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 al[8] reported that 28% of ACL injuries are induced by

The Orthopaedic Journal of Sports Medicine, 6(11), 2325967118805399
DOI: 10.1177/2325967118805399

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.[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[32] and Erickson et al[10] demonstrated no protective effect of PKBs against lateral impacts. A study by Baker et al[2] showed little to no protective effect for the medial collateral ligament (MCL). By contrast, France and Paulos,[13] Hangalur et al,[17] and Paulos et al[31] 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,[9,11,18,19,22,28,29] survey studies about a potentially protective effect have strongly debated the issue.[33] A retrospective study of American football players could not support the thesis that prophylactic braces prevent injuries.[20] Indeed, bracing can even lead to an increasing incidence of ACL injuries or ankle and foot injuries in football teams.[15,36,44] A survey among elite Swedish ice hockey players also questioned the protective effect of the PKB.[43] By contrast, a survey study among off-road motorcycle riders by Sanders et al[37] and a study among skiers by Sterett et al[41] found fewer injuries among the study participants wearing a PKB.

In an in vivo study, Beynnon et al[6] 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.[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.[21] Bing et al[7] 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.[14] Therefore, the effect of impacts on the knee joint and possible protective effects of PKBs have been investigated by Baker et al[2] and Erickson et al.[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[14] 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.[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[32] and Hangalur et al[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,[10] 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 dilated, 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[16] and

*Address correspondence to Steffen Paul Hacker, MEng, Institute of Orthopaedic Research and Biomechanics, Centre for Trauma Research, Ulm University Medical Centre, Helmholtzstraße 14, 89081 Ulm, Germany (email: med.biomechanik@uni-ulm.de) (Twitter: @OrthoLabUlm).

†Institute of Orthopaedic Research and Biomechanics, Centre for Trauma Research, Ulm University Medical Centre, Ulm, Germany.

One or more of the authors has declared the following potential conflict of interest or source of funding: This work was supported by the German Federal Ministry of Education and Research through the cooperative research project SE_BURG (grant No. 16SV6313). AOSSM checks author disclosures against the Open Payments Database (OPD). AOSSM has not conducted an independent investigation on the OPD and disclaims any liability or responsibility relating thereto.

Ethical approval for this study was obtained from Ulm University (No. 207/16).

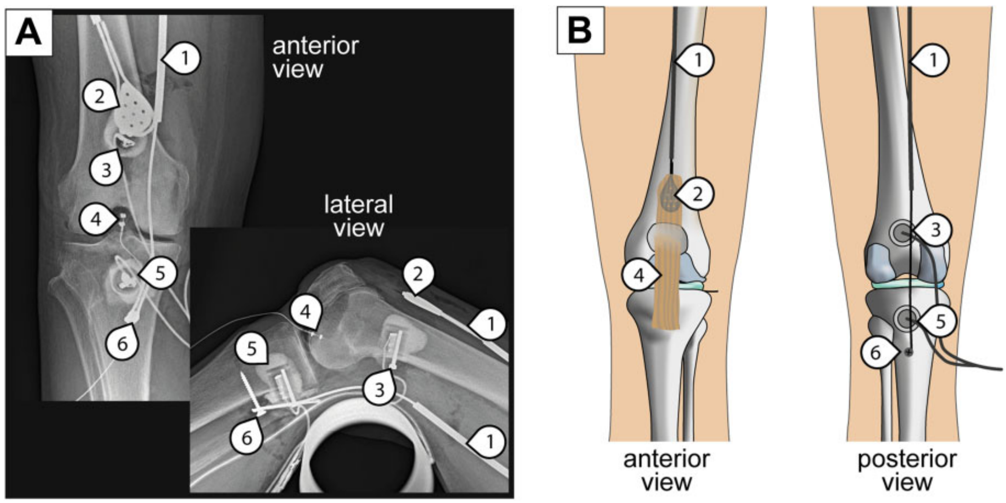

Figure 1. (A) Radiograph of the implanted sensors and muscle force application. (B) Schematic images of the implanted sensors. Two Bowden cables (1) were pulled through the soft tissue, with one of the cables anchored in the posterior tibia to simulate the hamstring muscle (6) and the other crimped to 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).

