, we established two additional parameters. Namely: (1) *in situ* slack, defined as the relative tibiofemoral motion from the neutral position of the joint to the tibial position where the ligament began to carry force; and (2) *in situ* stiffness, defined as the slope of the linear portion of the ligament force – tibial motion response.

In light of the association between variations in knee laxity from person-to-person and risk of ligament injury, we then used a cadaveric model to assess whether interspecimen variations in knee laxity were related to variations in engagement of important intra- and extra-articular ligaments, namely, the ACL, posterior cruciate ligament (PCL), and medial collateral ligament (MCL) in their primary stabilizing directions. We hypothesized that variations in parameters of ligament engagement would be associated with interspecimen variations in laxity of the tibiofemoral joint.

2. Materials and methods

This study was approved by the Hospital for Special Surgery institutional review board. 20 fresh-frozen human cadaveric knee specimens (mean age: 44.8 ± 13.5 years; range: 20–64; 14 male) were obtained from a non-profit anatomic donation organization. To prepare specimens for testing, soft tissues more than 10 cm from the joint line were removed. Skin, fat, and musculature (except for the popliteal muscle-tendon complex) were removed, leaving only the ligamentous and capsular tissues. A medial parapatellar arthrotomy was performed to confirm that ligaments and menisci were intact and to assure absence of gross chondral damage or prior surgery. Additionally, specimens underwent computed tomography imaging to ensure that they were free of osteophytes and other osseous abnormalities. Subsequently, the proximal femur and distal tibia were potted in cylindrical tubes of bonding cement (Bondo, 3M, St. Paul, MN, USA).

A six degrees-of-freedom serial robot (position repeatability: ±0.3 mm) instrumented with a universal force/moment sensor (Theta, ATI, Apex, NC, USA) was used to manipulate the specimens (ZX165U, Kawasaki Robotics, Wixom, MI, USA) (Fig. 1). The potted femur was rigidly fixed to the ground, and the potted tibia was mounted in full extension to the robot's end effector via custom fixtures. The specimens were covered in saline-soaked gauze to keep the soft tissues hydrated throughout testing (Viidik, 1973).

Anatomical landmarks were located using a three-dimensional digitizing arm with ±0.23 mm accuracy (G2X, MicroScribe, San Jose, CA, USA). These landmarks were the medial and lateral femoral epicondyles, the fibular insertion of the lateral collateral ligament, the superficial MCL about 1 cm distal to the joint line, and the distal tibia. These landmarks were used to define a coordinate system for the tibiofemoral joint such that all kinematics and loads were described using anatomical directions via established methods (Grood and Suntay, 1983; Imhauser et al., 2013).

Once mounted to the robot's end effector, the knee was flexed from full extension to 90° flexion in 1° increments using a previously described force-feedback algorithm that was implemented using custom MATLAB code (Mathworks, Natick, MA, USA) (Prisk et al., 2010). At each flexion angle, 10 N of compressive force was applied to the tibia to ensure bicondylar contact while all other forces and moments were minimized to within 5 N and 0.4 Nm, respectively. The position and orientation of the tibia relative to the femur, identified by this algorithm, served as the starting point from which all other loads were applied.

Soft tissues were preconditioned by determining the tibiofemoral kinematics in response to an anterior force of 134 N at 30° flexion and in response to rotatory loads (8 Nm of valgus torque with an additional 4 Nm of internal rotation torque) at 15° flexion; each trajectory was then repeated 10 times (Imhauser et al., 2013).

After preconditioning, neutral positions of the tibia relative to the femur were defined in the anterior-posterior (AP) and varus-valgus (VV) directions so that ligament engagement could be described relative to a consistently defined starting position in each direction across all 20 knees (Fig. 2). To determine an AP neutral position, 0, 10, 25, 50, 75, 100, and 134 N of anterior and posterior force were applied to the tibia with the knee held at 30° flexion and at 90° flexion. As the loads were applied, the tibia was free to translate and rotate in every direction save for flexion/extension, and the resulting kinematic trajectories were recorded. With these kinematics, the AP laxity profile, or applied load-displacement response for each knee, was characterized. Using a previously defined 'Kneedle' algorithm (Satopää et al., 2011), the points of maximum curvature in the anterior and posterior directions (i.e., the points where the joint transitioned from less stiff to more stiff) were identified. The AP neutral position was then defined as the midpoint of these transition points (Fig. 2). To determine a VV neutral position, 10 N of compressive force was applied to the tibia while it was held at 15° flexion and free to move in the other five degrees of freedom. This identified the tibiofemoral orientation in the frontal plane where bicondylar contact occurred.

The ACL was engaged by applying 134 N of anterior force to the tibia with the flexion angle fixed at 30° (Fig. 1B) (Daniel et al., 1985). The PCL and MCL were engaged by applying 134 N of posterior force and 8 Nm of valgus torque to the tibia, respectively (Fig. 1C, D) (Mauro et al., 2008; Schafer et al., 2016). The positions and orientations of the tibia relative to the fixed femur were recorded while the applied anteroposterior force or valgus torque was respectively increased to 134 N (0, 10, 25, 50, 75, 100, and 134 N) or 8 Nm (0, 1, 2, 3, 4, 5, 6, 7, and 8 Nm). Laxity was defined as the net translation or rotation of the tibia relative to the femur in the direction of the applied load in response to 134 N in both the anterior and posterior directions and 8 Nm in valgus. Laxity was measured relative to the aforementioned AP and VV neutral positions.

The principle of superposition was used to measure the force carried by each ligament *in situ* (Fujie et al., 1995; Woo et al., 1998). Specifically, the kinematics recorded during anterior, posterior, and valgus loading scenarios were repeated immediately before and after sectioning the ACL, PCL, and MCL, respectively. Performing vector subtraction on the reaction forces recorded by