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# Artificial Knee Joint and Ski Load Simulator for the Evaluation of Knee Braces and Ski Bindings

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**Abstract**

**Introduction**: Epidemiological studies show that severe knee injuries are prevalent in alpine skiing. Their incidence is related to ski boot and ski binding concept – both designed to prevent tibia fractures. To reliably protect the knee, ski bindings need a release mechanism which follows different release principles. Therefore, attempts are made to develop mechatronic concepts implementing additional criteria and to release the foot when critical loads at the knee are reached. One possibility to systematically manipulate external loads and to investigate the resulting stresses in the joint are experiments using an artificial leg. This paper describes the development and the evaluation of such kind of model ("leg surrogate") including a complex artificial knee joint. The evaluation includes tests concerning the reliability, sensitivity and plausibility of the surrogate.

**Method**: Tibia and femur consist of an aluminum bone imitate and are reconstructed based on human computerized tomography data. Human endoprosthesis are used as articulating surfaces for the tibial plateau, the femoral condyles, the trochlea as well as the patella. Ten steel ropes connected to a force measuring cell are incorporated simulating the muscle. The muscle volume is imitated by a three layer coat of thermoplastic. The artificial knee ligaments are instrumented with custom made elongation and force sensors. Leg surrogate presetting's can be varied trough the knee angle, hip angle, varus or valgus position, tension of the muscle and pretension of the ligaments. A test rig enables t quasi static application of external loads to the leg surrogate in any combination about the x, y and z-axis.

**Results**: The leg surrogate delivers reproducible measurements with a maximum variation of 2.7%. It allows to simulate different conditions like muscle tension or hip angles and to record their influence on the knee ligaments. The plausibility checks performed indicate, that the leg surrogate represents the behavior of the human knee to a large extend.

**Conclusion**: The new leg surrogate allows to simulate not only alpine skiing injury but also other load situations. It therefore can be used to systematically investigate critical load situations to the knee and the prevention effect of safety devices such as mechatronic ski bindings or knee protection devices like preventive knee braces.



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*Keywords:* knee joint, load simulation, skiing, leg surrogate, knee injury, ACL

## 1. Introduction

The most injured body part in alpine skiing is the knee [1]. Above all, injuries of the anterior cruciate ligament and the medial collateral ligament are prevalent [2]. Because of the ski acting as a lever, the knee loads in skiing are higher than in most other sport disciplines. The high injury rate is also related to the ski bindings that are based on a mechanical design and developed to prevent tibia fractures. Their release mechanism and release value fail to protect the knee reliably [2, 3]. Therefore concepts of mechatronic ski bindings were developed. Their release algorithms however need understanding of the relationship between forces and moments at the ski binding, kinematics of the knee and hip joint in interaction with the leg's muscle activation and the

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resulting loading of relevant knee structures, such as the ACL. As ethical reasons make it impossible to apply critical loads to the human being, the application of biomechanical leg models is a logical approach to determine the responses of the knee structures according to sport specific loading situations. In this publication a biomechanical leg model for testing ski specific loading situations is presented.

## 2. Former and current knee surrogates

Since the early Nineties a variety of leg surrogates were presented. They can be classified in two categories. One measuring the occurring forces in the knee ligaments [4–6] and the others analyzing the range of motion of the tibia and femur [7–10]. The surrogate of France et al. [6] is the most sophisticated one. Besides the upper and lower leg, it integrates a rudimental upper body, pelvis, hip and ankle. The surrogates of Liu et al. [10], Mitternacht [11] and Hochmann [9] consist of the upper and lower leg only, neither simulating the knee joint nor any corresponding structure. Each of the surrogates [4–8] incorporate anterior (ACL) and posterior (PCL) cruciate ligament, the medial (MCL) and the lateral (LCL) collateral ligament, simulated by Nylon, Teflon or other polymers. France et al. [6] measure tension force in all four ligaments whereas the other surrogates are limited to recording MCL force. Only France's and Cawley's [8] model disposes the patella femoral joint, the others either neglect it or represent it rudimental. The test rig of some of the artificial legs only allow an extended knee position [4, 5, 10], while the others permit the application of different knee flexion angles. Four of the surrogates have mechanical or actuator controlled muscles made of steel ropes [4, 6–8]. Two surrogates are able to control the muscle volume with a pneumatic system [9, 11].

Most of the above mentioned surrogates where developed to investigate the protection potential of knee braces, which are used in American football to prevent players from MCL injuries. With these surrogates however it is not possible to apply multi-directional loads as they occur in alpine skiing thus they cannot simulate typical skiing injury mechanisms. Further the existing leg surrogates have implemented ligaments as a single bundle only, even though many studies have shown, that the individual bundles of a ligament have different functions [13, 14]. It is well known and many studies [i.e. 17-20] report that muscles have significant influence on the knee stiffness. However with a maximum of four implemented muscles the referred surrogates cannot simulate muscle influence properly. Even if the position of the upper body plays an important role in the injurie mechanisms, only France et al. [6] and Daley et al. [5] modeled the hip joint. All the referred surrogates are neither able to systematically analyze knee injuries in alpine skiing nor to proof the effect of safety equipment like ski bindings or knee braces.

