To brace or not to brace?
Positive and negative effects of knee braces

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by
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“The surgeon thought there was no danger of ruining the knee further by playing, because there wasn’t that much left to ruin.”

Michael Real about Dick Butkus, middle linebacker in American football

“He’ll barely be able to walk by the time he’s forty.”

A doctor about Joe Namath, winner of the Superbowl III

“As is usual, widespread usage came first, with the research following later.”

Bruce Reider, Editor of the American Journal of Sports Medicine
Content

Nomenclature ............................................ V
1 Introduction ........................................... 1
   1.1 Knee structures .................................... 1
   1.2 Trauma patterns .................................... 5
   1.3 Knee braces ........................................ 8
   1.4 Aims of the work .................................. 10
2 Knee kinematics ....................................... 12
3 Brace kinematics ....................................... 16
4 Effect of a misaligned knee brace on ACL strain .... 20
5 Effect of knee braces vs. impacts or moments ......... 23
6 Conclusion ............................................. 27
7 Abstract ............................................... 29
8 Bibliography ........................................... 31
9 Originals with copyright and permission notices .... 40
Acknowledgements ....................................... 68
Curriculum vitae ........................................ 69
List of Publications ..................................... 70
## Nomenclature

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
<th>Unit</th>
</tr>
</thead>
<tbody>
<tr>
<td>ACL</td>
<td>anterior cruciate ligament</td>
<td></td>
</tr>
<tr>
<td>ε</td>
<td>strain</td>
<td>%</td>
</tr>
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<td>FHA</td>
<td>finite helical axis</td>
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<td>LCL</td>
<td>lateral collateral ligament</td>
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<td>MCL</td>
<td>medial collateral ligament</td>
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<tr>
<td>PCL</td>
<td>posterior cruciate ligament</td>
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<td>PKB</td>
<td>protective knee braces</td>
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1 Introduction

The history of protective knee braces (PKB) is closely linked to the history of American football. With the professionalization and growing investments in the sport and lucrative television contracts in the early 70s, the physical integrity of the athletes became an important economic factor [27]. Thus, the pressure for the athletes to perform and withstand injuries or pain increased [123]. In those days, knee injuries had the highest incidence and severity of all injuries in American football [23, 106]. Thus, knee braces that were originally designed to stabilize the knee after surgery were customized and modified to protect the athletes’ knees. In 1949, quarterback “Broadway” Joe Namath won the Superbowl III with his custom made knee brace, which helped him overcome his severe knee problems. Consequently, “football programs with deep pockets purchased [knee braces] for prophylactic use” [69, 123, 124]. Still today, athletes wear prophylactic braces to protect the important structures of the knee as well as patients who wear them during rehabilitation.

The aim of this thesis is to investigate positive and potential negative side effects of prophylactic knee bracing.

1.1 Knee structures

The knee is a complex joint with unique kinematics. It is stabilized by several ligaments and structures, which protect the knee under various loading conditions.

The anterior cruciate ligament (ACL) is anchored in the medial surface of the lateral epicondylus of the femur and the area intercondylaris anterior of the tibia (figure 1) and is divided into three bundles: intermediate, anteromedial and posterolateral bundle [4]. They have a length of around 31±3 mm and rotate against each other during motion from flexion to extension [111, 131]. In-vivo studies by Beynnon et al. found that the ACL strain increased by 6 % during a motion from 90° to 0° flexion angle [12]. Another study Beynnon et al. found that an anterior load of 150 N increased ACL strain by around 6 % in the non-weight bearing knee, whereas the weight bearing knee (subject standing) was not affected by the anterior load at all.
Fleming et al. found that an internal torque of 10 Nm leads to an increased ACL strain by approximately 5% and that an external torque of 10 Nm also increased the strain by approximately 2% [45]. This illustrates that the ACL inhibits an anterior shift of the tibia relative to the femur as well as tibial rotation [12, 76]. Besides providing mechanical stability, the ACL plays an important role in the proprioceptive feedback system of the knee, acting as a sensor for the joint position [75, 95].

The posterior cruciate ligament (PCL) is divided into the anterolateral and posteromedial bundle, which are both anchored in the lateral surface of the medial epicondylus and the area intercondylaris posterior of the tibia, and counteracts posterior tibial movement. With reference to recent studies, LaPrade et al. described the relationship between these two bundles as “synergistic and codominant” [1, 68, 78, 84]. The PCL is located contrary to the ACL and inhibits a posterior shift of the tibia relative to the femur as well as tibial rotation [58, 78, 87].

The medial and lateral collateral ligament (MCL and LCL) stabilize the knee joint against laxity [128] and are twice as effective as the cruciate ligaments in controlling rotary laxity [146]. The MCL can be divided in three layers, the superficial MCL, the deep MCL, and the posterior oblique ligament, which have an average width of 15 mm and an average length of 110 mm [99, 119]. It is located at the adductor
tubercle of the medial epicondyle of the femur and the medial condyle of the tibia. At 25° knee flexion, the MCL provides 78 % of the restraining forces against medial opening of the joint space, which indicates that the MCL is a major stabilizer against valgus moments [57]. A study by Robinson et al. found a material strength for the superficial MCL of 534 N, for the deep MCL 194 N and for the posterior oblique ligament 425 N with a failure at an extension of 10.2, 7.1 and 12.0 mm respectively [127].

The medial and lateral menisci are semilunar shaped structures located between the epicondyles of the femur and the tibial plateau (figure 2). Both menisci are anchored on the tibial plateau, with the lateral meniscus being more mobile than the medial one, as it is not as attached to the knee capsule as the medial meniscus [102]. They compensate the incongruent shape of the articular surface of femur and tibia resulting in up to 99 % of axial load transmission as found by Seedhorn et al. [134]. In the absence of the menisci, the incongruent shape of the femoral epicondyle and the tibial plateau lead to a significantly higher contact pressure [102]. This shows that the menisci distribute the axial force equally between femoral epicondyles and tibial plateau, which explains the increased risk for osteoarthritis after partial or total meniscal resection [60]. Additionally, Wu et al. found that patients who had undergone an ACL surgery and meniscal resection had significant higher active limitations than those with intact menisci [152]. The compensation of the incongruent shape also stabilizes the knee joint [3, 102]. However, the shock-absorbing effect of the menisci is still controversial [5].

Figure 2:
The lateral meniscus (1) and medial meniscus (2) are anchored on the tibial plateau. The anterior cruciate ligament (3) and posterior cruciate ligament (4) are anchored in the tibial plateau and femoral epicondyles.
Knee kinematics and finite helical axes

The kinematics of the knee joint provides six degrees of freedom, which are a result of the oval-shaped epicondyles [80, 101, 115, 137]. Thereby, the passive components of the knee motion are coupled to the flexion angle of the knee [147]. The acquisition of this complex kinematics in-vivo is impeded by soft tissue artifacts caused by muscle contraction or wobbling of tissue when marker-based camera systems are used [24-26, 36, 94]. Thus, the main insights about knee kinematics are derived from in-vitro experiments.

Besides flexion, the internal/external rotation of the tibia is the most prominent rotation in the knee. Albert was probably the first to systematically investigate the tibial rotation of the knee in 1878 [2]. He reported an external tibial rotation of up to 20° when moving the knee from 90° flexion to full extension with no muscle loads applied. Other research groups have built complex knee testing rigs to apply muscle loads and investigated the effect of various treatments on the in-vitro knee kinematics. These studies found varying external rotations depending on the test setup and applied loads between approximately -2° and 20° [18, 22, 29, 38, 83, 88, 98, 105, 120, 126, 147]. Some of these studies found that the external rotation occurs only between approximately 30° - 0° flexion, which led to the term “final rotation” or “screw home mechanism” to describe this effect [120]. The adduction / abduction angle in the knee changes by about 5° during flexion, as found in-vivo by Sharma et al. [137].

Describing knee joint kinematics with angles is error-prone due to the so called “cross talk effect” which is the misinterpretation of a rotation about one axis as a rotation about another axis [28]. This effect can be caused e.g. by soft tissue artefacts or bad marker placement [24, 125]. Besides the representation of pure knee angles,

![Finite helical axes of the knee joint. From 30° to 0° flexion angle, the axes is skewed in space.](image)
Blankevoort et al. proposed the calculation of finite helical axes (FHA) for the knee joint, with a method that was introduced by Spoor and Veldpaus [18, 142]. A helical axis is the rotation axis of an object performing a three-dimensional movement in space [150]. It can be seen as the three-dimensional representation of the angles. The inclination of the axis represents the ratio of angular movement around the rotation axis. For instance, a rotation of 10° flexion and 10° external tibial rotation results in an axis with an inclination of 45° in the frontal plane [18]. The FHA are rotation axes that are computed for every given increment of the flexion angle. The screw home mechanism of the tibia leads to a tilting of the FHA between 30° - 0° as found by Blankevoort et al. and Mannel et al. [18, 98] (figure 3). It is difficult to imitate this complex motion with a mechanical hinge, and it is questionable if knee braces are able to mimic the physiological knee kinematic. However, due to the unique knee kinematics and the complex interaction of tendons and ligaments together with high loads, the knee joint is prone to injuries.

1.2 Trauma patterns

In sports with fast cutting movements as in handball and squash or in sports with player-player interaction as in ice-hockey, American football or soccer, knee injuries and especially ACL ruptures are one of the most frequent injuries for athletes [32, 55, 92, 97, 109, 118, 135, 144] (figure 4 a). A study by Boden et al. reported that 28% percent of ACL injuries were induced by player-player contact, whereas 72% occurred without such interactions [19]. The mechanisms behind the injuries have been

![Figure 4:](image)

a) An exemplary duel between two soccer players with a lateral impact against the right knee of the red player.
b) Injury pattern of a „unhappy triad“: The anterior cruciate ligament, medial collateral ligament and the medial meniscus are torn
investigated by numerous studies and one typical frequently occurring injury pattern can be identified: A knee angle close to full extension combined with a tibial rotation moment and valgus stress [8, 79, 81, 86, 112, 114, 138]. During such combined loads, the tibia slides and rotates relative to the femur, resulting in high stress or even rupture of the MCL, ACL or medial meniscus, called an “unhappy triad” [50], (figure 4 b). A prevention program developed for basketball players by Henning tries to minimize the risk factors by teaching athletes to land with a bent knee after jumping instead of a straight knee and to stop motion in one direction with three steps instead of one step. He reported a decrease of ACL injuries by 89 % [56]. There is a discrepancy of injury incidences between genders. Arendt et al. found that female soccer players have a two times higher risk for ACL ruptures than the male counterpart [7]. Silver et al. reported a 2–10 fold higher risk for ACL injuries for female athletes [140]. Besides gender, an increasing age cannot be seen as a risk factor, as young athletes between 1525 years have the highest risk for an ACL injury [56]. In sports, other factors such as a high friction of the shoe sole in handball players or dry playing surface in football were identified as additional risk factors [52, 133].

PCL tears commonly occur together with ACL or MCL injuries [144]. A hyperflexion of the knee joint is the most common mechanism of PCL injury leading to isolated posterior cruciate ligament tear [46]. Fanelli et al. reported that 57 % of PCL injuries were caused by trauma and 33 % were sports related and that men are more likely to experience a PCL tear with a rate of 73 % [42]. However, a PCL injury can also occur during mundane activities such as the removal of boots [33]. Anyhow, with a prevalence of only 2 % of all high school knee injuries, a PCL injury is rather unlikely compared to an ACL injury that comprises 25 % of all knee injuries [144].

Meniscal injuries are the second most frequent injury in the human knee after ACL tears [96]. In a prospective study by Nielsen et al., the authors found that 27 % of the meniscal injuries were associated with sports, and as Fox et al. reported, most commonly without contact between players [47, 110]. A study by Baker et al. among 243 subjects found a particularly high risk for soccer players and people with high body mass index [11]. Another study by Baker et al. found that meniscal injuries occur four times more often in the medial than in the lateral menisci and that during various sports, meniscal injuries happen three times more often in men than in women [9]. A study by Metcalf et al., in which 1485 meniscus tears were investigated, showed that 98 % of the injuries in the medial meniscus were located near the posterior horn and that the most common tear pattern was horizontal in both medial and lateral menisci [104]. A link to ACL injuries was found by Frobel et al., who reported that 38 % of patients with an ACL injury had an associated meniscus tear [51].

The medial collateral ligament is the structure in the knee that is injured most frequently among American high school students with a rate of 36 % as found by
Swenson et al. [144]. A study by the Union of European Football Associations (UEFA) reported that MCL injuries are a common injury among professional soccer players with an incidence of 5% among all injuries which occur approximately 2.5 times per 1000 match hours [39]. The most common location for an MCL injury is the tibial and femoral insertion [72, 119]. Additionally, Phisitkul et al. concluded that a MCL rupture is a risk factor for a subsequent ACL injury [119].

