The Effect of the Shoe-Surface Interface in the Development of Anterior Cruciate Ligament Strain

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The shoe-surface interface has been implicated as a possible risk factor for anterior cruciate ligament (ACL) injuries. The purpose of this study is to develop a biomechanical, cadaveric model to evaluate the effect of various shoe-surface interfaces on ACL strain. There will be a significant difference in ACL strain between different shoe-surface combinations when a standardized rotational moment (a simulated cutting movement) is applied to an axially loaded lower extremity. The study design was a controlled laboratory study. Eight fresh-frozen cadaveric lower extremities were thawed and the femurs were potted with the knee in 30 deg of flexion. Each specimen was placed in a custom-made testing apparatus, which allowed axial loading and tibial rotation but prevented femoral rotation. For each specimen, a 500 N axial load and a 1.5 Nm internal rotation moment were placed for four different shoe-surface combinations: group I (AstroTurf-turf shoes), group II (modern playing turf-turf shoes), group III (modern playing turf-cleats), and group IV (natural grass-cleats). Maximum strain, initial axial force and moment, and maximum axial force and moment were calculated by a strain gauge and a six component force plate. The preliminary trials confirmed a linear relationship between strain and both the moment and the axial force for our testing configuration. In the experimental trials, the average maximum strain was 3.90, 3.19, 3.14, and 2.16 for groups I–IV, respectively. Group IV had significantly less maximum strain ($p<0.05$) than each of the other groups. This model can reproducibly create a detectable strain in the anteromedial bundle of the ACL in response to a given axial load and internal rotation moment. Within the elastic range of the stress-strain curve, the natural grass and cleat combination produced less strain in the ACL than the other combinations. The favorable biomechanical properties of the cleat-grass interface may result in fewer noncontact ACL injuries.
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Keywords: noncontact ACL injuries, shoe-surface interface, ACL strain

1 Introduction

The incidence of ACL injuries was reported between 80,000–250,000 ruptures per year in the United States annually [1–4]. Most of these (58–70%) are noncontact injuries occurring in young athletes 15–25 years of age (50%) [3–7]. Risk factors include environmental, anatomical, hormonal, and neuromuscular [3,4]. Within the environmental category, the role of the shoe-surface interaction with particular athletic maneuvers has been questioned.

There is epidemiologic evidence that increased traction at the shoe-surface interface may lead to improved sports performance at the expense of an increased ACL injury risk [3,4,8–11]. In a Norwegian registry of handball players, a high level of friction correlated with an increase in the number of ACL injuries [8,12]. Powell and Schootman [13] looked at injuries in the NFL during the 1980s and concluded that there was a higher risk of ACL sprain on artificial turf but only in certain game situations such as punts and kickoffs. In a later study, Scranton et al. [14] looked at NFL game exposures and found almost five times greater incidence of ACL injury on grass versus turf. However, for practice sessions, the reverse was true. When taken in total (practice and games), an incidence density ratio was calculated that revealed a 90% increase in ACL injuries on artificial turf per 1000 athlete exposures. More recently, Parekh et al. [15] reported a trend toward more ACL injuries (per 1000 athletic exposures) on turf surfaces (risk=0.0508) versus grass surfaces (0.0404) in a cohort of professional football players. While suggesting that an artificial surface may contribute to ACL injuries the data are ultimately mixed due to confounding variables such as weather conditions, field wear, accurately representing exposures, different footwear, and having an insufficient number of injuries [16–18]. These factors have led some researchers to evaluate the problem from a biomechanics perspective.

Torg et al. [19] was among the first to investigate these issues and defined a "release coefficient" based on the peak torque that develops at the shoe-surface interface. His experimental model employed a stainless steel shaft in a prosthetic foot oriented vertically and supported by two bearing systems. A cleated football shoe was affixed to the prosthetic foot and the shaft loaded vertically with 100 lbs. The release coefficient was calculated as the moment (ft–lbs)/vertical force (lbs). This model led to a redesign of cleated (soccer and football) shoes by determining a range of safe release coefficients for specific shoe-surface combinations.

Several other studies have attempted to evaluate the interaction of athletic shoe gear and playing surfaces by measuring the torques and frictional resistances. Andreasson et al. [20] developed a similar biomechanical model to assess the torque and simulated sliding that develops between sport shoes and artificial turf. They found that the torque generated was based not only on the frictional force but also on the distribution of the cleat material particularly at the ball and heel of the shoe. Cawley et al. [21] used a different biomechanical model and found that several shoe-surface interfaces developed a significant nonlinear increase in

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frictional resistance with an increase in axial load. Recently, Livesay et al. [22] developed a testing device to evaluate five different playing surfaces [22]. The highest peak torques were developed by the grass shoe-FieldTurf™ and the turf shoe-AstroTurf™ interfaces. Furthermore, these surfaces exhibited a higher rotational stiffness (the rate at which torque develops) then the grass shoe-grass interface. To date, no standardized method of evaluating the shoe-surface interface with regard to injury has emerged, which may explain some of the conflicting results investigators have demonstrated.

In 2005, an expert panel was convened at the Hunt Valley II Meeting to discuss the prevention of noncontact ACL injuries. The panel's consensus was that the evidence implicating the influence of environmental factors on the incidence of noncontact ACL injuries was "confusing and mixed" [4]. While many experts agreed that the increased coefficient of friction at the shoe-surface interface was likely to increase the incidence of ACL injury, studies have yet to definitively show a link between increased friction and increased strain on the ACL sufficient to cause injury. Many studies are flawed methodologically or lack the number of ACL injuries to make compelling arguments. The panel concluded that the investigation of environmental risk factors and their effects on noncontact ACL injury is an area, which requires further study and integration of biomechanical and epidemiologic data. Currently, the biomechanical studies evaluating the shoe-surface interface have only investigated the loading conditions at the level of the foot. Different maximum torques or rates of torque have been demonstrated based on the specific characteristics of the shoe-surface interface. In our opinion, this data represent circumstantial evidence with regards to the loading conditions at the knee. To date, the investigators are unaware of any biomechanical study that has addressed the effects of changing the shoe-surface interface on knee injury. The goal of this study is to quantitatively analyze the effects of the shoe-surface interface in the development of ACL strain during a simulated cutting motion.

