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Ankle joint contact loads and displacement in syndesmosis injuries repaired with Tightropes compared to screw fixation in a static model

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ABSTRACT

Background: The effect of syndesmotic fixation on restoration of pressure mechanics in the setting of a syndesmotic injury is largely unknown. The purpose of this study is to examine the contact mechanics of the tibiotalar joint following syndesmosis fixation with screws versus a flexible fixation device for complete syndesmotic injury.
Methods: Six matched pairs of cadaveric below knee specimens were dissected and motion capture trackers were fixed to the tibia, fibula, and talus and a pressure sensor was placed in the tibiotalar joint. Each specimen was first tested intact with axial compressive load followed by external rotation while maintaining axial compression. Next, syndesmotic ligaments were sectioned and randomly assigned to repair with either two TightRopes® or two 3.5 mm cortical screws and the protocol was repeated. Mean contact pressure, peak pressure, reduction in contact area, translation of the center of pressure, and relative talar and fibular motion were calculated. Specimens were then cyclically loaded in external rotation and surviving specimens were loaded in external rotation to failure.
Results: No differences in pressure measurements were observed between the intact and instrumented states during axial load. Mean contact presure relative to intact testing was increased in the screw group at 5 Nm and 7.5 Nm torque. Likewise, peak pressure was increased in the TightRope group at 7.5 Nm torque. There was no change in center of pressure in the TightRope group at any threshold; however, at every threshold tested there was significant medial and anterior translation in the screw group relative to the intact state.
Conclusion: Either screws or TightRope fixation is adequate with AL alone. With lower amounts of torque, the TightRope group appears to have contact and pressure mechanics that more closely match native mechanics.

Introduction

Ankle injuries are very common and may be associated with syndesmotic injuries. Syndemostic injuries can be purely ligamentous or associated with ankle fractures. There are an estimated 15 syndesmosis injuries per 100,000 persons each year, and syndesmotic injuries are reported in up to 24% of ankle sprains [1–3]. Traditionally, syndesmosis injuries are instrumented using screws leading to rigid fixation once an acceptable reduction is accomplished. More recently, alternative methods such as suture button fixation have been developed which theoretically mimic normal physiology more closely than screws. Some authors theorize that allowing more physiologic motion may facilitate faster ligamentous healing [4] and possibly allow earlier weight-bearing and improved clinical outcomes [5–8].

Regardless of the instrumentation used, the goal of treating syndesmosis injuries is to obtain an anatomic reduction and to recreate the native biomechanics of the syndesmosis in order to reduce the risk of post traumatic arthritis. Thus, numerous biomechanical studies have examined differences between screw fixation and suspensory fixation mainly focusing on pull out strength, torque to failure, subsequent diastasis between the tibia and fibula, and fibular motion [9–15]. However, there is very little data examining the effect syndesmosis instrumentation has on pressure distribution at the tibiotalar joint [16]. As little as 1 mm of talar shift can result in a decrease in 40–42% of the contact surface area between the tibia and talus emphasizing the importance of obtaining an anatomic reduction [17,18]. Hunt et al. (2015)

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reported a shift and increase in mean tibiotalar contact pressure in a cadaveric model simulating a syndesmotic injury [19]. The effect of screw fixation and/or suspensory fixation on restoration of pressure mechanics in the setting of a syndesmotic injury remains largely unknown.

The purpose of this study is to examine the contact mechanics of the tibiotalar joint following syndesmosis fixation with screws versus a suture button device for complete syndesmotic injury. Using a cadaveric injury model, the null hypothesis was that there is no difference between the two implants in restoration of baseline anatomic pressure distributions. Secondary outcome measures include relative motion of the fibula, tibia, and talus and strength of the implants.

Materials and methods

Six matched pairs of cadaveric below knee specimens (mean age: 57 ± 8 yrs; range: 41–64 yrs; 2 female, 4 male) were obtained from ***. Specimens were stored in a −20 °C prior to dissection and testing. Each specimen underwent two freeze thaw cycles, the first for dissection and preparation, and the second for mechanical testing.

