ACL Reconstruction Decision Support Through Personalized Simulation of the Lachman Test and Custom Activities

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Summary

Objectives: The objective of the proposed approach is to develop a clinical decision support system (DSS) that will help clinicians optimally plan the ACL reconstruction procedure in a patient specific manner.

Methods: A full body model is developed in this study with 23 degrees of freedom and 93 muscles. The knee ligaments are modeled as non-linear spring-damper systems and a tibiofemoral contact model was utilized. The parameters of the ligaments were calibrated based on an optimization criterion. Forward dynamics were utilized during simulation for predicting the model's response to a given set of external forces, posture configuration and physiological parameters.

Results: The proposed model is quantified using MRI scans and measurements of the well-known Lachman test, on several patients with a torn ACL. The clinical potential of the proposed framework is demonstrated in the context of flexion-extension, gait and jump actions. The clinician is able to modify and fine tune several parameters such as the number of bundles, insertion position on the tibia or femur and the resting length that correspond to the choices of the surgical procedure and study their effect on the biomechanical behavior of the knee.

Conclusion: Computational knee models can be used to predict the effect of surgical decisions and to give insight on how different parameters can affect the stability of the knee. Special focus has to be given in proper calibration and experimental validation.

Keywords: Virtual Physiological Human, ACL reconstruction, Simulation, Digital Patient, Dynamics

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1. Introduction

The human knee joint is distinguished by its complex, three dimensional geometry and multibody articulations that generate complex mechanical responses under moderate loads (1). The knee joint compliance and stability required for optimal daily function are provided by various articulations, the menisci, ligaments and muscle forces. A complete understanding of knee joint biomechanics significantly improves the prevention and treatment of knee joint disorders and injuries. Total knee arthroplasty and prosthetic ligament replacement are two examples that directly benefit from such knowledge (2) (3).

Computational subject specific knee modeling is important in predicting the reactions of a human knee under various loading conditions. The modeling techniques can help to predict biomechanics performance, to prevent injury and to customize treatment methodology (4). Moreover, biomechanical simulation helps in the prediction of the internal loads of the knee complex during specific activities. The latter can clearly help as a decision support mechanism, as far as surgical reconstruction is concerned, by evaluating the macroscopic effect of surgical decision in a simulated manner.

Concerning anterior cruciate ligament (ACL) tear, even if surgical reconstruction practice can be nowadays considered advanced, it is mainly based on expert judgment and less on objective biomechanical evaluation of the patient's physiology and everyday activities, while patient-specific surgery customization is very rare (5). With realistic ligament models and computational modeling techniques, ACL injuries could be potentially analyzed and optimally reconstructed using simulated decision support.

1.1 Related Work

Currently, there are three alternative modeling approaches of interest: finite element analysis (FEA) (2), multi-rigid body analysis or their combination (6). Multibody systems can be used in two different ways: as forward or as inverse models. In forward dynamics the physical forces and moments (e.g. gravitation and external forces) are the given quantities. When combined with geometric data (e.g. length of body segment) and mechanical parameters (e.g. moments of inertia and center of mass) the resulting body movement can be estimated.

During the last decades, several studies related to the modeling of the knee complex have been presented in the literature, utilizing analytical models with different degrees of sophistication and accuracy. They have mainly attempted to model the tibiofemoral joint (7), (8), (9), while a few studies have aimed at modeling the patellofemoral joint (10), (11), (12). Most of them concentrate on modeling partly the knee joint and don't take into account the full body in their simulations. Moreover, less studies consider the impact of the muscle contribution on the laxity of the tibiofemoral joint (9).

Various methods have been used to represent the ligaments in

computational models including finite element techniques and elastic springs (2). Modeling the ligaments as elastic springs is the most computationally efficient method and several approaches have looked at how these elastic springs should be defined (7), (8). Some of them have shown that ligaments have a non-linear toe region which occurs because of the initial crimping of the fibers. This toe region ends when all of the fibers have become taut (13). At that point the ligament behaves as a linear spring with a stiffness parameter, k. The proposed framework uses a non-linear one dimensional spring-damper method to define the cruciate and collateral ligaments.

2 Objective-Contribution

Patient specific and pre-surgical objective evaluations are two of the main goals in the present study. The proposed scheme aims to provide a feasibility study on how simulated virtual physiological humans (VPH) could help towards the aforementioned grand challenge.

