The Role of Calcaneofibular Ligament (CFL) Injury in Ankle Instability: Implications for Surgical Management

Abstract

Background: Acute inversion ankle sprains are among the most common musculoskeletal injuries. Higher-grade sprains, including anterior talofibular ligament (ATFL) and calcaneofibular ligament (CFL) injury, can be particularly challenging. The precise impact of CFL injury in ankle instability is unclear.

Hypothesis/Purpose: We hypothesized that CFL injury will result in decreased stiffness, peak torque, and increased talus and calcaneus motion, as well as alter ankle contact mechanics when compared to the uninjured ankle and the ATFL only injured ankle in a cadaveric model.

Study Design: Controlled Laboratory Study

Methods: Ten matched-pairs of cadaver specimens with a pressure sensor in the ankle joint and motion trackers on the fibula, talus, and calcaneus were mounted on an Instron with 20° of ankle plantar flexion and 15° of internal rotation. Intact specimens were axially loaded to body weight, then underwent inversion along the anatomic axis of the ankle from 0° to 20°. The ATFL and CFL were sequentially sectioned and underwent inversion testing for each condition. Linear mixed models (LMMs) were used to determine significance for stiffness, peak torque, peak pressure, contact area, and inversion angles of the talus and calcaneus, relative to the fibula across the three conditions.

Results: Stiffness and peak torque did not significantly decrease after sectioning the ATFL, but decreased significantly after sectioning the CFL. Peak pressures in the tibiotalar joint decreased and mean contact area increased significantly following CFL release. There was significantly more inversion of the talus and calcaneus as well as calcaneus medial displacement with weight-bearing inversion after sectioning the CFL.

Conclusions: The CFL contributes considerably to lateral ankle instability. Higher-grade sprains that include CFL injury result in significant decreases in rotation stiffness, peak torque, substantial alteration of contact mechanics at the ankle joint, increased inversion of the talus and calcaneus, and increased medial displacement of the calcaneus.

Clinical Relevance: Repair of the CFL should be considered during lateral ligament reconstruction when injured, and there may be a role for early repair in high-grade injuries to avoid intermediate and long-term consequences of a loose or incompetent CFL.

Key Terms: Ankle, Ligaments; Ankle Instability; Ankle Sprain; ATFL; CFL

What is known about the subject:

The ATFL and CFL are both important lateral ankle stabilizers in internal rotation and inversion. While there is a trend towards worse outcomes in combined ATFL and CFL injuries, there is still a lack of knowledge concerning the implications of insufficiency of the CFL as well as the possible relevance of its respective repair. Additionally, there is no current consensus amongst the Orthopaedic community whether the CFL should be repaired in high-grade ankle sprains. Hence, biomechanical studies, particularly in weight-bearing conditions are highly required.

What this study adds to existing knowledge:

This study presents the first biomechanical study examining the influence of the ATFL and CFL during weight-bearing inversion injury conditions concerning both joint stability and kinematics. Sequentially greater inversion of the talus and calcaneus was noticed with progressive ligament injury (ATFL alone followed by combined ATFL and CFL insufficiency). This study suggests that the CFL plays a more significant role in ankle joint stability and contact mechanics when compared to the ATFL, and that repair of the CFL should be considered during lateral ligament reconstruction. A CFL-deficient ankle has significantly different joint mechanics than the intact ankle, and there may be an important role for early repair of the CFL in high-grade ankle sprains.

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Introduction

Acute inversion ankle sprains are among the most common musculoskeletal injuries in both athletes and non-athletes. The incidence in the United States is 30,000 ankle sprains/day and accounts for 7-10% of emergency room visits.[4, 8, 9] It is estimated that 25-40% of all sports-related injuries involve the ankle.[8, 15] Non-operative management of acute ankle sprains is appropriate for the majority of ankle sprains. However, it is estimated that 20% of severe ankle sprains will lead to chronic ankle instability, diminished athletic performance, and further joint injuries.[20]