Lateral collateral ligament injuries are relatively uncommon, especially isolated LCL tears are rare [74]. Instead, LCL injuries often occur together with injuries in the cruciate ligaments [139]. As the mechanism behind LCL injuries, a varus moment with internal or external moments together with external moments were identified, which induce a higher strain in the LCL than internal moments [85, 139].

As can be seen, the knee is exposed to extreme loading conditions and various injury patterns exist. Prophylactic knee braces are designed to protect the knee joint from risk factors and unphysiological loads. Avoiding knee injuries would also save patients expensive rehabilitation and long-term consequences as well as absence from work.

**ACL rehabilitation and long-term consequences**

After ACL reconstruction surgery, it usually takes six months for athletes until they can return to moderate training [82]. In professional soccer, Bizzini et al. recommend a return to reduced soccer practice after 16-24 weeks after surgery [17]. In a study with more than 500 patients who performed sports on a competitive level, Ardern et al. found that only 33% of these athletes attempted sports within 12 month postoperatively and 9% stated that they had given up due to knee problems [6]. Even after complete rehabilitation, an instability in the knee can remain, which might lead to subsequent meniscal damage or ostheoarthritis [30, 93, 113]. The Danish ACL registry reported a revision rate of 4.1% for overall ACL reconstructions and even 8.7% for patients under the age of 20 [90]. After ACL reconstruction, patients need at least 18 month until the full proprioception of the knee is restored [73]. Thus, an ACL rupture entails a mechanical instability of the knee together with an impairment of the motion coordination [44, 75]. Seitz et al. reported that 65% of ACL-deficient patients sustained a secondary meniscal injury within 2.5 years after the initial injury [136]. Keene et al. also found an increasing incidence of meniscal tears in patients with ACL insufficiency [77]. All these findings underline the importance of protecting the knee so that ACL injuries do not occur in the first place. One method to protect the knee might be the use of prophylactic knee braces, which are claimed to protect the knee.
1.3 Knee braces

In order to prevent the knee from trauma or to protect the knee during rehabilitation, knee braces were developed by numerous manufacturers. Most of the braces consist of two frames that can be attached via Velcro® fasteners to thigh and shank. Both frames are connected via two hinge joints on each side of the knee, which are either designed as a simple hinged joint, a more complex gear joint with usually two gears (Figure 5) or based on the principle of a crossed four-bar chain. The number of Velcro® stripes as well as the order in which the strips are fastened vary among different manufacturers.

Three types of knee braces can be distinguished [100, 148, 151] (figure 6):

![Figure 5: Three x-ray images of typical designs of brace hinges: 4titude (Donjoy, Carlsbad, USA), Genu Arexa (Otto Bock, Duderstadt), Xforce (Sporlastic, Nürtingen).](image)

![Figure 6: Three different kinds of knee braces. a) functional brace, b) rehabilitative brace, c) prophylactic brace.](image)
• Functional braces are claimed to stabilize the knee joint, e.g. for subjects with ACL deficiency to reduce strain in the ACL and protect from unphysiological rotations (tibial rotations and adduction/abduction).

• Rehabilitative braces provide controlled range of motion for the knee after surgery or for conservatively treated patients.

• Prophylactic braces are claimed to absorb unphysiological loads during sports or other activities in order to protect the knee joint from injury.

The classification of braces is not always clear and many braces claim benefits for various indications.

**Protective effect of bracing**

The results of numerous prospective studies are controversial. Rovere et al. found a higher incidence of knee injuries among football players when braces were worn together with a higher rate of ACL injuries and thus questioned the efficacy of bracing in football [130]. In another study, Hewson et al. investigated the effect of bracing in college football at the University of Arizona and found no effect of knee braces or a stricter sanctioning of fouls that were originally supposed to decrease the risk for knee injuries [70]. Hansen et al. reviewed medical records at the University of Southern California and concluded that the so-called “Anderson knee stabler” – one of the first protective single-sided braces – reduces collateral ligament injuries and meniscus injuries [65]. In contrast, a study with more than 500 athletes by Grace et al. found more injuries in the group wearing single-hinged braces than for the unbraced control group and a higher number of injuries of the foot and ankle for athletes who wore braces [54]. Other prospective studies among American football players showed also no effect of bracing in reducing knee injuries [37, 141]. After braces became established in American football, they were also used in other contact and non-contact sports. A retrospective study among 600 Swedish elite ice-hockey players also questioned the protective effect of the PKB [145]. By contrast, a study among skiers by Sterett et al. and a survey study among off-road motorcycle riders by Sanders et al. found less injuries among the study participants wearing a PKB and recommend its use [132, 143]. To sum up, retrospective and prospective studies, which analysed the number of injuries of athletes, are inconsistent but rather support the hypothesis that PKB are not affective to protect the knee during sports [56, 100, 122].

In addition to the retrospective studies, numerous experimental studies have been conducted to investigate the effect of bracing against specific structures like the ACL. Again, these studies are inconsistent: In two individual in-vitro studies, Paulos et al. and Errickson et al. exposed human limb specimens to lateral impacts. Both found no
protective effect of PKBs for the medial collateral ligament (MCL) or the ACL [41, 117]. A similar study by Baker et al. showed only little protective effect of PKBs for the MCL against lateral impacts with the knee in full extension and no effect in 20° flexion [10]. By contrast, France et al., Hangalur et al. and Paulos et al. found a protective effect for the ACL or MCL using cadaver specimens during lateral impacts [48], drop landings [64] and for lateral impacts using a mechanical surrogate limb [116]. Other in-vitro studies found that bracing might be beneficial in reducing displacement or to provide stability against internal torques [31, 149]. The research group of Beynnon et al. investigated the effect of bracing on the ACL in-vivo. They found some protection for the ACL of the braced knee against low anterior shear loads or internal torques of 5 Nm and found no adverse effects of bracing [12, 16, 45]. A protective effect for the ACL against external torques or varus moments in the weight bearing and non-weight bearing knee was not found in-vivo [45].

Studies investigating the long-term effect after knee surgeries show explicitly that bracing is not beneficial during rehabilitation: Feller et al. found no positive effect of bracing for patients four month after ACL reconstruction in terms of passive and active flexion deficits, anterior laxity, isometric hamstring and quadriceps strength [43]. Muellner et al. also found no superior effect of bracing versus bandaging for patients after ACL surgery for the stability of the knee and concluded that bracing is not mandatory after ACL reconstruction [107]. Brandsson et al. and Harilainen et al. investigated the effect of bracing up to two years after ACL reconstruction. They found no differences between the braced and unbraced group in terms of function, knee laxity or isokinetic muscle torque [20, 67]. Even after five years, Harilainen et al. found no difference between the group that wore a brace after the surgery for 12 weeks and the group that was mobilized immediately after surgery [66]. Thus, a review by Martinek et al. concluded that there is no effect of bracing in the long-term after ACL reconstruction [100].

1.4 Aims of the work

Research about bracing was popular in the 1980s, when their use was widespread among professional football players [124]. However, even though various studies about the potential protective effect of bracing exist, there is still a gap of knowledge about the exact mechanism of the knee-brace-interaction. Thus, the aim of this thesis is to help answering the question: To brace or not to brace?

The interaction of knee joint and brace is influenced by their kinematics. Obtaining exact knee kinematics in-vivo is challenging due to soft tissue artefacts induced by skin movement or wobbling muscles or fat tissue. Thus, most research groups collect kinematic data by using an in-vitro approach. Unfortunately, varying test setups create
a high variation in observed kinematics. Therefore, a meta-analysis with 19 studies was conducted to answer the following question:

- What is the mean kinematics of the knee joint and is it influenced by the test-setup?

The results of this study were published in the Journal of Biomechanics (Hacker SP, Ignatius A, Dürselen L. The influence of the test setup on knee joint kinematics – A meta-analysis of tibial rotation. Journal of Biomechanics. 2016) [61]. In contrast to the complex anatomy of the knee, the mechanical hinge joints of the braces are designed rather simply with rigid frames. This raises the question:

- What are the kinematics of different types of knee braces and can they adapt to the physiological knee motion?

Six knee braces of different manufacturers were investigated. The results of this unpublished study showed a clear difference between knee and brace kinematics. Additionally, braces might be misaligned due to poor patient compliance or slippage, which might even increase kinematic differences. These differences must be compensated by soft tissue movement or otherwise constraining forces will be induced on the knee. As braces are actually supposed to protect the knee and especially the ACL, rather than causing ligamentous stress, an *in-vitro* study was performed to answer the question:

- Does a misaligned knee brace stress the ACL?

Therefore, 8 human limbs were tested in a knee simulator and kinematics together with ACL strain were recorded. The study was approved by the Ethics Committee Ulm (no. 207/16). The results of this study were published in Prosthetics and Orthotics International (Hacker SP, Schall F, Ignatius A, Dürselen L. The effect of knee brace misalignment on the anterior cruciate ligament – an in-vitro experimental study. Prosthetics and Orthotics International. 2019) [62]. In the literature, *in-vitro* studies showed a protective effect of bracing against lateral impacts at the height of the impact. However, during contact sports, the direction as well as the height of impacts on the knee vary. Thus, we asked:

- Does a knee brace protect the knee against impacts from different directions in different heights or against internal/external tibial moments?

Therefore, a study was performed with 8 human limbs with ACL strain sensor as well as acceleration sensors implanted in the knee. The results were published in the Orthopaedic Journal of Sports Medicine (Hacker SP, Schall F, Niemeyer F, Wolf N, Ignatius A, Dürselen L. Do prophylactic knee braces protect the knee against impacts or tibial moments? - An in-vitro, multi-sensorial study. 2018) [63]. The study was approved by the Ethics Committee Ulm (no. 207/16).

In summary, this work investigated potential protective and negative side effects of knee bracing.
2 Knee kinematics

Initially, in order to compare knee and brace kinematics as a preliminary work, a meta-analysis was conducted to investigate knee kinematics.

Obtaining knee kinematics in-vivo is error-prone due to soft tissue artefacts caused by muscle contraction or wobbling of skin and fat tissue [24, 25, 35]. Thus, accurate data for knee kinematics was mostly recorded in-vitro. Unfortunately, the test setup of different studies vary, which raises the question if the results of different studies can be compared and whether a universal knee kinematic exists that can be identified throughout all study designs. To answer these questions, a meta-analysis with 19 publications was performed.

A Pubmed and IEEE Xplore database search was performed using the keywords: “knee”, “tibial” or “tibiofemoral” in combination with “rotation”, without “walking” and “gait”. This search lead to 390 articles. Only publications that provided an internal/external rotation as a function of the flexion angle from at least 0° to 90° flexion with adequate resolution in the published graph and intact knee were selected. This led to 19 publications that were used for the meta-analysis [2, 18, 22, 29, 34, 38, 40, 58, 83, 88, 91, 98, 103, 105, 108, 126, 137, 147]. The neutral curve of the internal/external rotation curve was obtained using image segmentation. To overcome the problem of different offsets in the curves due to different methods for angle calculation, the gradient of the curves was computed. This gradient was used for further computations. First, a mean tibial rotation was computed with the integral of the gradient curve for the 14 studies that obtained internal/external rotation in-vitro. Second, the difference between the two gradient curves of each combination of the 19 publications was computed.

To compare the different test setups, four test groups were defined: method of angle calculation, loading condition, system for data acquisition and testing rig design. Each group was divided into various subgroups and each publication classified to one subgroup for each group respectively, (table 1). Two publications were defined as being similar in one particular group, if they were both in the same subgroup, but not in the subgroup “miscellaneous” respectively.
The mean internal rotation of the 14 *in-vitro* studies is shown in figure 7. To simplify the curve, a tangent at 0° flexion and one line that averages the external rotation from 60° to 90° flexion was inserted. This approximation intersects at 30° flexion, which might be used to define the start of the screw home mechanism. To compute specific points of the curve, a fitting function was computed with a goodness of fit of r-square equals 0.9999:

\[
f(\alpha) = 11.95 \times \sin(0.0244 \times \alpha - 0.0720) + 1.394 \times \sin(0.0673 \times \alpha + 0.5825)
\]

The correlation matrix in figure 8 shows the difference of two publications. The diagram is divided into two parts: The upper diagram visualizes differences between the gradient curves. The size of the circles represents the area between two gradient curves.