With the aforementioned limitations of existing leg surrogates in mind, the aim of this study was to develop and evaluate a novel knee surrogate that overcomes the restrictions and which is able to measure loads in knee ligaments under skiing typical loading situations [15].

## 3. Technical realization

### *3.1. Design & construction*

The surrogate simulates an artificial right leg (Fig 1b). It incorporates the bones of thigh and lower leg, the hip, six knee ligaments, ten muscles and the muscle volume.

Tibia head, femur head and the head of the fibula are made of cast aluminum (Fig 1c). The geometries of these bones are based on CT-data of a 35 year old man with healthy knee. The data-set is scaled to a 50-percentile German man between 30 and 40 years. Aluminum hollow shafts model the femur shaft and the tibia. The articulating surfaces involve the tibio femoral joint and the patella femoral joint and are realized by prosthetic components (Genesis UNI and Journey, Smith & Nephew, London, UK). The patella is made of two components, with a medical implant acting as articulating surface fixed to an aluminum shell. This shell enables the attachment of the muscle ropes and the patella tendon made of polyethylene (KoSa® hochfest, telos, Marburg, Germany). An alloy "pelvis" is pivot-mounted on a shaft providing to set hip angle within a range between -20° to + 90°. A negative value corresponds to a backward position of the upper body and vice versa.

In total ten muscles are modeled. Eight of the thigh (m. biceps femoris, m. semimembranosus, m. semitendinosus, m. vastus lateralis, m. vastus intermedius, m. rectus femoris, m. vastus medialis longus, m. vastus medialis obliquus) as well as two of the calf (m. gastrocnemius lateralis, m. gastrocnemius medialis). The insertion- and attachment points as well as the lines of action are implemented in the anatomical correct position. The muscles are imitated by 2.5 mm thick steel ropes and are instrumented with force sensors (0-5 kN, K-100 ATP Messtechnik, Ettenheim, Germany). Muscle force for each muscle can be set independently by a tensioning unit.

The ligament apparatus of the knee for which synthetic ligaments made of polyester (LARS Ligaments, Arc sur Tille, France) are used, includes the ACL, PCL, MCL and LCL. Further the two functional bundles of the ACL (posterolateral (PL) and anteromedial (AM)) and the PCL (anterolateral (AL) and posteromedial (PM)), are realized. All ligaments are inserted according to anatomical landmarks and attached to a screw to set its preload according to literature data [16–19]. The AM ACL, PL ACL and the MCL are instrumented with custom made tension force sensors integrated in the bone (range: up to 1.5 kN for the ACL and PCL bundles and 500 N for the MCL). To measure the elongation of the ligaments, a special displacement sensor [20] is applied to all knee ligaments.

As the surrogate shall not only be used for studying ligament forces in specific load situations but also to test orthopedic products, e.g. knee braces, a muscle volume is incorporated providing a functional correct attachment of these braces. The muscle volume consists of three layers. The outer one made of thermoplastic, the middle one of polyurethane with a shore hardness type A of 13 and the inner one of stiff thermoplastic enabling the attachment to the bones. The coat is assembled in a way that it does not interfere with the ligaments and the muscles.