Inversion force of the ankle with the foot in plantarflexion is the most common mechanism of ankle ligament injury.[13] Two of the most important ligaments in the ankle's lateral ligament complex during acute lateral ankle injury are the anterior talofibular ligament (ATFL) and calcaneofibular ligament (CFL). The ATFL is responsible for restricting internal rotation of the talus in the mortise and inversion during plantar flexion. The ATFL is the most often injured ligament in acute ankle sprains with a failure load at around 138 N, which is reported to be 2 to 3.5 times lower than the failure of the CFL.[2, 19, 29, 30] In a cadaver model, Bahr et al. measured the maximum force in the ATFL to be 76±23 N and the highest load in the CFL to be 109±28 N in a cadaver model.[3] This ATFL load is 55% of the 138 N failure load and the CFL is 22% to 39% of this failure load. High-grade ankle sprains include both the ATFL and CFL. The CFL is nearly exclusively responsible for resistance to inversion during dorsiflexion in the neutral state. During plantarflexion, the CFL resists inversion alongside the ATFL, and also acts as a stabilizer of the subtalar joint.[16] In an estimated 50-70% of high grade ankle sprains, it is thought that following ATFL elongation, the stronger CFL becomes stretched until it fails at around 345 N.[2, 12]

For patients who fail conservative management for high-grade sprains, the gold standard surgical procedure is the lateral ligament repair first described by Broström.[6] Recently, arthroscopic techniques to repair the ATFL have emerged as clinically effective in the short term.[26] The impact of CFL injury in ankle instability is unclear and there is variability in current practices in terms of whether the CFL is repaired during lateral ligament repair. For example, some surgeons suggest that repair of the CFL is unnecessary, yet a survey of an international consensus group indicates that 80% of respondents routinely repair the CFL during a lateral ligament repair procedure.[1, 23] Some authors do not advocate repairing the CFL based on biomechanical data and clinical outcomes data.[21, 22] Contributing to the lack of consensus on the necessity of repairing the CFL are limited biomechanical data in the literature examining what role the CFL plays in lateral ankle stability. The objective of this study was to evaluate the impact of CFL injury on ankle joint stability and biomechanics. We hypothesized that CFL injury will result in decreased stiffness, decreased peak torque, and increased talus and calcaneus motion, as well as alteration of ankle contact mechanics when compared to the uninjured ankle and the ATFL only injured ankle in a cadaveric model.

Methods

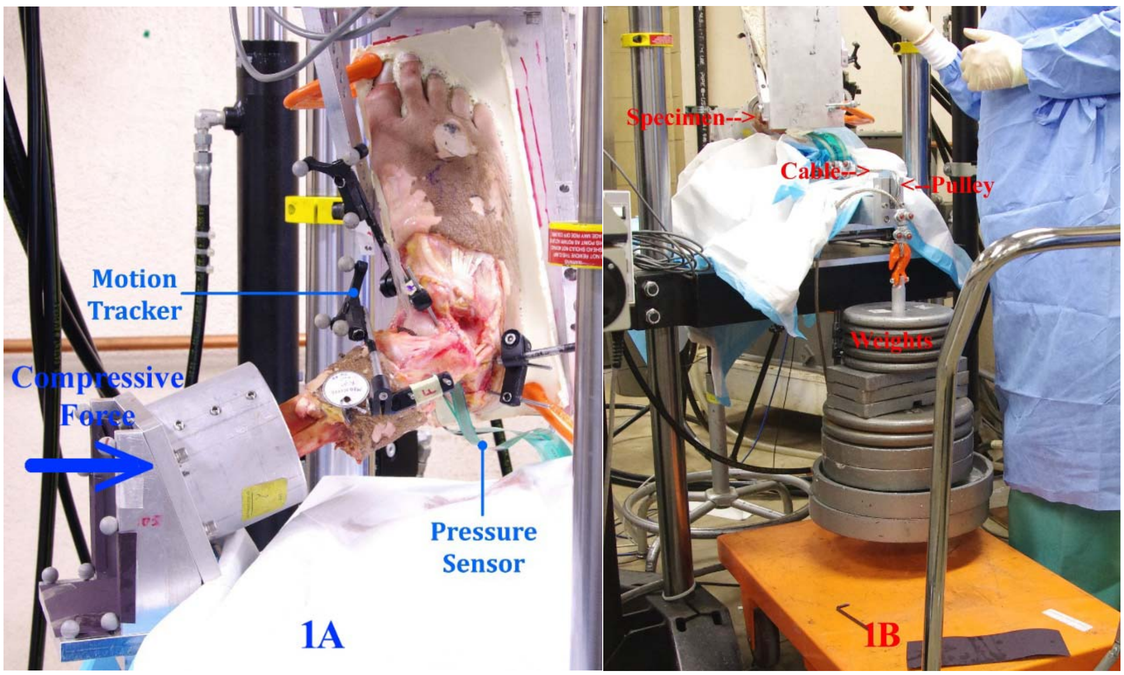
Ten matched pairs of fresh frozen human cadaveric specimens from mid-tibia to toe tip, (5 male, average age 51.4 years, range 38-60; 5 female, average age 53.8 years, range 32-64) were obtained for experimentation from a tissue bank. This project followed all Institutional Review Board requirements in our institution for cadaver laboratory research. Previous studies have established the use of fresh frozen specimens compared to specimens not frozen, as there was little effect on the gross biomechanical properties of the ligaments and other connective tissues due to freezing.[25, 31] Each specimen was transected at the mid-shaft tibia/fibula. All specimens were evaluated visually and radiographically for signs of gross deformity, previous operation, fracture, and rheumatoid arthritis. Specimens were wrapped in moist gauze and placed in a -20°C freezer for storage. The specimens were thawed at room temperature on the day they were prepared and tested. The proximal 4” of soft tissue was removed from the tibia and fibula. The fibula was rigidly fixed to the tibia with a 4.5 mm cortex screw. The proximal 3” of the tibia/fibula was potted with an epoxy (SmoothCast 321; Smooth-On, Inc., Easton, PA, USA) in a 3” diameter round tube. To facilitate approach to the tibiotalar joint, the extensor digitorum longus, tibialis anterior, extensor hallucis longus, and Achilles tendons were sectioned.[17] The plantar surface was secured in an epoxy bed with one additional screw for fixation in the calcaneus. The skin and soft tissue covering the ATFL and CFL were carefully removed without damaging either ligament.

