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Use of distraction loading to estimate subject-specific knee ligament slack lengths

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ABSTRACT

Knee ligaments guide and restrain joint motion, and their properties influence joint mechanics. Inverse modeling schemes have been used to estimate specimen-specific ligament properties, where external joint forces are assumed to balance with internal ligament and contact forces. This study simplifies this assumption by adjusting experimental loads to remove internal contact forces. The purpose of this study was to use novel experimental loading in an inverse modeling scheme to estimate ligament slack lengths, perform validation using additional loading scenarios, and evaluate sensitivity to the applied loading. Joint kinematics and kinetics were experimentally measured for a set of load cases. An optimization scheme used a specimen-specific forward kinematics model to estimate ligament slack lengths by minimizing the residual between model and experimentally measured kinetics. The calibrated model was used for a form of validation by evaluating non-optimized load cases. Additionally, uncertainty analysis related kinetic errors to previously reported kinematic errors. The six DOF tibial reactions realized RMS errors less than 23 N and 0.75 Nm for optimized load cases, and 33 N and 2.25 Nm for the non-optimized load cases. The uncertainty analysis, which was performed using the optimized load cases, showed average kinetic RMS errors less than 26 N and 0.45 Nm. The model's recruitment patterns were similar to those found in clinical and cadaveric studies. This study demonstrated that experimental distraction loading can be used in an inverse modeling scheme to estimate ligament slack lengths with a forward kinematics model.

1. Introduction

Knee ligaments guide and restrain joint motion, and their recruitment during passive motions is an area of focus for healthy and diseased joints before and after surgical intervention. Ligaments have been shown to influence knee kinematics and contact mechanics when subject to both laxity (Baldwin et al., 2009; Dhaher et al., 2010; Lenhart et al., 2015) and dynamic (Smith et al., 2016) loading conditions. Computational knee models offer the ability to evaluate complex loading conditions that may be difficult or impossible to evaluate *in vitro* and *in vivo*. These models have been used to evaluate the effects of gait variations (Ardestani et al., 2015), injury (Ali et al., 2016), and surgical interventions (Thompson et al., 2011; Amiri and Wilson, 2012; Smith et al., 2016) on joint mechanics. Regardless of the modeling goal, specimen-specific ligament properties are needed to accurately reflect specimen-specific joint behavior (Ewing et al., 2015).

Inverse modeling schemes can be used to estimate specimen-specific ligament properties. These schemes typically utilize optimization to calibrate a computational knee model using experimentally measured joint loads and kinematics (Blankevoort and Huiskes, 1996; Baldwin et al., 2012; Ewing et al., 2015; Harris et al., 2016). Within this process, the optimization iteratively adjusts ligament properties to minimize the residual between model and experimental measurements. This approach assumes that, for a given joint position, the known external joint forces balance with unknown internal ligament and contact forces (Blankevoort and Huiskes, 1996). Regarding the boundary conditions, studies normally adopt a pseudo forward dynamics approach, where experimentally measured joint loads are applied and the model's kinematics are determined. Studies normally use a subset of the available six degrees of freedom (DOFs) in the knee in the optimization's objective function. For instance, Blankevoort and Huiskes (1996) included one rotational and translational

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DOF in their objective because it was assumed that the other DOFs were more dependent on the articular surface, and not sensitive to ligament slack length. Similarly, others have used optimization schemes where two (Baldwin et al., 2012) or three (Harris et al., 2016; Ewing et al., 2015) DOFs are included in the objective function. Regardless of the chosen boundary conditions and the objective, the goal is to minimize the differences between model and experimental force-displacement behavior. Ideally, the result is a mechanically consistent and predictive model of a given knee. As knee kinematics are influenced by ligament properties (Baldwin et al., 2009; Dhaher et al., 2010; Smith et al., 2016) as well as articular contact, there may be benefits to simplifying the approach by removing the complexity of contact. Eliminating contact forces would reduce the number of unknown internal joint loads, and leave ligament forces as the only unknowns in the inverse modeling scheme.