On the day of preparation, specimens were dissected to allow for access to the tibiotalar joint. Skin overlying the distal tibia was excised and the underlying Achilles tendon posteriorly and the tendons of extensor digitorum longus, tibialis anterior, and extensor hallucis longus anteriorly were transected. The muscle bellies were reflected superiorly and the anterior and posterior capsule was excised to allow exposure of the tibiotalar joint to allow for insertion of a pressure sensor into the tibiotalar articular surface.

Prior to mechanical testing and ligament sectioning, two pilot holes for later fixation were drilled to ensure adequate reduction of the syndesmosis after ligament sectioning. Specimens were randomly assigned fixation with either two solid Synthes 3.5 mm cortical screws or two suture button devices (TightRopes). Pilot holes for the designated instrumentation were placed 2 cm above the joint line and were drilled at a 30 ° angle from the coronal plane. For the TightRope group, pilot holes were drilled divergent from one another as described by the manufacturer [13–20].

Attachment of the specimen to the materials testing system was accomplished by inserting an aluminum rod (12. 7 mm diameter, 19 cm length) through the articular surface of the tibial plateau into the tibial diaphysis. The rod was secured to the tibia with two transversely oriented machine screws. Three wood screws were inserted into the calcaneus to augment fixation of the palmar surface of the foot in polymethylmethacrylate.

Mechanical testing was performed on an E10000 materials testing system fitted with a 10 K N/100 Nm biaxial load cell (Instron Corporation, Norwood, CA) (Fig. 1).

Proximally, the rod protruding from the tibial articular surface was attached to the test machine actuator. During mechanical testing, both axial and torsional loads were applied through the rod. Distally, the foot was mounted in neutral alignment on an xy-stage that permitted unrestricted translation of the shank relative to the foot in the axial plane.

Motion of the fibula and talus relative to the tibia was monitored throughout testing using a 3D Creator motion capture system (Boulder Innovation Group, Boulder, CO). Custom rigid body trackers were mounted to the tibia, fibula, and talus. The position and rotational accuracies of the trackers are approximately 0.1 mm and 0.2°, respectively. The long axis of the tibia was defined as the line formed by connecting the midpoints between the medial and lateral malleoli, distally, and the medial and lateral boarders of the tibial condyles, proximally. The tibial origin was located midway between the medial and lateral malleoli. The fibular origin was positioned midway between anterior and posterior points on the fibula, located 5 mm superior to the joint line. The talar origin was defined as the average of antero-medial, antero-lateral, postero-medial, and postero-lateral points, 5 mm inferior to the joint line.

Fig. 1. Experimental set-up used to apply combined axial and torsional loads across the ankle joint. Loads were applied through the rod inserted into the proximal tibial intramedullary canal. Distally, the foot was mounted in neutral alignment on an xy-stage that permitted unrestricted translation of the shank relative to the foot in the axial plane.

Specimens were first preconditioned in axial compression from 0 to 700 N for 10 cycles at 0.5 Hz followed by a hold at 700 N axial compressive load while cyclically loading to 7.5 Nm internal/external rotation at 0.25 Hz for 10 cycles.

Following preconditioning, a model 5033 pressure sensor (Tekscan Inc, Boston, MA) was inserted posterior to anterior into the tibiotalar joint. The pressure sensors were conditioned three times to 1200 N and calibrated using a 2-point power function (200 N and 800 N) using previously established techniques [19]. Two tacks were used to secure the sensor to the talus immediately anterior to the articular surface. The posterior end of the sensor was held in place manually during static testing. Care was taken to avoid crimping of the sensors.

Following static testing in the intact state, the AITFL, PITFL, interosseous ligament (IOL) and transverse ligament (TL), and the full deltoid ligament were sectioned. The interosseous ligament was transected all the way to the level of the proximal fibular head and both the deep and superficial deltoid was completely transected. The fibula was manually manipulated to ensure complete ligamentous disruption of the syndesmosis. Specimens were then instrumented with the fixation method selected during the initial specimen dissection. At the time of mechanical testing, a physiologic reduction of the ankle was closely approximated, as insertion of either the cortical screws or TightRopes forced the ankle back to its intact alignment. In the screws group, cortical screws were placed across all 4 cortices for each screw.

Specimens were loaded in axial compression to 700 N followed by a hold at 700 N for one second [19,21–23]. Specimens were then externally rotated at 5 deg/sec to 7.5 Nm torque while simultaneously maintaining axial compression, and then held at 7.5 Nm torque for one second prior to unloading. The torsional load level is approximately 75% of the threshold for discomfort during passive nonweightbearing [24]. Three trials were performed in the intact as well as destabilized/instrumented states. A recovery period of approximately two minutes was observed between each trial.