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 al[3] 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[2] 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

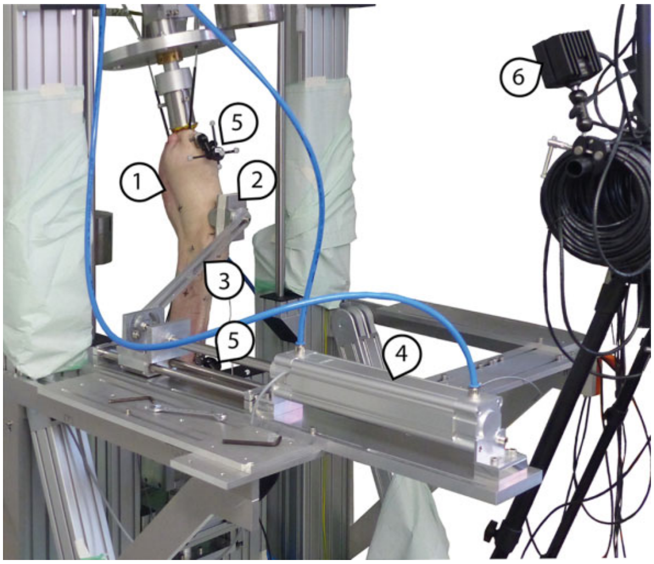

Figure 2. Test setup: The limb (1) was mounted in the testing rig, and the weight (2) on the lever arm (3) was accelerated by the pneumatic actuator (4), creating a frontal impact 100 mm above the center of the joint line. The optical markers (5) were tracked with a 3-dimensional camera system (6).

a total of (4 × 3 × 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·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_{res} of all 3 directions a_x, a_y, and a_z was computed with $a_{res} = \sqrt{a_x^2 + a_y^2 + a_z^2}$. 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).

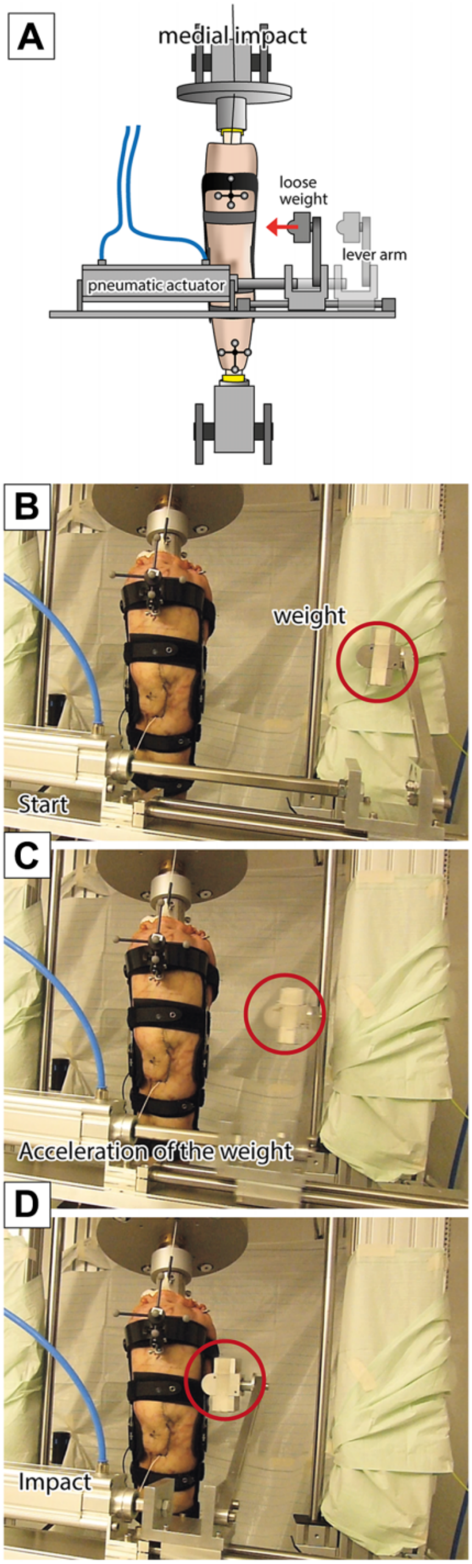

Figure 3. Sequence of a measurement: (A) Scheme of the test setup for a medial impact at a height of 100 mm above the center of the joint line. (B-D) High-speed recording (240 fps) of the impact. It can be seen in D how the weight is mounted with low friction on the lever arm as the position relative to the lever arm in B has changed.