### Table 1: Classification of the groups and their subgroups.

<table>
<thead>
<tr>
<th>Group</th>
<th>Method for angle calculation</th>
<th>Loading condition</th>
<th>System for data acquisition</th>
<th>Testing rig design</th>
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The mean tibial rotation computed as integral of mean gradient curve of 14 *in-vitro* studies with no torque applied. Dashed line: geometric approximation of screw home mechanism. Reprinted with permission from [61], p. 2985, Elsevier 2016.
The upper graph shows the difference between the gradient of tibial rotation of all 19 publications. The larger the diameter, the more different are the measured tibial rotations between these two publications. This can be understood as the difference between the magnitudes of tibial rotation between the two respective publications. The lower graph displays similarities or differences within the test setup. Each circle of the cloverleaf stands for one of four attributes of the test setup. Green indicates a similar test setup, while gray marks a difference. Reprinted with permission from [61], p. 2987, Elsevier 2016.
curves of two publications respectively: The larger the circle, the larger the difference between the measured tibial rotation curves of these two publications. The lower graph displays differences or similarities in the test setup of two publications. With this correlation matrix, all publications can be compared with regard to the measured tibial rotation and similarities within the test setup. It can be seen that there are only two combinations of publications with the same test setup (Grood, 1988 vs. Wilson, 2000 and Bull, 2008 vs. Merican, 2011). Additionally, there are research groups (e.g. Nagamune, 2008) who use a rather unique setup, whereas other test setups are more frequent in literature (e.g. Grood, 1988). As expected, publications with a good correlation to the mean tibial rotation (e.g. Li, 2008) overall have a good correlation with other publications. Anyhow, even if the test setup is the same, the comparability between the raw data of two different studies is limited and the collection of a control group is necessary.

The mean internal rotation of more than 130 specimens that was computed in this study clearly shows the screw home mechanism of the human knee joint of approximately 10°, starting at 30° flexion. To the author’s knowledge, this was the first meta-analysis providing an internal rotation as a function of flexion angle based on such a large data base. This shows that the knee performs a complex movement which has to be imitated by a knee brace. Otherwise, constraining forces on the knee might occur.

Thus, in the next step, the brace kinematics were investigated.
3 Brace kinematics

With the complex kinematics of the knee that was shown in the previous paragraph and the simple gear mechanism of brace hinges in mind, it can be assumed that there is a difference between both kinematics, which has to be compensated by either knee joint, brace or a combination of both. Therefore, the kinematics of five different knee braces (table 2) and the effect on soft tissue movement were investigated.

The finite helical axis and the external rotation of the braces were computed for three test setups:

- The knee brace were not worn by a subject and moved under no load by hand for five cycles.
- The brace was worn by a subject with the brace hinge correctly aligned according to the manufacturers’ guidelines.
- The brace was moved 50 mm distally to simulate hinge misplacement or migration respectively.

In setup (2) and (3), the subject performed five consecutive squats.

Four markers were placed on the thigh and shank frame of the brace respectively. A brace coordinate system was defined at 0° flexion angle according to Grood and Suntay’s [59] definition of a joint coordinate system in the knee joint (figure 8). Two planes were defined as being parallel to the sagittal plane with an offset of 50 mm in the medial and lateral direction. These planes are located close to the braces’ hinges (figure 9). Marker movement was recorded using nine 3D cameras (Prime 13, Optitrack, NaturalPoint Inc., USA) with a frame rate of 240 Hz and a mean error of less than 0.25 mm after calibration. The finite helical axes were computed with the method provided by Spoor and Veldpaus in combination with a quaternion-based method for the determination of the rotation matrix by Horn [71, 142].

All computed FHA for all three scenarios (unloaded, correct alignment, wrong alignment) of the braces are displayed in fig. 10. The FHA for the unloaded testing scenarios are aligned parallel throughout the movement from extension to flexion.
while performing a small circular translation from a minimum of 6 mm (GenuArexa) to a maximum of 10 mm (M.4s®). When worn by the subject (correct and wrong alignment), the computed FHA differ slightly in some cases: Whereas the axes are still aligned parallel, the translational movement changes from a smooth circular translation to a more edged translation (M.4s®, Genudyn). The maximum translation is still in the range of 6 mm (GenuArexa) to 10 mm (Genudyn). However, the physiological screw home mechanism of the knee joint, which would result in an erection of the FHA, could not be observed for any of the braces in any of the three testing scenarios.

The internal/external rotation of the knee brace is shown for all braces in relation to the physiological tibial rotation reference curve that was obtained in the meta-analysis

![Figure 9: Definition of the used coordinate system of the brace.](image)
Figure 10: The FHA of all knee braces: each row represents three test scenarios of one brace. Left: brace not worn by a subject but moved by hand. Middle: worn by a subject with the braces' hinge correctly aligned with the knee joint. Right: worn by a subject with the braces moved 50 mm to distal to simulate a migration worst case. Additionally, the intersection curve of each FHA with a plane parallel to the Y-Z-plane and a distance of 50 and 50 mm respectively is displayed.
The internal/external rotation of all braces with neutral loading and correct alignment is between -1.5° and 1.5°. The internal/external rotation of the shank frame is higher for some braces under wrong alignment with a maximum of 2.9° (Genudyn) at the maximum flexion angle.

To sum up, we found that from a kinematic point of view, the characteristics of the different braces vary only slightly even when the braces were worn misaligned. Thus, an essential difference in kinematics of braces designed for different indications could not be observed.

Even more essentially, the physiological movement of the knee joint is not imitated by any of the investigated braces. This behaviour could actually be expected considering the mechanism of the hinge and frame, which is apparently too rigid to adapt to misalignment of knee and brace rotation axis. This raises the question if a brace induces constraining forces on the knee joint, especially in the misaligned condition.
4 Effect of a misaligned knee brace on ACL strain

Knee brace misalignment and discomfort can be induced by brace slippage [121]. A study by Brownstein found a maximal migration of 11 mm after only 15 min of exercises [21]. This brace slippage combined with the misalignment of the finite rotation axes of knee and brace caused by the different kinematics might induce constraining forces and influence structures within the knee. As described above, the ACL is one of the most prominent structures in the knee and prone to injuries and tears. For that reason we asked if knee brace misalignment influences ACL strain [62].

To answer this question, an experimental study was performed in which eight human leg specimens were instrumented with an ACL strain sensor (DVRT, Microstrain, Williston, USA), additional Bowden cables to simulate muscle forces and reflective markers to obtain knee kinematics. After preparation, the cadaveric limbs were mounted in a testing rig and muscle forces (100 N hamstrings, 150 N quadriceps) were applied to provide some joint stabilization. Care was taken not to damage the soft tissue and the skin of the limb was carefully closed after the preparation.

The specimen legs then were moved to simulate five squats with a velocity of 20°/s from 10° to 60° flexion angle (figure 12) and the ACL strain and knee kinematics were recorded. The squats were performed in three conditions: 1) without brace, 2) correctly aligned brace (4titude, DJO Global, Inc., Vista, USA) and 3) brace migrated 20 mm distally. Data and statistical analysis were performed in Matlab 2017a using a Student’s paired t-test and a level of significance of alpha = 0.05 which is similar to a study performed by Beynnon et al. [13, 14].

After the first two specimens were tested, an analysis of the data revealed that the pins of the DVRT which were pierced through the ACL were too long, causing an impingement of the pins with the tibial plateau. This influenced the DVRT accuracy and was consequently corrected for the following tests. For this reason, all eight specimens were used for the kinematic analysis but only six for the ACL strain analysis.

We found that the ACL strain at 10° flexion angle was significantly reduced by
the correctly aligned brace to a mean of −1.5 % but it did not significantly reduce the ACL strain at 60° (figure 13). In the misaligned condition, the ACL tended to be more stressed than in the unbraced condition at 60°, with an ACL strain of −1.1 %. This effect was not statistically significant. The ACL strain in the misaligned condition at 60° was significantly higher than in the correctly aligned condition. There were no statistically significant differences between the kinematics for any of the three scenarios (figure 14).

Discussing the results, the overall gradient of the ACL strain vs. flexion curve was similar to the in-vivo findings of Beynnon et al., with a decreasing ACL strain from 10° to 40° flexion and an almost constant strain from 40° to 60° flexion [15]. In-vivo studies by Beynnon et al. and Fleming et al. investigated the effect of bracing on the ACL strain during internal and external moments on the tibia and anterior-posterior shear load in patients undergoing arthroscopic surgery [14, 45]. Both found that a correctly aligned brace reduced the ACL strain when no external loads were applied, which is similar to our findings. In addition, the findings of this study indicate that a misaligned brace neutralizes the stress-decreasing effect on the ACL of a correctly aligned brace. There might even be a trend of a negative effect on the ACL strain if the brace is misaligned. This being said, we conclude that the correct alignment of the brace is mandatory for providing a protective effect for the ACL.
This study showed that the positive effect of bracing for the ACL can be neutralized by misalignment. In a next step, we investigated if bracing protects the knee against impacts or tibial moments.
During sports, the knee joint is exposed to various unphysiological loading conditions and thus in high danger of being injured. Especially sports with player-player contacts or fast cutting manoeuvres, as can be seen in soccer, handball, squash or American football, have a high incident of knee injuries [92, 97]. In these sports, tackles or side impacts to the leg are main risk factors for injuries, which can occur from any side: medial, lateral, anterior, posterior [53]. Up to now, studies have only investigated the effect of lateral impact at the height of the center joint line [41, 49, 117]. Thus, to investigate whether a PKB has a protective effect against impacts from other directions, impact heights or tibial rotational torques, an in-vitro study was performed.

Eight leg specimens were used in this study (age 70 ± 12 years; Science Care, USA). These specimens were instrumented with two acceleration sensors, which were implanted in the femur and tibia, an ACL strain sensor on the anteromedial bundle of the knee and retroreflective markers to obtain knee kinematics (figure 15). During the preparation of the leg, care was taken to leave the soft tissue intact and the skin was closed again after the sensor placement. One additional Bowden cable was attached to the tibial and another Bowden cable was sewed to the longitudinally split quadriceps tendon. The equipped knee was mounted in a knee simulator at 30° flexion angle. Moderate muscle forces (100 N hamstrings, 150 N quadriceps) were applied to stabilize the knee joint and a 2 kg weight was shot onto the knee with a velocity of 1 m/s from medial, lateral, anterior, posterior direction in three heights (center joint line, and 100 mm superior and inferior) in the braced and unbraced condition (figure 16). Afterwards, internal and external torques of 5 Nm at 30° and 60° flexion angle were applied. Data and statistical analysis was performed in Matlab 2017a using a nonparametric Mann-Whitney-U-test and a level of significance of alpha = 0.05. We recorded the acceleration within the femoral and tibial bones, which indicates
how much impact energy is transferred directly to the joint. As this energy has to be absorbed by the ligaments and tendons, a high acceleration measured inside the bones indicates a high load on the structures of the knee. We found that bracing significantly reduced the inner joint acceleration for medial and lateral center impacts (figure 17). A protective effect of bracing for the inner joint acceleration could not be seen for anterior and posterior impacts except for inferior anterior impacts. Interestingly, bracing tended to increase ACL strain for all anterior impacts at every height. Furthermore, we found that bracing did not influence ACL strain or tibial rotation for internal or external moments.
While the acceleration sensors measured high differences in the accelerations inside the bones, the observed effects in the ACL strain were rather small. This might be explained with the stabilizing effect of the muscle forces together with the subcritical impact energy of 1 Joule, which leads to the assumption that the effect of bracing is probably more prominent during contact sports with higher impact energy. Even so, our results indicate trends and demonstrate that there is no simple answer to the research question whether prophylactic braces actually do protect the knee during sports. For lateral and medial impacts, bracing tends to reduce the ACL strain, which is in agreement with a study by Paulos et al. [116]. Erickson et al. found a reduced impact force for braced knees during lateral impacts but no significant protective effect for the ACL [41]. Our finding of higher ACL strain for anterior impacts in the braced condition might be explained with the mechanical coupling effect of the brace: In braced condition, the mechanical coupling between thigh and shank might prevent a posterior translation of the tibia so that instead, an extension torque is induced. This knee extension torque might have had the same effect as an actual knee extension, which was shown to increase ACL strain [15].

It is notable that there is a high difference in the effect of anterior vs. posterior impacts on the acceleration within the knee joint. This can most likely be explained by the higher amount of soft tissue at posterior that apparently absorbs most of the impact energy, whereas an impact from the anterior more or less hits the bone directly. Regarding internal/external torques, bracing had no significant effect on the ACL strain. These findings are in agreement with a study by Fleming et al. who also found no significant reduction of bracing in the weightbearing knee against internal/external tibial moments [45].

In conclusion, the effect of bracing depends on the direction and height of the impact and is partly positive, negative or neutral. Thus, we cannot advice bracing for every loading scenario. In fact, looking at the absorbed impact energy, kneepads or padding that help absorb the impact energy might be more beneficial than knee braces.
Figure 17: (graph is rotated 90°) Results of all parameters for lateral and medial impacts at the height of the center of the joint line and 100 mm above and below. Green indicates the unbraced, blue the braced condition. The acceleration within the bone is reduced partially, whereas the effect on the ACL strain change is only significant for high medial impacts. Bracing does not influence the knee kinematics. Data is displayed as the median and lower and upper quartiles.