Following static testing in the destabilized/instrumented state, the pressure sensor was removed for fatigue testing. Specimens were loaded in axial compression to 700 N followed by cyclic external rotation from 0 to 7.5 N m at 1 Hz for 1000 cycles while maintaining axial compression. Surviving specimens were loaded in axial compression to 700 N followed external rotation at 5 deg/sec to failure. Load and displacement data from the Instron test system, kinematic data from the motion capture system, and pressure readings were synchronized and monitored at 50 Hz.

All reported motion was calculated relative to unloaded intact state (~20 N axial compressive load). The average of the three trials of intact and instrumented static testing were used in the subsequent data analysis. Mean contact pressure, peak pressure, and contact area were calculated during intact and instrumented static tests during axial loading (AL) alone and combined AL with 2.5 Nm external rotation (ER), 5 Nm ER, and 7.5 Nm ER. Additionally, medial-lateral (M–L) and anterior-posterior (A–P) center of pressure (COP) translations during combined AL + ER torsional loading were calculated relative to the AL only state.

Fibular and talar kinematics were calculated for static, cyclic, and load to failure tests. Cyclic creep was defined as the difference in fibular ER between cycle 1000 and cycle 20. Peak torque and ER at peak torque were calculated during the load to failure tests. Comparisons between intact and instrumented configurations for each group were made using paired t-tests, while comparisons between instrumented groups were made with Welch's t-tests. Significance was set at $p < 0.05$.

Results

A total of 11/12 specimens completed testing. The one specimen that failed was in the TightRope and failed prior to reaching 7.5 Nm torque during instrumented static testing and was not included in any data reporting or analysis.

Pressure Mechanics

No differences in contact pressure, peak pressure, or contact area were observed between the intact and instrumented states during AL alone for either of the test groups.

Mean contact pressure relative to intact testing increased with larger ER torques for both test groups, however statistically significant differences were only observed in the screw group at 5 Nm and 7.5 Nm torque ($p < 0.05$) (Fig. 2a). Likewise, peak pressure relative to intact testing increased with larger ER torques in both groups, although the only significant finding was at 7.5 Nm torque in the TightRope group ($p < 0.05$) (Fig. 2b). Contact area was significantly reduced in both test groups for all ER torque levels (Fig. 2c), with no significant difference between groups.

There was no significant change in the translation of the COP compared to intact testing for all torque levels in the TightRope group, however both the M–L and A–P COP translated significantly for all torque levels in the screw group (Fig. 3a and b).

Comparing pressure mechanics of the TightRope and screw groups, there was no difference in mean contact pressure, peak pressure, contact area, or COP translation for any of the loading levels analyzed.

Translations of the Fibula and Talus

During AL alone, no significant A–P or M–L fibular translations were observed for either test group relative to the intact state (Table 1). Moreover, no differences in translations were found between TightRope and screw groups during AL alone.

During combined AL and ER, the magnitude of fibular posterior translation increased significantly in the TightRope group relative to intact testing as ER torque increased ($p < 0.05$ for ER $\geq$ 5 Nm), while translations were relatively constant for all torque levels in the screw group. For ER torques of at least 5 Nm, significantly more fibular A–P translation was observed in the TightRope group compared to the screw group ($p < 0.05$).

Similar trends were observed in fibular M–L translation, with the fibula displacing postero-medially with respect to the intact state during ER in the TightRope group ($p < 0.05$ for all ER torque levels); however no significant differences in fibular M–L translation were observed between TightRope and screw groups for any torque levels.

Fibular superior-inferior displacements were less than 1 mm for both test groups compared to testing in the intact state, for all load levels. Aside from anterior displacement of the talus relative to intact in the screw group (1.2 ± 1.8 mm, $p = 0.153$), no mean talar translations greater than 1 mm were observed relative to intact testing for any of the loading levels analyzed for either test group. No significant differences in talar translations were observed when comparing TightRope and screw groups.

Rotation of the Fibula and Talus

During AL alone, significantly more fibular external rotation was observed in the screw group when compared to the intact state ($p < 0.05$) (Table 1). No difference in fibular or talar external rotation was observed in the TightRope group during AL alone.