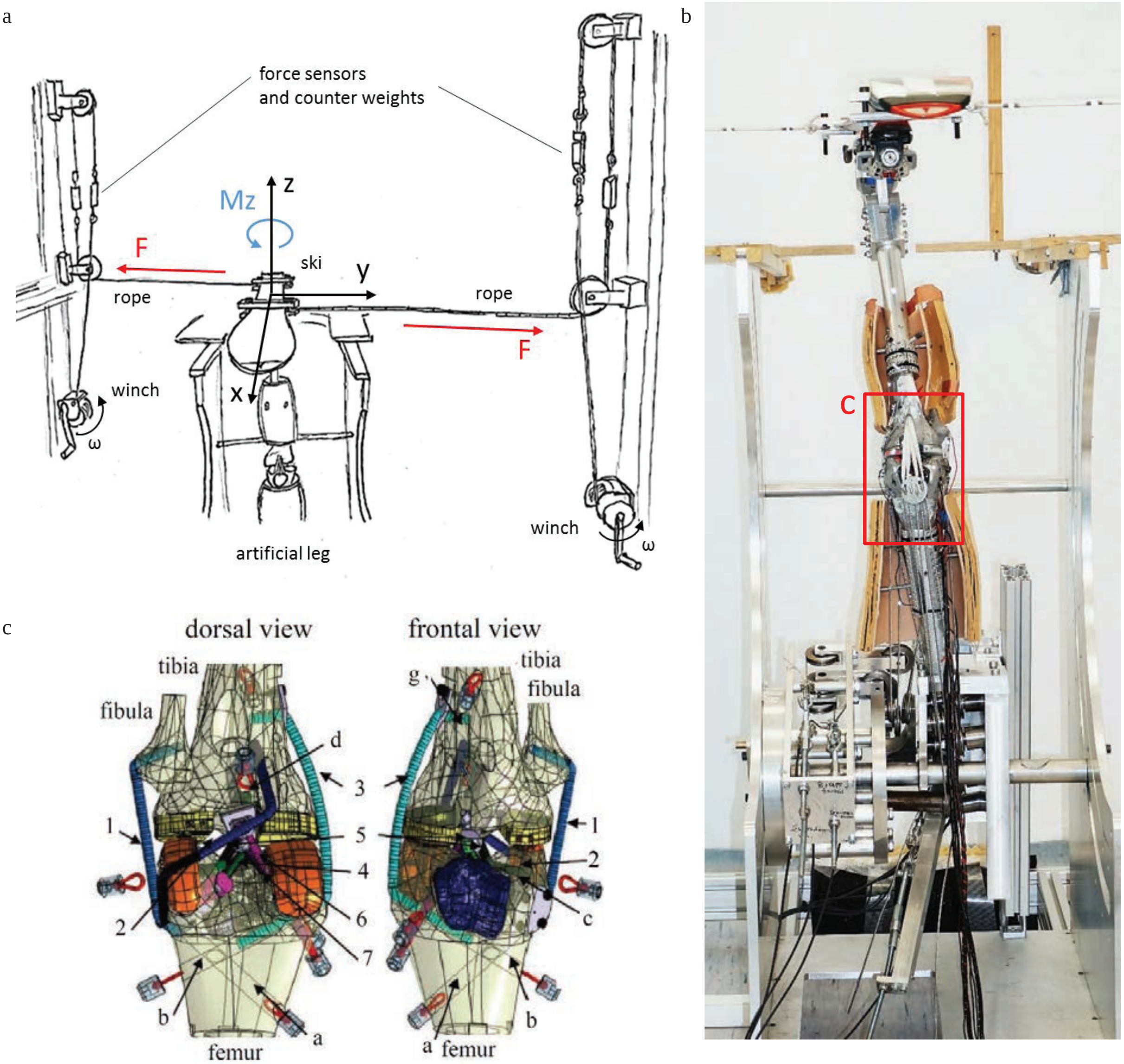


Fig 1. (a) Principle of the load application of the test rig. With the same principle torques around the x and y axis and forces in all spatial directions, as well as combined loads can be applied. (b) Knee surrogate (c) CAD data of the knee, frontal and dorsal view. Numbers indicate following structures: 1) LCL, 2) popliteus tendon, 3) MCL, 4) PL ACL, 5) AM ACL, 6) PM PCL, 7) AL PCL; letters indicate the fixation channels for the ligaments: a) LCL, b) MCL, c) PL ACL, d) AM ACL, e) PM PCL, f) AL PCL g) popliteus tendon

### *3.2. Mode of operation*

The simulation of typical skiing loads is achieved by applying external loads to the ski fixed to the surrogate's "foot". The method corresponds to the ASTM standard (F504 – 05(2012) which describes a 'standard test method for measuring the quasi-static release moments of alpine ski bindings'. Thus the surrogate is surrounded by an aluminum frame (Fig. 1a). Attached on the frame are winches which are used to apply loads to the ski via ropes. Different positioning of the four winches enables the application of any combination of quasi-static moments and forces around all three spatial axis of the ski. The ropes are instrumented with force sensors (range 1 kN, K-25 ATP Messtechnik, Ettenheim, Germany). The rotational and transversal displacement of all components of the leg-ski-system is recorded by a marker based video system. Plotting the applied moments against the rotational displacement allows the definition of the rotational stiffness of the knee [Nm/°].

The artificial leg is fixed in a carrier frame. The steel shaft representing the hip axis, is attached to an aluminum plate on each side. Moving the shaft in a crescent-shaped notch allows the setting of the knee flexion angle in the range of 5°-135°.

The setting of different muscle forces and thus applying axial loads on the knee makes it possible to preset the knee stiffness as well as to study the influence of the muscle tension to knee ligament tension force.

## 4. Evaluation

The test rig was tested for reproducibility, sensitivity and validity. All the test settings are listed in the following Table 1. For more detailed information concerning the test methods see [15]. The evaluation results given here refer to AM ACL and PL ACL only. For a comprehensive evaluation of the other ligaments see [15].