* p < 0.05. Reprinted with permission from [63], p. 6, SAGE 2018 distributed under CC BY-NC-ND 4.0, https://creativecommons.org/licenses/by-nc-nd/4.0/
6 Conclusion

Various manufacturers claim that prophylactic knee braces protect the knee against various loading conditions. Even though these orthotic devices have been on the market for more than 40 years, it is still controversial if knee braces are protective against loading conditions such as impacts or rotational moments at all. Thus, the aim of this thesis was to identify positive and negative effects of bracing on the knee.

In a first step, the kinematics of knee and brace were analyzed. Therefore, a meta-analysis was performed to identify the physiological knee movement and averaging the kinematics of more than 130 specimens out of 19 publications. The kinematics of five knee braces were investigated in three loading scenarios: neutral, correct aligned and misaligned. The latter might be caused by poor patient compliance or by brace slippage due to muscle contraction during walking or sport activities. We found that there is a difference between knee and brace kinematics regarding the physiological tibial rotation of the knee, which is not mimicked by the brace. Additionally, the kinematics of the five braces were almost the same, meaning that there is no difference in the braces’ kinematics depending on the indications claimed by the manufacturers. Thus, the only relevant differences between braces for different indications is the strapping of the Velcro® fasteners. Finally, the frame of the brace is too stiff to allow an adaption of the kinematics if the brace is misaligned.

With this in mind, we investigated the effect of a correctly aligned and misaligned brace on the ACL. We found that a correctly aligned brace reduces ACL strain during a cyclic movement, while a misaligned brace slightly increases ACL strain and neutralizes the positive effect of the correctly aligned brace on the ACL completely. These results underline that it is important to train patients how to tighten a knee brace correctly and to check the correct alignment of the brace frequently. Additionally, this might be an indicator that a brace that is meant to protect the knee might induce constraining forces on the ACL.

Finally, we investigated if braces are protective against impacts from all sides and tibial rotational torques. For this, a new method was developed with two acceleration sensors that were implanted in the femur and tibia to record the energy that is trans-
ferred to the inner joint. We found that the brace protects the knee against lateral and medial impacts but might be disadvantageous against anterior impacts. Additionally, the soft tissue absorbs a lot of the impact energy and might even be more protective than the brace. Finally, we found that bracing does not protect the knee against internal or external torques. This can be explained by the soft tissue movement that inhibits a rotational stabilization of the brace.

To answer the question “To brace or not to brace?”, we summarize that knee braces do not mimic the physiological knee movement, they have no protective effect on the ACL if misaligned and have no effect or even a negative effect for anterior and posterior impacts as well as internal/external torques. Only against medial and lateral impacts and in the correct aligned condition, braces can decrease the impact energy and the stress on the ACL. With that in mind, patients after ACL surgery might benefit from the stabilizing effect against varus/valgus moments and the stress-decreasing effect of bracing. Athletes with intact ACL have to question if the potential benefit of bracing is worth the possible risk of constraining forces on the knee. Especially with the success of studies in mind that have shown the positive effect of prevention training to reduce knee injuries and the protective effect of well-trained musculature, there might be more effective ways than bracing to protect the knee.
Abstract

Prophylactic knee braces are used in rehabilitation after surgery or during sports and are claimed to protect the knee from unphysiological loads. In this work, the effect of prophylactic knee braces on the knee was investigated to answer the question, if braces do really protect the knee or might even have negative side effects. First, the kinematics of knee and brace were compared to improve the understanding of the knee-brace interaction and derive further research questions for this thesis. Therefore, a meta-analysis was performed to obtain the kinematics of the knee using the kinematic data presented in 19 publications. The mean kinematics of these publications showed a tibial rotation during a movement from 30° flexion to full extension. Additionally, the kinematics of five knee-braces was obtained under various loading conditions. The comparison of the kinematic data of these two studies revealed that knee braces do not mimic the physiological screw home mechanism of the knee joint.

This led to the question if wearing a brace influences knee kinematics or if constraining forces on the structures of the knee might be induced. Thus, a prophylactic brace was attached to eight leg specimens, which were mounted in a knee simulator and were moved from extension to flexion while ACL strain and kinematics were recorded during five motion cycles. The attachment of the brace was varied from correctly aligned to misaligned, simulating a worst case brace slippage that might occur during sports or daily activities. It was found that a knee brace reduced ACL strain when correctly aligned. In contrast, a misaligned knee brace increased the ACL strain compared to the correctly aligned condition. This indicates possible negative effects of bracing if the knee brace is misaligned and underlines the importance of good patient compliance when prescribing a knee brace.

Furthermore, it was investigated if prophylactic knee braces protect the knee against impacts or rotational moments that typically occur during contact and non-contact sports. Therefore, a 2 kg weight was shot onto eight leg specimens with and without a brace. The position and height of the impacts was varied resulting in 24 impacts on each specimen. Inner joint acceleration was obtained using two custom build sensors
which were implanted in the femur and tibia respectively as well as ACL strain and knee kinematics. Afterwards, internal and external moments were applied to the specimens in braced and unbraced condition at 30° and 60° flexion angle. The results showed that the effect of bracing for impacts is inconsistent. While the knee brace protected against medial and lateral impacts, bracing tended to increase ACL strain during anterior impacts. It was also found that soft tissue absorbs a lot of the impact energy, reducing the effect of the impacts on inner joint acceleration and ACL strain. An effect of bracing on ACL strain or knee kinematics for the induced internal and external torques was not observed.

To sum up, a correctly aligned brace reduces ACL strain, whereas brace slippage aggravates the mismatch of knee and brace kinematics leading to increased ACL stress. Protective knee braces protect the knee against medial and lateral impacts, but for all other loading scenarios tested in this work, bracing had no or even a negative effect. Thus, bracing might protect injured or repaired ACL during rehabilitation against varus/valgus moments, but might not be beneficial for prophylactic uses during sports or daily activities were a combination of different load scenarios act on the knee.
8 Bibliography


Bibliography - 34


9 Originals with copyright and permission notices

Own contribution: planning of study, data acquisition, data analysis, visualization and interpretation of data, creating figures, writing publication

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The influence of the test setup on knee joint kinematics – A meta-analysis of tibial rotation

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A B S T R A C T

The human knee is one of the most investigated joints in the human body. Various test setups exist to measure and analyse knee kinematics in vitro which differ in a wide range of parameters. The purpose of this article is to find an answer to the question if the test setup influences the kinematic outcome of studies and to what extend the results can be compared. To answer this question, we compared the tibial rotation as a function of flexion angle presented in 19 published studies. Raw data was extracted via image segmentation from the graphs depicted in these publications and the differences between the publications was analysed. Additionally, all test setups were compared regarding four aspects: method for angle calculation, system for data acquisition, loading condition and testing rig design. The resulting correlation matrix shows the influence of the test setup on the study outcome. Our results indicate that each study needs to collect its own reference data. Finally, we provide a mean internal rotation as a function of flexion angle based on more than 140 specimens tested in 14 different studies.

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1. Introduction

The investigation of joint kinematics is a wide field in biomechanical testing. For the knee joint, a lot of studies exist to investigate for example the outcome of surgical procedures, the influence of muscle forces and ligaments or the effect of various loading conditions on the joint’s motion. Such studies measure the complex kinematics of the knee joint, that has three rotational and three translational degrees of freedom (DOF). Since all DOF are coupled to flexion, knee kinematics can be seen as a function of the flexion angle (Wilson et al., 2000). Towards the end of the rotation of the tibia from flexion to extension the tibia performs an external rotation, also known as “final rotation” or “screw-home mechanism” (Piazza and Cavanagh, 2000). The magnitude of this external rotation is an indication for instability or injury of the knee and is also affected by the shape of the femoral condyles and tibial plateau (Kärrehom et al., 1988; Sharma et al., 2012). Typically, studies compare the influence of loading conditions, injuries, repair strategies etc. on tibial rotation in comparison to a reference measurement of an unloaded or healthy knee respectively. So, the influence of the specific research question can be compared to a neutral condition. This raises the question if these already measured references can be used for new studies by other research groups with potentially varying test setup and if the test setup has an influence on the study outcome.

One of the differences in the test setup is the used testing rig for in vitro measurements. In most instances one of the following three designs is used: a vertical oxford rig, a horizontal rig or a robotic arm, Fig. 1a–c). These testing rigs vary in the part of the leg that is rigidly attached to the frame and possible loading conditions. This becomes apparent when the neutral path of the knee is determined. With a robotic arm, typically a passive path of motion, which is the path of minimum required forces, is identified (Diermann et al., 2008; Li et al., 2008), whereas the other testing rigs apply an axial load and/or muscle forces to the joint and one or both segments of the knee joint can move freely (D’Lima et al., 2001; Müller et al., 2009).

Additionally, several techniques exist for data acquisition, such as marker based 3D-camera systems (Mannel et al., 2004; Merican et al., 2011). With the markers placed on anatomical landmarks, the movement is described with an Euler angle approach or as suggested by the ISB with a coordinate system using a floating axis, Fig. 1d) and e). Alternatively, knee joint motion can be measured with the testing rig respectively robotic arm itself (Diermann et al., 2009; Li et al., 1999; Reuben et al., 1989). Another possibility to describe knee joint motion is to calculate the instantaneous rotation axis of the knee, the so called finite helical axis (Fig. 1f)) (Blankevoort et al., 1990; de Lange et al., 1990).

We performed a meta-analysis and compared the published reference curve of the tibial rotation of 19 studies with varying test setups. We analysed tibial rotation of all studies and offer a function that can be used to compute tibial rotation as a function of flexion

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Fig. 1. a)-c): Different types of rig designs. a) Vertical Oxford rig: Femur and tibia are mounted one upon the other, both can rotate. b) Horizontal “floating” rig design: One segment is fixed while the other segment can rotate freely. c) Robotic arm: One segment is fixed while the other is moved by a robotic arm. Forces are measured with a universal force sensor (UFS) attached to the mount of the robotic arm. d)-f) Different types of measurement techniques: d) Euler angles (or so called cardan angles), with a coordinate system that is fixed within the moving segment. e) The definition according to Grood and Suntay (1983) with a floating axis perpendicular to two fixed axis within each segment respectively. f) The instantaneous rotation axes of the tibia for finite rotations steps from 0° to 90° flexion, called finite helical axis (FHA).

Fig. 2. Overview procedure for comparing publications.
angle. Additionally, we compared each study against each other in order to find differences or similarities within different settings. Our meta-analysis indicates what extend measurements between different test setups can be compared and if a study should collect its own reference measurements.

2. Methods

2.1. Design of meta-analysis

In this review we compared the tibial rotation curve of the knee joint presented in 19 published studies. Each study was selected with regard to the published data and a uniform rotation representation (tibial rotation as a function of flexion angle). To compare the magnitude of the final knee rotation without the influence of offsets that can occur due to different methods for angle calculation, the gradient of the raw data was computed. For every combination of the 19 publications the difference, meaning the area between the gradient curves of each of two publications respectively, was determined. Together with an overview of each of the utilised test setups, a correlation matrix visualises the results of this meta-analysis. Finally, a mean tibial rotation of 14 in vitro studies was computed and used to determine a best-fitting function, Fig. 2.

2.2. Selection of publications

A Pubmed and IEEE Xplore database search was performed using the keywords: “knee”, “tibial” or “tibiofemoral” in combination with “rotation”. The keywords “walking” and “gait” were excluded from search, as gait analysing studies are susceptible to soft tissue artefacts, leading up to 13° error in tibial rotation and the maximum of flexion during gait reaches only 40°–60° (Feinschmidt et al., 1997). This search lead to 390 articles. Additional publications were found as citations within these articles. Only publications that provided an internal/external rotation as a function of the flexion angle from at least 0° to 90° flexion with adequate resolution in the published graph were selected. The authors of the studies indicated that only knee joints were used without signs of ligament insufficiencies or were in intact condition respectively. This lead to a total of 19 publications that were used for this meta-analysis, Fig. 2a (Albert, 1879; Blankevoort et al., 1990; Bull et al., 2008; Churchill et al., 1998; D’Lima et al., 2001; Diermann et al., 2009; Elias et al., 1999; Grood et al., 1988; Kwak et al., 2000; Li et al., 1999, 2008; Lo et al., 2008; Mannel et al., 2004; Merican et al., 2011; More et al., 1993; Nagamune et al., 2008; Reuben et al., 1989; Sharma et al., 2012; Wilson et al., 2000) (Table 2).