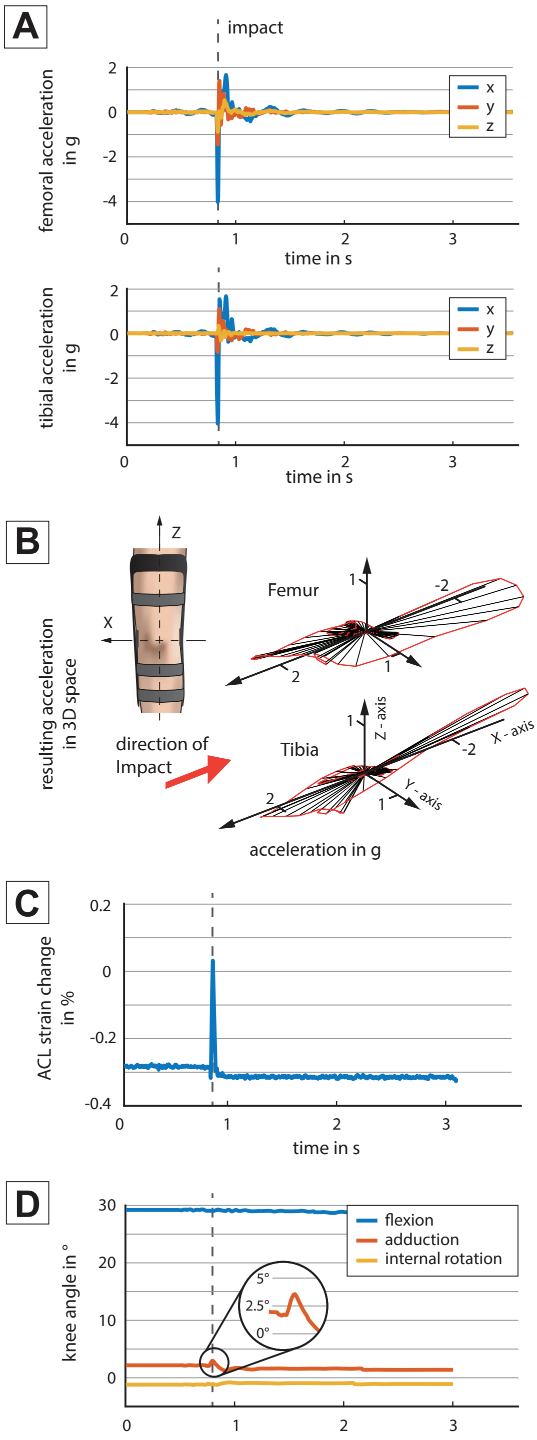

Figure 4. Example of an impact from the lateral direction at the height of the center of the joint line and in the braced condition. The dashed line marks the time of impact. (A) Raw data of the femoral and tibial acceleration sensors. (B) The 3-dimensional plot shows that maximal acceleration occurs 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.

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, $s = \frac{l_{impact} - l_0}{l_0} \times 100$, with l_{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