Both fibular and talar external rotation increased with higher ER torque. There was significantly greater ER of the fibula and talus in both the TightRope and screw groups compared to testing in the intact state for all ER torque levels ($p < 0.05$).

For all load levels, no significant differences in fibular or talar external rotation was observed when comparing TightRope and screw groups.

Cyclic Properties

A single specimen in the screw group dislocated during cyclic testing. Comparing the surviving specimens in both test groups, there was significantly higher peak external rotation (TightRopes: 8.7 ± 2.3 deg vs screws: 3.3 ± 2.9 deg, $p = 0.013$) and greater external rotation creep (TightRopes: 2.3 ± 1.7 deg vs screws: −0.4 ± 1.9 deg, $p = 0.042$) of the fibula in the TightRope group compared to the screw group. No differences in talar rotation were observed during cyclic testing.

Rotation to Failure

There was no difference in the torque to failure between the test groups (TightRopes: 17.1 ± 6.6 Nm vs screws: 15.7 ± 10.5 Nm, $p = 0.814$); however, fibular external rotation at failure was significantly higher in the TightRopes (TightRopes: 11.7 ± 1.9 deg vs screws: 4.5 ± 5.4 deg, $p = 0.039$). No difference in talar external rotation was observed during load to failure tests.

Discussion

The results of this biomechanical study demonstrate no difference in tibiotalar contact mechanics between ankles instrumented with either two 3.5 syndesmosis screws or two TightRopes compared to intact specimens with AL applied alone. With the addition of ER to the AL, neither group was able to restore the native ankle contact mechanics; however, at lower ER torque levels the TightRope group more similarly matched the location of the COP and the mean contact pressure. With regards to fibular motion, the TightRope group allowed for greater fibular translations while neither group controlled axial rotation of the fibula. More flexible fixation of the syndesmosis may allow for more anatomic syndesmotic kinematics and thus more anatomic