2.3. Segmentation of graphs

To extract data from published graphs, an open source program for image segmentation was used (Engauge Digitizer, version 5.1). Each graph was scaled to full screen, saved as PNG-file to prevent compression artefacts and imported into Engauge Digitizer. The maximal error of this procedure was determined with reference curves and found to be less than 0.1° per data point in the worst case. The tibial rotation angle vs. flexion angle was then segmented and exported as CSV-file. The CSV-file contained the raw data pairs of flexion and internal rotation of each publication respectively, Fig. 2b. The number of segmented data points equals the number of nodes that are found in the corresponding graph.

2.4. Data analysis

The analysis was done using Matlab R2014b (The MathWorks, Natick, US). Raw data of each of the 19 segmented data sets were imported and linearly interpolated from 0° to 90° flexion angle α in 1° increments at every integer degree \( i = 0 \ldots 90° \) (Fig. 2c)). This resulted in an internal rotation \( \beta_i \) for every flexion angle of each publication \( j \) respectively.

Internal rotation angles can have an offset depending on the used method for angle calculation (Elias et al., 1999; Sharma et al., 2012). To compare the magnitude of the final knee rotation without the influence of offset compensation, the gradient of the raw data was computed (Fig. 2d)).

\[
\beta_i - \beta_{i-1} \\
\alpha_i - \alpha_{i-1}
\]

In order to obtain a mean internal rotation curve that can be used e.g., as neutral boundary condition for FEM models, a mean gradient curve \( g \) was computed and integrated (Fig. 3e) using all in vitro studies that do not apply tibial torque, resulting in 14 studies.

\[
\bar{g}_j = \frac{1}{14} \sum_{i=1}^{14} g_{ij}
\]
3. Results

3.1. Gradient curves show screw home mechanism

The raw data of the 19 publications is presented in Fig. 3a). The resultant mean gradient and standard deviation of the tibial rotations of all publications is displayed in Fig. 3b). The gradient gives an impression, if the tibia performs an internal or external rotation and is an indication about the ratio of tibial rotation and flexion. As the screw home mechanism is also named final rotation, we defined the internal and external rotation with respect to a knee movement from 90° flexion to full extension. In that case, an external rotation has a positive gradient and an internal rotation a negative gradient. These curves are offset independent. Notably, the standard deviation does not vary considerably from 15° to 90° flexion, with a maximum of 0.34 at 0°, a minimum of 0.80 at 56° and an overall mean of 0.15. The largest difference between publications is found between 15° flexion and full extension. Interestingly half of the publications measured an internal rotation of the tibia from 90° to 60° flexion angle, whereas the other half measured an external rotation. Additionally, all publications consistently found an external rotation for a flexion angle smaller than 30° (Table 2).

3.2. Best-fit curve of mean tibial rotation

The mean internal rotation of the 14 in vitro studies with no torque applied was computed as the integral of the mean gradient curve of 14 in vitro studies with no torque applied. Dashed line: geometric approximation of the mean tibial rotation angle smaller than 30°, with a goodness of fit of r-square: 0.9999.

Table 2

<table>
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<th>System for data acquisition</th>
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</table>

3.3. Correlation between publications is independent of test setup

The correlation between two publications is defined as the difference between both gradient curves, which is equal to the area between both curves. As an example, the internal rotation raw data of two publication pairs with the highest and lowest difference are displayed in Fig. 5a) respectively. The gradient of these four publications and the corresponding differential areas are shown in Fig. 5b). This graph gives an impression of the large differences that can occur between the internal rotation characteristics assessed by two different researchers.

The correlation matrix for the difference of two publications is shown in Fig. 6. The upper diagram visualises differences between the gradient curves. The size of the circles represents the area between two gradient curves of two publications respectively: The larger the circle, the larger is the difference between the measured tibial rotation curves of these two publications. The mean difference of the published tibial rotation gradient between two publications is 170, with a minimum of 5.3 (Li et al., 1999 vs. D’Lima et al., 2001) and a maximum of 39.1 (Nagamine et al., 2008 vs. Albert, 1879). Both examples are shown in Fig. 5. The mean difference of tibial rotation gradient between one publication and the mean gradient is 116.

The lower graph visualises differences and similarities in the test setup. Each circle of the “cloverleaf” represents one group of attributes: The upper circle represents conventions for angle calculation, the left one the method of angle calculation, the right one the design of the testing rig and the lower circle represents the system for data acquisition. The circles are colour coded, with green colour corresponding to closer reading of all publications is displayed in Fig. 3b). The gradient gives an impression, if the tibia performs an internal or external rotation and is an indication about the ratio of tibial rotation and flexion. As the screw home mechanism is also named final rotation, we defined the internal and external rotation with respect to a knee movement from 90° flexion to full extension. In that case, an external rotation has a positive gradient and an internal rotation a negative gradient. These curves are offset independent. Notably, the standard deviation does not vary considerably from 15° to 90° flexion, with a maximum of 0.34 at 0°, a minimum of 0.80 at 56° and an overall mean of 0.15. The largest difference between publications is found between 15° flexion and full extension. Interestingly half of the publications measured an internal rotation of the tibia from 90° to 60° flexion angle, whereas the other half measured an external rotation. Additionally, all publications consistently found an external rotation for a flexion angle smaller than 30° (Table 2).

3.2. Best-fit curve of mean tibial rotation

The mean internal rotation of the 14 in vitro studies with no torque applied was computed as the integral of the mean gradient curve of 14 in vitro studies with no torque applied. Dashed line: geometric approximation of the mean tibial rotation angle smaller than 30°, with a goodness of fit of r-square: 0.9999.

Table 2

<table>
<thead>
<tr>
<th>Author, Year</th>
<th>Number of specimens</th>
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<th>Loading condition</th>
<th>System for data acquisition</th>
<th>Testing rig design</th>
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indicating a similar test setup, while grey is marking a difference. The highest congruence can be found in the method for data acquisition with 57 publication combinations using the same technique. 46 combinations conform in the applied loads, 48 in the used testing rig and 27 in the method for angle calculation. With this correlation matrix all publications can be compared with regard to the measured tibial rotation and similarities within the test setup.

As can be seen there are only two combinations of publications with the same test setup (Grood, 1988 vs. Wilson, 2000 and Bull, 2008 vs. Merican, 2011). Additionally, there are some publications (e.g., Nagamune et al., 2008) using a rather unique setup, whereas other setups are more frequent in literature (e.g., Grood, 1988). As expected it can be seen that publications with a good correlation to the mean tibial rotation (e.g., Li, 2008) have an overall good correlation with other publications. Furthermore, with this graph an individual case analysis can be performed, e.g., applying an internal torque on the reference path does not necessarily improve correlation (Diermann, 2009 vs. Blankevoort, 1990).

4. Discussion

In this meta-analysis we compared the reference curves of the tibial rotation of 19 publications with respect to the used test setup. These neutral curves typically serve as a reference to compare the influence of various loading conditions, injuries or repair strategies in the corresponding studies. Our findings show that only few publications measured a similar tibial rotation. One reason might be the different test setup, which differs in terms of testing rig design, applied loads, method for data acquisition and the computational method for angle calculation. Of all 19 publications and 171 combinations of test setups, only 2 similar setup combinations were found.

As suspected, there are publications that have a similar outcome with a similar test setup, e.g. Churchill, 1998 vs. D’Lima et al., 2001. But, surprisingly a higher correlation in the test setup does not necessarily lead to a smaller difference in the measured tibial rotation as can be seen at Bull et al. (2008) vs. Merican et al. (2011), who have the same test setup but only an average similarity. In contrast, Reuben et al. (1989) and Li et al. (2008) show a good congruence of mean tibial rotation, but with a different test setup. To sum up, there is no trend or rule apparent that can explain the variations dependent on the groups that can be clustered. Hence, the reason for the difference in tibial rotation assessed by various researchers is not fully understood. To our opinion, high interindividual anatomical variations may lead to the varying knee joint kinematics found by the authors of the considered studies.

Another explanation for the fact that no relation between test setup and measured tibial rotation could be found, might be an insufficient definition of the groups or subgroups, which could be a limitation of this study. Unfortunately, it is difficult to define more than four main groups, as the information provided by the publications about the specific test setup is limited. However, we believe that the current definitions are a decent compromise between an exact classification of the test setup and a reasonable number of subgroups.

A comparison between the standard deviations of the publications and of the computed mean internal rotation would have also been interesting. Due to non uniformly defined coordinate systems used in the selected publications the offset compensation varies. Therefore, the standard deviation for the mean internal rotation is not explicitly defined and not reasonable to compute. Nevertheless, the offset corrected standard deviation of the gradient curves (Fig. 3) clearly elucidates the diversity of the published tibial rotation curves. However, this result can be seen as a clear indication that a reference curve has to be collected for each individual study respectively.

Unfortunately, the number of DOF is not clearly visible in all publications. Since constraints would alter knee joint kinematics and could possibly explain the absence of the screw home mechanism in some publications, it is also important to accurately describe the experimental setup in publications. We propose a brief description of the following keywords: rig design, system for data acquisition, coordinate system/definition of axis, loading conditions, type of specimen, constraints and algorithms for data analysis. There are more attributes that might effect a difference in measurements than the four groups we defined. For example: age, gender and individual anatomy can also influence the outcome of studies. Furthermore, varus and valgus angles of the knee joint were not included in our model, because only a small number of publications measured these rotations as a function of flexion angle.

The presented mean internal rotation of more than 130 specimens being tested clearly shows the screw home mechanism of the human knee joint. To our knowledge, this is the first meta-analysis that can provide internal rotation as a function of flexion angle on such a large data base. Unfortunately, due to offset compensation it is not possible to compute a standard deviation for this curve. But nevertheless, the provided function can be of

Fig. 5. To show the high variability of tibial rotation characteristics assessed by different researchers the data of two pairs of publications exhibiting the lowest and highest correlation are displayed exemplarily: a) Tibial rotations of the publications with the highest (dotted lines) and lowest correlation (dashed lines). b) Gradient of the curves displayed in a. The areas highlighted in dark and light grey respectively represent the difference between the gradient curves of two publications.
use for other groups to validate experimental results or to compute analytic knee models.

5. Conclusion

Our meta-analysis of 19 studies shows that it is extremely important for in vitro studies to collect their own reference data respectively. Even if the test setup is the same, the comparability between the raw data of two different studies is limited and the collection of a control group necessary. None of the test setups proved to be advantageous or inferior in terms of difference in mean tibial rotation. Additionally, we provide an average tibial rotation curve as a function of knee flexion angle showing the typical screw home mechanism that occurs when extending the knee. Clinically, the highly variable data should not be used as a
standard, which we believe is due to the interindividual anatomical differences of human knee joints. However, this data can be used e.g. for defining boundary conditions for finite element models.

Conflict of Interest

None.

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References


Own contribution: planning of study, performing experiments, data acquisition, data analysis, visualization and interpretation of data, creating figures, writing publication

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Do Prophylactic Knee Braces Protect the Knee Against Impacts or Tibial Moments?

An In Vitro Multisensory Study

Steffen Paul Hacker,*† MEng, Florian Schall,† MEng, Frank Niemeyer,† PhD, Nicolas Wolf,† MEng, Anita Ignatius,† DVM, and Lutz Dürselen,† PhD

Investigation performed at the Institute of Orthopaedic Research and Biomechanics, Centre for Trauma Research, Ulm University Medical Centre, Ulm, Germany

Background: Knee braces are prescribed by physicians to protect the knee from various loading conditions during sports or after surgery, even though the effect of bracing for various loading scenarios remains unclear.

Purpose: To extensively investigate whether bracing protects the knee against impacts from the lateral, medial, anterior, or posterior directions at different heights as well as against tibial moments.

Study Design: Controlled laboratory study.

Methods: Eight limb specimens were exposed to (1) subcritical impacts from the medial, lateral, anterior, and posterior directions at 3 heights (center of the joint line and 100 mm inferior and superior) and (2) internal/external torques. Using a prophylactic brace, both scenarios were conducted under braced and unbraced conditions with moderate muscle loads and intact soft tissue. The change in anterior cruciate ligament (ACL) strain, joint acceleration in the tibial and femoral bones (for impacts only), and joint kinematics were recorded and analyzed.

Results: Bracing reduced joint acceleration for medial and lateral center impacts. The ACL strain change was decreased for medial superior impacts and increased for anterior inferior impacts. Impacts from the posterior direction had substantially less effect on the ACL strain change and joint acceleration than anterior impacts. Bracing had no effect on the ACL strain change or kinematics under internal or external moments.

Conclusion: Our results indicate that the effect of bracing during impacts depends on the direction and height of the impact and is partly positive, negative, or neutral and that soft tissue absorbs impact energy. An effect during internal or external torque was not detected.