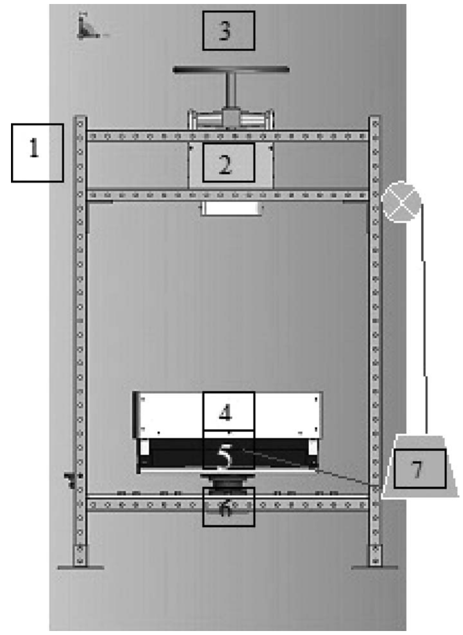

Fig. 1 Graphic depiction of the novel testing device: (1) Unistrut (steel) is the supporting beam of the testing device; (2) testing cube allows superior and inferior translation while preventing axial rotation; (3) turn screw allows the application of an axial load; (4) turf box houses the different athletic surfaces; (5) six component force plate, which calculate forces and moments in the *x*, *y*, and *z* planes; (6) lazy susan/potentiometer allows axial rotation of the surface and calculates the angle; and (7) pulley with weights, which creates a moment about the shoe-surface interface

2 Materials and Methods

This study used a cadaveric based experimental model to evaluate the role of the shoe-surface interface on ACL strain. The independent variable is the shoe-surface interface. Loads generated at the shoe-surface interface during the simulated cutting movement were transmitted up the kinetic chain of the cadaveric lower limb to the knee joint and generated a strain in the ACL. The primary dependent (outcome) variable is the maximum strain in the ACL and secondary outcomes include maximal loads and moments at the shoe-surface interface.

2.1 Testing Apparatus. Each potted cadaveric specimen was attached to a custom shear constrained loading assembly (Figs. 1 and 2). This was composed of the Unistrut Steel Framing (Wayne, MI), which formed the base and the columns. It also consisted of a $6\times6\times6$ in.^{3} (15.24 cm$\times$15.24 cm$\times$15.24 cm) steel cube in which the potted specimens were mounted. To create an axial load the testing cube was attached to the testing apparatus via a mounting rail and frictionless ball bearing design. This allowed axial translation of the testing cube while preventing any rotation. The testing cube was attached to a turn screw, which used a displacement mechanism in order to generate an axial load.

At the base of the testing apparatus a wooden box that housed the different surfaces was secured to a slightly larger steel box via 1 1/2 in. (3.81 cm) nuts and bolts. This steel box rested on a six component force plate (Bertec Corp., Columbus, OH), which measured both forces and moments in the *x*, *y*, and *z* planes. The force plate was attached to a lazy susan and potentiometer, which allowed axial rotation but prevented translation. In order to generate a standard moment several pulleys were affixed to the testing apparatus. In addition, a bar and traction rope were connected to the lazy susan. The rope was then attached to several sandbags and extended over the most superior aspect of the testing apparatus. When the lazy susan was unlocked and the weights were dropped, this generated torque within the axial plane of the force plate.

2.2 Cadavers. Eight cadaveric lower extremities were obtained for this study. Any cadaver with a positive Lachman examination or varus, valgus, or anterior/posterior instability was excluded from this study. In addition, on intra-articular inspection

