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Identification of optimal laxity tests to load individual parts of knee ligaments

Michael Skipper Andersen[1] and Dennis Pedersen[1]

[1]Department for Materials and Production, Aalborg University, Denmark
Email: msa@mp.aau.dk, Web: http://www.biomechanics.mp.aau.dk/

INTRODUCTION

Knee instability can arise for several reasons, including ligament injuries, osteoarthritis or joint replacements and can lead to pain, joint degradation, decreased mobility, and an overall reduced quality-of-life [1].

To advance the understanding of basic joint mechanics and to develop patient-specific interventions for joint stability, it is important to gain insight into the mechanical properties of the knee ligaments. This information can also be used in computational knee models applicable to study knee mechanics and how the knee is affected by surgical or non-surgical interventions for instance to treat knee instability related to ligament reconstruction and knee arthroplasty.

The primary approach to investigate the mechanical properties of ligaments is to perform in *vitro tests* [2] as there currently are no way to measure these tissues *in vivo*. To overcome this, recent studies have proposed to employ laxity measurements to assess the overall joint laxity and from these measurements estimate the ligament properties using optimization-based techniques [3]. Currently, these studies have applied laxity tests that closely resemble those that clinicians use for manual tests, e.g. internal/external rotation, varus/valgus etc. While these loading directions are easy to understand and perform manually, it is not given that these are the best load cases in order to identify individual ligament properties.

We have developed a novel technology capable of applying any load case while capturing the knee translations and rotations using biplanar x-rays [4]. This technology enables both loads along the usually applied directions and combined load cases. This facilitates designing experimental procedures optimized for the purpose of identifying mechanical properties of ligaments *in vivo*.

Therefore, the purpose of this study was to identify laxity tests that results in the largest force in a desired ligament bundle while minimizing the loads in all other ligaments and keeping the load magnitudes on the same level as currently applied.

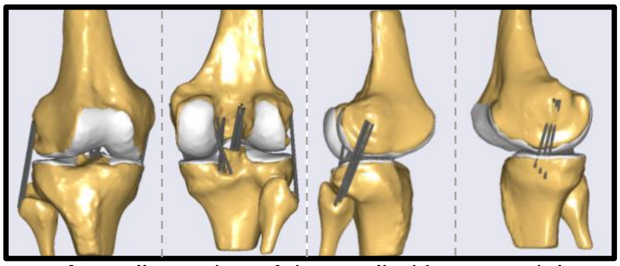

Fig 1: Illustration of the applied knee model.

METHODS

To gain an understanding of how external loads affect the loads of the knee ligaments, we apply a subject-specific knee model (Fig. 1).

A series of Medical Resonance Imaging (MRI) scans of the right leg of a female subject (27 year-old, 1.72 m, 61 kg) were obtained. From these, we segmented the bone and articular cartilage of the femur and tibia, and the anterior cruciate (ACL), posterior cruciate (PCL), medial collateral (MCL), and lateral collateral (LCL) ligaments. For each ligament, the origin and insertions were identified. Bony landmarks for the definition of anatomical coordinate systems according to ISB recommendations were also identified.

The model was developed in the AnyBody Modeling System v. 7.1 (AMS, AnyBody Technology, Denmark). The femur was fixed relative to the global coordinate system and we modelled the tibiofemoral joint using the Force-dependent Kinematics (FDK) approach [5]. The contact between the femoral and tibial articular cartilage was modelling using an elastic foundation model (pressure modulus of 10 GNm^{-3}) and the ligaments modelled as multiple line elements each with a slack, toe and linear region [6]. The ACL was modelled using four elements whereas PCL, MCL and LCL were modelled with three each. The ligament properties of were adapted from the literature [6].

As inputs to the model, we provided the knee flexion angle as well as an applied force and an applied moment to tibia. The force was applied to the tibial tuberosity and the moment as a pure moment in the tibial ISB anatomical coordinate system.