Clinical Relevance: Bracing in contact sports with many lateral or medial impacts might be beneficial, whereas athletes who play sports with rotational moments on the knee or anterior impacts may be safer without a brace.

Keywords: knee; brace; ACL strain; acceleration; kinematic analysis

During sport activities, the knee joint is highly stressed and thus at a particular risk of injuries. Sports with rapid cutting and player-to-player contact, as in American football, soccer, handball, squash, or skiing, have an especially high incidence of knee injuries.26,27 The mechanisms behind the injury have been investigated by numerous studies, such that a typical injury pattern can be identified: a knee angle close to full extension combined with a valgus and tibial rotation moment.1,23-25,30,39 During such combined loads, the tibia slides in the frontal plane and rotates internally or externally relative to the femur, which may cause an anterior cruciate ligament (ACL) rupture, one of the most frequent knee injuries during sports.20,35,38,42

To guard the knee joint against excessive loads during sports, protective knee braces (PKBs) have been developed by various manufacturers. They consist of 2 rigid frames that are strapped around the thigh and shank, respectively, and that are usually connected via a polycentric hinge. This construction is designed to absorb stress affecting the knee joint while allowing the athlete full range of motion during sports. A study by Boden et al9 reported that 28% of ACL injuries are induced by
player-to-player contact, whereas 72% occurred without such interaction. Therefore, it is unsurprising that athletes wear PKBs in various sports, particularly contact sports such as American football.\textsuperscript{15,34}

Numerous in vitro and survey studies on the effectiveness of (prophylactic) bracing have been performed. In vitro studies using human specimens by Paulos et al\textsuperscript{32} and Erickson et al\textsuperscript{10} demonstrated no protective effect of PKBs against lateral impacts. A study by Baker et al\textsuperscript{2} showed little to no protective effect for the medial collateral ligament (MCL). By contrast, France and Paulos,\textsuperscript{13} Hangalur et al,\textsuperscript{17} and Paulos et al\textsuperscript{34} found a protective effect for the ACL or MCL using cadaveric specimens or a mechanical surrogate limb. In several surveys and prospective studies, the long-term effect of bracing has been investigated in sports and for postoperative use after surgery. While postoperative studies showed explicitly that bracing has no positive effect in the long term after knee surgery,\textsuperscript{9,11,18,19,22,28,29} survey studies about a potentially protective effect have strongly debated the issue.\textsuperscript{33} A retrospective study of American football players could not support the thesis that prophylactic braces prevent injuries.\textsuperscript{20} Indeed, bracing can even lead to an increasing incidence of ACL injuries or ankle and foot injuries in football teams.\textsuperscript{15,36,44} A survey among elite Swedish ice hockey players also questioned the protective effect of the PKB.\textsuperscript{35} By contrast, a survey study among off-road motorcycle riders by Sanders et al\textsuperscript{37} and a study among skiers by Sterett et al\textsuperscript{38} found fewer injuries among the study participants wearing a PKB.

In an in vivo study, Beynon et al\textsuperscript{8} implanted a Hall effect strain transducer in 13 participants and applied an anterior shear force to the knee joint. They found a stress-shielding effect of the PKB for low forces (<100 N) and a 5-N m internal moment.\textsuperscript{6} Another in vivo study used video fluoroscopy to investigate the effect of bracing and found no difference between the braced and unbraced conditions in tibial translation.\textsuperscript{41} Bing et al\textsuperscript{39} found a decreased knee flexion angle in the braced knee for a stop-jump task and concluded that knee braces might prevent ACL injuries.

With contact, in sports such as American football, soccer, and ice hockey, impacts to the lower leg due to tackles are the main risk factors for injuries.\textsuperscript{14} Therefore, the effect of impacts on the knee joint and possible protective effects of PKBs have been investigated by Baker et al\textsuperscript{2} and Erickson et al.\textsuperscript{10} Unfortunately, these studies only simulated lateral impacts at the height of the brace’s hinge and their influence on ACL and/or MCL stress. Giza et al\textsuperscript{34} reported that the force on the leg due to tackles does not only occur from the lateral direction but may also happen from the medial, anterior, or posterior directions.

To summarize, experimental studies and systematic reviews about the protective effect of bracing are controversial and incomplete.\textsuperscript{28,33} The question remains of whether a PKB has a protective effect against impacts from other directions than just the lateral that occur during contact sports. With the results of Paulos et al\textsuperscript{32} and Hangalur et al\textsuperscript{17} in mind, we hypothesized that bracing might be protective against any impacts from all sides.

METHODS

The ACL strain change and knee joint acceleration were obtained in 8 leg specimens (2 male, 6 female; mean age, 70 ± 12 years; Science Care) during lateral, medial, anterior, and posterior impacts. Based on the results of a similar study,\textsuperscript{1} power analysis was performed with the type I error set to 5% and 80% power to calculate a sample size of 6, which was increased to 8 specimens because the standard deviation in this study might be greater. This study was approved by a local ethics committee.

Pilot Study and Preparation

In total, 10 human fresh-frozen specimens were used in this study. One specimen was used to test various implementation techniques for the sensors and to evaluate mounting in the testing rig as well as positioning of the impact cylinder. A second cadaveric knee was used in a preliminary test in which we examined the knee with impacts from medial, lateral, posterior, and anterior at 3 heights with knee flexion angles of 30° and 60° to determine the influence of the knee flexion angle on the test results. Because there were no significant differences in the ACL strain change or absorbed energy between 30° and 60° of knee flexion, we decided to perform the impact tests only with an angle of 30° to reduce the number of impacts and prevent the knee from degeneration.

The 8 remaining specimens were used for the main study. After thawing overnight, two 3-axis acceleration sensors (MPU9250; InvenSense) were implanted into the femur and tibia to determine acceleration within the joint (Figure 1). Therefore, the skin was opened from the posterior, soft tissue was carefully diluted, and 2 boreholes with a diameter of 20 mm and a depth of 25 mm were drilled 30 mm above and below the intercondylar notch. The acceleration sensors were protected by custom-made, single-use casings made of polyoxymethylene that were cemented into the holes using plaster. The casings were designed to allow easy removal and reuse of the sensors after the tests. Care was taken to orient the sensors parallel to the axis of the joint coordinate system defined by Grood and Suntay\textsuperscript{16} and

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Ethical approval for this study was obtained from Ulm University (No. 207/16).
at a reproducible distance of 60 mm from each other using a custom-made implanting tool.

To achieve moderate stabilization in the knee joint, muscle flexors and extensors were simulated using 2 Bowden cables. One was pulled through the anterior and 1 pulled through the posterior thigh’s soft tissue. To simulate the hamstring muscle, 1 steel wire was pulled through the corresponding Bowden cable and anchored approximately 10 mm below the tibial acceleration sensor with a wall anchor (Duopower 4 × 35 mm; Fischer). To simulate the quadriceps muscle, another cable was pulled through the anterior Bowden cable and crimped to a perforated metal plate. The quadriceps tendon was split longitudinally; the perforated plate was placed between and sewn into place using simple interrupted stitches that were threaded through the holes of the plate and the tendon (Figure 1).

To determine the ACL strain change, a differential variable reluctance transducer (DVRT; M-DVRT-9; LORD MicroStrain) was used. Notchplasty was performed to implant the DVRT, while care was taken not to damage the ACL. The DVRT was pinned at the anteromedial bundle of the ACL and secured with cross-stitches to prevent DVRT slippage during impacts. During the insertion of the DVRT, the knee was held at 30° of flexion, and the ACL was carefully palpated to ensure that the DVRT was pinned in a prestrained section of the anteromedial bundle. After implanting the sensors and wire cables, the soft tissue was carefully returned into place, and the skin was closed. The femur and tibia were transected with a saw at a distance of approximately 450 mm from the joint space. To allow for potting, a total of 100 mm of soft tissue at the distal and proximal ends was removed, the skin was tied together, and the bones were embedded in a polymer based on methyl methacrylate (Technovit 3040; Kulzer). Two Schanz screws (Orthofix SRL) were drilled into the femur and tibia, respectively, to mount a coordinate system with retroreflective markers for the kinematic 3-dimensional (3D) analysis. The coordinate system was aligned parallel with the anatomic axis of the tibia and femur, as defined by Grood and Suntay,16 using a lockable ball joint.

Test Setup and Procedure

The limb was mounted in a knee testing rig at 30° of flexion, and moderate muscle forces (quadriceps: 150 N; hamstring: 100 N) were applied (Figure 2). This is equivalent to a ratio of 0.66, which was reported by Besier et al2 to be the approximate co-contraction index. The proximal femur was mounted to a Cardan joint (universal joint), allowing flexion/extension and adduction/abduction and thus simulating the hip joint. The distal tibia was mounted to another Cardan joint, allowing flexion/extension, adduction/abduction, and internal/external tibial rotation and thus simulating the ankle. A pneumatic actuator (DNCI-63-300-PA; Festo) was used to accelerate a 2-kg weight to a velocity of 1 m/s, creating a reproducible and nondestructive energy of 1 J at the point of impact. For this, a catapult-like test setup was developed: A lever arm was linearly accelerated by a pneumatic actuator and stopped 10 mm before the skin surface. Therefore, the weight, which was mounted on the lever arm with a linear bearing, shot onto the knee joint, creating the impact (Figure 3). The weight was shot onto the limb from 4 directions (anterior, posterior, medial, and lateral) and at 3 heights (center of the joint line and 100 mm below and above it). Each experiment was conducted in a braced and unbraced condition, resulting in

Figure 1. (A) Radiograph of the implanted sensors and muscle force application. (B) Schematic images of the implanted sensors. Two Bowden cables (1) were pulled through the soft tissue, with one of the cables anchored in the posterior tibia to simulate the hamstring muscle (6) and the other crimped to a perforated plate (2) and sewn to the quadriceps tendon to simulate the quadriceps. A borehole was drilled in both the posterior femoral and the tibial bones, and an acceleration sensor was cemented into each hole (3 and 5). Notchplasty was performed, and the differential variable reluctance transducer strain sensor was pinned in the anteromedial bundle of the anterior cruciate ligament (4).
a total of \((4 \times 3 \times 2) 24\) impacts. After each impact, it was ensured that the soft tissue was not compromised.

Three braces (4titude; DJO Global), sized small, medium, and large, were available for this study. The appropriate size of the brace was determined for each knee separately, and the corresponding brace was then fastened as described in the manual, while care was taken to fasten the Velcro strips consistently. For each impact, the femoral and tibial acceleration sensor signals were recorded as multiples of gravity \((g)\) in the \(x, y,\) and \(z\) directions with 250 Hz and the ACL strain change with the DVRT with 1000 Hz. The movement of the markers on the coordinate systems attached to the femur and tibia was obtained with 240 Hz using nine 3D cameras (Prime 13; NaturalPoint).

After completion of the impact experiments, an internal and subsequently external tibial rotation moment of 5 N\(\cdot\)m was applied at 30° of flexion in the braced and unbraced conditions. Finally, the knees were moved to 60° of flexion, and the tests were repeated, leading to a total of 8 experiments. During these 8 tests, the ACL strain change and knee kinematics were recorded.

Data and Statistical Analyses

All data and statistical analyses were performed in Matlab R2017a (MathWorks). The data of both acceleration sensors required offset compensation because of gravity. The resulting acceleration \(a_{\text{ros}}\) of all 3 directions \(a_x, a_y,\) and \(a_z\) was computed with \(a_{\text{ros}} = \sqrt{a_x^2 + a_y^2 + a_z^2} .\) The maximal acceleration induced by the impact was used for statistical analysis. For plausibility analysis, the acceleration vector was computed for every time step and visualized in a 3D plot (Figure 4B).

Figure 3. Sequence of a measurement: (A) Scheme of the test setup for a medial impact at a height of 100 mm above the center of the joint line. (B-D) High-speed recording (240 fps) of the impact. It can be seen in D how the weight is mounted with low friction on the lever arm as the position relative to the lever arm in B has changed.
The DVRT sensor was calibrated using a materials testing machine (Zwick). The resulting polynomial calibration curve was used to compute the ACL length between the 2 insertion points of the DVRT. The change in ACL strain was computed for the braced and unbraced knees, 

\[ \frac{l_{\text{impact}} - l_0}{l_0} \times 100 \]

with \( l_{\text{impact}} \) being the length of the DVRT during the impact in the braced and unbraced conditions, respectively, and \( l_0 \) being the DVRT length of the unloaded and unbraced knee. In both conditions, the strain change was computed with reference to the length of the DVRT in the unbraced condition and at 30° of flexion, similar to studies by Yasuda et al.\(^{45} \) and Erickson et al.\(^{10} \) Thus, a positive strain change is equal to elongation of the ACL, and a negative value represents relaxation of the ACL relative to the unloaded and unbraced knee.