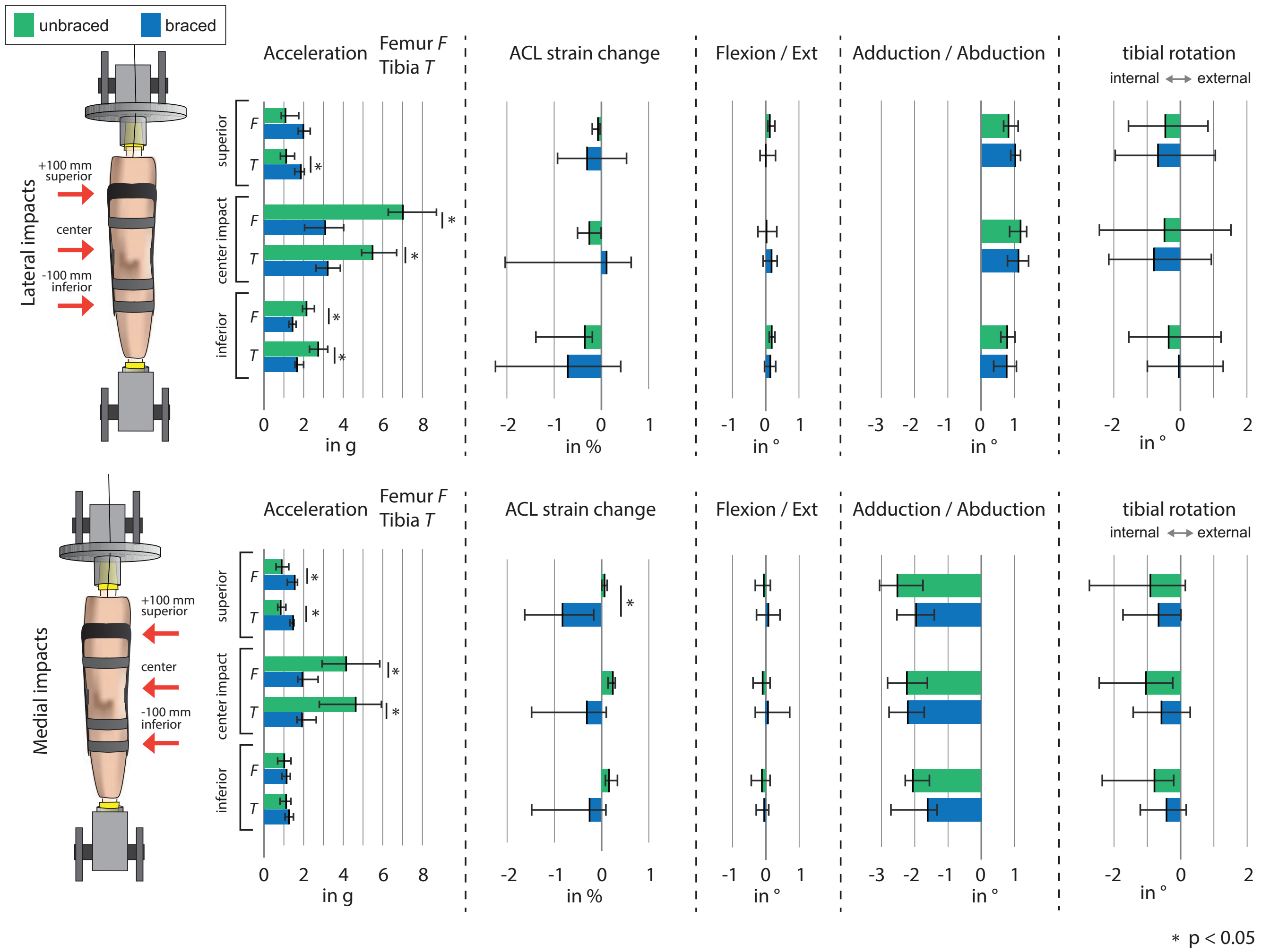

Figure 5. Results of all parameters for lateral and medial impacts of 1 J at the height of the center of the joint line and 100 mm above and below at 30° of flexion. Green indicates the unbraced condition and blue the braced condition. Acceleration within the bone was reduced in some cases, whereas the effect on the anterior cruciate ligament strain change was only significant for high medial impacts. Bracing did not influence kinematics. Data are displayed as the median and interquartile range. *$P < .05$.

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 0.9g to 1.5g and from 0.8g to 1.4g, respectively ($P = .049$ and $P = .021$, respectively). Acceleration induced by medial impacts on the center of the joint line was significantly reduced from 4.1g to 1.9g ($P = .015$) for the femoral sensor and from 4.6g to 1.9g ($P = .007$) 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° and 2.5°. 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 2.6g for the unbraced knee to 1.3g 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° and 1.5° in the unbraced and braced conditions, respectively.