Adjusting experimental laxity-style loading to apply joint distraction, while also removing articular contact, results in a joint's force-displacement behavior being primarily driven by ligaments. This approach is not without precedent, as a gap balanced surgical technique for total knee arthroplasty (TKA) may use distraction forces to tension the joint's soft tissue restraints (D'Lima and Colwell, 2017). Similar experimental distraction loads applied to a resected specimen could be used in an optimization framework to estimate ligament properties (Blankevoort and Huiskes, 1996; Baldwin et al., 2012; Ewing et al., 2015; Harris et al., 2016). In addition, removing articular contact allows for a computationally efficient forward kinematics model to be used, where it can be assumed that the passive six DOF kinetics are primarily due to soft tissue restraint. Including all DOFs when calibrating ligament properties is important because it has been shown that the limits of passive joint motion are not related within a knee (Roth et al., 2015).

Utilizing experimental distraction loading in the absence of articular contact with established inverse modeling methods may be a robust and computationally efficient approach to estimate ligament properties. Previous work, which also focused on the envelope of motion, has shown that joint kinematics are more sensitive to ligament slack length than stiffness (Baldwin et al., 2009). This surgically relevant loading state provides an opportunity to limit joint restraint to the ligaments and to reduce the complexity of the joint model needed for calibration. The purpose of this study was to use novel experimental loading in an inverse modeling scheme to estimate ligament slack lengths, perform validation using additional loading scenarios, and to evaluate sensitivity to the applied kinematics. Note the sensitivity analysis was performed to relate the kinetic errors found in this current study to kinematic errors reported in previous studies.

2. Methods

2.1. Specimen preparation

Experimental joint distraction tests were conducted on one knee specimen (male, age 62, with a BMI of 21). The specimen was prepared following the Open Knee(s) protocol (Erdemir et al., 2015). Following initial preparation and MR imaging, the specimen was dissected by an orthopedic surgeon (BNS), which included removal of the skin, muscle, patella and 1 cm thick distal and posterior femoral bone cuts (Fig. B.1). See Appendix A.1 for further details regarding specimen preparation and digitization.

2.2. Experimental testing

A series of five laxity-style distraction loading profiles were applied to the specimen with a six DOF robot (Robotpod 2000, Mikrolar, Hampton NH, USA) with a custom alignment fixture (Fig. B.1). Each loading profile was repeated at 0°, 30°, 60°, and 90° knee flexion. Loads were applied with respect to the tibia's fixed coordinate system, and kinematics were defined with respect to the tibiofemoral joint coordinate system (Grood and Suntay, 1983). All off-axis loads were minimized throughout each test (Table B.1). Joint kinematics and 6 DOF tibial loads were measured throughout testing. Justification of experimental loads can be found in Appendix A.2 (Nagai et al., 2014).

2.3. Rigid body model

The MR images were used to generate specimen-specific joint geometry. Osseous femur and tibia surfaces were segmented, and digitized points of the resected femoral surfaces were used to digitally resect the femoral surface. Ligaments were modeled as 11 fiber bundles (Fig. B.2), and their insertion sites were defined using the specimen's MR images. The anterior and posterior cruciate ligaments were modeled as four bundles, the anteromedial and posterolateral anterior cruciate ligament (amACL, plACL) (Duthon et al., 2006) and the anterolateral and posteromedial posterior cruciate ligament (alPCL, pmPCL) (Anderson et al., 2012). The medial collateral ligament (MCL) was modeled as three bundles, with two composing the proximal and distal superficial MCL (psMCL, dsMCL), and one bundle defined the deep MCL (dMCL) (LaPrade, 2007). Additional ligament bundles included the lateral collateral ligament (LCL), the popliteofibular ligament (PFL) (LaPrade et al., 2003), the anterolateral ligament (ALL) (Claes et al., 2013), and the oblique popliteal ligament (OPL) (LaPrade, 2007; Hedderwick et al., 2017).