We introduced five FDK degrees-of-freedom (DOF) that were allowed to equilibrate according

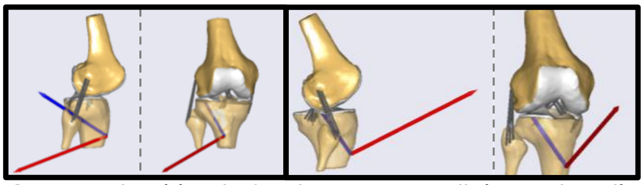

Fig 2: Optimal loads for the anteromedial ACL bundle. The first two figures from the left show one load case and the next two, the other. The red and blue arrows show the force and moment vectors, respectively.

to the applied loads, ligament and contact forces. A reaction moment around the knee flexion axis was included to simulate a fixated knee flexion as typically done during laxity tests. To identify what we will denote the optimal load cases to strain each bundle of the ligaments, we set up the following optimization problem:

$$\max_{\mathbf{x}^1,\ \mathbf{x}^2} \left(F_i^2 - F_i^1\right)^2 - \frac{1}{N-1}\sum_{j=1, j\neq i}^{N} \left(F_j^2 - F_j^1\right)^2 \qquad (1)$$

where $\mathbf{x}^1$ and $\mathbf{x}^2$ denote two load cases (i.e. each containing the knee flexion angle and the applied force and moment on tibia). F_k^1 and F_k^2 denote the force in the *k*th ligament bundle under load case $\mathbf{x}^1$ and $\mathbf{x}^2$, respectively. The first term in the equation specifies the squared difference in the ligament bundle of interest (*i*th) from which the second term subtracts the average force in all other ligament bundles. N denotes the number of ligament bundles.

To solve this optimization problem, we applied a Monte Carlo sampling approach. First, we applied 104.000 random load cases, containing the knee flexion angle, and tibial forces and moments and recorded the resulting ligament forces. The knee angle was sampled in the interval [0°:90°], and the tibial forces and moments with magnitudes of [0 N: 150 N] and [0 Nm: 10 Nm], respectively. The full 0° to 360° rotation around two axes were sampled for the direction of both the tibial force and moment vectors. Subsequently, we searched through all samples for the two load cases that would result in a maximization of the optimization problem in (1).

RESULTS AND DISCUSSION

Load cases for all ligaments were found but for the sake of brevity, we only show results for ACL and PCL (Table 1). Fig 2. illustrates the identified load cases for the anteromedial bundle of ACL.

For the anteromedial, anterolateral and posteromedial ACL bundles, the optimal knee angles were between 4.5° and 22.1° and with a clear anterior and posterior force difference between the two loads for each ligament. The posteromedial bundle of ACL, however, required one load at 7.3° and the other at 93.2° but still with a clear anterior and posterior load difference. The PCL bundles on the other hand required markedly different load cases both in terms of knee flexion angles, forces and moments.

CONCLUSIONS

In this study, we identified optimal loads to strain individual ligament bundles. This information can be applied to develop future measurement protocol, ultimately leading to a better assessment of knee ligament properties *in vivo.*

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Table 1: Optimal loads for the different branches (anteromedial (am), anterolateral (al), posteromedial (pm), posterolateral (pl) of ACL and anterior, mid and posterior of PCL). The forces and moments are reported in the tibial ISB coordinate system with standard abbreviations for anterior-posterior (ap), superior-inferior (si) etc.

Ligament	Knee flexion [°]	$\mathbf{F_{ap}}$ [N]	$\mathbf{F_{SI}}$ [N]	$\mathbf{F_{ML}}$ [N]	$\mathbf{M_{vv}}$ [Nm]	$\mathbf{M_{IE}}$ [Nm]	$\mathbf{M_{FE}}$ [Nm]
ACL am	6.0	-120.0	-44.3	-28.0	-8.4	5.3	-0.6
	22.4	95.8	65.0	-92.6	-5.4	6.1	-0.2
ACL al	15.6	-128.8	0.9	76.4	0.4	5.5	2.1
	11.2	126.0	-25.1	-43.8	-2.1	7.6	4.9
ACL pm	7.4	137.9	16.8	-19.5	-1.1	7.4	-4.9
	93.2	-74.3	-29.6	-114.4	-0.3	4.2	2.2
ACL pl	7.4	137.9	16.8	-19.5	-1.1	7.4	-4.6
	4.5	-120.0	8.2	-87.2	-1.2	4.6	-0.2
PCL a	82.1	-13.2	10.0	12.2	0.5	-1.4	-0.6
	49.7	-18.5	95.3	10.6	0.8	-0.7	1.1
PCL m	38.6	-91.9	-47.6	102.9	1.1	1.3	0.3
	15.7	-12.0	85.7	0.5	5.3	3.9	-1.7
PCL p	0.5	120.6	20.6	74.6	1.9	-1.3	1.2
	1.3	12.7	40.5	-0.5	-0.1	9.7	0.1