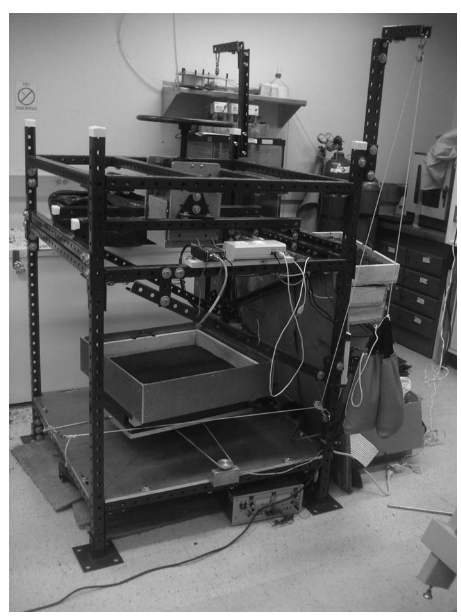

Fig. 2 Testing device

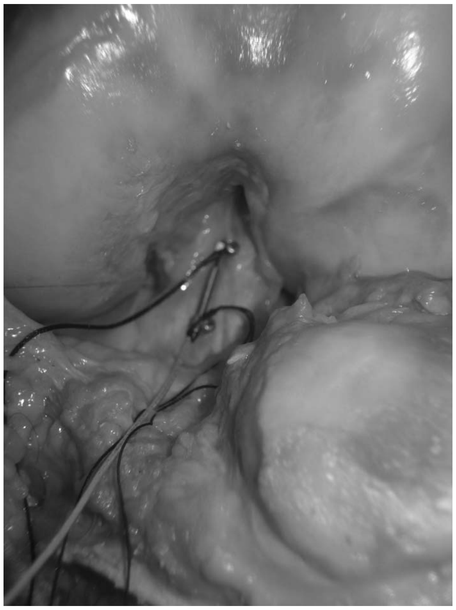

Fig. 3 A Microstrain DVRT is inserted into the anteromedial bundle of the ACL

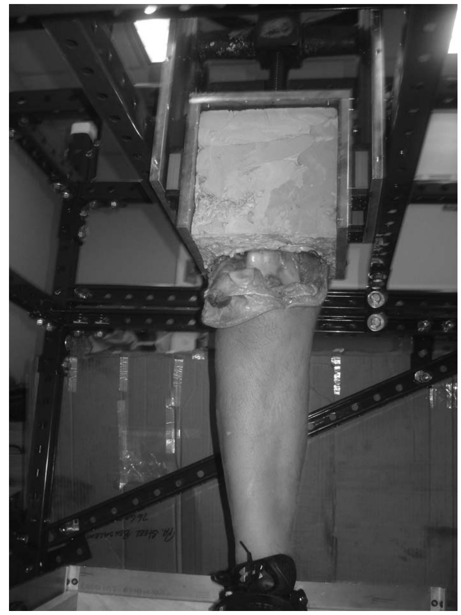

Fig. 4 Potted cadaver loaded into testing device

any cadavers, which had grade IV chondral changes or an injury to the ACL were not included in this study. The lower extremities were then dissected to remove all soft tissue attachments above the level of the medial and lateral epicondyles. The origins of the MCL, LCL, and capsule were all preserved. An oscillating saw was then used to amputate the femur 10 cm proximal to the medial and lateral epicondyles. This distance was chosen to accommodate the testing cube and is consistent with the amount of femur potted in prior experiments [23]. Three 1/4 in. (0.635 cm) screws were then placed in the femur equidistant from each other proximal to the level of the epicondyles; this was done to help control rotation. Finally, the femur was potted in 30 deg of knee flexion with Body Filler (Bondo Corp, Atlanta, GA). It was held in the appropriate position until the body filler hardened.

To approach the ACL, a medial parapatellar arthrotomy was performed. This confirmed the presence of an intact ACL and also allowed inspection of the articular surfaces. A strain gauge (Microstrain, Williston, VT) was placed in the midportion of the anteromedial bundle of the ACL under direct visualization (Fig. 3). It was a microminature differential variable reluctance transformer (DVRT), which had resolution up to 1.5 μm displacement. The DVRT was then attached to a 16 bit analog to digital conversion system (Measurement Computing Inc., Norton, MA) and transmitted to a PC laptop computer via a USB interface. Inputs included the six channels from the force plate and one channel from the potentiometer as well. The data collection was performed with TracerDaq Pro (Measurement Computing Inc., Norton, MA).

2.3 Surfaces. Three of the major playing surfaces for football and soccer (a sport which have a higher incidence of ACL injuries) are grass, AstroTurf™, and modern playing turfs, which have an infill. Each of the surfaces is composed of varying amounts of rubber, sand particles, and differently sized grass blades. All of these surfaces have different coefficients of friction. Fresh Kentucky Bluegrass sod, AstroTurf™, and a typical modern playing turf were analyzed for the purposes of this study. The grass had an average blade length of 2 in. (5.08 cm). The AstroTurf™ (SRI Sports, Augusta, GA) is a synthetic playing surface composed of coarse, monofilament knitted nylon fibers. It is essentially a carpet with 1/2 in. (1.27 cm) fibers on a 5 mm foam pad. The modern playing turf (Turfstore, Calhoun, GA) is composed of 2 in. polyethylene fibers and a crumb rubber infill. It has 3 lbs of infill per square foot of turf. Each of these surfaces was cut into a 2×2 ft^2 (60.96 cm $\times$ 60.96 cm) section and then secured to the turf box via well spaced screws to minimize the motion at the surface-plate interface. The surfaces were marked to register the center of the rotating platform.

2.4 Shoes. Two different shoe types were studied: a turf shoe and a cleated shoe. While most trainers and athletes alike agree that turf shoes are best for the AstroTurf™ surface and cleated shoes are best for the grass surface, there is no consensus on which shoes should be worn on the newer infill surfaces. For the purposes of this study we chose the Metal Mid Super Turf shoes and the Iso Mid D cleats, which are both 3/4 athletic shoes and made by Under Armour™ (Baltimore, MD). These are among the most common shoes used for the aforementioned playing surfaces. The cleats had seven (screw-in) grass cleats with a depth of 14.3 mm. The shoe sizes were measured so that each of the cadavers was tested with appropriately fitting shoes on all the playing surfaces.

2.5 Preliminary Trials. For each experiment, the potted specimen was placed in the testing cube (Fig. 4). In order to confirm appropriate calibration of our strain gauge, serial Lachman examinations were performed and the strains were recorded. Next, a specimen was placed in a testing cube and 15 lbs of weight was attached to the pulley. An axial load was then placed on the specimen using the turn screw and measured using the force plate. Due to stress relaxation of the viscoelastic structures within the cadaveric limb, the experiment was conducted only after the load level had reached a plateau at the desired axial load level. The sandbags were then released, a torque was created at the shoe-surface interface and the maximum strain was recorded (Fig. 5). The serial experiments were performed in this fashion with increasing axial loads from 100 N to 900 N at 100 N intervals. Next, an axial load of 500 N was chosen and serial moments from 0.5 Nm to 3.0 Nm (0.5 Nm intervals) were applied to the cadaveric limb. Finally, an axial load of 500 N and an initial moment of 1.5 Nm were chosen and then the experiment was performed in both internal and external rotation. Each of these trials was performed with the specimen wearing a turf shoe and using the AstroTurf™ surface. Two specimens were used for this portion of the experiment and then discarded.

2.6 Experimental Trials. Four shoe-surface interface combi-

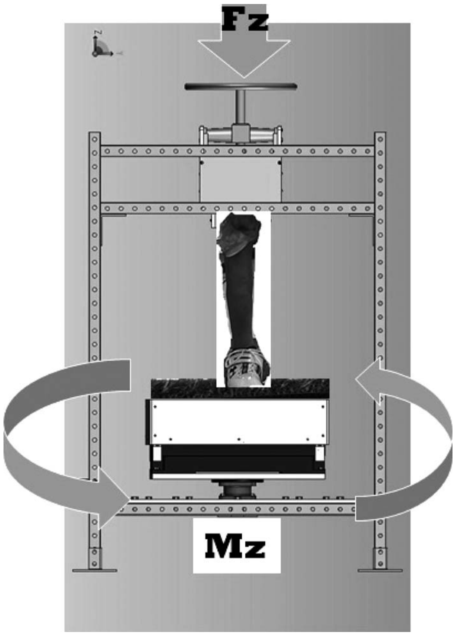

Fig. 5 Schematic drawing of the experiment: A shoe is placed on the potted cadaver and loaded into the testing assembly, an axial load is then placed followed by a moment about the axial plane and the ACL strain is recorded