The margins of each ligament bundle's femoral and tibial insertion were defined with four points, two on the femur and two on the tibia. These points defined two fibers, and 23 fibers were evenly spaced between the margins (Fig. B.2). Each fiber was modeled as a nonlinear spring (Blankevoort and Huiskes, 1996). Every spring in the ligament bundle had a uniform stiffness value, which summed to the equivalent stiffness of the ligament bundle (Table B.2, Amiri et al., 2007; Kia et al., 2016).

Experimental kinematics were applied to the geometry, and the length of each ligament fiber was used to calculate the magnitude of force carried by each fiber. A cylinder was used to approximate wrapping of the dsMCL around the medial tibial condyle, and the femoral surface was used to approximate wrapping of the psMCL, dMCL, ALL, and PFL. Each fiber's line of action was defined using the tibial and femoral insertion sites for non-wrapping ligaments, and wrapping ligaments accounted for femoral or tibial wrapping points.

Tibial reaction forces were calculated by summing the force carried by each ligament fiber along its line of action (Mommersteeg et al., 1996), and when applicable, the wrapping reaction forces. Tibial reaction moments were calculated with tibial insertion points, wrapping points, and each fiber's force magnitude and line of action.

2.4. Optimization

The slack length of each ligament bundle was estimated with a constrained optimization using a sequential quadratic programming method (Jones et al., 2001). The lengths of the two fibers that defined the margins femoral and tibial insertion were used to define the slack length of each fiber in a ligament bundle. The slack lengths of the remaining fibers in each bundle were defined using linear interpolation (Fig. B.2)). A total of 22 control variables were used in the optimization, and each control variable was bounded between 10 mm and 110 mm.

The optimization minimized the sum of the squared residual between the model (M) and experimentally (E) measured kinetics for two tests, the anterior-posterior drawer and varus-valgus tests at 0°, 30°, 60°, and 90° flexion (Eq. (1)).

$$\underset{\boldsymbol{x}}{\text{minimize}} \;\; f(\boldsymbol{x}) = \sum_{i=1}^{4}\sum_{j=1}^{18}\sum_{k=1}^{6}\left[w_k(M_{ijk}(\boldsymbol{x}) - E_{ijk})\right]^2 \text{subject to}$$
$$h(x_m) \geqslant 0.1, m = 1, \ldots, 22 \qquad (1)$$

where $\boldsymbol{x}$ is the set of slack lengths, i indicates the flexion angle of the test, j is the index of the point in a loading cycle (with 10 points in the anterior-posterior drawer test, and 8 points for the varus-valgus test), and k is the index of the kinetics DOF, and m is the index of the control variable that defines a fiber's slack length at the margin of a specific ligament. Forces and moments are weighted by $w_k = [1, 1, 1, 20, 20, 20]$, where forces have a weight of 1, and moments have a weight of 20. Weights were selected to equate 20 N of force error to 1 Nm of moment error.

There is one constraint ($h(x_m)$) for every control variable in Eq. (1). These constraints ensure that the fibers that define the margins of the femoral and tibial ligament bundle insertions (Fig. B.2) achieve a total force magnitude greater than or equal to 0.1 N throughout the loading cycle.

The optimization was repeated 1000 times with random initial guesses to addresses known issues with local minima (Ewing et al., 2015). The optimization solution with the lowest objective value was defined as the optimal set of slack lengths. These were the slack lengths used in the "calibrated" model. See Appendix A.3 for further description of the constraints and initial guesses.

Three experimental tests were excluded from the optimization to evaluate the performance of the optimization scheme. These tests were the distraction, kinetic plane, and internal-external rotation tests at 0°, 30°, 60°, and 90° flexion. The RMS error between the calibrated model and experimentally measured kinetics was calculated to evaluate the performance of the optimization. Additionally, the force magnitude carried by each ligament during the testing was calculated using the calibrated model.

2.5. Kinetic sensitivity

An uncertainty analysis was performed to relate the kinetic errors found in this current study to kinematic errors reported in previous studies(Blankevoort and Huiskes, 1996; Baldwin et al., 2009; Ewing et al., 2015; Harris et al., 2016). The uncertainty analysis simulated kinematic errors in the test that were used in the optimization. Kinematic error was simulated by adding perturbations to the experimentally measured kinematics, and these perturbed kinematics were applied to the calibrated model. A total of 1000 perturbations were randomly sampled from a normal distribution with a mean of zero, and standard deviations of 0.47 mm for translations and 0.40° for rotations (flexion was not perturbed), which are one third the minimum RMS errors reported by Harris et al. (2016).