Posterior impacts induced only small and nonsignificant acceleration at all 3 heights of impact (0.1g-1.1g). 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%

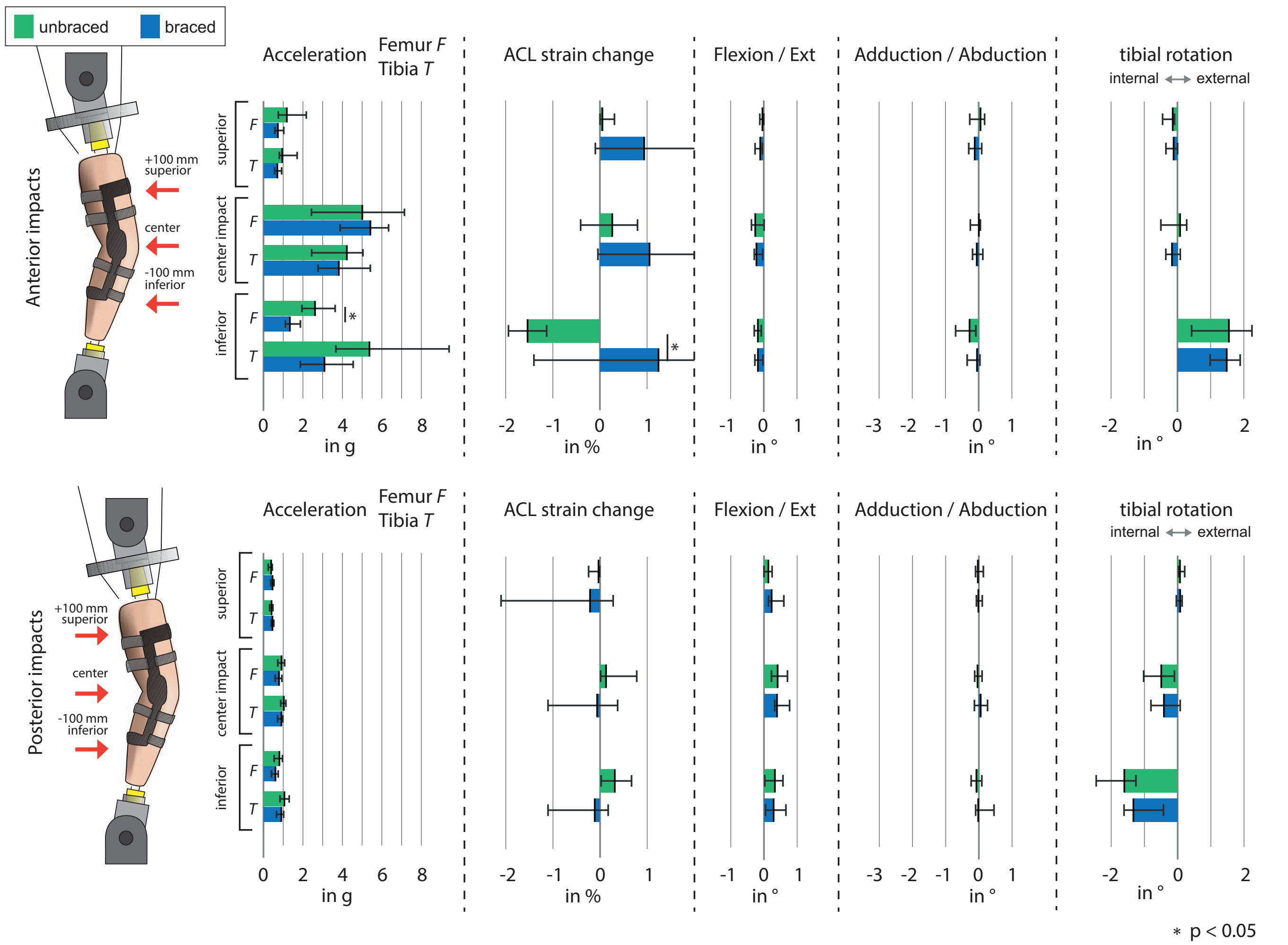

Figure 6. Results of all parameters for anterior and posterior impacts of 1 J at the height of the center of the joint line and 100 mm above and below at 30° of flexion. Green indicates the unbraced condition and blue the braced condition. Acceleration within the bone was only reduced for low anterior impacts. The anterior cruciate ligament strain change was increased for low anterior impacts on the shin. Bracing did not influence kinematics. Data are displayed as the median and interquartile range. **P* < .05.

without a brace and –0.11% with a brace. The posterior drawer mechanism consequently led to external rotation of 1.6° and 1.3° 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 in 30° or 60° 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.[2,4,6,12,31] 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