nations were utilized for this portion of the experiment: AstroTurf-turf shoe, modern playing turf-turf shoe, modern playing turf-cleat, and natural grass-cleat. A starting axial load of 500 N and an initial axial moment of 1.5 Nm was chosen, which produced an internal rotation torque of the tibia relative to the femur. The turf box was allowed to rotate until it reached 90 deg of rotation or until stopped by the constraints of the cadaver. This was confirmed by the potentiometer. The initial axial force, initial moment, maximum strain, maximum force, and maximum moment in the axial plane were all recorded using the strain gauge and the force plate. Five trials for each of the eight specimens on each shoe-surface interface (40 data sets for each shoe-surface interface) were performed in a repeated measures fashion. Before each trial a Lachman examination was conducted to confirm competency of the ACL and appropriate calibration of the strain gauge. Statistics performed with analysis of variance (ANOVA) and posthoc Bonferoni–Dunn tests with significance set at $p < 0.05$.

3 Results

3.1 Preliminary Trials. Each specimen had a competent ACL based on inspection and Lachman examination. Lachman examination produced an average strain of 4.3% (range of 1.25–6.38%). There was a proportional increase in strain in the ACL with increasing load (Fig. 6) until a level of 500 N was attained. Higher initial loads from this point produced a plateau in maximum ACL strain and then slowly began to decline. With respect to a constant

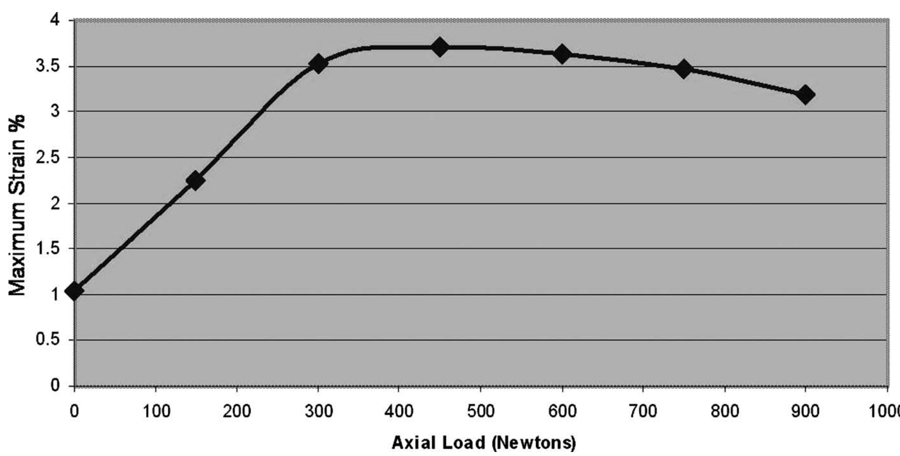

Fig. 6 Graph of strain (%) versus axial load: Note that at approximately 500 N the strain plateaus

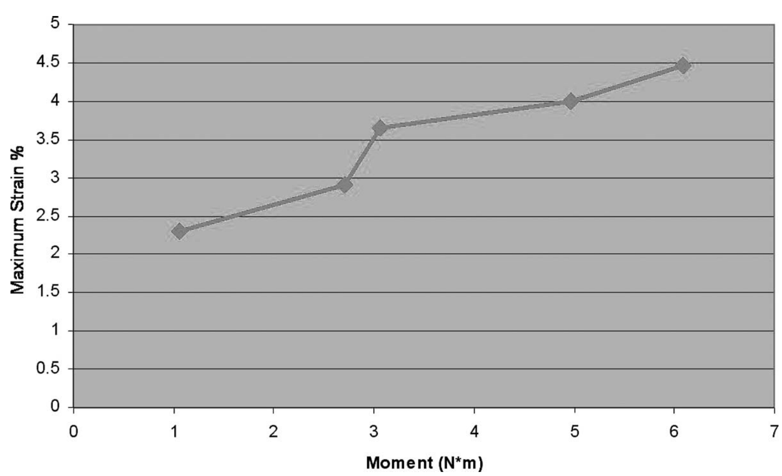

Fig. 7 Graph of strain (%) versus moment: Note that as moment increases so does ACL strain in a linear type pattern

Table 1 Data summary of the experiment trials

Group	Starting force avg. (N)	Starting moment avg. (Nm)	Max strain avg. (%)	Max force avg. (N)	Max moment avg. (Nm)
AstroTurf-turf shoes	494	1.9	3.90	918	32.1
Modern playing turf-turf shoes	471	1.7	3.19	759	34.1
Modern playing turf-cleats	477	1.4	3.14	725	39.7
Natural grass-cleats	461	1.6	2.16	732	37.5

initial load and increasing moment there was a linear relationship between moment and ACL strain throughout the experiment (Fig. 7). For the final portion of these trials a 500 N axial load and a 1.5 Nm initial moment were evaluated. Two specimens had these loading conditions for five trials of internal rotation and five trials of external rotation. For internal rotation a detectable, reproducible strain was produced in the ACL. (average of 4.05%, range of 0.88–7.17%) In contrast, for external rotation there was no detectable strain until the terminal portion of the experiment (75–90 deg of external rotation). The average strain was 0.03% (range of 0–0.17%). The difference between internal and external rotation was statistically significant ($p<0.001$).

3.2 Experimental Trials. The average age of the eight cadaveric specimens was 57.2 years (range of 54–61). Each specimen had an intact ACL based on Lachman examination and visual inspection. A detectable strain was produced within the ACL and recorded for each of the trials. The results for each of the 4 groups: AstroTurf-turf shoe, modern playing turf-turf shoe, modern playing turf-cleat, and natural grass-cleat are summarized in Table 1. The average starting force and starting moment for these experiments were 475.8 N (range of 461–491 N) and 1.65 N (range of 1.4–1.9 N), respectively. There were no statistically significant differences in the initial loading conditions between trials performed on the four different shoe-surface combinations.

The natural grass-cleat combination had a statistically lower maximum strain than any of the remaining three groups (Fig. 8). The AstroTurf-turf shoe was 80.2% greater ($p<0.001$), modern playing turf-turf shoe was 47.5% greater ($p=0.014$), and the modern playing turf-cleat was 45.1% greater ($p=0.022$). There were no statistically significant differences between each of the remaining groups.