Biomechanical testing was performed on a material testing system (Instron Model 1321 with 8500 controllers; Instron Corporation, Norwood, MA, USA). A 3D, 2 camera motion capture system (Innovision Systems Inc., Columbiaville, MI, USA) was used with custom reflective trackers each rigidly attached with two, 3.0 mm pins, to the fibula, talus, and calcaneus to record the motion of each bone during testing. A pressure measurement system (Model 5033 sensors; Tekscan Inc., Boston, MA, USA) was used to obtain intra-articular tibiotalar pressure data. The sensor was coated with petroleum jelly before being inserted into the ankle joint to minimize the shear forces on the sensor. The pressure sensor is 38.4mm long and 26.7mm wide. It contains 46 rows and 32 columns of 0.694 mm^2 sensels for a total of 1472 sensels. The sensor was inserted so that there were uncontacted sensles anterior, posterior, and lateral to the initial points of contact present on the sensor reading. In many cases, the medial edge of the sensor abutted the bony medial border of the joint. To calibrate the sensors, they were conditioned for 4 cycles to 1800 N, followed by a 10-point power law calibration. Conditioning and calibration cycles consisted of loading for 10 seconds, held at designated load for 30 seconds, unloaded over 10 seconds, and recovery for 2 minutes.[24]

Each specimen was mounted with the tibia horizontal onto the testing apparatus in 20° of plantarflexion and 15° of internal rotation, ensuring that the center of rotation of the tibiotalar joint was aligned with the rotation of axis of the actuator.[7, 14] The tibia was fixed to a platform on the base of the material testing system that was mounted on two linear bearings that allowed free motion in the anatomic superior/inferior direction. Specimens were axially loaded in compression to full body weight by running a cable horizontally from the platform that the tibia was fixed to over a pulley. Weights were hung on the cable equal to the body weight of each individual donor that was obtained from their donor summary report. Each ankle was preconditioned for 10 cycles from 0° to 10° of inversion at 0.25 Hz.[29] After preconditioning, a pressure sensor was inserted into the tibiotalar joint posteriorly to avoid crimping of the sensor **(Figure 1A, 1B)**. Each ankle was tested from 0° to 20° of inversion along the anatomic axis of the ankle at a rate of 5°/s for three cycles. The ATFL and CFL were then sequentially sectioned, and inversion testing was repeated for each of the following conditions: (1) intact; (2) ATFL-injury sectioning; and (3) CFL-injury sectioning. Data were collected at 25 Hz on a PC equipped with an analog to digital board and data acquisition software.