Table 1 Test protocol for the reproducibility, sensitivity and validation tests. a) Muscle forces were set in percentage of maximum muscle force as defined in [26]

| Test protocol | Knee angle [°] | Hip angle [°] | Muscle force [N] | External load | Rotation of Tibia around femur | Number of repetition for each setting | Resulting number of settings | Parameter | Tested force sensors |
|---|---|---|---|---|---|---|---|---|---|
| Reproducibility | 0 | 0 | 15%a) | quasi-static | external | 10 | 1 | mean of maximal force<br>Standard deviation (Std)<br>Variation coefficient (Var) | PL ACL<br>AM ACL |
| Test protocol | Knee angle [°] | Hip angle [°] | Muscle force [N] | External load | Rotation of Tibia around femur | Number of repetition for each setting | Resulting number of settings | Parameter | Tested force sensors |
| Sensitivity | | | | | | | | | |
| hip position | 15 | -20<br>+20 | neutral: 18% a)<br>Qmax: 22% a)<br>Ham-max: 22% a) | quasi-static | external<br>internal | 5 | 12 | mean of maximal force | PL ACL<br>AM ACL |
| muscle force | 15 | -20<br>+20 | neutral: 18% a)<br>Qmax: 22% a)<br>Ham-max: 22% | quasi-static | external<br>internal | 5 | 12 | mean of maximal force | PL ACL<br>AM ACL |
| valgus position | 15 | -20<br>+20 | - | quasi-static | external<br>internal | 5 | 4 | mean of maximal force | PL ACL<br>AM ACL |
| combined external load | 15 | -20<br>+20 | neutral: 18% a)<br>Qmax: 22% a)<br>Ham-max: 22% a) | quasi-static | external<br>internal | 5 | 12 | mean of maximal force | PL ACL<br>AM ACL |
| Validity and Plausibility | | | | | | | | | |
| Knee stiffness | 0-90 | 20 | neutral: 15% a) | quasi-static | external<br>internal | 5 | 12 | Stiffness of the knee joint [Nm/°] | PL ACL<br>AM ACL |
| Plausibility | 0-90 | 20 | neutral: 15% a) | quasi-static | external<br>internal | 5 | 12 | mean of maximal force | PL ACL<br>AM ACL |

Table 2 Comparison of Variation Coefficient of the knee surrogate and results of cadaver studies.

| Sensor | Results | Std KS | Var KS | Var Cadaver Studies |
|---|---|---|---|---|
| PL ACL force | 174.04 N | 0.17 N | 0.1% | 52.38% [21] |
| AM ACL force | 319.31 N | 8.72 N | 2.73% | 18.75% [22], 42.86% [21], 47.27% [23] |

*Reproducibility*

The variation coefficient of the PL ACL is 0.1%, the one of the AM ACL 2.7% , indicating comparable reproducibility as the test rig presented by Cawley et al. [7] with a variation coefficient of 2% and considerably better than the test rig of Brown et al. [4] who report a variation coefficient of 19%. Also compared to cadaveric or experimental studies, the variation coefficient of the current knee surrogate is much lower than from other studies (Table 2).

*Sensitivity*

The analysis of the test results in Table 3 shows that the knee surrogate is sensitive enough to detect force changes in the ligaments due to different settings of the hip, due to varying muscle forces, different valgus-position of the knee or changes of the combined external loads.

The correlation of higher loads in the ACL in a backward position combined with an increased muscle tension of the quadriceps corresponds with the results of in vivo studies by Boden et al. [24], Blackburn et al. [25] and Koyanagi et al. [26]. The measured values of the AM ACL with activated quadriceps muscles are in the same range as given by Li et al. [27]. As in a number of studies [27–30], an antagonistic behavior of the quadriceps to the ACL can be determined with our new leg surrogate.

Co-contraction of the hamstrings and the quadriceps (neutral setting) results in the lowest strain in the AM ACL and the PL ACL which corresponds to other studies [17, 33, 37]. Furthermore a protective effect of the hamstrings to the ACL as described by [31–33] can also be confirmed through our results in the sensitivity study performed.

In our measurements, a valgus position combined with an external rotation leads up to 39% higher loads in the AM ACL in a back weighted compared with the neutral one. A number of studies see a higher valgus stress as a risk factor for ACL ruptures [34–40]. The PL ACL reacts less sensitive on valgus stress, which is confirmed by other studies [41, 42].

The experiment to the combined load situation simulates a “landing back-weighted injury mechanism” in skiing. Skiers often respond to this back-weighted position by maximally activating the quadriceps muscle to regain balance. Our experiments to simulate this case results in the overall highest forces in both bundles of the ACL. The fact that combined loadings on the knee produce higher loads in the ACL is also reported in the studies of Durselen et al. [29] and Markolf et al. [43].