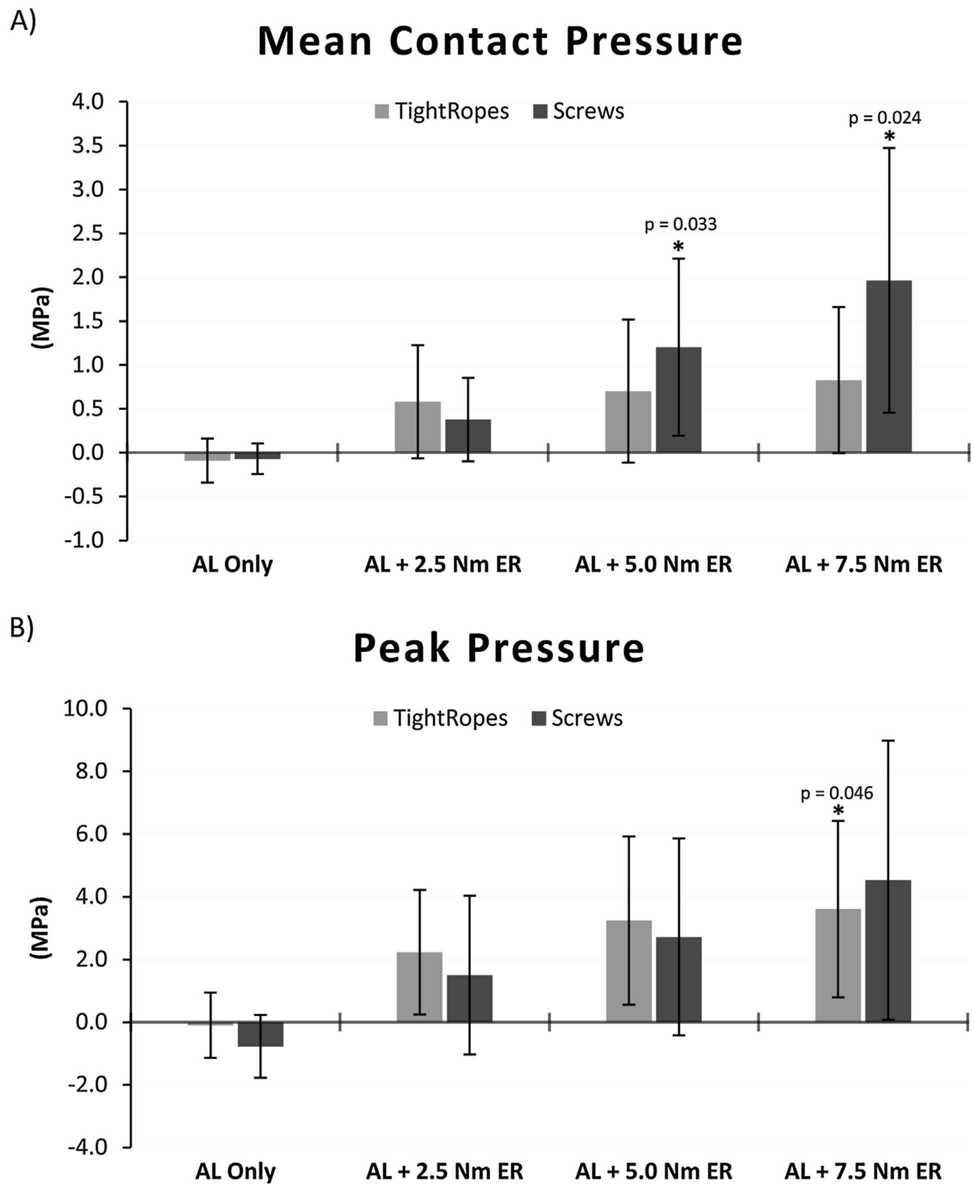

Fig. 2. Mean contact pressure (A), peak pressure (B), and reduction in contact area (C) during axial loading only (AL) and combined AL + ER. * Indicates significant difference between intact and instrumented states ($p < 0.05$).

pressure distributions at the tibiotalar joint explaining our results. Furthermore, the results suggest that both techniques are adequate when axial load alone is applied. However, with lower amounts of torque applied, the TightRope group appears to have contact and pressure mechanics that more closely match native mechanics. Previous studies demonstrate up to approximately 3Nm of ER torque may be experienced at the ankle, similar to the allowance that may be seen during straight line walking and turning during daily activities [25,26]. Thus, our findings suggest that early weightbearing and range of motion may be safe post-operatively, in terms of syndesmotic stabilization, regardless of the fixation technique.

Heretofore, little has been reported on the effect of fixation on restoration of native mechanics. Pereira et al (1996) previously analyzed tibiotalar contact area in different degrees of ankle flexion comparing specimens instrumented with two 4.5 mm screws to intact specimens, finding a decrease in contact area [16]. The authors suggest that rigid fixation of the fibula to the tibia alters the joint's ability to accommodate for changes in the position of the talus, thus altering the joint mechanics [16]. While the specimens in this study were tested in neutral flexion, there was no difference in contact pressure with AL alone for specimens fixed with screws or TightRopes. This difference may be due to the unconstrained model used by Pereira et al, which may have unintentionally introduced rotational torque to their model [16]. Our findings suggest that with AL alone there is little difference in pressure mechanics but the addition ER torque can lead to significant changes.

The findings in the present study are similar to previous reports suggesting greater fibular motion allowed with flexible fixation compared to screws [11,27]. Of note, however, there are other studies that suggest there is no difference in fibular translation between ankles fixed with flexible fixation compared to screws [13]. In addition, the results of this study demonstrate increased fibular rotation with both TightRope and screw fixation similar to Clanton et al (2017) suggesting that neither technique controls axial rotation of the fibular very well [27]. We know from in vivo data that the fibula tends to translate posteromedially relative to