The AstroTurf-turf shoe combination had a statistically higher maximum force than any of the remaining three groups (Fig. 9). The modern playing turf-turf shoe, modern playing turf-cleat, and natural grass-cleat were 17.2% less ($p=0.016$), 21.0% less ($p=0.002$), and 20.2% less ($p<0.003$), respectively. There were no statistically significant differences between each of the remaining groups.

With respect to maximum moment, there was only one significant difference between the groups. The modern playing turf-cleat combination had a 19% higher maximum moment than the AstroTurf-turf group ($p=0.032$).

4 Discussion

While there are many risk factors for ACL injury it is our contention that the shoe-surface interface needs to be more critically examined because it is easily modifiable. Many epidemiologic studies showed that it may play a role [3,4,16,18,19,24–31]. However, the various confounders have undermined the data and prevented the authors from making definitive conclusions. Several authors designed experiments looking at the torques that develop at the shoe-surface interface [19–22,27,32–37]. These authors have espoused that higher peak torques and rates of developing torque may lead to higher injury rates. However, none of these authors has quantified the loading conditions at the knee. It is unclear how the forces generated at the shoe-surface interface travel up the kinetic chain and may affect injury.

This was a pilot study designed to be a proof-of-concept experiment. This model allows the reproducible generation of ACL strain in the elastic range for a given internal rotation moment and axial load. The Lachman examination confirmed presence of the ACL and appropriate calibration of the strain gauge. While the examination was not controlled (i.e., it was performed manually), the values for strain obtained were consistent with those reported in literature for the Lachman examination [38–42]. The anterome-

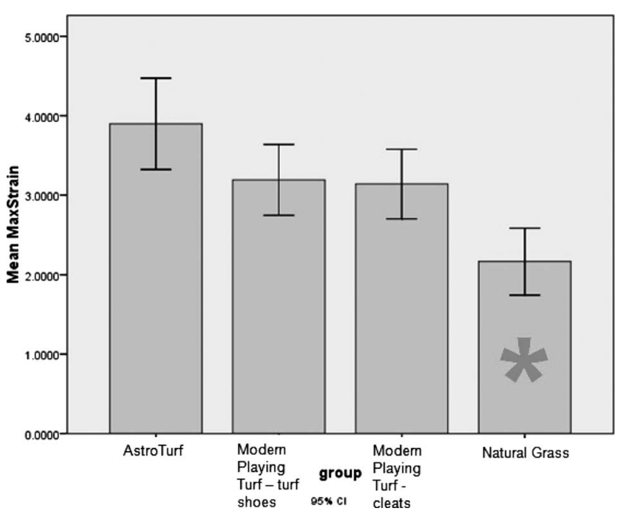

Fig. 8 Graph of the mean maximum strain in the ACL versus the shoe-surface interface: The red star indicates a statistically significant difference ($p<0.05$)

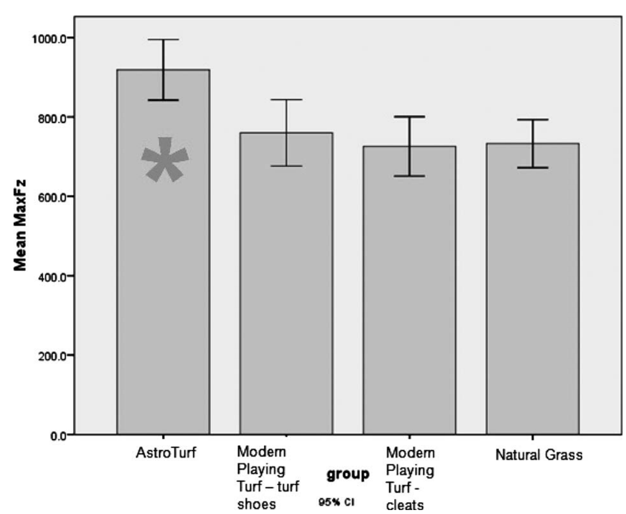

Fig. 9 Graph of the mean maximum load on the force plate versus the shoe-surface interface: The red star indicates a statistically significant difference ($p<0.05$)

dial bundle of the ACL was chosen due its accessibility and ability to accommodate the strain gauge. In addition, we wanted to compare our strain data with other studies, which also used the anteromedial bundle of the ACL [38,41–43]. Serial axial loads confirmed an increase in strain up to a level of 500 N. At this level the maximum strain attained in the ACL tapers. We believe that this is due to the significant increase in articular contact pressure, which in effect, protects the ACL for a given moment. This was supported by the potentiometer data, which revealed that for the 1.5 Nm moment, the ultimate rotation of the turf box was less (<90 deg) at higher axial loads (<90 deg). We also surmise that if we increase the moment, this plateau effect would be delayed to even higher axial loads. This is supported by our other preliminary data, which showed a proportional increase in maximum strain with increasing moments for a standard axial load.

With respect to rotation direction, we found a significant difference between internal and external rotation for this model. This was consistent with other cadaveric studies, which also found higher strains in internal rotation. Clinically, this may be explained by the anatomy of the ACL. External rotation in effect unwinds the ACL until terminal external rotation at which time the slack in the ACL is removed and strain is again perceptible [44]. Conversely, internal rotation continues to twist the ACL, which may explain why strain is detected earlier in the cutting maneuver.

Given our preliminary data, we chose a specific axial load and internal rotation moment to reproducibly create a detectable strain in the anteromedial bundle of the ACL. Using these parameters, the generation of strain in the ACL appears to change with different shoe-surface combinations. Specifically, the natural grass and cleat combination produced less strain in the ACL than the modern playing turf-cleat, modern playing turf-turf shoe, and AstroTurf-turf shoe combinations for a given axial load and moment. This did not correlate with the maximum moment appreciated by the force plate. This may occur for several reasons. Livesay and colleagues proposed that the rate of development of torque may be an important criterion for assessment of athletic fields [22]. The duration in which a noncontact ACL injury takes place is often a fraction of a second. In such a brief time period an appropriate neuromuscular response may not be feasible and rate of development of torque may be as important if not more important than the peak torque. We did not evaluate this in our study.