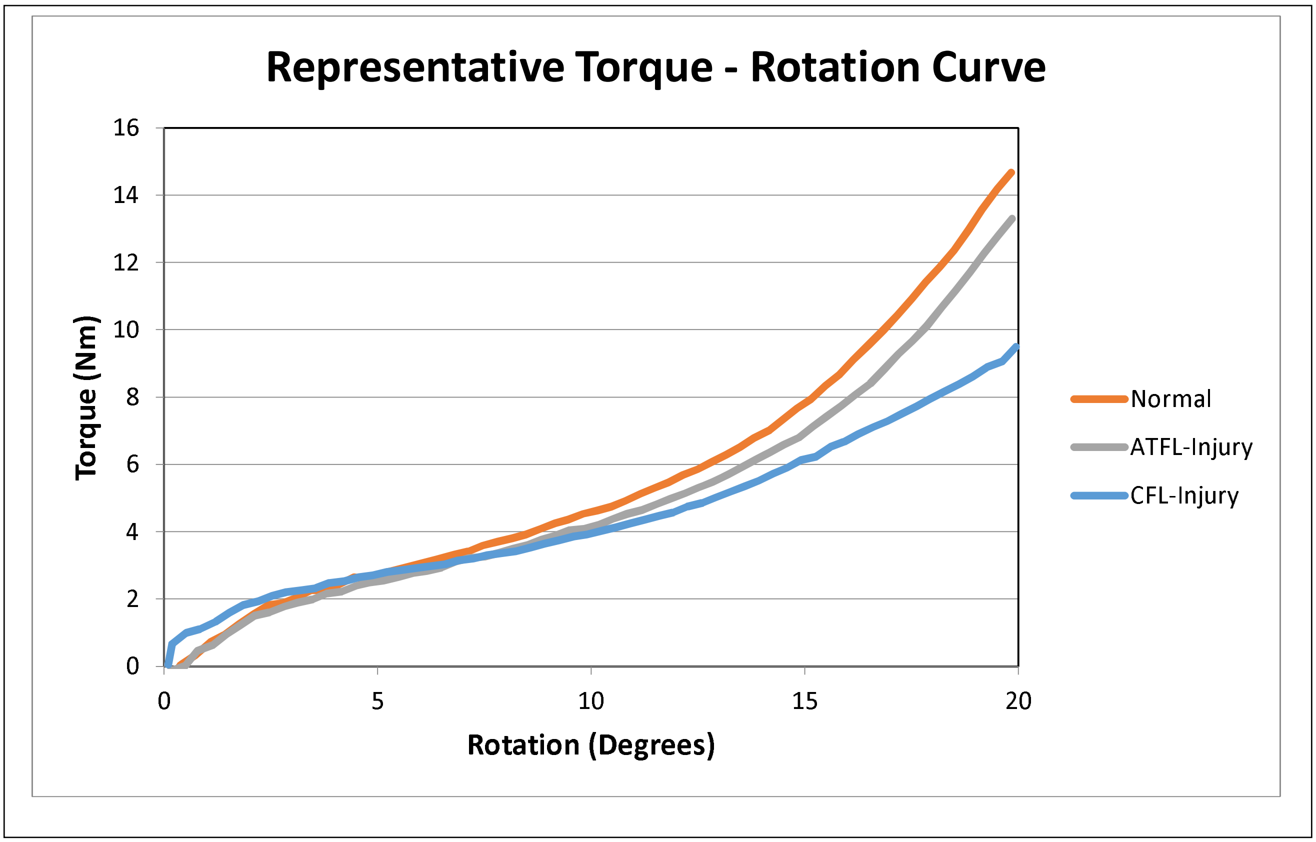
Figure 1(A). Test Setup. The ankle is in 20° of plantar flexion and internally rotated 15°. The platform the tibia is mounted to sits on linear bearings that allow free motion in the anatomic superior/inferior direction (horizontal in the figure). The cable that applies the axial compression force cannot be seen in the picture but it runs horizontally to the right of the picture where it runs over a pulley and weights are hung on the end. The motion trackers can be seen in the fibula and talus. **1(B).** Test setup showing the cable, pulley, and weights that create the body weight axial compressive force on the foot and ankle.

A. **B.**



Data Analysis: Stiffness was calculated from the slope of the torque/rotation curve from 5° to 15° rotation of the second cycle **(Figure 2)**. The peak torque at 20° ankle inversion was reported. Intra-articular tibiotalar peak pressure (MPa), mean contact area (mm^2), and center of force (mm) were recorded at 15 Hz using the pressure measurement system. The peak pressure frame of the second of three cycles of inversion was used for analysis of contact area, peak pressure, and center of force (COF) because this is when the inversion motion had the smoothest arc. The COF was reported as a single, static point in the peak pressure frame. The 3D motion capture camera system was used to assess the following: (1) the angle of inversion of the talus relative to the fibula; (2) the angle of inversion of the calcaneus relative to the fibula and; (3) the medial displacement of the calcaneus relative to the fibula.