The main contribution can be summarized as follows: the development of a realistic knee model, as a whole of a full 3D multibody model, comprised of 10 ligaments along with a tibiofemoral contact[1]. Furthermore, the contribution of the muscles is considered during simulations, as they play an important role in the stability of the knee (1). Moreover, the model is customizable, meaning it can be further parameterized based on patient specific in vivo measurements. Last but not least, the model can be used in simulation of different tasks, such as walking, running and jumping making it a valuable tool for assessment of pre-surgical decision.

In the scope of this study, different parameters are presented for patient specific parameterization, while the model can simulate different task. Namely, 1) the generic model can be scaled accordingly to the subject, changing its mass and size parameters of the body segments, 2) the kinematics of the knee joint can be altered or restricted, 3) the muscle properties can be adjusted (e.g. muscle length, maximum force, points of action and insertion, etc.), 4) ligaments parameters such as rest or reference length, stiffness, insertion position, which can be directly measured in vivo from MRI scans or used in the DSS framework.

3 Methods

3.1 Ligaments

The knee ligaments, which attach the femur to the tibia or fibula, are very important to stabilize the knee joint and prevent knee injuries. There are four main ligaments in the knee joint: 1) the anterior cruciate ligament (ACL), which is a primary restraint to anterior tibial translation and secondary restraint to tibia rotation, 2) the posterior cruciate ligament (PCL), which mainly restrains posterior translation of the tibia, 3) the medial collateral ligament (MCL), which counteracts valgus instability, and 4) the lateral collateral ligament (LCL), which primarily restrains varus stress of the knee joint and resists tibial external rotation.

According to the literature, the ligament has non-linear and elastic properties and the tension is mainly a function of its length. Based on (7), for low strains the function exhibits non-linearity. Its behavior becomes linear for strains higher than a certain level. The force-strain curve is described by:

$$f_l = \begin{cases} 0, & e \leq 0 \\ \frac{1}{4}k\,^{e^2}/_{e_l}, & 0 \leq e \leq 2e_l \\ k(e - e_l), & e > 2e_l \end{cases}$$

where, k is the ligament stiffness and $2e_l$ is the limit at which the ligament moves from the non-linear region to linear region. The strain e is described by:

$$e = \frac{(L - L_0)}{L_0}$$

where, L is the length of the ligament and L_0 is the zero-load length which is defined as the length of the ligament when it first becomes taut. In addition, the resting length is defined by:

$$L_0 = \frac{L_r}{e_r + 1}$$

where, L_r is the reference length of the ligament, usually measured at a reference posture (e.g. extension) and e_r is the reference strain at the reference posture.

Besides the non-linear ligament force-strain relation, we integrated an extra damping force, which is proportional to the ligament's shortening-lengthening rate. Thus we can constrain the oscillation of the knee joint by choosing appropriate damping coefficients for every ligament bundle using the formula:

$$c_{critical} = \sqrt{4km}$$

for critical damping, where k is the stiffness and m is the total mass that connects to the spring. The damping forces compensate for the unknown knee joint's tissue parameters. The latter are very hard to estimate using typical clinical equipment. Moreover, damping is a common practice and necessary for achieving stability (11).

Initial values of the parameters for the ligaments (reference length, reference strain and stiffness) were

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[1] Model and simulation data are open source and can be downloaded at: https://simtk.org/home/multikneedss

chosen based on (12). The parameters that are under investigation are summarized in Table 1 along with their initial values.

Table 1 Initial parameters of the ligaments Lr reference length, k stiffness and er reference strain (a: anterior, p: posterior, i: inferior, D: deep, r: reference).

Name	L_r(cm)	k(kN)	e_r
aACL	3.23	1.5	0.02
pACL	2.47	1.6	0.01
aPCL	2.58	2.6	-0.23
pPCL	2.52	1.9	0.02
aMCL	7.22	2.5	0.02
iMCL	7.31	3.0	0.04
pMCL	8.8	2.5	0.02
aDMCL	3.63	2.0	-0.08
pDMCL	3.72	4.5	0.03
LCL	5.59	2.0	0.02

3.2 Model

A full body musculoskeletal model is used in this study based on the one proposed in (14). It consists of 12 rigid segments, 23 degrees of freedom (DOFs) and 93 muscles partially illustrated in Figure 1. The model has been modified in order to enable a flexion-extension, varus-vargus, tibial and femoral rotations, anterior-posterior, medial-lateral and inferior-superior translations, that are of significant importance in the proposed framework. The passive forces of the ligaments and the extra damping forces were integrated as described in subsection 3.1 Ligaments. In addition to the main muscles, extra patella-tendon muscles were inserted since their contribution to the stability of the joint is not negligible.

