Inversion-Eversion moments supported by TayCo Braces

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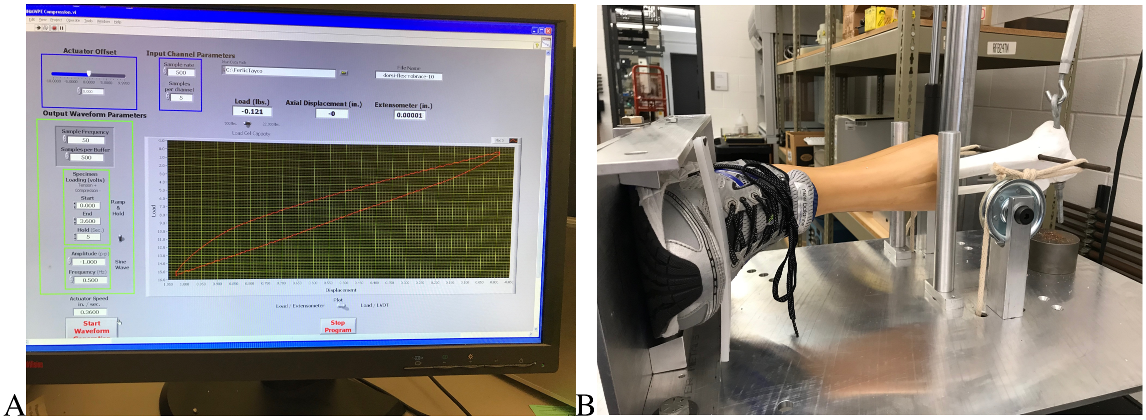
Methods

Ankle stiffness was measured in inversion and eversion using a custom fixture and an MTS 8500 load frame. The foot-ankle-tibia model was mounted horizontally in the load frame. For the unbraced condition and for the TayCo braces, a shoe was placed on the foot, and the shoe gripped against a foot plate. A ¼” rod was placed through the foot plate fixture and the posterior portion of the shoe sole to further prevent motion relative to the fixture.For the walking boot, the boot was bolted to the foot plate. For all tests, an axial load of 37 lbs. was applied to the foot-ankle-shank complex via a rope and pulley (Fig. 1).

Fig. 1. A) Anterior view of test configuration for eversion of the left ankle with a long TayCo brace. B) Posterior view of the same setup. C) Weights on a cord were used to apply an axial load to the tibia during testing through a pulley system. D) The walking boot was bolted to the foot plate to maintain a rigid connection.

The straps of the TayCo brace were pulled very tightly around the ankle, tibia, and foot. The bladder of the walking boot was inflated, but no control over the amount of inflation was used. The inflation was the same for all tests.

The distal end of the tibia was displaced vertically (lateral or medial relative to the foot), and the force and displacement were recorded at 100 samples per second. A total displacement of 1 inch was set in order to achieve linear-force displacement measurements without damaging the foot model or the braces (Fig. 2).


UHMWPE Compression.vi

Actuator Offset: 0.000

Input Channel Parameters

Sample rate: 500

Samples per channel: 5

Main Data Path: C:\FerlicTayco

File Name: dorsi-flex-nobrace-10

Load (lbs.): -0.121

Axial Displacement (in.): -0

Extensometer (in.): 0.00001

Output Waveform Parameters

Sample Frequency: 50

Samples per Buffer: 500

Specimen Loading (volts) Tension + Compression -

Start: 0.000

End: 3.600

Hold (Sec.): 5

Amplitude (p-p): -1.000

Frequency (Hz): 0.500

Actuator Speed in. / sec.: 0.3600

A B

Fig. 2. A) The load displacement data collected from the material testing system. The loading curve is linear, while the unloading curve is nonlinear, reflecting the viscoelasticity of the foam limb. B) The limb in the deflected position.

Inversion and eversion tests with no brace, the two TayCo braces, the CAM walking boot, a Breg Ultra brace, and an Aircast were performed with the foot in 0° of flexion. For all tests, an axial load of 37 lbs. was applied to the leg model via a rope and pulley (Fig. 3). During testing, the straps of the braces were pulled very tightly, and the shoe was tied very tightly around the leg model.

For the unbraced condition, Breg brace, and Aircast, ¼ inch lag screws were inserted through the sole of the shoe and into the heel of the foot to minimize motion of the foot within the shoe (Fig. 3F and G). A ¼" rod was inserted through the foot plate fixture and the posterior portion of the shoe sole to further prevent motion relative to the fixture.