Table 3: Average values ($\bar{x}$) and standard deviation (σ) of the maximum force values of the PL ACL and the AM ACL for different settings for the hip angle and different preset muscle forces and a valgus setting.

| PL ACL | hip angle +20° | hip angle -20° | combined loading (hip angle -20°) |
|---|---|---|---|
| **external rotation** | $\bar{x}_{max} \pm \sigma\ [N]$ | $\bar{x}_{max} \pm \sigma\ [N]$ | $\bar{x}_{max} \pm \sigma\ [N]$ |
| Neutral | 165.3 ± 0.6 | 158.6 ± 0.2 | 157.9 ± 0.2 |
| Quadriceps max | 182.7 ± 1.7 | 168.6 ± 0.9 | 177.5 ± 1.0 |
| Hamstrings max | 167.8 ± 0.4 | 176.4 ± 0.7 | 174.2 ± 1.5 |
| valgus | 176.8 ± 1.1 | 155.6 ± 0.0 | 156.4 ± 0.2 |
| **internal rotation** | | | |
| Neutral | 171.5 ± 1.1 | 161.0 ± 0.8 | 176.1 ± 1.9 |
| Quadriceps max | 185.0 ± 0.8 | 179.6 ± 0.4 | 189.2 ±1.9 |
| Hamstrings max | 178.2 ± 0.5 | 170.8 ± 0.6 | 182.3 ± 1.8 |
| valgus | 155.0 ± 0.1 | 155.1 ± 0.0 | 155.1 ± 0.1 |
| **AM ACL** | | | |
| **external rotation** | $\bar{x}_{max} \pm \sigma\ [N]$ | $\bar{x}_{max} \pm \sigma\ [N]$ | $\bar{x}_{max} \pm \sigma\ [N]$ |
| Neutral | 289.4 ± 1.2 | 288.8 ± 4.6 | 316.5 ± 20.4 |
| Quad-max | 405.5 ± 5.6 | 470.0 ± 16.8 | 473.8 ± 13.4 |
| Ham-max | 357.7 ± 3.7 | 370.2 ± 2.9 | 337.8 ± 23.7 |
| valgus | 394.0 ± 5.7 | 470.8 ± 6.6 | 454.6 ± 11.0 |
| **internal rotation** | | | |
| Neutral | 190.5 ± 1.2 | 185.5 ± 0.2 | 292.1 ± 16.4 |
| Quadriceps max | 239.6 ± 10.0 | 364.1 ± 6.0 | 457.3 ± 7.1 |
| Hamstrings max | 230.7 ± 2.7 | 197.2 ± 1.3 | 288.9 ± 14.8 |
| valgus | 184.8 ± 0.1 | 183.8 ± 0.4 | 391.1 ± 13.1 |

Table 4: Torques needed to rotate the tibia by 8° (T8) and 12° (T12) for knee flexion angles (KF) from 0° to 90° for external (e) and internal (i) rotation.

| KF [°] | | T 8° [Nm] | T 12° [Nm] |
|---|---|---|---|
| 0 | e | 10.9 | 17.3 |
| | i | 8.9 | 13.3 |
| 15 | e | 10.0 | 16.9 |
| | i | 4.1 | 6.8 |
| 30 | e | 14.9 | 22.3 |
| | i | 12.6 | 18.9 |
| 45 | e | 14.2 | 24.2 |
| | i | 10.6 | 14.0 |
| 60 | e | 14.5 | 21.1 |
| | i | 14.1 | 22.0 |
| 90 | e | 12.5 | 18.9 |
| | i | 11.9 | -- |

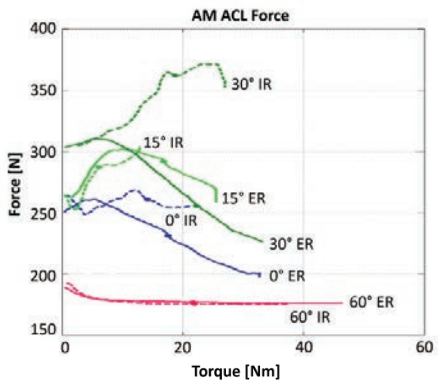


Fig 2. Exemplary force curves for the AM ACL with respect to the applied torque for the knee flexion angles 0°, 15°, 30° and 60° for internal (IR) and external rotation (ER).

*Proof of validity and plausibility*

**Knee stiffness:** An overview of the results is given in Table 4. The average stiffness of the artificial knee is for internal rotation 1.3 (±0.4) Nm/°, and for external rotation 1.6 (±0.2) Nm/°. Compared with results of studies with human subjects [44–48] the knee surrogate shows higher rotational stiffness. Three reasons may lead to that difference. As the knee surrogates ankle has only one degree of freedom (flexion and extension), the applied torque acts directly on the knee. Alam et al. [44] figured out that a measurement of the rotation angle at the foot instead of measuring it at the knee induces 103% higher values of rotation. Several studies indicate that an activated state of the muscles leads to an increase of the stiffness of 123% up to 250% depending on the knee flexion angle [49–51]. It can be assumed that the surrogate's artificial articulating surfaces have the greatest influence on the stiffness. The friction coefficient of these are estimated to be five times higher compared to the real human knee. The prostheses might also have a slightly different pivot point than in reality.