The magnitude of the muscle force depends on its activation level as well as its force-generation properties defined by force-fiber length and force-fiber velocity

relationships (15). The widely known Hill-type muscle model is defined in a generic fashion as two differential equations:

$$\dot{a} = f(u, a, t_{act}, t_{dact})$$
$$f_M = g(a, l_M, v_M, F_O; \theta)$$

where, $u \in [0, 1]$ is the excitation of the muscle, $a \in [0, 1]$ is the activation level, which is modeled as a first order differential equation and t_{act}, t_{dact} are the activation and deactivation time constants respectively. As for the muscle force f_M, it is a function of the activation level a, the muscle length l_M, muscle shortening velocity v_M, the maximum isometric force F_O and are other parameters θ depending on the muscle's type.
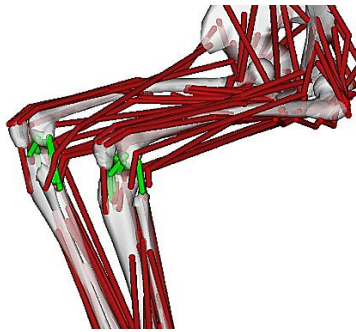

Figure 1 A portion of the developed model (the muscles are marked with red and the ligaments with green).

The tibiofemoral contact was modeled using elastic foundation model (16) (17), which uses a mesh to represent arbitrary surfaces in contact and calculates deformations and forces using a simplified bed of springs elastic model.

3.3 Simulation

The general dynamic equation of motion of the full multibody system is described by:

$$M(q)\ddot{q} = R(q)F_M + V(q, \dot{q}) + G(q) + F(q, \dot{q})$$

where, $q, \dot{q}, \ddot{q} \in R^N$ correspond to position, speed and acceleration respectively with N the degrees of freedom of the system, $M(q) \in R^{NxN}$ is the system mass matrix,

$V(q, \dot{q}) \in R^N$ represents the centrifugal and Coriolis forces vector, $G(q) \in R^N$ is the gravity acting on the body segments, $F(q, \dot{q}) \in R^N$ is a vector of the external forces acting on the system, $R(q) \in R^{NxM}$ is the moment arm matrix (18) and $F_M \in R^M$ is a vector of muscle forces with M the number of muscles.

In the above equation, if the motion is known $(q, \dot{q}, \ddot{q})$ along with the external forces $(F(q, \dot{q}))$, it can be solved for the muscle force (or joint torques and forces) in an inverse manner. If the external forces and torques are known, the equation can be solved in a forward manner for acceleration.

$$\ddot{q} = M^{-1}(q)\left(R(q)F_M + V(q, \dot{q}) + G(q) + F(q, \dot{q})\right)$$

In that case the above equation can be integrated twice in order to derive the produced motion trajectories. The equations above are solved numerically using 4th order Runge Kutta Merson integrator. It should be mentioned that in this study we use OpenSim framework (15), as a basis for our simulations.

3.4 Lachman Test

The Lachman test is a clinical test used to diagnose injury of the ACL. It is recognized as reliable, sensitive, and usually superior to the anterior drawer test (20). The patient should be relaxed for this test, especially the tested extremity. The examiner places the tested leg into about 20 degrees of flexion, by placing the examiner's knee under the patient's thigh. One hand is used to stabilize the distal femur near the joint line on the anterior side, while palpating the joint line. Then the thumb of the other hand is placed on the anterior side of the tibia and the fingers grasp the posterior side of the tibia near the joint line. Finally, quick posterior-to-anterior directed forces are applied through the tibia (21). There should be a firm end-feel. A positive test is

excessive movement or the lack of a firm end-feel. An alternate method involves holding the femur and tibia without the examiner's knee under the patient's thigh. During the procedure it is important that the correct joint angle is used for this test, since a position closer to full extension, naturally has less anterior translation of the tibia and can result in a false endpoint (21).
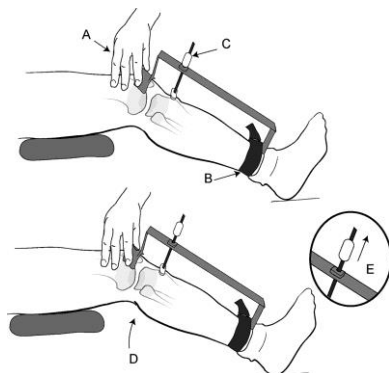

Figure 2 Rolimeter configuration during Lachman Test.