2.6. Statistical Analysis

The root mean square (RMS) error was calculated between the calibrated model (M) and experimental (E) results in all six kinetic DOF. Individual RMS errors were calculated for the optimization results, tests not included in the optimization as well as individual flexion angles and specific tests (Eq. (A.1)). For the kinetic sensitivity analysis, (1) the average range of calculated kinetics, and (2) the average RMS error due to kinematic perturbations was calculated (Eq. (A.2)). See Appendix A.4 for further description.

3. Results

For the tests that were included in the optimization (anterior-posterior drawer and varus-valgus), the RMS error of the joint kinetics across all flexion angles was 5.3 N, 22.0 N and 17.1 N for the medial, anterior, and superior loads respectively, and 0.71 Nm, 0.61 Nm, and 0.34 Nm for the extension, valgus, and internal rotation moments respectively (Figs. B.3 and B.4). Of the 1000 optimizations, 51 had a solution where every slack length was within 0.1 mm of the corresponding slack lengths that yielded the overall lowest objective value.

The individual tests that were in the optimization show variability between flexion angles. The distraction force error was less than 28 N, with the highest values for the anterior-posterior drawer and valgus tests occurring at 30° flexion (Table B.3). The peak anterior tibial reaction load during the posterior drawer test was consistently underestimated across flexion angles (Fig. B.3). The posterior tibial reaction force during the anterior drawer test was overestimated at 90° flexion, and underestimated at 0° flexion (Fig. B.3).

Tests not included in the optimization (distraction, kinetic plane, and internal-external rotation) showed similar RMS error values to the test included in the optimization in all DOF except for internal rotation moments. For the tests excluded from the optimization, the RMS error of the joint kinetics across all flexion angles was 5.4 N, 16.2 N and 32.0 N for the medial, anterior, and superior loads respectively, and 0.82 Nm, 0.48 Nm, and 2.22 Nm for the extension, valgus, and internal rotation moments respectively. The distraction test, which has relevance during TKA procedures, showed similar RMS error values as the tests that were included in the optimization. The RMS error of the joint kinetics across all flexion angles for the distraction test was 2.6 N, 13.0 N and 14.0 N for the medial, anterior, and superior loads respectively, and 0.31 Nm, 0.40 Nm, and 0.36 Nm for the extension, valgus, and internal rotation moments respectively (Fig. B.5).

The tests that were not included in the optimization also showed variability across flexion angles. For the kinetic plane and distraction tests, the highest errors in anterior force occurred at 90° flexion, and the highest distraction force errors occurred at 0° flexion (Table B.4). The internal-external rotation test demonstrated the largest difference in distraction force (Table B.4). For the internal-external rotation tests, the model performed better when the applied torque was lower. Across flexion angles, the RMS distraction force error was 15.0 N at $\pm$ 1 Nm internal rotation torque, and 46.6 N at $\pm$ 5 Nm internal rotation torque.

The ligaments showed varied recruitment between tests. The amACL carried the most load at the peak of the anterior drawer tests, however the plACL carried little force throughout all of the tests (Fig. B.6). At the peak of the posterior drawer test, the LCL carried the most load at 0° flexion, and the alPCL carried the most load at 30° and 60° and 90° flexion (Fig. B.6). The LCL carried the highest loads during the varus distraction tests, and the psMCL carried more load than the dsMCL during the valgus tests. During the distraction tests, the amACL and alPCL showed increasing recruitment with increasing flexion (Figs. B.6 and B.7). The alPCL showed zero recruitment during the distraction test at 0° flexion, however it had the highest superior force at 90° flexion.