Figure 2. Typical Torque-Rotation curve of the same specimen in the Normal, ATFL-injury, and CFL-injury state.



Statistical Analysis: All analyses were performed using SAS 9.4. (SAS Institute Inc. Cary, NC, USA). Student's t-test with Bonferroni correction was used to compare the differences in COF (mm) across the three conditions; a p-value of < 0.017 was regarded as statistically significant. Linear mixed model regression analyses were used to compare ankle peak torque (N·m) and stiffness (N·m/deg) across the three conditions. Linear mixed model regression analyses were also used to determine significance for peak pressure (MPa), contact area (mm^2), the inversion angles (in degrees) of the talus and calcaneus relative to the fibula, as well as the medial displacement (in mm) of the calcaneus relative to the fibula across the three conditions; a p-value < 0.05 was regarded as statistically significant.

Results

Stiffness and Peak Torque

Mean stiffness and peak torque values for the three conditions can be found in **Table 1**. When compared to the intact condition, the difference in mean stiffness for the CFL-injury condition was significant (p = 0.0002). Similarly, the mean difference in stiffness between the ATFL-injury and CFL-injury conditions was also significant (p = 0.0075). There was no significant difference in mean stiffness when comparing the ATFL-injury and intact conditions (p = 0.2254) **(Appendix A)**. When comparing the CFL-injury and intact conditions, the mean difference in peak torque was significant (p < 0.0001). When comparing the CFL-injury and ATFL-injury conditions, the mean difference in peak torque was also significant (p = 0.0012). However, there was no significant difference in mean peak torque when comparing the ATFL-injury and the intact condition (p = 0.3371) **(Appendix A)**.

Table 1. Stiffness (N·m/deg) and Peak Torque (N·m)

Condition	Mean (SD)	95% Confidence Interval	
		Lower Bound	Upper Bound
Stiffness (N·m/deg)			
Normal	0.67 (0.38)	0.49	0.85
ATFL-injury	0.61 (0.35)	0.45	0.78
CFL-injury	0.49 (0.33)	0.34	0.64
Peak Torque (N·m)			
Normal	16.03 (8.37)	11.99	20.06
ATFL-injury	15.46 (7.82)	11.80	19.11
CFL-injury	12.22 (7.57)	8.68	15.77

Peak Pressure, Contact Area, and Center of Force (COF)

Mean peak pressure and contact area values for the three conditions can be found in **Table 2**. When comparing the CFL-injury and the intact condition, the mean difference in peak pressure was significant (p = 0.0003). Similarly, when comparing the CFL-injury and ATFL-injury conditions, the mean difference in peak pressure was also significant (p= 0.002). However, there was no significant difference in mean peak pressure when comparing the ATFL-injury and intact conditions (p = 0.4848) **(Appendix B)**. When comparing the CFL-injury and intact conditions, there was a significant difference in mean contact area (p= 0.0084). When comparing the CFL-injury and ATFL-injury conditions, the results also showed that there was a significant difference (p = 0.0037). However, there was no significant difference in mean contact area when comparing the ATFL-injury and intact conditions (p= 0.7587) **(Appendix B)**.

Table 2. Peak Pressure (MPa) and Contact Area (mm^2)

Condition	Mean (SD)	95% Confidence Interval	
		Lower Bound	Upper Bound
Peak Pressure (MPa)			
Normal	19.56 (13.13)	13.41	25.70
ATFL-injury	18.89 (12.94)	12.83	24.94
CFL-injury	15.72 (9.76)	11.15	20.28
Contact Area (mm^2)			
Normal	137.58 (49.12)	114.59	160.57
ATFL-injury	135.27 (44.76)	114.32	156.22
CFL-injury	158.31 (65.80)	127.52	189.11

Center of Force (COF)

Representative COF images can be found in **Figure 3**. During the ATFL-injury condition, the COF moved 0.76 mm medially, relative to the intact condition (p = 0.008). While there was a net movement of 0.99 mm medially from the intact condition to the CFL-injury condition, this was not significant (p = 0.059). During the ATFL-injury condition, the COF moved 0.32 mm anterior relative to the intact condition (p = 0.773). During the CFL-injury condition, the COF moved 1.03 mm posterior, relative to the ATFL-injury condition, resulting in a net movement of 0.71 mm, posterior from the intact condition to the CFL-injury condition (p = 0.009) **(Appendix B)**.

Figure 3. Representation of COF during Inversion

Motion Capture Data

All mean values from the motion capture data can be found in **Table 3**.