A second explanation may be the result of the complex interaction between the bottom of the shoe and the top of the surface during the cut. Two important variables with respect to this interaction are the coefficient of friction and the coefficient of restitution. The coefficient of friction is closely related to torque and studies showed that there is a higher incidence of ACL injuries with surfaces that have a higher coefficient of friction [8]. However, the coefficient of restitution may also play an important role [45]. The coefficient of restitution is defined as the ability of a field to absorb shock. It is measured by using the G-Max value where one "G" represents one unit of gravity. The United States Consumer Products Safety Commission (USCPSC) [19,45] determined that fields with a G-Max of greater than 200 are unsafe for athletic play. However, to date, it is unclear how the interplay of these two properties affects injury rates. It seems plausible that the traction, which develops is a combination of not only the coefficient of friction but also how hard the surface may be. This is also supported by climatic studies, which assert that temperature, assuming dry conditions, may alter the hardness of a particular surface and injury rates [29].

In our study, this concept of restitution can be seen in our maximum load data. The AstroTurf-turf shoe combination was the stiffest construct. The turf shoe is only permitted to displace the AstroTurf several millimeters. As a result, the maximum force was statistically higher than any of the other three combinations, which were all more flexible. Clinically, we observed that this stiffer construct allowed less vertical displacement of the foot for the same vertical load and likely increased the intra-articular pressures. The pliability of modern playing turf and grass allow for a far greater displacement into the turf for a given load and likely contribute to a lower maximum force. The generation of lower intra-articular pressures may be protective against knee injury.

With respect to shoe type, there was no difference in any of the dependent variables measured with respect to a simulated cut made with cleats or turf shoes on the modern playing turf. This suggests that for these two particular shoe types the risk of injury to the ACL may depend more on the playing surface than the shoes. It is important to note that although the depth of the cleats (14.3 mm) was slightly larger than the depth of the studs on the turf shoes (10 mm), the displacement into modern playing turf was similar for a given axial load. This may explain many of the similarities between the two groups.

This study had several limitations. Beynnon et al. [43,46,47] did a number of human studies in which a strain gauge was placed in vivo and the impact of various activities an ACL strain were evaluated. While this type of study may produce the most accurate information on the subject, the inherent risks of our study to patients precluded us from conducting such an experiment and led us to develop a cadaveric model. We chose to study ACL injuries due to their prevalence and devastating consequences. Using a cadaveric model we only account for the static stabilizers of the knee. Several authors demonstrated that the dynamic stabilizers of the knee can also play a role in ACL injury [48,49]. Furthermore, in athletes the forces while making a cut are generated proximally while we chose to generate our forces distally to aid in the logistics of the experiment. We only tested within the elastic range and did not test in the range, which would cause plastic deformation and injury. We utilized this method in order to conserve specimens and perform a repeated measures analysis. Ultimately, the shoe-surface profile, which leads to ACL failure, will be of the most value. We also only tested rotation, while most authors assert that the ACL injury mechanism is more complex and may include valgus and translation as well [50].

This cadaveric model was able to demonstrate that performing a cut on certain shoe-surface combinations causes significantly more strain in the ACL and thus has the potential to be more deleterious to the knee. This may also provide an explanation for the increased soreness and muscle fatigue that was reported when playing on artificial surfaces [51]. While the kinematics of the ACL injury mechanism clearly involves a complex array of motions, it is our assertion that the shoe-surface interface does as well. As such, only studying the effects at the shoe-surface interface do not accurately represent our outcome of interest as clinicians, namely the loading conditions at the anatomic structures of the lower extremity. While the bulk and the current kinematic profile of the experimental apparatus do not lend itself to be a commonly used surrogate to test the safety of shoe-surface interfaces it does provide important information with respect to these outcomes of interest. Future studies will continue to attempt simulation of the ACL injury mechanism and correlate strain data with the physical properties of the shoe-surface interface in addition to epidemiologic data in order to help establish standards for ideal shoe-surface combinations.

5 Disclosures

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References

[1] 1996, Center for Disease Control and Prevention: National Hospital Discharge Survey.

[2] Daniel, D. M., and Fritschy, D., 1994, “Anterior Cruciate Ligament Injuries,” *Orthopedic Sports Medicine: Principles of Practice*, J. DeLee and D. J. Drez, eds., Saunders, Philadelphia, PA, pp. 1313–1361.

[3] Griffin, L. Y., Agel, J., Albohm, M. J., Arendt, E. A., Dick, R. W., Garrett, W. E., Garrick, J. G., Hewett, T. E., Huston, L., Ireland, M. L., Johnson, R. J., Kibler, W. B., Lephart, S., Lewis, J. L., Lindenfeld, T. N., Mandelbaum, B. R., Marchak, P., Teitz, C. C., , and Wojtys, E. M., 2000, “Noncontact Anterior Cruciate Ligament Injuries: Risk Factors and Prevention Strategies,” J. Am. Acad. Orthop. Surg., **8**(3), pp. 141–150.

[4] Griffin, L. Y., Albohm, M. J., Arendt, E. A., Bahr, R., Beynnon, B. D., Demaio, M., Dick, R. W., Engebretsen, L., Garrett, W. E., Jr., Hannafin, J. A., Hewett, T. E., Houston, L. J., Ireland, M. L., Johnson, R. J., Lephart, S., Mendelbaum, B. R., Mann, B. J., Marks, P. H., Marshall, S. W., Myklebust, G., Noyes, F. R., Powers, C., Shields, C., Jr., Shultz, S. J., Silvers, H., Slauterbeck, J., Taylor, D. C., Teitz, C. C., Wojtys, E. M., and Yu, B., 2006, “Understanding and Preventing Noncontact Anterior Cruciate Ligament Injuries: A Review of the Hunt Valley II Meeting, January 2005,” Am. J. Sports Med., **34**(9), pp. 1512–1532.

[5] Agel, J., Arendt, E. A., and Bershadsky, B., 2005, “Anterior Cruciate Ligament Injury in National Collegiate Athletic Association Basketball and Soccer: A 13-Year Review,” Am. J. Sports Med., **33**(4), pp. 524–530.