Uncertainty analysis revealed large ranges of possible kinetics, though a relatively consistent average response due to perturbed kinematics. Simulated kinematic errors for the anterior-posterior drawer and varus-valgus distraction tests across all flexion angles resulted in an average range of 24.3 N, 121.2 N, and 186.2 N in the medial, anterior, and superior directions respectively, and 2.86 Nm, 2.86 Nm, and 0.91 Nm in extension, varus, and internal rotation moment respectively (Fig. B.8). The average RMS error (Eq. (A.3)) between the calibrated and perturbed model across all 1000 perturbations was 2.7 (SD 1.5) N, 15.6 (SD 10.1) N and 25.9 (SD 19.6) N in the medial, anterior and superior directions respectively, and 0.38 (SD 0.19) Nm, 0.43 (SD 0.25) Nm, and 0.12 (SD 0.07) Nm in extension, varus, and internal rotation moment respectively.

4. Discussion

Utilizing joint distraction loading offers the opportunity to evaluate ligament recruitment without the confounding effects of articular contact. This study used joint distraction loads in an optimization scheme to calibrate a forward kinematics knee model, where the objective was to minimize the residual between model and experimental kinetics. These methods are similar to other studies that have estimated ligament properties (Blankevoort and Huiskes, 1996; Baldwin et al., 2012; Ewing et al., 2015; Harris et al., 2016), however the current study minimized the residual between model and experimental kinetics for six DOF instead of for two or three kinematic DOF. Including all available DOF in the inverse modeling scheme is in accordance with Roth et al. (2015) who concluded that all DOF of interest should be considered when evaluating soft tissue restraint. The six DOF tibial reaction forces and moments had RMS errors less than 23 N and 0.75 Nm, respectively, for the tests included in the optimization.

Ligament recruitment patterns found in this current study during distraction loading were similar to those found in previous studies of patients diagnosed with osteoarthritis (Ma et al., 2017) as well as nonarthritic cadaveric specimens (Mihalko and Krackow, 1999; Nowakowski et al., 2012). Results showed that the PCL was not recruited at 0^{o} flexion, and was increasingly recruited with flexion (Fig. B.7). Previous studies have shown that resection of the PCL significantly increases the flexion gap, but does not have a significant affect on the extension gap (Ma et al., 2017; Mihalko and Krackow, 1999; Nowakowski et al., 2012). This indicates that the PCL contributes more to joint restraint in flexion than in extension, which is similar to behavior seen in the calibrated model during distraction tests (Fig. B.7). This similarity is especially encouraging considering the distraction test was not included in the optimization. Additionally the ACL and PCL carried the highest loads during the anterior and posterior drawer tests, respectively, and the LCL and sMCL carried the highest loads during the varus and valgus tests, respectively. This finding increased confidence in the optimization's performance because the calibrated model's recruitment patterns matched expected trends during these tests.

This study presented an optimization scheme with the potential to improve repeatability and robustness of predictive knee simulation. This scheme randomly selected initial guess values to avoid introducing bias to the optimization's solution. This increased confidence that the calibrated model's recruitment patterns were a result of the optimization's performance, and not influenced by a judicious selection of initial guess values. Additionally, Ewing et al. (2015) discussed how there can be multiple sets of parameters that are local solutions for the optimization. This current study attempted to address this issue by repeating the optimization 1000 times with different randomly selected initial guess values. It was found that 5.1% of the optimization iterations converged to a similar set of slack length values. Additionally, the optimization in this study used constraints that ensured that every control variable can have an impact on the objective function, which is applicable to an inverse modeling scheme that uses either a forward kinematics or a forward dynamics model. The estimated ligament slack lengths (and prestrains, AAppendix B.1) were generally within the range of values reported in similar inverse modeling studies (Ewing et al., 2015; Harris et al., 2016; Baldwin et al., 2012). However the prestrain values can vary between specimens within the same study by 24% (Ewing et al., 2015). Future work will address the repeatability and performance of the optimization framework for both data rich (i.e. research) and clinical applications, which may present limitations in both the depth and quality of the available data for model development.