**Plausibility:** For the PL ACL the highest loads are measured for internal rotation at 0° and 30° (179.0 N resp. 160.0 N). This corresponds to a load increase of 10.7%. Similar results can be found in the cadaveric studies of Wu et al. [42] (12.3% increase) and Seon et al. [41] (18.5% increase).

For the AM ACL (Fig. 2) the increase adds up to 27.7% and is slightly higher than in the two mentioned studies with 20.6% [42] and 19.4 % [41]. The highest forces causes the internal rotation at 30° (371.8 N) and 15° (302.0 N). Seon et al. [41] and Wu et al. [42] however measure higher forces at 30° than at 15°.

## 5. Conclusion and future prospects

The results of the evaluation of the artificial leg show that the new developed leg surrogate allows high reproducible measurements, reacts sensitively to settings of different parameters like muscle stiffness and delivers plausible results concerning the knee stiffness.

Therefore the device can be used to analyze complex injury mechanism like those occurring in alpine skiing. This publication provides a proof of concept. The interpretation of the results with respect to the consequences for the skier and possible injuries, as well as investigations of more specific load situations on the knee is part of Nusser [15] and our future work.

Ongoing development steps are heading towards an electronically controlled muscle tension mechanism as well as the realization of an electric motor control for the application of the external forces, thus simplifying the handling of the novel measurement system.