For the CAM walker, the boot was bolted to the foot plate near the heel and ball of the foot. The bladder of the walking boot was inflated, but no control over the amount of inflation was used; the inflation was the same for all tests.

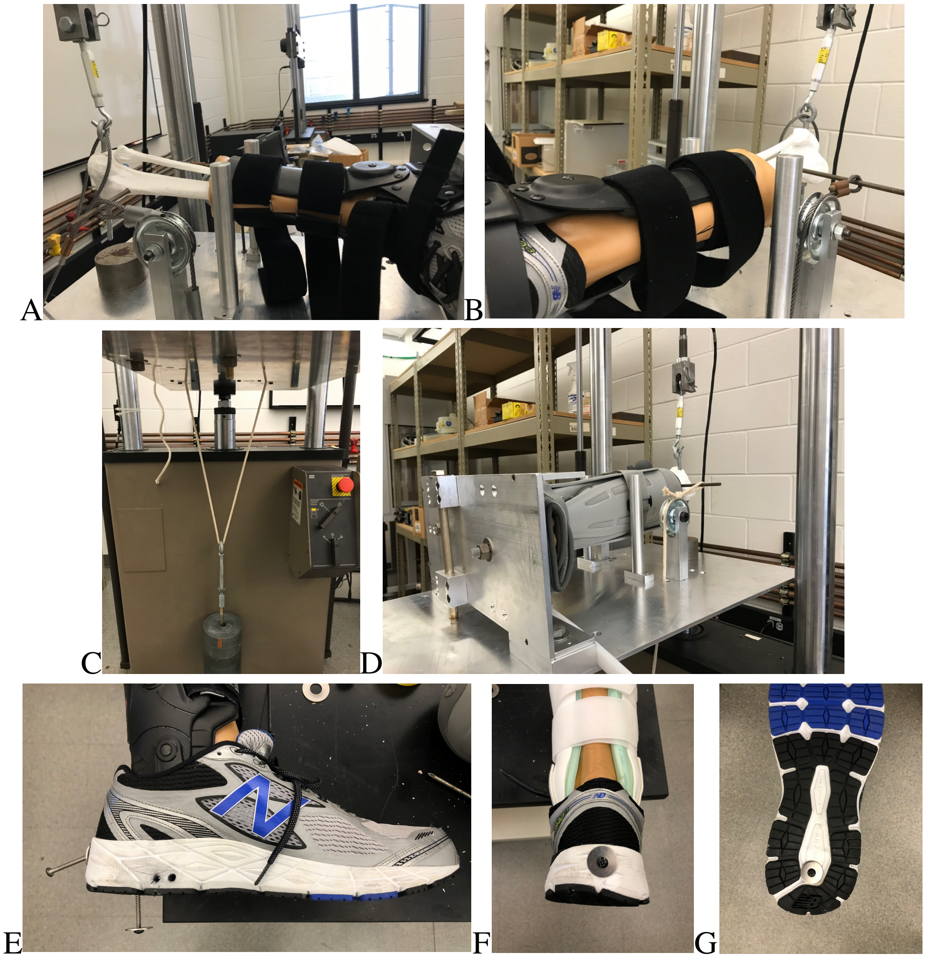


Fig. 3: A) Anterior view of test configuration for eversion of the left ankle with a long TayCo brace. B) Posterior view of the same setup. C) Weights on a cord were used to apply an axial load to the tibia during testing through a pulley system. D) The walking boot was bolted to the foot plate to maintain a rigid connection. E) F) G) Side, back, and bottom view of screws through the shoe and into the heel.

Inversion measurements were also performed with the foot in 20° plantar flexion with no brace and with the Long TayCo brace (Fig. 4).