With the markers on the coordinate systems that were aligned with the knee axes, the angles were computed as described by Grood and Suntay.\(^{16} \) For statistical analysis, the maximal angle change was used. A data set example of a lateral impact at the height of the center of the joint line is displayed in Figure 4.

Statistical analysis was performed in Matlab with the built-in statistical toolbox. The data were pairwise analyzed with a nonparametric Mann-Whitney U test. \( P < .05 \) was considered statistically significant. The sample size was \( n = 8 \) for every tested combination. All results are presented as the median with interquartile range.

RESULTS

Each of the 8 specimens was tested in all 32 loading scenarios, including 24 impacts and 8 rotational moments. The data were grouped by loading scenario and are displayed in Figures 5, 6, and 7. All data are shown as the median and interquartile range.

Impacts

Acceleration recorded within the femoral and tibial bones indicates how much energy is transferred directly to the joint or otherwise is absorbed by soft tissue. This energy leads to a horizontal movement of the joint, impeded by ligaments and tendons stressed at the moment of impact. Therefore, a high acceleration indicates a high load on the knee and its structures. A 3D analysis of acceleration demonstrated that the main acceleration always occurred in the direction of the impact.

Lateral impacts induced a maximal acceleration within the femoral and tibial bones of 7.0g and 5.5g, respectively. It was significantly reduced by bracing to 3.1g and 3.2g, respectively (\( P = .002 \) and \( P < .001 \), respectively). While bracing also significantly reduced acceleration within the bone for inferior impacts from 2.1g to 1.4g (femur) and from 2.7g to 1.7g (tibia) (\( P = .038 \) and \( P = .021 \), respectively), tibial acceleration induced by an impact superior to the tibia was significantly increased from 1.1g to 1.8g (\( P = .015 \)). ACL strain during lateral impacts was not significantly reduced for the braced condition. However, there was a tendency for bracing to reduce ACL strain for

![Figure 4. Example of an impact from the lateral direction at the height of the center of the joint line and in the braced condition. The dashed line marks the time of impact. (A) Raw data of the femoral and tibial acceleration sensors. (B) The 3-dimensional plot shows that maximal acceleration occurs in the direction of the impact. (C) The anterior cruciate ligament in the braced condition is more relaxed than in the unbraced condition, as indicated by an offset of approximately 0.3%. (D) Only adduction is affected by a lateral impact, whereas flexion and internal rotation are unchanged.](image-url)
superior and inferior impacts. Kinematic analysis demonstrated no significant differences.

Medial superior impacts caused significantly increased femoral and tibial acceleration in the braced condition from 0.9\textsuperscript{g} to 1.5\textsuperscript{g} and from 0.8\textsuperscript{g} to 1.4\textsuperscript{g}, respectively (\(P = .049\) and \(P = .021\), respectively). Acceleration induced by medial impacts on the center of the joint line was significantly reduced from 4.1\textsuperscript{g} to 1.9\textsuperscript{g} (\(P = .015\)) for the femoral sensor and from 4.6\textsuperscript{g} to 1.9\textsuperscript{g} (\(P = .007\)) for the tibial sensor. An effect for inferior medial impacts could be detected. The ACL strain change was significantly reduced during superior impacts, from +0.07\% in the unbraced condition to −0.82\% with the brace (\(P = .019\)). The impacts caused adduction of between 1.6\(^\circ\) and 2.5\(^\circ\). However, a significant effect of bracing on knee angles could not be detected.

The effect of anterior impacts on joint acceleration was not significantly different in the braced or unbraced condition, with 1 exception: The femoral sensor recorded significantly decreased acceleration for inferior impacts from 2.6\textsuperscript{g} for the unbraced knee to 1.3\textsuperscript{g} for the braced knee (\(P = .029\)). There was a consistent trend toward a greater ACL strain change for all impacts in the braced condition (all \(P < .17\)). This effect was even statistically significant for inferior impacts, with an ACL strain change from −1.5\% (unbraced) to 1.2\% (braced) (\(P = .028\)). The kinematics were unaffected by anterior impacts except for an inferior impact, which resulted in internal rotation of 1.6\(^\circ\) and 1.5\(^\circ\) in the unbraced and braced conditions, respectively.

Posterior impacts induced only small and nonsignificant acceleration at all 3 heights of impact (0.1\textsuperscript{g}-1.1\textsuperscript{g}). Posterior inferior impacts had the opposite effect on ACL strain compared with anterior inferior impacts. Additionally, the effect of bracing for posterior inferior impacts approached statistical significance (\(P = .059\)): ACL strain was 0.32\% for both the femoral and tibial sensors.
without a brace and \(-0.11\%\) with a brace. The posterior drawer mechanism consequently led to external rotation of \(1.6^\circ\) and \(1.3^\circ\) in the unbraced and braced conditions, respectively.

**Tibial Moments**

There was no significant effect of bracing on the ACL strain change or kinematics for internal or external moments because of considerable variation of the data. Even so, Figure 7 illustrates that the flexion angle of the knee influences the effect of internal and external moments on the ACL strain change: for example, \(-1.6\%\) versus \(-0.3\%\) for the braced knee with external moments at \(30^\circ\) of flexion. Additionally, the ACL was relaxed under external moments and stressed under internal moments. However, the magnitude of tibial rotation under internal/external moments was influenced by neither the brace nor the flexion angle.

**DISCUSSION**

Prophylactic knee braces are designed to protect the joint during contact and noncontact sports in which rapid cutting maneuvers and player-to-player contact lead to various loading scenarios affecting the knee. To answer the question of whether knee braces can protect the knee during these inconsistent conditions, we extensively tested the effect of lateral, medial, anterior, and posterior impacts at the height of the center of the joint line and 100 mm inferior and superior, as well as under internal and external moments. As outcome parameters, the ACL strain change and joint acceleration were recorded. The use of acceleration sensors within the bone is novel in this context and extends the data analysis by a new outcome parameter compared with similar studies. The acceleration sensors provided reproducible and reliable results, whereas the observed effects on ACL strain were rather limited. This can be explained by the subcritical impact...
energy of 1 J and the stabilizing effect of muscle forces, both of which caused strain in the ACL to change only slightly. Even so, our results indicate trends and demonstrate that there is no simple answer to the research question of whether prophylactic braces actually do protect the knee during sports. Thus, our hypothesis that a brace protects the knee against impacts from any direction has to be partly accepted.

The PKB tends to reduce acceleration within the femoral and tibial bones. Therefore, the structures of the knee are less stressed as the magnitude of horizontal tibial movement relative to the femur is reduced. Only for superior impacts is acceleration greater in the braced condition for lateral and medial impacts, which might be explained by a shock-transmitting effect of the brace to the center of the knee.

Our results indicate that for lateral and medial impacts, bracing tends to reduce ACL strain. These findings are in agreement with a study by Paulos et al, who found a mean reduction of the ACL peak load of 38.9% for lateral impacts in the braced condition. In their study, an overcritical impact with 60 J (270-lb impact mass with 2.18 mph) was induced at the center of the joint line of a surrogate limb model. Differences in the models (cadaveric limb vs mechanical surrogate) and testing conditions make comparisons difficult between the Paulos et al study and ours. Another study, by Erickson et al, found a reduction of impact force for braced knees during lateral impacts but no significant protective effect for ACL strain.

Our finding that bracing increases ACL strain for anterior impacts is contrary to an in vivo study by Fleming et al, who found a significant reduction of ACL strain for anterior shear loads in patients undergoing arthroscopic surgery under nonweightbearing and weightbearing conditions. The larger ACL strain change in the braced condition might be explained by the mechanical coupling effect of the brace: In the unbraced condition, an anterior impact leads to posterior translation of the tibia relative to the femur. In the braced condition, mechanical coupling between the thigh and shank possibly reduced or even prevented the posterior tibial shift. Additionally, as the kinematic chain was closed in our test setup, the anterior impact induced an extension torque on the knee, possibly having the same effect on the ACL as a knee extension movement, which has been shown to increase ACL strain.

When comparing the acceleration of anterior and posterior impacts, it is noticeable that the posterior impacts had only a limited effect on the ACL as well as on acceleration. This can be explained by the soft tissue (hamstring, gastrocnemius, adipose tissue) that apparently absorbs most of the impact energy, whereas an inferior impact from the anterior more or less directly hits the bone.

Kinematic analysis showed no effect of bracing on kinematics for subcritical impacts for all 12 combinations of impact directions and heights. These findings extend those of Baker et al, who reported no change in external tibial rotation of the braced or unbraced knee at 20° of flexion for lateral impacts.

In the present study, bracing slightly and nonsignificantly increased the median strain change for 5-N-m internal moments and did not significantly decrease the strain change for external moments. Therefore, bracing had no
significant effect on ACL strain in the knee with moderate muscle forces applied for internal and external torques of 5 N.m. These findings are in agreement with the above-mentioned study by Fleming et al., who also found no significant reduction with bracing in the weightbearing knee. Only in the nonweightbearing knee with applied internal torque did they find a significant reduction of ACL strain when braced. Therefore, compressing the knee, by applying bodyweight and/or muscle forces, might have the same protective effect on the ACL against internal/external torques as bracing. Furthermore, our result that the amplitude of tibial rotation under internal and external torques is unchanged for the unbraced and braced conditions underlines the nonprotective effect of bracing on ACL strain versus subcritical torques. It can be assumed that soft tissue displacement counters the stabilizing effect of a brace.

As with similar in vitro studies, this study has some limitations. Although we carefully implanted the DVRT sensor and secured it with additional cross-stitches to prevent sensor slippage, the large interquartile range of the results of some conditions is conspicuous. One explanation is probably the complex inhomogeneous strain pattern of the filaments of the anteromedial bundle that makes it difficult to place the strain sensor in exactly the same segment of the bundle in each specimen.

The use of only 1 PKB type by 1 manufacturer is another limitation of this study. However, the overall design of the PKBs with regard to strapping, the hinge mechanism, and padding is similar throughout various manufacturers, so this influence can be assumed to be rather limited. Additionally, even though we used braces in 3 sizes (small, medium, or large) to ensure the best fit of the brace, small variations in strap tensioning and application of the brace to a cadaveric limb with dead muscles might have led to a higher variance in the data. Additionally, it has been documented that women are more prone to ACL tears during sports than men. As there were more female than male specimens in this study (ratio, 3:1), the actual effect of impacts might be smaller for male athletes.

CONCLUSION

The effect of bracing depends on the direction and height of the impact and is partly positive, negative, or neutral. Bracing might be beneficial in contact sports with many lateral or medial impacts, whereas sports with rotational moments on the knee or anterior impacts might be safer without a brace. With this being said, simply to protect from impacts, knee pads or padding that helps absorb impact energy may be more beneficial.

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Own contribution: planning of study, performing experiments, data acquisition, data analysis, visualization and interpretation of data, creating figures, writing publication

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The effect of knee brace misalignment on the anterior cruciate ligament: An experimental study

Steffen P Hacker, Florian Schall, Anita Ignatius and Lutz Dürselen

Abstract
Background: Protective knee braces are used for rehabilitation or prevention. Due to poor patient compliance or slippage, the brace might be misaligned with the knee axis.
Objectives: Does a misaligned knee brace stress the anterior cruciate ligament?
Study design: It is an experimental study.
Methods: A strain sensor was implanted on the anterior cruciate ligament in eight limbs. The limbs were mounted in a knee simulator, muscle forces were applied and a cyclic motion from 10° to 60° flexion was performed under three conditions: unbraced, braced and with a misaligned brace.
Outcome measures: The outcome measures were anterior cruciate ligament strain and three-dimensional kinematics of the knee joint.
Results: The correctly aligned brace significantly reduced the anterior cruciate ligament strain at 10° compared to the unbraced condition from 0% to −1.54% (standard deviation = 1.4). The misaligned brace neutralised the effect of bracing to −0.06% (standard deviation = 1.1) anterior cruciate ligament strain. At 60° flexion angle, bracing had no statistically significant effect on the anterior cruciate ligament strain compared to the unbraced knee: −2.58% (standard deviation = 0.8) versus −1.64% (standard deviation = 1.0). The anterior cruciate ligament in the misaligned braced knee at 60° flexion with a strain of −1.1% (standard deviation = 0.9) was significantly more stressed than in the correctly aligned condition. An effect of bracing on knee kinematics was not detected.
Conclusion: A correctly aligned knee brace reduced anterior cruciate ligament strain. By contrast, a misaligned brace tended to increase the anterior cruciate ligament strain compared to the unbraced knee.

Clinical relevance
The correct alignment of the brace was identified as a key factor decisively influencing the effectiveness of bracing.