Talus inversion: When comparing the CFL-injury condition to the intact condition, the mean difference in the inversion angle was significant ($p < 0.0001$). Additionally, the mean difference in the inversion angle was also significant when comparing the CFL-injury and ATFL-injury conditions ($p = 0.0021$). There was no significant difference when comparing the intact and ATFL-injury conditions ($p = 0.1215$) **(Appendix C)**.

Calcaneus inversion: When comparing the CFL-injury and intact condition, the mean difference in the inversion angle was found to be significant ($p < 0.0001$). The mean difference in the inversion angle was also significant when comparing the CFL-injury and ATFL-injury conditions ($p = 0.0016$). However, the mean difference in inversion angle when comparing the intact and ATFL-injury conditions was not significant ($p = 0.2887$) **(Appendix C)**.

Medial displacement of calcaneus: Additionally, when comparing the mean medial displacement between intact and ATFL-injury conditions, as well as the ATFL-injury and CFL-injury conditions, these differences were not found to be significant either ($p = 0.2721$ and $p = 0.5639$, respectively) **(Appendix C)**.

Table 3. Motion Capture Measurements

Condition	Mean (SD)	95% Confidence Interval	
		Lower Bound	Upper Bound
Talus Inversion Angle (°)			
Normal	4.39 (4.73)	1.65	7.12
ATFL-injury	4.89 (4.98)	2.02	7.77
CFL-injury	5.98 (5.52)	2.79	9.16
Calcaneus Inversion Angle (°)			
Normal	13.12 (2.87)	11.46	14.78
ATFL-injury	13.70 (3.33)	11.77	15.62
CFL-injury	15.58 (4.33)	13.08	18.08
Medial Displacement of Calcaneus (mm)			
Normal	8.22 (4.93)	5.91	10.52
ATFL-injury	9.36 (8.19)	5.53	13.19
CFL-injury	9.96 (8.47)	6.00	13.93

Discussion

The goal of this study was to determine the role of the ATFL and CFL in inversion ankle stability. These data support the hypotheses that the CFL plays a significant role in ankle joint stability during load-bearing inversion conditions. Stiffness and peak torque decreased significantly only after sectioning of both ATFL and CFL. Peak pressures in the tibiotalar joint decreased significantly only following CFL release, and mean tibiotalar contact area significantly increased only following CFL release. Motion capture data showed a significant increase in inversion angle of both the calcaneus and talus after sectioning the CFL but not after sectioning the ATFL. While the data did not show significant increases in the calcaneus medial displacement in both the ATFL-injury and CFL-injury condition, there was a trend.

The ATFL and CFL are considered the primary lateral ankle stabilizers. The current study examined their role in inversion only. In another study examining the role of the ATFL and CFL on ankle stability, Ziai et al. examined internal rotation in a cadaver model, in which they measured the torque necessary to internally rotate the tibia 30° intact and with both the ATFL and CFL sectioned.[32] They found that sectioning both the ATFL and CFL significantly reduced the torque necessary to achieve 30° degrees of internal tibia rotation. These studies demonstrate the important role that both the ATFL and CFL play on ankle stability in both inversion and internal rotation.

The individual role that the ankle joint and subtalar joint play in the stiffness and peak torque measurements made in the current study may explain why there were no significant differences in stiffness or peak torque between the Normal and the ATFL-injury while there were significant differences between the Normal and CFL-injury. The ankle joint primarily allows for plantar/dorsiflexion and the subtalar joint primarily allows for inversion/eversion. When the ATFL was sectioned, the lateral and medial malleolus maintained most of the inversion stiffness and peak torque that the ankle joint contributes to overall stiffness and peak torque. When the ATFL was sectioned, the inversion angle only increased 0.50° for the talus and 0.58° for the calcaneus, which did not result in an overall significant change in stiffness or peak torque. When the CFL was sectioned, the inversion angle increased 1.59° in the talus and 2.46° in the calcaneus. This resulted in a significant decrease in the stiffness and peak torque. These results are similar to the results of Bahr et al.[3] They tested the foot and ankle with a 375 N compressive joint load and 3.4 N·m inversion torque. After sectioning the ATFL, the tibiocalcaneal motion increased approximately 1° and the tibiotalar motion increased approximately 2°. After sectioning both the ATFL and CFL, the tibiocalcaneal motion increased approximately 8° and the tibiotalar motion increased approximately 15°. In addition, the non-significant changes in the ATFL-injury may be due to the differences in stiffness of the ATFL and CFL. Attarian et al. showed in a typical load deflection curve that the CFL is stiffer than the ATFL, approximately 40 N·m compared to 25 N·m, respectively.[2] Sectioning the less stiff ATFL first resulted in smaller changes in stiffness and peak torque than when the more stiff CFL was sectioned.