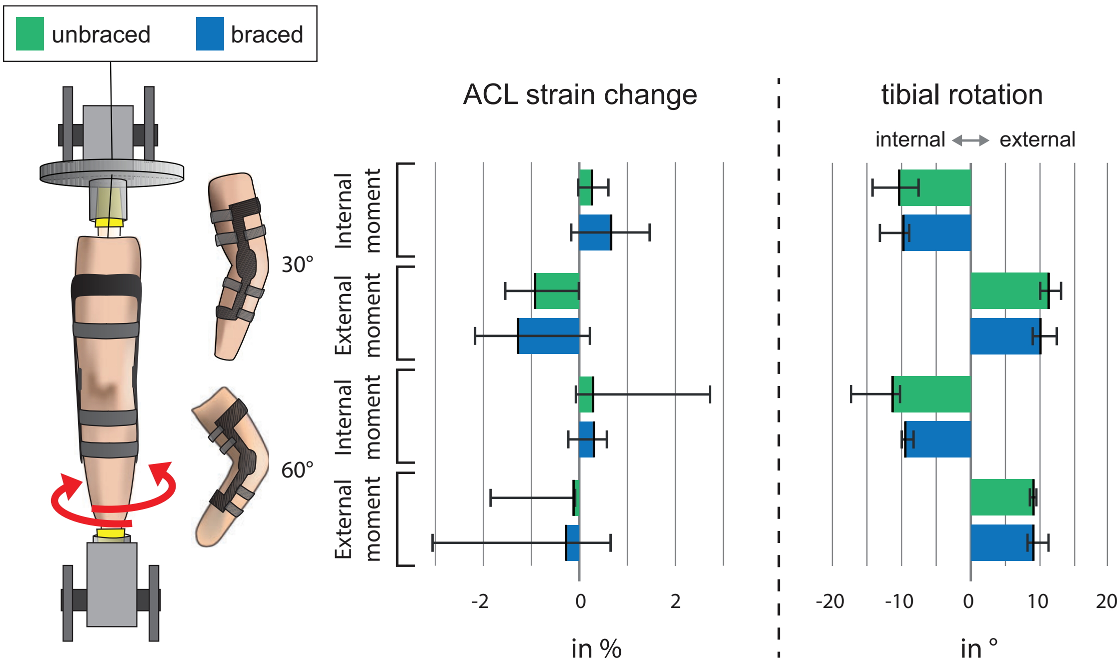

Figure 7. Results of all parameters for internal and external tibial moments of 5 N·m in 30° and 60° of flexion. Green indicates the unbraced condition and blue the braced condition. Neither the anterior cruciate ligament strain change nor kinematics was significantly altered by bracing. Data are displayed as the median and interquartile range.

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,[31] 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[31] study and ours. Another study, by Erickson et al,[10] 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,[12] 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.[5]

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,[2] 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 abovementioned study by Fleming et al,[12] 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.[40] 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.