## References


[1] Ruedl G, Linortner I, Schranz A, Fink C, Schindelwig K, Nachbauer W et al. Distribution of injury mechanisms and related factors in ACL-injured female carving skiers. Knee Surgery, Sports Traumatology, Arthroscopy 2009;17:1393–8.
[2] Senner V, Lehner S, Nusser M, Michel F. Skiausrüstung und Knieverletzungen beim alpinen Skifahren im Freizeitsport: Eine Expertise zum gegenwärtigen Stand der Technik und deren Entwicklungspotenzial. Bern: BfU; 2014.
[3] Senner V, Michel FI, Lehner S, Brügger O. Technical possibilities for optimising the ski-binding-boot functional unit to reduce knee injuries in recreational alpine skiing. Sports Engineering 2013;16: 211–28.
[4] Brown TD, Van Hoeck, J E, Brand RA. Laboratory evaluation of prophylactic knee brace performance under dynamic valgus loading using a surrogate leg model. Clinics in sports medicine 1990;9:751–62.
[5] Daley BJ. A Parametric Design Evaluation of Lateral Prophylactic Knee Braces. Journal of Biomechanical Engineering 1993;115:131.
[6] France EP, Paulos LE, Jayaraman G, Rosenberg TD. The biomechanics of lateral knee bracing: Part II: Impact response of the braced knee. The American Journal of Sports Medicine 1987;15:430–8.
[7] Cawley PW, France EP, Paulos LE. Comparison of rehabilitative knee braces: A biomechanical investigation. The American Journal of Sports Medicine 1989;17:141–6.
[8] France P & Cawley P. Mechanical Testing of Functional Knee Braces: An Evaluation of the BREG FUSION XT versus Selected Custom and Off-TheShelf Functional Knee Braces
[9] Hochmann D, Kraft M. Development of validated testing procedures for knee braces. Berlin.
[10] Liu SH, Lunsford T, Gude S, Vangsess C.T.Comparison of Functional Knee Braces for Control of Anterior Tibial Displacement. Clinical Orthopaedics and Related Research 1994;303.
[11] Mitternacht J. Abschlussbericht: Ermittlung der stabilisierenden Eigenschaften eines Prototyps Knieorthese Bort auf der Basis eines Kniesimulators.
[12] Steinbrück A, Milz S, Woiczinski M, Schröder C, Utzschneider S, Jansson V et al. Anatomie und Biomechanik des Patellofemoralgelenks: Physiologische Gegebenheiten und Veränderungen nach Implantation einer Knietotalendoprothese. Der Orthopäde 2011;40:848, 850-2, 854.
[13] Brucker PU, Lorenz S, Imhoff AB. Aperture Fixation in Arthroscopic Anterior Cruciate Ligament Double-Bundle Reconstruction. Arthroscopy: The Journal of Arthroscopic & Related Surgery 2006;22:1250.e1-1250.e6.
[14] Hosseini A, Qi W, Tsai T, Liu Y, Rubash H, Li G. In vivo length change patterns of the medial and lateral collateral ligaments along the flexion path of the knee. Knee surgery, sports traumatology, arthroscopy official journal of the ESSKA 2014.
[15] Nusser Michaela. Entwicklung, Evaluation und Anwendung eines Kniesimulators zur Ermittlung der auftretenden Kräfte im Knie unter skitypischen Belastungssituationen. Ph.D Dissertation. Technical University of Munich, Faculty Sport and Health Sciences; submitted 2015.
[16] LaPrade RF, Ly TV, Wentorf FA, Engebretsen L. The Posterolateral Attachments of the Knee. The American Journal of Sports Medicine 2003;31:854–60.
[17] LaPrade RF, Engebretsen AH, Ly TV, Johansen S, Wentorf FA, Engebretsen L. The Anatomy of the Medial Part of the Knee. The Journal of Bone & Joint Surgery 2007;89:2000–10.
[18] Lorenz S, Elser F, Brucker PU, Obst T, Imhoff AB. Radiological evaluation of the anterolateral and posteromedial bundle insertion sites of the posterior cruciate ligament. Knee surgery, sports traumatology, arthroscopy official journal of the ESSKA 2009;17:683–90.
[19] Zantop T, Wellmann M, Fu FH, Petersen W. Tunnel positioning of anteromedial and posterolateral bundles in anatomic anterior cruciate ligament reconstruction: anatomic and radiographic findings. The American Journal of Sports Medicine 2008;36:65–72.
[20] Nusser M, Fehle A, Senner V. Preliminary Studies for Validation of a Novel Sensor Fiber to Measure Forces in Artificial Knee Ligaments. Procedia Engineering 2012;34:236–41.
[21] Gabriel MT, Wong EK, Woo SL, Yagi M, Debski RE. Distribution of in situ forces in the anterior cruciate ligament in response to rotatory loads. Journal of Orthopaedic Research 2004;22:85–9.
[22] Wascher DC, Markolf KL, Shapiro MS, Finerman GA. Direct in vitro measurement of forces in the cruciate ligaments. Part I: The effect of multiplane loading in the intact knee. The Journal of Bone & Joint Surgery 1993;75:377–86.
[23] Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GA, Slauterbeck JL. Combined knee loading states that generate high anterior cruciate ligament forces. Journal of orthopaedic research official publication of the Orthopaedic Research Society 1995;13:930–5.
[24] Boden BP, Torg JS, Knowles SB, Hewett TE. Video analysis of anterior cruciate ligament injury: abnormalities in hip and ankle kinematics. The American Journal of Sports Medicine 2009;37:252–9.
[25] Blackburn JT, Padua DA. Sagittal-plane trunk position, landing forces, and quadriceps electromyographic activity. Journal of Athletic Training 2009;44:174–9.
[26] Koyanagi M, Shino K, Yoshimoto Y, Inoue S, Sato M, Nakata K. Effects of changes in skiing posture on the kinetics of the knee joint. Knee surgery, sports traumatology, arthroscopy official journal of the ESSKA 2006;14:88–93.
[27] Li G, Rudy T, Sakane M, Kanamori A, Ma C, Woo S. The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ foreces in the ACL. Journal of Biomechanics 1998;32:395–400.
[28] DeMorat G, Weinhold P, Blackbrun T, Chudik S, Garrett W. Aggressive Quadriceps Loading Can Induce Noncontact Anterior Cruciate Ligament Injury. The American Journal of Sports Medicine 2004;32:477–83.
[29] Durselen L, Claes L, Kiefer H. The Influence of Muscle Forces and External Loads on Cruciate Ligament Strain. The American Journal of Sports Medicine 1995;23:129–36.
[30] Withrow TJ, Huston LJ, Wojtys EM, Ashton-Miller JA. The relationship between quadriceps muscle force, knee flexion, and anterior cruciate ligament strain in an in vitro simulated jump landing. The American Journal of Sports Medicine 2006;34:269–74.
[31] Barone M, Senner V, Schaff P. ACL Injury Mechanism in Alpine Skiing: Analysis of an Accidental ACL Ruptur. In: Johnson RJ, editor. Skiing Trauma and Safety: Twelfth Volume. 100 Barr Harbor Drive, PO Box C700, West Conshohocken, PA 19428-2959: ASTM International; 1999, p. 63–81.
[32] Blackburn JT, Norcross MF, Cannon LN, Zinder SM. Hamstrings stiffness and landing biomechanics linked to anterior cruciate ligament loading. Journal of Athletic Training 2013;48:764–72.
[33] More RC, Karras BT, Neiman R, Fritschy D, Woo SL, Daniel DM. Hamstrings-an anterior cruciate ligament protagonist: An in vitro study. The American Journal of Sports Medicine 1993;21:231–7.
[34] Berns G, Hull M, Patterson H. Strain in the Anteromedial Bundle of the Anterior Cruciate Ligament Under Combination Loading. Journal of Orthopaedic Research 1992;10:167–76.
[35] Krosshaug T, Nakamae A, Boden BP, Engebretsen L, Smith G, Slauterbeck JR et al. Mechanisms of anterior cruciate ligament injury in basketball: video analysis of 39 cases. The American Journal of Sports Medicine 2007;35:359–67.