The uncertainty analysis provided context for the amount of error between the calibrated model and experimental kinetics. A conservative amount of kinematic error from a published forward dynamics model (Harris et al., 2016) resulted in a large amount of error in this study's calibrated forward kinematics model (Fig. B.8). The range of kinetic errors were nearly three times that found between the calibrated model and experimental kinetics in all six DOF, with notably large ranges for anterior, superior, and varus reactions. This shows that the model's force-displacement relationship is sensitive to the kinematic inputs, and that small kinematic perturbations can yield kinetic errors that are considerably larger than the RMS errors found between the calibrated model and experimental kinetics. This indirect evaluation was necessary because, to our knowledge, no other study has used experimental distraction loading to calibrate a forward kinematics model.

There are several opportunities for improvement in this study. First, the methods should be validated by evaluating multiple specimens, which would provide the opportunity to assess the general applicability of the proposed approach. Given the wide variability in ligament properties, joint laxities found in previous studies, and the arthritic state of the specimen, the error results reported in this current study may not be applicable to all knee specimens. Additionally, the optimization did not include ligament stiffness. This reduced the number of control variables in the optimization, which is supported by previous work that has shown laxity based joint kinematics to be more sensitive to ligament slack length than stiffness (Baldwin et al., 2009). Another consideration is the way that ligament geometry and slack length were defined. It was assumed that a linear distribution of fibers can represent ligament mechanics, and it was also assumed that slack length varies linearly across the bundle. Additionally, the results from the internal-external rotation tests showed that improvements can be made in the model. The model captured joint kinetics at low torques, however it failed to reproduce experimental kinetics as the applied internal or external torque increased. This could be due to the assumed ligament representation, but it may also be influenced by the model's approximation of wrapping. The collateral ligaments likely have some wrapping around the tibia, and the ligaments would exert a reaction force on the tibial wrapping surface, as well as the insertion. The dsMCL was the only ligament in the model that wrapped around the tibia, and this assumption neglects the tibial reaction loads that the ALL and deep MCL and proximal sMCL could contribute to the total tibial loads. Furthermore, the use of a cylinder to approximate wrapping around the tibia would neglect the effects of osteophytes on ligament length estimations (Pottenger et al., 1990). Future work will certainly evaluate whether addressing these assumptions results in improved model performance.

Future work should also evaluate the effects of potential model to experiment registration errors and the representation of ligaments (Naghibi Beidokhti et al., 2017). Errors in registration could lead to a misalignment of the femoral and tibial coordinate systems, and this could introduce systematic error to the model's kinematics. The presented sensitivity analysis resulted in a wide range of possible kinematic errors for the calibrated model, though evaluation of registration based kinematic uncertainty on the optimization process itself should also be performed. Additionally, ligament geometry and material behavior was simplified to a bundle of nonlinear springs. This simplification will likely have an impact on the kinetics calculated from the forward kinematics model. Future work should evaluate the effect that different representations, from continuum to spring based, of ligament geometry and material behavior have on model performance.

This study demonstrated that experimental distraction loading can be used in existing inverse modeling schemes to estimate ligament slack lengths with a forward kinematics model. Previous inverse modeling approaches have assumed that external joint forces balance with internal ligament and articular contact forces (Blankevoort and Huiskes, 1996). The current study simplified that assumption by adjusting the external loads to remove articular contact, and assumed that external joint forces balance with internal ligament forces. This adjustment to experimental protocol likely focused the joint's force-displacement behavior on the ligaments, which was desirable when calibrating ligament properties. This approach could possibly lead to more reliable estimation of ligament properties for controlled research studies, and it may also be adaptable to a clinical setting. Previous clinical studies have coupled joint distraction devices and surgical navigation to provide intraoperative measures of distraction force and the resulting kinematics (Matsumoto et al., 2006; Heesterbeek et al., 2010; Heesterbeek et al., 2017; Swank et al., 2007). The current study's inverse modeling scheme demonstrated promising results for recreation of knee-specific kinematic-kinetic response.

Declaration of Competing Interest

Authors acknowledge that there is no conflict of interests.

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Appendix A. Supplementary material

Supplementary data associated with this article can be found, in the online version, at https://doi.org/10.1016/j.jbiomech.2019.04.040.

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