REFERENCES

1. Arnold JA, Coker TP, Heaton LM, Park JP, Harris WD. Natural history of anterior cruciate tears. *Am J Sports Med*. 1979;7(6):305-313.
2. Baker BE, VanHanswyk E, Bogosian SP, Werner FW, Murphy D. The effect of knee braces on lateral impact loading of the knee. *Am J Sports Med*. 1989;17(2):182-186.
3. Besier TF, Lloyd DG, Achland TR. Muscle activation strategies at the knee during running and cutting maneuvers. *Med Sci Sports Exerc*. 2003;35(1):119-127.
4. Beynnon BD, Johnson RJ, Fleming BC, et al. The effect of functional knee bracing on the anterior cruciate ligament in the weightbearing and nonweightbearing knee. *Am J Sports Med*. 1997;25(3):353-359.
5. Beynnon BD, Johnson RJ, Fleming BC, Stankewich CJ, Renström PA, Nichols CE. The strain behavior of the anterior cruciate ligament during squatting and active flexion-extension: a comparison of an open and a closed kinetic chain exercise. *Am J Sports Med*. 1997;25(6): 823-829.
6. Beynnon BD, Pope MH, Wertheimer CM, et al. The effect of functional knee-braces on strain on the anterior cruciate ligament in vivo. *J Bone Joint Surg Am*. 1992;74(9):1298-1312.
7. Bing Y, Daniel H, Jennifer P, William L, Donald TK, William EG. Immediate effects of a knee brace with a constraint to knee extension on knee kinematics and ground reaction forces in a stop-jump task. *Am J Sports Med*. 2004;32(5):1136-1143.
8. Boden BP, Dean GS, Feagin JA Jr, Garrett WE Jr. Mechanisms of anterior cruciate ligament injury. *Orthopedics*. 2000;23(6):573-578.
9. Brandsson S, Faxén E, Kartus J, Eriksson BI, Karlsson J. Is a knee brace advantageous after anterior cruciate ligament surgery? *Scand J Med Sci Sports*. 2001;11(2):110-114.
10. Erickson AR, Yasuda K, Beynnon B, Johnson R, Pope M. An in vitro dynamic evaluation of prophylactic knee braces during lateral impact loading. *Am J Sports Med*. 1993;21(1):26-35.
11. Feller J, Bartlett J, Chapman S, Delahunt M. Use of an extension-assisting brace following anterior cruciate ligament reconstruction. *Knee Surg Sports Traumatol Arthrosc*. 1997;5(1):6-9.
12. Fleming BC, Renstrom PA, Beynnon BD, Engstrom B, Peura G. The influence of functional knee bracing on the anterior cruciate ligament strain biomechanics in weightbearing and nonweightbearing knees. *Am J Sports Med*. 2000;28(6):815-824.
13. France EP, Paulos LE. In vitro assessment of prophylactic knee brace function. *Clin Sports Med*. 1990;9(4):823-841.
14. Giza E, Fuller C, Junge A, Dvorak J. Mechanisms of foot and ankle injuries in soccer. *Am J Sports Med*. 2003;31(4):550-554.
15. Grace TG, Skipper BJ, Newberry JC, Nelson MA, Sweetser ER, Rothman ML. Prophylactic knee braces and injury to the lower extremity. *J Bone Joint Surg Am*. 1988;70(3):422-427.
16. Grood ES, Suntay WJ. A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. *J Biomech Eng*. 1983;105(2):136-144.
17. Hangalur G, Brenneman E, Nicholls M, Bakker R, Laing A, Chandrashekar N. Can a knee brace reduce the strain in the anterior cruciate ligament? A study using combined in vivo/in vitro method. *Prosthet Orthot Int*. 2016;40(3):394-399.
18. Harilainen A, Sandelin J. Post-operative use of knee brace in bone–tendon–bone patellar tendon anterior cruciate ligament reconstruction: 5-year follow-up results of a randomized prospective study. *Scand J Med Sci Sports*. 2006;16(1):14-18.
19. Harilainen A, Sandelin J, Vanhanen I, Kivinen A. Knee brace after bone-tendon-bone anterior cruciate ligament reconstruction: randomized, prospective study with 2-year follow-up. *Knee Surg Sports Traumatol Arthrosc*. 1997;5(1):10-13.
20. Hewson GF, Mendini RA, Wang JB. Prophylactic knee bracing in college football. *Am J Sports Med*. 1986;14(4):262-266.
21. Jalali M, Farahmand F, Mousavi SME, et al. Fluoroscopic analysis of tibial translation in anterior cruciate ligament injured knees with and without bracing during forward lunge. *Iran J Radiol*. 2015;12(3): e17832.
22. Kartus J, Stener S, Köhler K, Sernert N, Eriksson BI, Karlsson J. Is bracing after anterior cruciate ligament reconstruction necessary? A 2-year follow-up of 78 consecutive patients rehabilitated with or without a brace. *Knee Surg Sports Traumatol Arthrosc*. 1997;5(3): 157-161.
23. Kim SY, Spritzer CE, Utturkar GM, Toth AP, Garrett WE, DeFrate LE. Knee kinematics during noncontact anterior cruciate ligament injury

as determined from bone bruise location. *Am J Sports Med*. 2015;43(10):2515-2521.