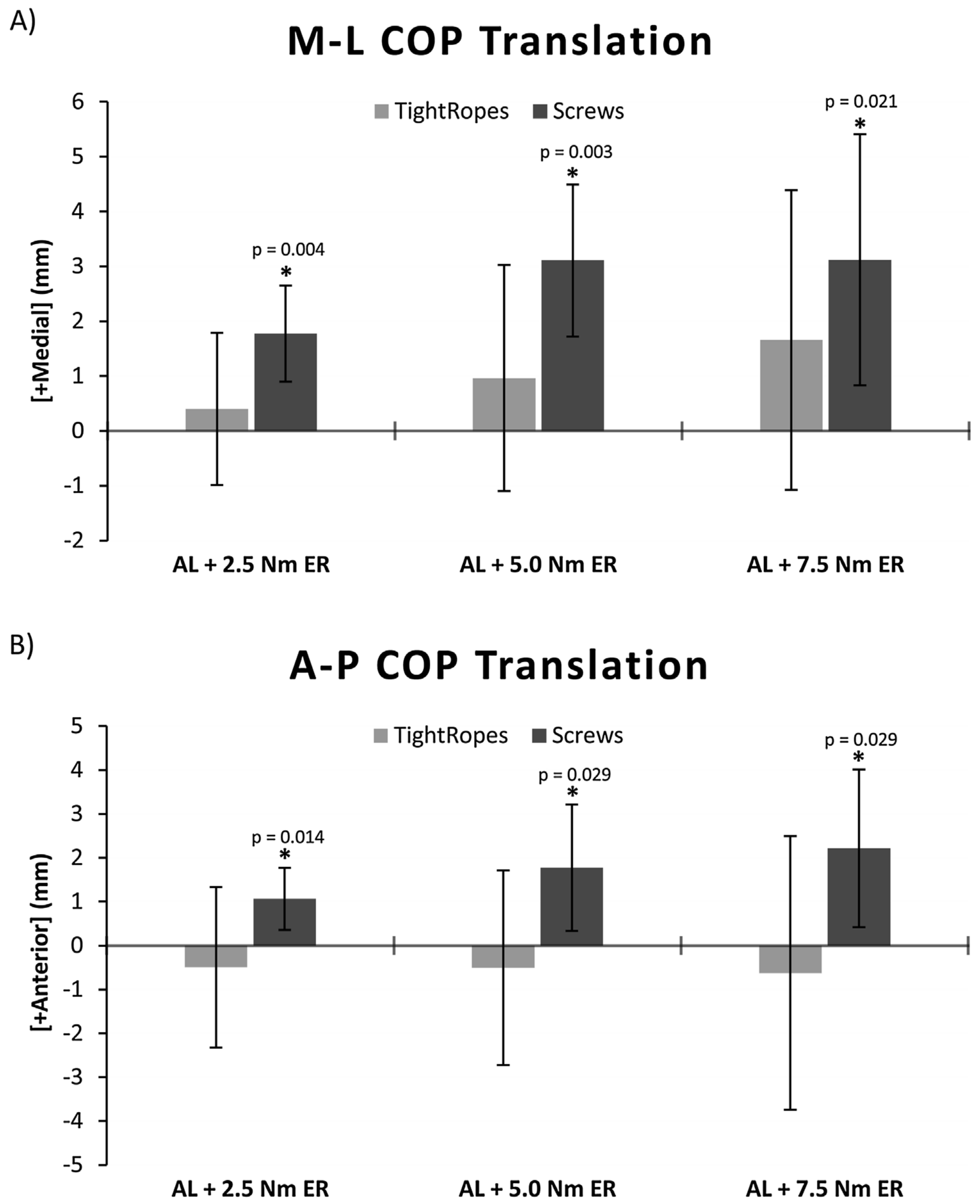

Fig. 3. M-L (A) and A–P (B) center of pressure (COP) translation during combined axial loading (AL) + ER. Data reported relative to the intact state. * Indicates significant difference between intact and instrumented states ($p < 0.05$).

the tibia during athletic activities that involve jumping and pivoting. There is very little lateral translation of the fibula during sports activities, but the significant amount of posteromedial translation that occurs during these activities would suggest an advantage for a flexible device that could restore normal or near-normal motion while mitigating risk of abnormal translations [28].

Lastly, we found similar failure torque in both groups. These findings are similar to previous studies [10,13], however, Ebramzadeh et al (2013) found greater failure torques than those found in our study using a single TightRope and 3.5 mm screw (32.9 Nm +/− 8 Nm and 30.1 Nm +/− 9.6 Nm respectively) [9]. This may be due to the extensive soft tissue dissection required to access the tibiotalar joint for pressure monitoring. The capsule and local tendons likely contribute to resisting torque applied at the joint and excising them would transfer more stress to the instrumentation and leading to lower torque at failure. This may explain why our findings demonstrated lower failure torques despite using two implants.