The Lachman test is imperative for the prognosis, follow-up, and scientific comparison of anterior cruciate ligament surgery results. The most commonly used tools, that can quantify the Lachman test results, by measuring the respective displacements are the KT-1000, KT-2000 (MEDmetric, San Diego, Calif., USA), and the Stryker knee laxity testers (22). An alternative and less expensive device, the Rolimeter knee tester (Aircast Europa, Neubeuern, Germany) has also shown exact and simple quantification of anterior knee joint instability (22). The Rolimeter knee tester is a reliable device for quantifying knee joint laxity, and is sensitive enough to identify anterior cruciate ligament deficiency (23). Consequently, the Rolimeter provides a cost effective and simple operating device for quantifying anterior knee joint instability.

Recently, the diagnostic efficiency of all these devices has been confirmed in several investigations. Statistical evaluation showed no significant difference in the

exactitude of measurement between the Rolimeter and the KT-1000 arthrometer which currently is the gold standard for such measurements (24) (25).

As for the model's posture during simulation, care was taken to be configured according to the Lachman test description. An external force is applied to the posterior aspect of the proximal tibia. During simulation we measure the anterior displacement, the muscles' contributions, the passive forces of the ligaments and their strain.

Results

The presented framework aims to provide decision support with respect to several rehabilitation parameters, like surgical reconstruction or not, artificial ligament surgical positioning, during performance of different activities. In order to perform the experiments, the model has to be quantified especially when it comes to the parameterization of ligaments. In the following subsections, we present our results with respect to the optimization or the parameter estimation scheme and clinical potential of the proposed framework.

4.1 Clinical Data

Experimental data were collected in order to be used during model calibration. The data originate from 30 young patients, age between 20-28 years old, average height and weight 1.78m and 78.3kg respectively with an excellent physical condition. All patients were amateur football players, who were suffering from ACL tear, where optimal surgery and rehabilitation was necessary. Different morphometric measurements were carried, along with the MRI recordings and the pre-surgical Lachman test in order to scale and calibrate the model according to the patient.

Measurements of the tibiofemoral gap along with the length of the ligaments Table 2 have been obtained from the MRI scans.

Similarly, the stiffness coefficient of the ligaments was estimated based on the area of the ligament using the same recordings and was in agreement with the literature (7) (26). Measurements of the anterior displacement with respect to various magnitudes of the applied force were performed on patients with deficient knee, in the context of the pre-surgical Lachman test.

4.2 Optimization Scheme and Parameter Estimation

The objective of the parameter estimation module is to calibrate the model's ligament parameters in order to produce a similar force-displacement curve as the one produced from the pre-surgical Lachman test, neglecting the contribution of ACL bundles. We employed an optimization scheme based on the error function, defined as the distance between the experimental and the model's displacement after simulation. The error function can be mathematically expressed by:

$$error = \sum w_i (d_m(f_i) - d_e(f_i))^2$$

where, $d_m(f_i)$ is the model's displacement function with respect to the input force f_i, $d_e(f_i)$ is the experimental displacement and w_i are non-negative weights. Larger weights are assigned to larger displacements, since their contribution to the aggregate error is more significant.

In Table 2 the average, min and max values for the four ligaments collected from different patients based on their MRI recordings are in agreement with (27). Consequently, these measurements were used as bound values for the search space of the optimization procedure. In contrast to Table 1, where we defined the initial values of the ligament model, which were used as seed values. The optimization procedure was performed several times,

restarting the process with different initial conditions to avoid for sub-optimal solutions and the best overall solution was stored with the corresponding parameter values.

Table 2 Average, min and max values for the five ligaments determined from MRI measurements from 30 young patients.