[6] Gottlob, C. A., and Baker, C. L., Jr., 2000, “Anterior Cruciate Ligament Reconstruction: Socioeconomic Issues and Cost Effectiveness,” Am. J. Orthop., **29**(6), pp. 472–476.

[7] Gottlob, C. A., Baker, C. L., Jr., Pellissier, J. M., and Colvin, L.,1999 “Cost Effectiveness of Anterior Cruciate Ligament Reconstruction in Young Adults,” Clin. Orthop. Relat. Res., **367**, pp. 272–282.

[8] Myklebust, G., Maehlum, S., Holm, I., and Bahr, R., 1998, “A Prospective Cohort Study of Anterior Cruciate Ligament Injuries in Elite Norwegian Team Handball,” Scand. J. Med. Sci. Sports, **8**(3), pp. 149–153.

[9] Nigg, B. M., and Segesser, B., 1988, “The Influence of Playing Surfaces on the Load on the Locomotor System and on Football and Tennis Injuries,” Sports Med., **5**(6), pp. 375–385.

[10] Orchard, J., Seward, H., McGivern, J., and Hood, S., 1999, “Rainfall, Evaporation and the Risk of Non-contact Anterior Cruciate Ligament Injury in the Australian Football League,” Med. J. Aust., **170**(7), pp. 304–306.

[11] Orchard, J. W., and Powell, J. W., 2003, “Risk of Knee and Ankle Sprains Under Various Weather Conditions in American Football,” Med. Sci. Sports Exercise, **35**(7), pp. 1118–1123.

[12] Myklebust, G., Maehlum, S., Engebretsen, L., Strand, T., and Solheim, E., 1997, “Registration of Cruciate Ligament Injuries in Norwegian Top Level Team Handball. A Prospective Study Covering Two Seasons,” Scand. J. Med. Sci. Sports, **7**(5), pp. 289–292.

[13] Powell, J. W., and Schootman, M., 1992, “A Multivariate Risk Analysis of Selected Playing Surfaces in the National Football League: 1980 to 1989. An Epidemiologic Study of Knee Injuries,” Am. J. Sports Med., **20**(6), pp. 686–694.

[14] Scranton, P. E., Jr., Whitesel, J. P., Powell, J. W., Dormer, S. G., Heidt, R. S., Losse, G., and Cawley, P. W., 1997, “A Review of Selected Noncontact Anterior Cruciate Ligament Injuries in the National Football League,” Foot Ankle Int., **18**(12), pp. 772–776.

[15] Parekh, S. G., Sennett, B. J., and Carey, J. L., 2005, “Epidemiology and Outcomes of Anterior Cruciate Ligament Injuries in the National Football League,” AAOS Annual Meeting, Washington, DC.

[16] Meyers, M. C., and Barnhill, B. S., 2004, “Incidence, Causes, and Severity of High School Football Injuries on FieldTurf Versus Natural Grass: A 5-Year Prospective Study,” Am. J. Sports Med., **32**(7), pp. 1626–1638.

[17] Nicholas, J. A., Rosenthal, P. P., and Gleim, G. W., 1988, “A Historical Perspective of Injuries in Professional Football. Twenty-Six Years of Game-Related Events,” JAMA, J. Am. Med. Assoc., **260**(7), pp. 939–944.

[18] Ramirez, M., Schaffer, K. B., Shen, H., 2006, “Injuries to High School Football Athletes in California,” Am. J. Sports Med., **34**(7), pp. 1147–1158.

[19] Torg, J. S., Quedenfeld, T. C., and Landau, S., 1974, “The Shoe-Surface Interface and Its Relationship to Football Knee Injuries,” J. Sports Med., **2**(5), pp. 261–269.

[20] Andreasson, G., Lindenberger, U., Renstrom, P., and Peterson, L., 1986, “Torque Developed at Simulated Sliding Between Sport Shoes and an Artificial Turf,” Am. J. Sports Med., **14**(3), pp. 225–230.

[21] Cawley, P. W., Heidt, R. S., Jr., Scranton, P. E., Jr., Loose, G. M., and Howard, M. E., 2003, “Physiologic Axial Load, Frictional Resistance, and the Football Shoe-Surface Interface,” Foot Ankle Int., **24**(7), pp. 551–556.

[22] Livesay, G. A., Reda, D. R., and Nauman, E. A., 2006, “Peak Torque and Rotational Stiffness Developed at the Shoe-Surface Interface: The Effect of Shoe Type and Playing Surface,” Am. J. Sports Med., **34**(3), pp. 415–422.

[23] Loh, J. C., Fukuda, Y., Tsuda, E., Tsuda, E., Steadman, R. J., Fu, F. H., and Woo, S. L.-Y., 2003, “Knee Stability and Graft Function Following Anterior Cruciate Ligament Reconstruction: Comparison Between 11 o’clock and 10 o’clock Femoral Tunnel Placement. 2002 Richard O’Connor Award Paper,” Arthroscopy: J. Relat Surg., **19**(3), pp. 297–304.

[24] Arendt, E., and Dick, R., 1995, “Knee Injury Patterns Among Men and Women in Collegiate Basketball and Soccer. NCAA Data and Review of Literature,” Am. J. Sports Med., **23**(6), pp. 694–701.

[25] Arendt, E. A., Agel, J., and Dick, R., 1999, “Anterior Cruciate Ligament Injury Patterns Among Collegiate Men and Women,” J. Athl. Train., **34**(2), pp. 86–92.

[26] Bradley, J. P., Klimkiewicz, J. J., Rytel, M. J., and Powell, J. W., 2002, “Anterior Cruciate Ligament Injuries in the National Football League: Epidemiology and Current Treatment Trends Among Team Physicians,” Arthroscopy: J. Relat. Surg., **18**(5), pp. 502–509.

[27] Lambson, R. B., Barnhill, B. S., and Higgins, R. W., 1996, “Football Cleat Design and Its Effect on Anterior Cruciate Ligament Injuries. A Three-Year Prospective Study,” Am. J. Sports Med., **24**(2), pp. 155–159.

[28] Levy, I. M., Skovron, M. L., and Agel, J., 1990, “Living With Artificial Grass: A Knowledge Update. Part 1: Basic Science,” Am. J. Sports Med., **18**(4), pp. 406–412.