Keywords
Knee, brace, anterior cruciate ligament, misalignment

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Background
Knee braces are frequently prescribed by physicians to prevent the knee joint from damage after surgery. This is particularly true following reconstruction of the anterior cruciate ligament (ACL), although the long-term effectiveness of bracing after ACL surgery could not be demonstrated in various studies.1–5 In addition, athletes participating in contact sports or sports with fast cutting may wear prophylactic knee braces to protect against ACL ruptures, one of the most prominent injuries in athletes.6–8 Studies about a protective effect of these braces against unphysiological loadings in sports are controversial.9–14 However, in vitro studies did demonstrate that bracing could protect the ACL and medial collateral ligament against lateral impacts.15,16 Furthermore, a study by Tomescu et al.17 supported the use of a brace with a dynamic tensioning system for ACL-deficient knees.

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Knee kinematics is complex, with six degrees of freedom, including flexion/extension, adduction/abduction and internal/external rotation. After flexion and extension, tibial rotation is the most prominent rotation. This rotation is coupled to the flexion angle, such that the knee performs an external rotation of approximately 10° when it is extended from 30° to 0° flexion angle. As a result of this overlapping motion of flexion/extension and internal/external tibial rotation, the physiological finite rotation axis is skewed over a rotation of 30° to 0° flexion angle.

In contrast to the knee, brace kinematics is less complex. In general, braces consist of two rigid frames that are connected via a polycentric hinge and strapped around the thigh and shank, respectively. Because of the polycentric hinge, the tibial frame performs a translational movement relative to the femoral frame during flexion. In a pretest, we assessed the rotation axis of knee braces and found that the finite rotation axis moved translationally within the sagittal plane and did not tilt in contrast to the finite rotation axis of the physiological knee joint. This result could be expected, because of the simple brace-joint mechanics, which are equal on both sides. Therefore, it cannot mirror the tibial rotation of the knee, typically occurring between 30° and extension, so that the brace is misaligned during this specific motion (Figure 1).

Another effect that causes misalignment and leads to discomfort issues is brace slippage. A study by Brownstein found a maximal migration of 11 mm after only 15 min of exercises. The misalignment of the rotation axes of knee and brace caused by the different kinematics and through brace slippage might influence structures within the knee. This research question was investigated by Regalbuto et al., where the misalignment of brace and knee joint induced mechanical forces in the frame of the brace. The authors concluded that this could lead to adverse internal joint mechanics. Another study by Singer and Lamontagne found no effect of brace migration on the kinematics or net joint moments.

However, the effect of brace misalignment on the inner knee structures, particularly the ACL, has not yet been specifically investigated. With the different kinematics of the physiological knee and brace in mind, we performed this study to answer the question if a misaligned knee brace stresses the ACL.

**Materials and methods**

ACL strain and knee joint kinematics were obtained in eight leg specimens (2 males, 6 females; age 70 ± 12 years; Science Care, Phoenix, AZ, USA) during five consecutive squats simulated in a knee simulator. Based on the results of another study investigating the ACL strain in the braced and unbraced knee, an a priori power analysis was performed with type 2 error set to 5% and 80% power to calculate a sample size of seven, which was increased to eight specimens. This research was approved by the Ethics Committee at Ulm University (No. 207/16).

**Preparation**

Eight fresh frozen specimens were used for this study. Specimens were thawed overnight at room temperature. To measure ACL strain, a differential variable reluctance transducer (DVRT; Microstrain, Williston, VT, USA) was used. The knee capsule was opened, and muscles and fat tissue were carefully dilated. A notch plastic surgery was performed to avoid impingement of the DVRT with the intercondylar roof. The DVRT was then pinned on the anteromedial bundle of the ACL and secured with cross-stitches to avoid sensor slippage. To simulate muscle forces, two Bowden cables were pierced through the posterior and anterior soft tissue of the thigh, respectively. A steel cable was pulled through the anterior Bowden cable and crimped to a perforated metal plate. This plate was then sutured to the patella tendon to simulate the quadriceps muscles. Another cable was pulled through the posterior Bowden cable and anchored in the tibia 30 mm below the joint space to simulate hamstring muscles. During the entire preparation, the soft tissue was maintained intact as far as possible. Following placement of the sensors, soft tissue and skin were carefully sutured at the position of the insertion.

To assess knee kinematics, two Schanz screws were drilled one each into the tibial and femoral bone, respectively, and a lockable ball joint was mounted onto each pair of screws. A coordinate system with reflective markers was

![Figure 1. Exemplary scheme of the rotation axes of the knee joint (left) and a functional brace (right). The physiological rotation axes are slightly tilted in the vertical plane between 0° and 30° flexion angle, whereas the rotation axes of the brace are parallel throughout the entire flexion.](image-url)
attached to the ball joint and aligned with the anatomical joint axes defined by Grood and Suntay.26 Finally, both the tibia and femur were dissected at a distance of approximately 500 mm from the joint space. Both bones were cleared of soft tissue at the dissection cut, skin was laced up and the bare tibial and femoral bones were both cemented in polymethyl methacrylate (Technovit®; Kulzer, Wehrheim, Germany) (Figure 2).

**Test setup**

The prepared limbs were mounted in a knee simulator. Moderate muscle forces were applied using the previously implanted steel cables with a force of 100 N simulating hamstring muscles and a force of 150 N simulating quadriceps muscles. The mechanical hip joint of the knee simulator consists of a Cardan angle that allows flexion/extension and abduction/adduction. The ankle is imitated by another Cardan ankle with all three rotational degrees of freedom.

The knee was flexed from 10° to 60° and back to 10° flexion angle, simulating the range of motion during normal walking,27 for five cycles in unbraced, braced and braced but 20 mm distal migrated conditions with a speed of 20°/s (Figure 3). The brace (4TTITUDE; DJO Global, Inc., Vista, CA, USA) was strapped according to the manufacturer’s instructions.

**Data analysis**

ACL strain was recorded at 1000 Hz using custom-developed software in LabVIEW (National Instruments; Austin, TX, USA). The kinematics was recorded using 9 three-dimensional cameras (Prime 13, Optitrack; NaturalPoint Inc., Corvallis, OR, USA) at 240 Hz with an error of <0.3 mm. The kinematics was computed as described by Grood and Suntay,26 with a custom script in MATLAB 2017a (The MathWorks, Natick, USA).

The DVRT sensor continuously recorded the ACL length during five cycles, with only the final cycle used for data analysis. The ACL strain $\varepsilon$ was computed as $\varepsilon = \frac{(l_i - l_0)}{l_0} \times 100$, with $l_i$ the length of the DVRT at every time frame $i$ and $l_0$ the length of the DVRT in the unbraced condition at 10° flexion. Therefore, the strain was computed with reference to the unbraced knee for all three conditions.

Statistical analysis was performed in MATLAB with the built-in statistical toolbox. The data were analysed according to Beynnon et al.25,28 with a Student’s paired $t$-test and a level of significance of alpha = 0.05.

After the first two specimens were tested, we analysed the data and found that the pins of the DVRT that were pierced through the ACL were too long, causing an impingement of the pins with the tibia plateau. This influenced the DVRT accuracy and was consequently corrected for the following tests. For this reason, we decided to exclude the first two specimens from the ACL strain analysis. Therefore, all eight specimens were used for the kinematic analysis but only six for the ACL strain analysis. All data are presented as the mean and standard deviation (SD).

**Results**

**ACL strain**

The ACL strain at 10° flexion angle was significantly reduced by the correctly aligned brace to a mean of $-1.54\%$ (SD = 1.4, 95% confidence interval (CI) = $-2.9\%$ to $-0.1\%$) (Figure 4). The misaligned brace slightly, but non-significantly, decreased the ACL strain at 10° flexion to a mean of $-0.06\%$ (SD = 1.1). The ACL in the unbraced condition at 60°, with a mean of $-1.64\%$ (SD = 1.0, 95% CI = $-2.7\%$ to $-0.6\%$), was significantly more relaxed than at 10° flexion. Bracing did not significantly reduce the ACL strain.
compared to the unbraced condition at 60°, with −2.58% (SD = 0.8). In the misaligned condition, the ACL tended to be more stressed at 60°, with an ACL strain of −1.13% (SD = 0.9), compared to the unbraced knee. This effect was not statistically significant. The ACL strain in the misaligned condition at 60° was significantly higher than in the correctly aligned condition (95% CI = −2.1% to −0.1%).

The ACL strain as a function of the flexion angle is presented in Figure 5. Bracing reduced the mean strain of the ACL during the complete cycle compared to the unbraced condition. By contrast, the misaligned brace tended to induce an increasing ACL strain and was significantly higher than the ACL strain with the correctly aligned brace from 50° to 60° flexion angle.

**Discussion**

In the present study, the ACL strain and kinematics were obtained in six and eight limb specimens, respectively, during five cyclic squats in the unbraced, braced and 20 mm distally misaligned conditions. The overall gradient of the ACL strain versus flexion curve was similar to the in vivo findings of Beynnon et al.,29 with a decreasing ACL strain from 10° to 40° flexion and an almost constant strain from 40° to 60° flexion. However, the maximal stress release of the ACL during flexion in the unbraced condition was lower in the present study than that of Beynnon et al., with 1.5% versus approximately 4%. This might be explained by the applied muscle loads, which were most likely higher in vivo compared to the loads that we were able to simulate in vitro. However, our results are similar with the in vitro study of Dürselen et al.,30 who reported a linear strain release of 2% between 10° and 40° flexion without muscle forces.
In vivo studies by Beynnon et al.\textsuperscript{25} and Fleming et al.\textsuperscript{31} investigated the effect of bracing on the ACL strain during internal and external moments on the tibia and anterior–posterior shear load in patients undergoing arthroscopic surgery. Both found that a correctly aligned brace reduced the ACL strain when no external loads were applied, which is similar to our findings. In addition, our results are in line with those of Tomescu et al.,\textsuperscript{17} who found that bracing significantly reduced the ACL strain by 83\% in a double-leg squat and 38\% in single-leg squat in vitro.

Unfortunately, knee brace misalignment occurs frequently. This results from migration of the brace during daily exercises, as found by Brownstein,\textsuperscript{22} or by poor patient compliance, such that the brace is already misaligned during placement of the brace when fastening the Velcro strips. For this reason, we investigated the effect of a brace that was 20 mm distally migrated during a normal gait cycle, simulating a worst-case scenario. Our results indicate that a misaligned brace neutralises the stress-decreasing effect on the ACL of a correctly aligned brace. Consequently, correct alignment of a knee brace was identified as a key factor that decisively influences the effectiveness of bracing.

Bracing had no effect on the kinematics of the knee. This might be explained by soft tissue movement, which compensated the effect of bracing for rotational degrees of freedom. Typically, ACL strain is induced by anterior translation or internal rotation of the tibia. Because the kinematic analysis showed no effect of the misaligned brace on internal/external rotation, we concluded that the higher ACL strain in the misaligned braced knee is most likely explained by the anteriorly directed shear force induced by the brace.

A limitation of this study are the small muscle forces compared to the physiological loading condition. However, we intentionally used this approach, so that the effects of bracing became more apparent as the muscle forces had only slightly stabilised the knee. In addition, because the change of the ACL strain during knee flexion is higher in vivo than in vitro, we also expect a larger effect of bracing in the living knee.

Another limitation is that the soft tissue in the present in vitro study was not as tensed as living tissue, even though we stretched the skin tightly to prevent wobbling. However, softer tissue can more effectively compensate misalignment than tensed tissue, and thus the negative effect of bracing might be even higher in the living.

**Conclusion**

Our results indicate that a correctly aligned knee brace does reduce ACL strain during squatting, particularly in a knee position close to extension. By contrast, a misaligned brace significantly increased the ACL strain compared to the correctly aligned brace. These findings emphasise the importance of educating patients and athletes how to place a knee brace correctly and to be aware of brace slippage. Non-compliance might result in a negative effect of bracing on the ACL.

![Figure 4. ACL strain at 10° flexion and 60° flexion for the three tested conditions: unbraced, braced and braced but 20 mm distally misaligned. The ACL strain at 10° flexion for the unbraced knee was defined as 0\% baseline. Error bars represent standard deviation. *p < 0.05.](image)

![Figure 5. ACL strain as a function of the knee flexion angle for the three tested conditions: unbraced, braced and braced but 20 mm distally misaligned. Error bars represent the standard deviation at several data points.](image)
Author contribution
SPH, FS and LD conceived and designed the experiments; SPH and FS performed the experiments; SPH, FS and LD analysed the data; and SPH, AI and LD wrote the publication.

Declaration of conflicting interests
The author(s) declared no potential conflicts of interest with respect to the research, authorship, and/or publication of this article.

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Figure 6. Kinematic analysis of the knee joint motion: flexion, adduction and external rotation for the three tested conditions. There are no statistically significant differences between the kinematics of each condition.


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Curriculum vitae

Curriculum vitae hidden for reasons of data protection
List of Publications