Ligament	Avg. Length (cm)	Min (cm)	Max (cm)
ACL	3.8	2.3	4.1
PCL	3.5	3.2	3.8
MCL	9.5	6	12.7
LCL	5.5	5	6

In Figure 3 we show the experimental anterior-posterior displacement with respect to the applied force from Lachman test and the model's response after optimizing the ligament parameters and neglecting the contribution of aACL and pACL bundles. We can see that after optimizing for the parameters of the ligaments we were able to calibrate the model in order to achieve a similar response for the given external forces. For the specific patient experimental anterior displacement for 100N was measured to be 8mm and our model response after optimization is seen to be 7.9mm. The overshoot transient at the beginning of the model's response time series is due to the assumption made by modeling the ligament as a spring-damper component and does not convey clinical information.

4.3 Clinical DSS

Table 3 presents the parameters that can be fine-tuned by the user in order to investigate their effect on the dynamics of the knee. Indicatively, the parameters under investigation in the following analysis are the number of bundles used for reconstruction and the insertion coordinates of the aACL on the tibia. Ligament strain is estimated for various potential values of the aforementioned parameters. Most of the parameters presented in Table 3 can be pre-surgically measured and used as an input to the model.

Table 3 Indicates potential variables that can be currently modified so as to investigate in a simulated manner their effect on the biomechanical behavior of the knee.

Parameter type	Variable
Ligament Insertion Position	Tibia, Femur
Ligament Parameters	Stiffness, Reference Parameters
Model Customization	Weight, Height, Kinematics of the Knee Joint
Muscle Parameters	Activation, Contraction Dynamics
Actions	Simulated Activities

For example, based on the ligaments' geometry from the MRI recordings the stiffness can be well approximated, along with the reference length (11), while the reference strain is difficult to measure and thus derived from the literature (8) (7) (28). Moreover, the model can be further personalized based on MRI scans (29), accounting for the skeletal segment geometry, the mass parameters of the body segments (mass, inertia and center of mass), the knee joint axis of rotation and the muscular parameters (muscle length and muscle maximum force based on the muscle cross section etc.). Initially, we provide two different knee models where the ACL is reconstructed using one or two bundles respectively. Simulation of the Lachman test to both models was employed in order to estimate the response of the models that are

illustrated in Figure 4. Using a single anterior ACL bundle is enough to constrain the anterior displacement of 2.7mm for 100N and posterior bundle doesn't contribute much the anterior displacement which was expected. In addition, a variation of the initial resting length is made on both bundles. Results show that the response of the model is very sensitive to changes of the ligament resting length, which is in agreement with (11). Additionally, the resting length parameter of the bundles can be adjusted, as illustrated in Figure 4, and an optimal solution can be chosen specifically per patient.

Moreover, as illustrated in Figure 5 the passive forces of LCL and aDMCL ligaments during the Lachman test simulation are the main restrictors in the absence of ACL bundles, which is in agreement with (30). In both diagrams we can see the strain with or without ACL bundles, where the peak passive force for LCL increases 11% and for aDMCL 67% during the absence of the ACL.

Extraction of comparative results in cases involving variables that cannot be explicitly observed, like for example the ligament strains in our case, is always a challenge. In order to produce comparable results a reference model for the simulations of knee flexion-extension is utilized. The reference model represents an intact knee with prescribed anterior-posterior and inferior-superior displacement during flexion which is based on the previous work of (31) and is widely used in the biomechanics community. The location where the aACL is attached to the tibia is defined on a two dimensional surface, forming a regular octagon with 10mm apothem. For every one of the eight possible positions a forward simulation was performed for a motion of knee flexion from 80 degrees to full extension at 0 degrees.

A potential decision criterion could be defined as the distance

between the reference strains (intact knee model) and the strains produced by the simulation for every ligament with respect to the different attachment positions of the aACL. Figure 6 presents the comparison between the reference strains and the strains produced for every one of the different insertion positions per ligament. In Table 4 a summary of the distance error per position and per ligament is provided and we marked the smallest errors with bold. The distance error is defined by aligning the reference strain and the simulated strain using dynamic time wrapping for knee flexion-extension angle and summing the absolute offset between them. Based on Table 4 we conclude that insertion position 8 produces the best overall matched strains with respect to the reference strains.