24. Krosshaug T, Nakamae A, Boden BP, et al. Mechanisms of anterior cruciate ligament injury in basketball. *Am J Sports Med*. 2007;35(3):359-367.
25. Leppänen M, Pasanen K, Kujala UM, et al. Stiff landings are associated with increased ACL injury risk in young female basketball and floorball players. *Am J Sports Med*. 2016;45(2):386-393.
26. Loës M, Dahlstedt LJ, Thomée R. A 7-year study on risks and costs of knee injuries in male and female youth participants in 12 sports. *Scand J Med Sci Sports*. 2000;10(2):90-97.
27. Majewski M, Susanne H, Klaus S. Epidemiology of athletic knee injuries: a 10-year study. *Knee*. 2006;13(3):184-188.
28. Martinek V, Friederich NF. To brace or not to brace? How effective are knee braces in rehabilitation? [in German]. *Orthopade*. 1999;28(6):565-570.
29. Muellner T, Alacamlioglu Y, Nikolic A, Schabus R. No benefit of bracing on the early outcome after anterior cruciate ligament reconstruction. *Knee Surg Sports Traumatol Arthrosc*. 1998;6(2):88-92.
30. Olsen O-E, Myklebust G, Engebretsen L, Bahr R. Injury mechanisms for anterior cruciate ligament injuries in team handball. *Am J Sports Med*. 2004;32(4):1002-1012.
31. Paulos LE, Cawley PW, France EP. Impact biomechanics of lateral knee bracing. *Am J Sports Med*. 1991;19(4):337-342.
32. Paulos LE, France EP, Rosenberg TD, Jayaraman G, Abbott PJ, Jaen J. The biomechanics of lateral knee bracing, part I: response of the valgus restraints to loading. *Am J Sports Med*. 1987;15(5):419-429.
33. Pietrosimone BG, Grindstaff TL, Linens SW, Uczekaj E, Hertel J. A systematic review of prophylactic braces in the prevention of knee ligament injuries in collegiate football players. *J Athl Train*. 2008;43(4):409-415.
34. Popovich C, Deutsch J. Coaching high school American football offensive linemen to reduce injury. *Int J Phys Ed Sports Health*. 2016;3(3):472-474.
35. Rothenberg P, Grau L, Kaplan L, Baraga MG. Knee injuries in American football: an epidemiological review. *Am J Orthop (Belle Mead NJ)*. 2016;45(6):368-373.
36. Rovere GD, Haupt HA, Yates CS. Prophylactic knee bracing in college football. *Am J Sports Med*. 1987;15(2):111-116.
37. Sanders MS, Cates RA, Baker MD, Barber-Westin SD, Gladin WM, Levy MS. Knee injuries and the use of prophylactic knee bracing in off-road motorcycling. *Am J Sports Med*. 2011;39(7):1395-1400.
38. Seil R, Nührenbörger C, Lion A, Gerich T, Hoffmann A, Pape D. Knieverletzungen im handball. *Sport Orthop Sport Traumatol*. 2016;32(2):154-164.
39. Shimokochi Y, Shultz SJ. Mechanisms of noncontact anterior cruciate ligament injury. *J Athl Train*. 2008;43(4):396-408.
40. Silvers HJ, Mandelbaum BR. Prevention of anterior cruciate ligament injury in the female athlete. *Br J Sports Med*. 2007;41(suppl 1):i52-i59.
41. Sterett WI, Briggs KK, Farley T, Steadman JR. Effect of functional bracing on knee injury in skiers with anterior cruciate ligament reconstruction. *Am J Sports Med*. 2006;34(10):1581-1585.
42. Swenson DM, Collins CL, Best TM, Flanigan DC, Fields SK, Comstock RD. Epidemiology of knee injuries among US high school athletes, 2005/06–2010/11. *Med Sci Sports Exerc*. 2013;45(3):462-469.
43. Tegner Y, Lorentzon R. Evaluation of knee braces in Swedish ice hockey players. *Br J Sports Med*. 1991;25(3):159-161.
44. Teitz CC, Hermanson BK, Kronmal RA, Diehr PH. Evaluation of the use of braces to prevent injury to the knee in collegiate football players. *J Bone Joint Surg Am*. 1987;69(1):2-9.
45. Yasuda K, Erickson AR, Beynnon BD, Johnson RJ, Pope MH. Dynamic elongation behavior in the medial collateral and anterior cruciate ligaments during lateral impact loading. *J Orthop Res*. 1993;11(2):190-198.