The current study can be compared to other studies in the literature that also reported inversion stiffness results from tests with the foot in 20° of plantarflexion and 15° of inversion.[7, 14] For example, Giza et al. tested the ankles after sectioning the ATFL and CFL and repairing them, while Brown et al. tested the ankles after sectioning and repairing only the ATFL.[7, 14] However, neither study tested the intact ankle; they only tested the repaired ankles that showed stiffness that is less than the stiffness found in the current study. In addition, neither study conducted testing with load-bearing inversion. Giza et al. showed a stiffness of the repaired ankle ranging from 0.4 N·m/deg to 0.45 N·m/deg, while Brown et al. reported a stiffness of 0.315 N·m/deg and 0.417 N·m/deg.[7, 14] However, the current study reports the stiffness of the ATFL deficient ankle being 0.615 N·m/deg and the stiffness of the ATFL/CFL deficient ankle being 0.49 N·m/deg. The reported stiffness in the current study is larger than that found in the two other studies because a weight-bearing force was applied across the joint during testing, simulating weight-bearing inversion conditions. This force, intended to simulate the typical injury mechanism of weight-bearing inversion, increases the friction across the joint resulting in higher stiffness.

The alteration in the location of COF was an important finding in this study. It is known that repeated ankle injuries can increase risk of cartilage damage with further injury. While incompetent ligaments can certainly increase the risk of more severe injury, alteration of the location of forces in the tibiotalar joint during load-bearing inversion suggest that risk can be increased even in sub-injury conditions. Our data suggest a movement of the COF medially toward the medial shoulder of the talar dome, which has been reported as the most common location of osteochondral lesions of the talus.[11] Since talar OCDs are commonly identified in patients with ankle injuries, the COF may play a role in the etiology or exacerbation of these lesions. The study by Prisk et al. measured the COF during ankle inversion in the intact and CFL-injury state.[27] They found the COF to move medially and anteriorly while the current study found the COF to shift medially and posteriorly. This difference may be due to the different loading conditions. Prisk et al. used a 200 N axial compressive force and 4.5 N·m of inversion. The current study applied a compressive axial load of donor body weight (ranging from 400 N to 1112 N) and inversion to 20°, which was 16.0 N·m and 12.2 N·m, for intact and CFL-injury, respectively.

There are several limitations to this study. With the use of cadavers, the complex muscle forces and ground reaction forces that cross the ankle joint in vivo were not simulated. Additionally, we were only able to test in one configuration, 20° plantarflexion 15° internal rotation; however, this has been shown to be the most common position of the ankle during lateral ankle injuries.[13] Furthermore, only the ATFL and CFL were examined in this study. The posterior talofibular ligament (PTFL) also contributes to lateral ankle instability but was not examined in this study because it is less commonly injured in isolated ankle sprains. In addition, we did not incorporate injury to the interosseous ligament or other ligaments that stabilize the subtalar joint (that are often injured in high-grade sprains) in order to isolate the impact of CFL injury on the ankle joint only. In addition, in order to gain access to the tibiotalar joint to insert the pressure sensors, the extensor digitorum longus, tibialis anterior, extensor hallucis longus, and Achilles tendons were sectioned. However, these structures are not considered lateral ankle stabilizers and should not have influenced the results. The accuracy of Tekscan sensor has been show to decrease with repeated measures and may have affected the results. Jansson et al. showed that a Tekscan sensor calibrated in a dry environment and tested in either a humid or wet environment recorded 100% or 95% of the initial load at 0.75 hours.[18] Each specimen in the current study was completed within 0.25 hours, from start to finish.

Conclusion

Evolving lateral ankle instability surgical techniques focus on the importance of restoring the ATFL. However, the results of this biomechanical study under weight-bearing conditions, suggest that the CFL plays an important role in the stability of both the ankle and subtalar joints, and in tibiotalar contact mechanics.

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Figure 1A. Test Setup. The ankle is in 20° of plantar flexion and internally rotated 15°. The platform the tibia is mounted to sits on linear bearings that allow free motion in the anatomic superior/inferior direction (horizontal in the figure). The cable that applies the axial compression force cannot be seen in the picture but it runs horizontally to the right of the picture where it runs over a pulley and weights are hung on the end. The motion trackers can be seen in the fibula and talus. B. Test setup showing the cable, pulley, and weights that create the body weight axial compressive force on the foot and ankle.