[36] Freudiger S, Friederich N. Critical Load Cases for Knee Ligaments at Skiing - An Engineering Approach. In: Johnson RJ, Zucco P, Shealy JE, editors. Skiing trauma and safety. West Conshohocken, Pa.: ASTM; 2000, p. 160–174.
[37] Maes R, Adrianne Y, Rémy P. Knee ligament injuries in downhill skiing: Epidemiology and etiopathogenic hypothesis [Incidence croissante des traumatismes ligamentaires du genou lors de la pratique du ski alpin: Épidémiologie et hypothèses étiopathogéniques]. Revue Medicale de Bruxelles 2002;23:87–91.
[38] Myer GD, Ford KR, Hewett TE. The effects of gender on quadriceps muscle activation strategies during a maneuver that mimics a high ACL injury risk position. Journal of electromyography and kinesiology official journal of the International Society of Electrophysiological Kinesiology 2005;15:181–9.
[39] Nguyen A, Shultz SJ, Schmitz RJ, Luecht RM, Perrin DH. A Preliminary Multifactorial Approach Describing the Relationships Among Lower Extremity Alignment, Hip Muscle Activation, and Lower Extremity Joint Excursion. Journal of Athletic Training 2011;46:246–56.
[40] Olsen O. Injury Mechanisms for Anterior Cruciate Ligament Injuries in Team Handball: A Systematic Video Analysis. American Journal of Sports Medicine 2004;32:1002–12.
[41] Seon JK, Gadikota HR, Wu J, Sutton K, Gill TJ, Li G. Comparison of single- and double-bundle anterior cruciate ligament reconstructions in restoration of knee kinematics and anterior cruciate ligament forces. The American Journal of Sports Medicine 2010;38:1359–67.
[42] Wu J, Seon JK, Gadikota HR, Hosseini A, Sutton KM, Gill TJ et al. In situ forces in the anteromedial and posterolateral bundles of the anterior cruciate ligament under simulated functional loading conditions. The American Journal of Sports Medicine 2010;38:558–63.
[43] Markolf KL, O'Neill G, Jackson SR, McAllister DR. Effects of applied quadriceps and hamstrings muscle loads on forces in the anterior and posterior cruciate ligaments. The American Journal of Sports Medicine 2004;32:1144–9.
[44] Alam M, Bull, Anthony M J, Thomas Rd, Amis AA. Measurement of rotational laxity of the knee: in vitro comparison of accuracy between the tibia, overlying skin, and foot. The American Journal of Sports Medicine 2011;39:2575–81.
[45] Lorbach O, Wilmes P, Theisen D, Brockmeyer M, Maas S, Kohn D et al. Reliability testing of a new device to measure tibial rotation. Knee Surgery, Sports Traumatology, Arthroscopy 2009;17:920–6.
[46] Muaidi QI, Nicholson LL, Refshauge KM. Do elite athletes exhibit enhanced proprioceptive acuity, range and strength of knee rotation compared with non-athletes? Scandinavian journal of medicine & science in sports 2009;19:103–12.
[47] Shultz SJ, Shimokochi Y, Nguyen A, Schmitz RJ, Beynnon BD, Perrin DH. Measurement of varus-valgus and internal-external rotational knee laxities in vivo-Part I: assessment of measurement reliability and bilateral asymmetry. Journal of orthopaedic research official publication of the Orthopaedic Research Society 2007;25:981–8.
[48] Tsai AG, Musahl V, Steckel H, Bell KM, Zantop T, Irrgang JJ et al. Rotational knee laxity: reliability of a simple measurement device in vivo. BMC musculoskeletal disorders 2008;9:35.
[49] Wojtys, E. M., Huston, L. J., Schock HJ, Boylan JP, Ashton-Miller JA. Gender Differences in Muscular Protection of the Knee in Torsion in Size-Matched Athletes. The Journal of Bone & Joint Surgery 2003;85:782–9.
[50] Hull, M.L., Johnson, C. Axial Rotation of the Lower Limb Under Torsional Loading: I. Static and Dynamic Measurements in Vivo. In: Johnson RJ, Mote CD, Binet M, editors. Skiing Trauma and Safety: Seventh International Symposium. 100 Barr Harbor Drive, PO Box C700, West Conshohocken, PA 19428-2959: ASTM International; 1989, p. 277–290.
[51] Louie JK, Mote CD. Contribution of the musculature to rotatory laxity and torsional stiffness at the knee. Journal of Biomechanics 1987;20:281–300.