Inverse simulations are utilized for different activities based on their kinematics and recordings of the external forces acting on the model. Two different motions were used namely normal gait and jump. The kinematics and ground reaction forces used for the gait are retrieved from the dataset of OpenSim (19) and for the jump action from the work of (9). In Figure 7, left we present the results for the gait simulation and on the right for the jump action. The parameter under investigation for both simulations is the strain of the aACL ligament for the eight different insertion positions. It should be emphasized that our approach is not limited to the specific motion nor the strain parameter. Different motion trajectories and parameters can be easily investigated. Finally, several other surgical decision variables Table 3 can be tested so as to optimally plan the ACL reconstruction procedure in a more comprehensive manner.

5. Discussion-Conclusion

A computational model of both the kinematics and dynamics of the human knee is developed that is subsequently simulated for custom activities. The proposed scheme demonstrates how VPH simulations can be used in a pre-surgical step for optimal planning of several parameters related to the surgical procedure. The effect of the chosen parameters on the motor behavior of the knee can be estimated through the proposed simulation scheme, thus leading to a powerful clinical decision support system.

There are many alternative and notable models available in the literature. Some remarks and comparisons for the modeling decisions that we made in the current study follow. First of all we didn't choose to utilize FEA (6) (8) (7) (13), because the goal of the current study was to evaluate different surgical decisions such as the optimal insertion position of the ACL ligament, where the most important factor was the relative motion of the segments, the contribution of the muscles and less on the tibiofemoral contact model or the approximation of the stress-force of the ligaments. Secondly, in contrast with (7) (11) (13) (2) we chose to utilize a 3D full body multibody model making it suitable for the simulation of different activities, thus permitting to study the impact of different surgical decisions on a specific task. Finally, we believe that muscles play an important role in the stability of the knee, and its dynamics were sensitive to muscle forces, thus neglecting their contribution can potentially lead to misleading results.

Even if the potential use of such a framework is evident, there are several issues that still need to be considered. Even if, by definition, the proposed scheme aims to provide relative evaluations of different rehabilitation treatment strategies, an extensive clinical validation of the model and the simulation results are still necessary. Sensitivity and uncertainty analysis (32) (23) can also be performed jointly with remodeling to improve the overall assessment and robustness of the current model.

There is of course space for further improvement, which will enhance the accuracy and the predictability of the proposed computational knee model. First of all, customization of the model based on in vivo measurements is an important factor in patient specific healthcare, where geometric representation of the knee can be reconstructed based on MRI scan (29) (33). The ligament model can be further improved by taking into account wrapping surfaces or by modeling more realistically the insertion and origin sites spanning the entire area and not just a single point. A more accurate and comprehensive model of the knee contact adopted in our work can be investigated to account for the contribution of the menisci, which have a significant effect on distributing the tibiofemoral contact forces and thus their modeling and integration in the proposed framework is a major research direction for future work.

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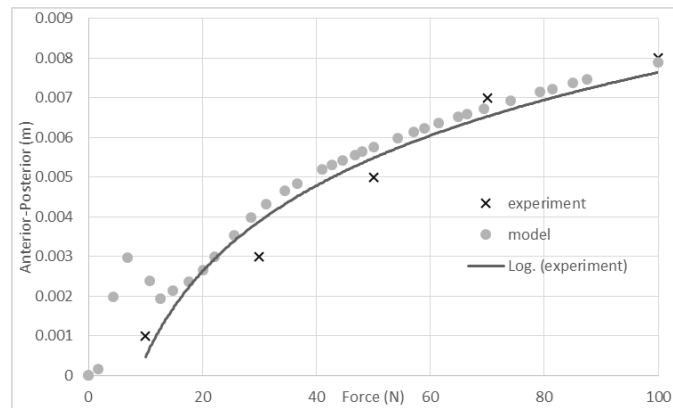
Figure 3 *Experimentally measured vs. model's response for the Lachman test. The diagram shows the anterior-posterior displacement with respect to the force applied to the posterior aspect of the proximal tibia.*
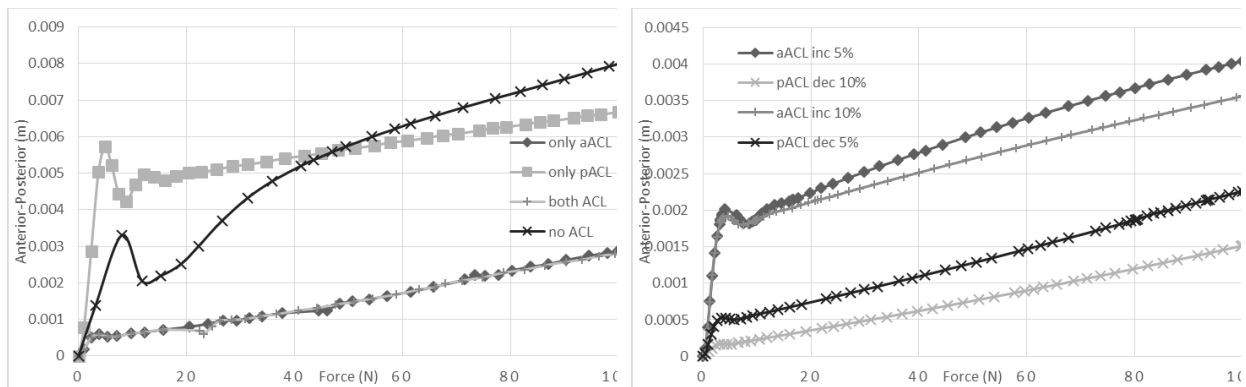