[29] Orchard, J., 2002, “Is there a Relationship Between Ground and Climatic Conditions and Injuries in Football?” Sports Med., **32**(7), pp. 419–432.

[30] Orchard, J. W., Chivers, I., Aldous, D., Bennell, K., and Seward, H., 2005, “Rye Grass is Associated With Fewer Non-Contact Anterior Cruciate Ligament Injuries Than Bermuda Grass,” Br. J. Sports Med., **39**(10), pp. 704–709.

[31] Skovron, M. L., Levy, I. M., and Agel, J., 1990, “Living With Artificial Grass: A Knowledge Update. Part 2: Epidemiology,” Am. J. Sports Med., **18**(5), pp. 510–513.

[32] Bonstingl, R. W., Morehouse, C. A., and Niebel, B. W., 1975, “Torques Developed by Different Types of Shoes on Various Playing Surfaces,” Med. Sci. Sports, **7**(2), pp. 127–131.

[33] Bowers, K. D., Jr., and Martin, R. B., 1975, “Cleat-Surface Friction on New and Old AstroTurf,” Med. Sci. Sports, **7**(2), pp. 132–135.

[34] Culpepper, M. I., and Niemann, K. M., 1983, “An Investigation of the Shoe-Turf Interface Using Different Types of Shoes on Poly-Turf and Astro-Turf: Torque and Release Coefficients,” Ala. J. Med. Sci., **20**(4), pp. 387–390.

[35] Heidt, R. S., Jr., Dormer, S. G., Cawley, P. W., Scranton, P. E., Loose, G., and Howard, M., 1996, “Differences in Friction and Torsional Resistance in Athletic Shoe-Turf Surface Interfaces,” Am. J. Sports Med., **24**(6), pp. 834–842.

[36] Rheinstein, D. J., Morehouse, C. A., and Niebel, B. W., 1978, “Effects on Traction of Outsole Composition and Hardnesses of Basketball Shoes and Three Types of Playing Surfaces,” Med. Sci. Sports, **10**(4), pp. 282–288.

[37] Stanitski, C. L., McMaster, J. H., and Ferguson, R. J., 1974, “Synthetic Turf and Grass: A Comparative Study,” J. Sports Med., **2**(1), pp. 22–26.

[38] Beynnon, B., Howe, J. G., Pope, M. H., Johnson, . J., and Fleming, B. C., 1992, “The Measurement of Anterior Cruciate Ligament Strain In Vivo,” Int. Orthop., **16**(1), pp. 1–12.

[39] Beynnon, B. D., Johnson, R. J., Fleming, B. C., Renstrom, P. A., Nichols, C. E., Pope, M. H., and Haugh, L. D., 1994, “The Measurement of Elongation of Anterior Cruciate-Ligament Grafts In Vivo,” J. Bone Jt. Surg., Am. Vol., **76**(4), pp. 520–531.

[40] Beynnon, B. D., Uh, B. S., Johnson, R. J., Fleming, B. C., Renstrom, P. A., and Nichols, C. E., 2001, “The Elongation Behavior of the Anterior Cruciate Ligament Graft In Vivo. A Long-Term Follow-Up Study,” Am. J. Sports Med., **29**(2), pp. 161–166.

[41] Fleming, B. C., Beynnon, B. D., Tohyama, H., Johnson, R. J., Nichols, C. E., Renström, P., and Pope, M. H., 1994, “Determination of a Zero Strain Reference for the Anteromedial Band of the Anterior Cruciate Ligament,” J. Orthop. Res., **12**(6), pp. 789–795.

[42] Howe, J. G., Wertheimer, C., Johnson, R. J., Nichols, C. E., Pope, M. H., and Beynnon, M. S., 1990, “Arthroscopic Strain Gauge Measurement of the Normal Anterior Cruciate Ligament,” Arthroscopy: J. Relat Surg., **6**(3), pp. 198–204.

[43] Beynnon, B. D., and Fleming, B. C., 1998, “Anterior Cruciate Ligament Strain In-Vivo: A Review of Previous Work,” ASME J. Biomech., **31**(6), pp. 519–525.

[44] Arms, S. W., Pope, M. H., Johnson, R. J., Fischer, R. A., Arvidsson, I., and Eriksson, E., 1984, “The Biomechanics of Anterior Cruciate Ligament Rehabilitation and Reconstruction,” Am. J. Sports Med., **12**(1), pp. 8–18.

[45] 2008, United States Consumer Product Safety Commission, American Society for Testing and Materials—Athletic Fields.

[46] Beynnon, B. D., Fleming, B. C., Johnson, R. J., Nichols, C. E., Renstrom, P. A., and Pope, M. H., 1995, “Anterior Cruciate Ligament Strain Behavior During Rehabilitation Exercises In Vivo,” Am. J. Sports Med., **23**(1), pp. 24–34.

[47] Beynnon, B. D., Pope, M. H., Wertheimer, C. M., Johnson, R. J., Fleming, B. C., Nicholas, C. E., and Howe, J. G., 1992, “The Effect of Functional Knee-Braces on Strain on the Anterior Cruciate Ligament In Vivo,” J. Bone Jt. Surg., Am. Vol., **74**(9), pp. 1298–1312.

[48] DeMorat, G., Weinhold, P., Blackburn, T., Chudik, S., and Garrett, W., 2004, “Aggressive Quadriceps Loading Can Induce Noncontact Anterior Cruciate Ligament Injury,” Am. J. Sports Med., **32**(2), pp. 477–483.

[49] Fleming, B. C., Renstrom, P. A., Ohlen, G., Johnson, R. J., Peura, G. D., Beynnon, B. D., and Badger, G. J., 2001, “The Gastrocnemius Muscle is an Antagonist of the Anterior Cruciate Ligament,” J. Orthop. Res., **19**(6), pp. 1178–1184.

[50] Shimokochi, Y., and Shultz, S. J., 2008, “Mechanisms of Noncontact Anterior Cruciate Ligament Injury,” J. Athl. Train., **43**(4), pp. 396–408.

[51] 2006, National Football League Players Association: Playing Surfaces Opinion Survey.