A. **B.**

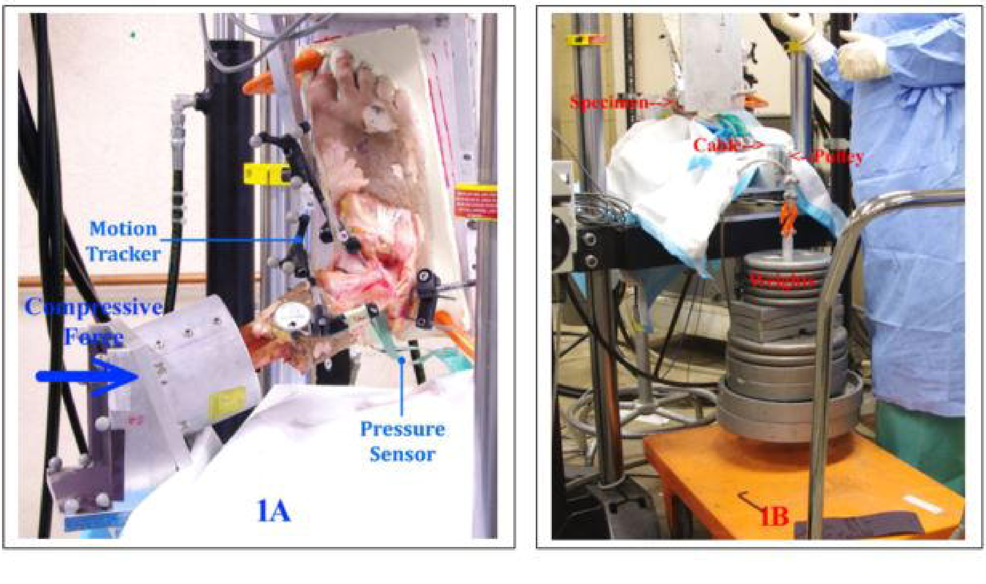

Figure 2. Typical Torque-Rotation curve of the same specimen in the Normal, ATFL-injury, and CFL-injury state.

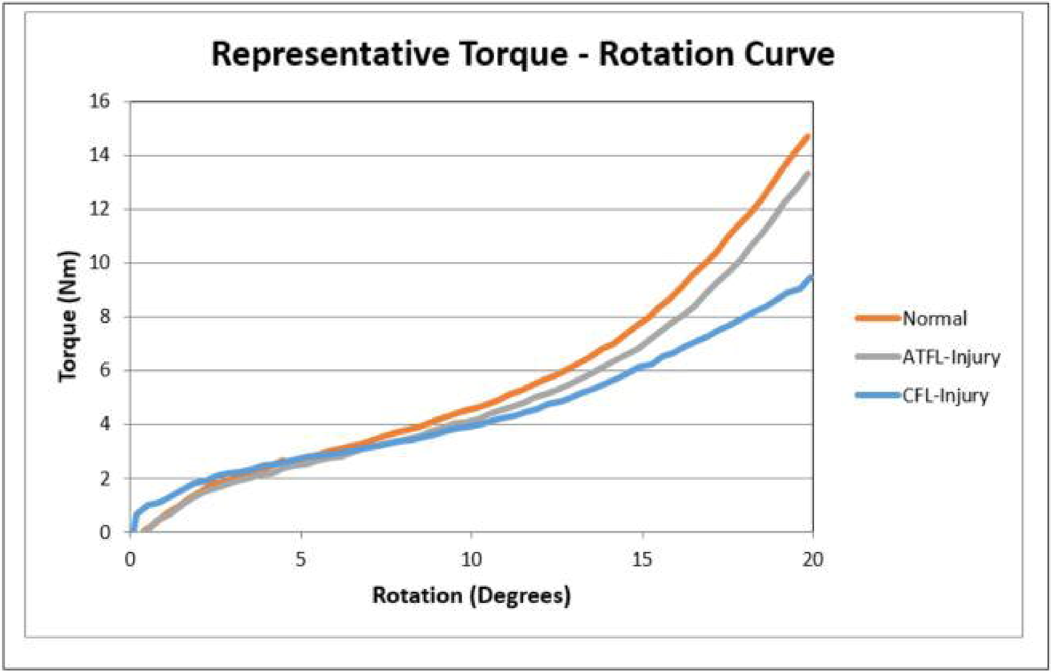

Figure 3. Representation of COF during Inversion