Figure 4 *The diagrams show the anterior-posterior displacement with respect to the force applied to the posterior aspect of the proximal tibia during the Lachman test simulation for different surgical choices. On the left diagram the response of the model is investigated based on the insertion of single or double bundles for the ACL ligament. On the right diagram an investigation is made by adjusting the restring length of the two bundles, by increasing or decreasing the initial resting length.*
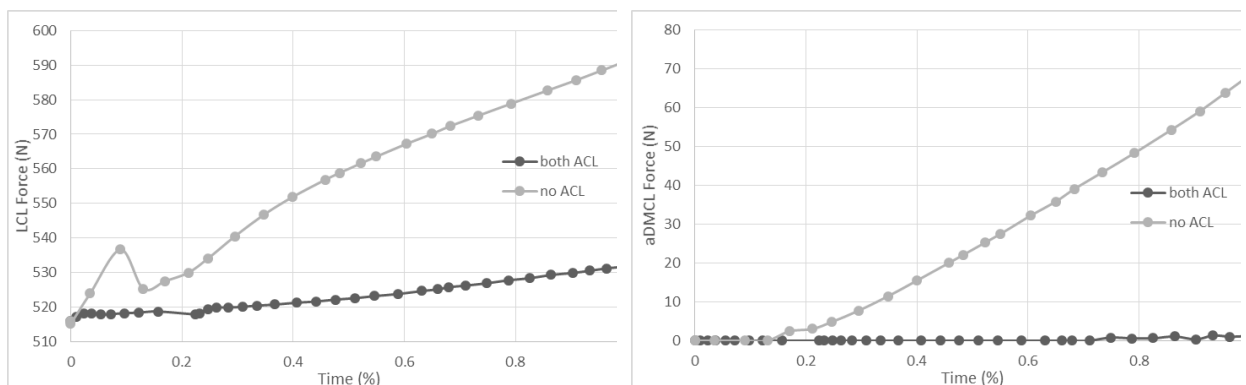

Figure 5 *Simulation of the strain force for LCL (left) and aDMCL (right) during the Lachman test which are the main restrictors in the absence of ACL bundles.*
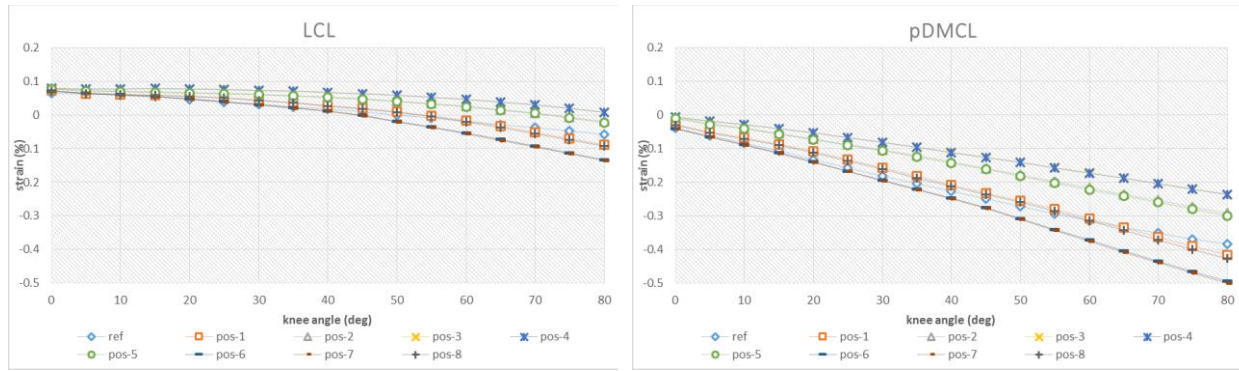
Figure 6 Comparison between simulated strains for all ligaments with respect to the reference strain of a healthy knee flexion motion. Investigation of 8 different attachment positions of the aACL on the tibia and their effect on the simulated strains curves.