There are several limitations to this study model. Despite drilling holes with all ligaments intact, fibular alignment was on average approximately four degrees externally rotated for the screw group and approximately two degrees externally rotated for the TightRope group, relative to the unloaded intact reference state, which may affect the results. This may be explained by small allowances in rotation allowed by the instrumentation during implantation. Additionally, during reduction, the syndesmosis was manually compressed which may have also introduced small changes in rotation. The clinical relevance of such small variations in axial rotation is likely inconsequential [29]. In order to access the tibiotalar joint significant soft tissue dissection was required which would create a more unstable model than would likely be found in vivo. Amongst previous biomechanical studies there exists a great amount of variability in testing protocols (loading protocols, devices used, number of devices used, storage, etc.), it is therefore difficult to directly compare results among studies. We chose to use two TightRopes and two screws, due to the loading protocols

Table 1
Fibular translations along with external rotation (ER) of the fibula and talus during axial loading alone (AL) and AL + ER up to 7.5 Nm torque. Values are reported relative to the unloaded intact state. Data is presented as mean ± SD (p-value: instrumented vs. intact). P-values comparing TightRope versus screw groups are presented for each load level. Positive translations/rotations are in the direction listed first (ex: anterior displacements are positive for "A–P Displacement").

		Fibula			Talus
Applied Load	Group	A-P Displacement (mm)	M-L Displacement (mm)	Ext-Int Rotation (Deg)	Ext-Int Rotation (Deg)
AL only	TightRopes	0.3 ± 0.7 (0.360)	0.1 ± 0.5 (0.661)	1.1 ± 1.5 (0.177)	−1.1 ± 1.2 (0.115)
	Screws	0.1 ± 0.8 (0.716)	0.2 ± 1.4 (0.767)	3.1 ± 2.5 (0.028)	−0.7 ± 1.1 (0.199)
	p-value	0.638	0.907	0.137	0.546
AL + 2.5 Nm ER	TightRopes	−0.8 ± 0.9 (0.108)	0.7 ± 0.5 (0.037)	3.1 ± 1.6 (0.012)	2.1 ± 1.7 (0.051)
	Screws	0.2 ± 0.5 (0.437)	0.2 ± 1.5 (0.765)	3.4 ± 2.7 (0.028)	0.7 ± 0.9 (0.093)
	p-value	0.067	0.458	0.867	0.153
AL + 5 Nm ER	TightRopes	−2.3 ± 1.6 (0.031)	1.4 ± 0.5 (0.003)	4.8 ± 2.4 (0.010)	6.8 ± 4 (0.020)
	Screws	0.5 ± 0.8 (0.237)	−0.1 ± 1.4 (0.883)	3.4 ± 2.4 (0.017)	2.4 ± 2 (0.033)
	p-value	0.013	0.050	0.354	0.069
AL + 7.5 Nm ER	TightRopes	−2.6 ± 1.9 (0.041)	1.2 ± 0.4 (0.002)	5.1 ± 2.4 (0.009)	7.6 ± 3.9 (0.012)
	Screws	0.1 ± 0.9 (0.825)	0.1 ± 1.2 (0.850)	4.1 ± 2 (0.004)	5 ± 4.2 (0.032)
	p-value	0.033	0.075	0.468	0.320

we were unable to test specimens using a single implant. Axial and external rotation forces were applied to the cadaveric specimens; however, the forces experienced at the tibiotalar joint during physiologic activity and injury are more complex and not likely to be fully represented by testing protocols used. Securing specimens to the testing machine in the same orientation was difficult and minor variations in orientation of the specimens relative to the machine likely resulted. Testing a larger number of specimens would minimize such variations and should be considered in future studies. One specimen did not survive testing up to 7.5 Nm of torque. The reason for failure was unclear. Lastly, a cadaveric model was used to simulate in vivo mechanics and cannot properly simulate the effects of syndesmotic healing at different time points seen in the postoperative setting.

Conclusions

Either screws or TightRope fixation is adequate with AL alone. With lower amounts of torque, the TightRope group appears to have contact and pressure mechanics that more closely match native mechanics. These findings may have implications for proper implant selection in active patients, early postoperative weight-bearing protocols, and perhaps reduction in the risk of developing post-traumatic arthritis. Further research is needed to determine the clinical implications of these findings.

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