Table 4 The error between simulated and reference strains for every ligament and for every insertion position of the aACL on the tibia (the minimum errors are marked with bold). The error is defined as the accumulated absolute value between reference and simulated curve after alignment of the time series with respect to knee flexion-extension angle. Minimum values show that for the concrete insertion position the ligament strain curve is the most closest to the reference strain curve of an intact knee during flexion.

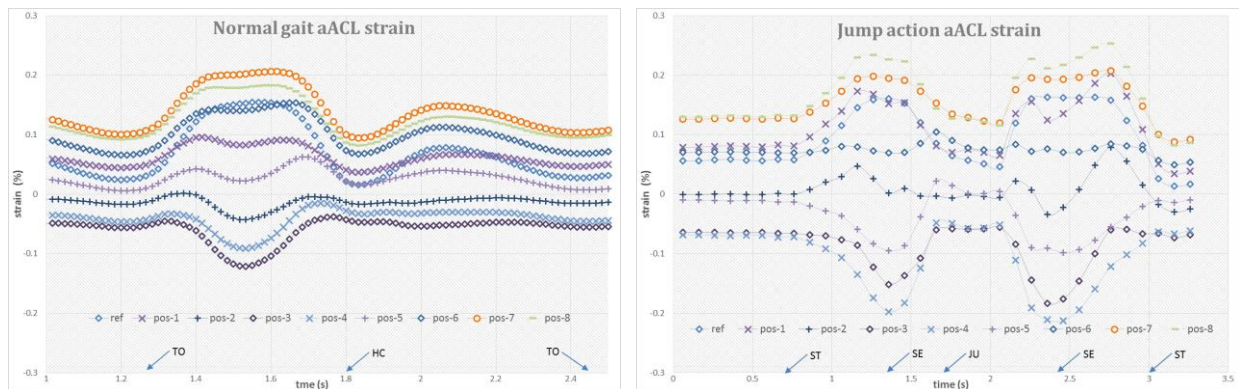| Position	aACL	pACL	aPCL	pPCL	aMCL	iMCL	pMCL	aDMCL	pDMCL	LCL	Total
1	0.638	0.795	0.831	0.766	**0.252**	**0.254**	**0.217**	0.438	0.509	**0.320**	5.019
2	0.390	3.988	1.968	0.697	0.942	0.841	0.773	2.114	2.279	0.955	14.948
3	1.855	5.409	2.151	1.379	1.344	1.207	1.094	2.983	3.222	1.422	22.068
4	1.752	5.390	2.159	1.358	1.336	1.199	1.088	2.968	3.205	1.412	21.867
5	**0.322**	3.786	1.848	**0.689**	0.908	0.813	0.745	2.018	2.175	0.926	14.230
6	0.724	1.653	0.878	1.066	0.680	0.649	0.559	1.232	1.334	0.847	9.620
7	0.768	1.776	0.912	1.070	0.718	0.684	0.590	1.311	1.414	0.889	10.131
8	0.769	**0.637**	**0.718**	0.764	0.256	0.260	0.218	**0.415**	**0.489**	0.342	**4.870**

Figure 7 Simulation of normal gait (left) and jump action (right) for different insertion positions of the aACL on the tibia. For the gait simulation the annotations are TO: toe off and HC: heel contact. For the jump action the annotations are ST: stand, SE: seated and JU: jump postures.