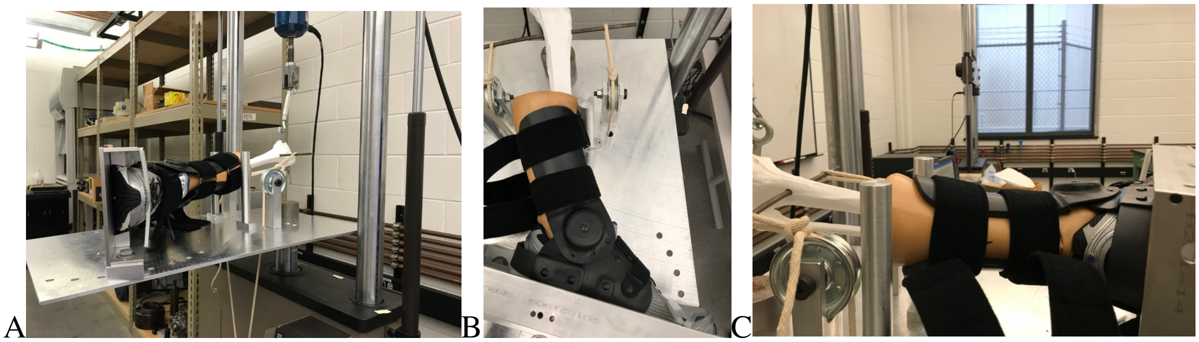


Fig. 4. A) Anterior view of the test configuration with the ankle in 20° of flexion. B) Top view of the same configuration. C) Posterior view.

For all tests, the force displacement curves were highly linear (Fig. 4). The slope of the cure was determined by linear regression, resulting in measurement of the applied force per inch of displacement. The distance from the ankle joint to the point of load application was 15 inches, and the force was converted to a moment by multiplying the force by the 15 inch moment arm. The displacement was similarly converted to an angle using the approximation $\sin(\theta) \approx \theta$ (in radians) for small angles. The total angle of ankle version was less than 4°, for which the error in this approximation is less than 0.004°.

Since the foam bone ankle is only a representation of a true ankle, the best measure of the effect of the braces is to determine the difference in stiffness between the unbraced ankle and the braced conditions. Since all of the force-displacement relationships were linear, the principal of superposition applies, and the contributors to the stiffness can be decomposed additively.

Inversion

The walking boot provided the greatest contribution to stiffness in inversion, followed by the long TayCo brace and the short TayCo brace (Fig. 5). The walking boot contributed 56.6% greater inversion resistance than the short TayCo brace and 18.2% more than the long TayCo brace. The long brace provided 32.5% more resistance than the short brace.

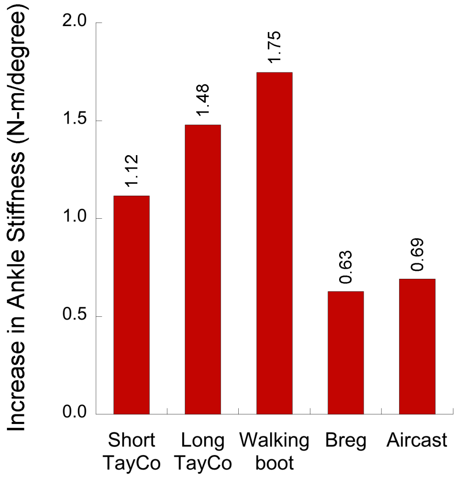


Fig. 5. The walking boot increased the resistance to inversion by the greatest degree. Both TayCo braces increased the inversion resistance by twice as much as the Breg or Aircast.

Eversion

The long TayCo brace provided the greatest resistance to eversion. The contribution to the eversion resistance was 54.7% higher than the walking boot, and nearly twice as high as the short TayCo brace. The walking boot contributed 26.3% greater eversion resistance than the short TayCo brace (Fig. 6).

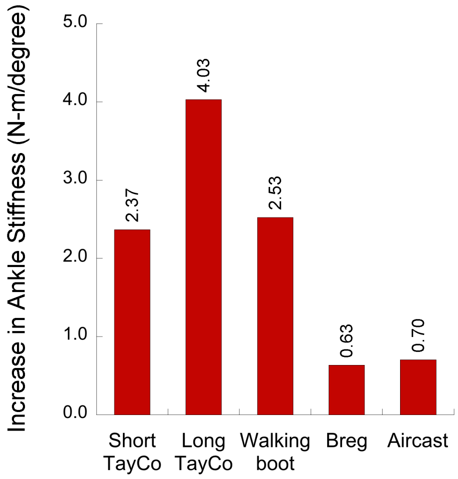


Fig. 6. The long TayCo brace increased the resistance to eversion by nearly twice as much as the short TayCo brace. The walking boot was similar to the short TayCo brace. The Breg and Aircast braces provided much lower support in eversion.

Plantar flexed foot

Only the long TayCo brace was tested with a plantar flexed foot, because the walking boot does not allow flexion of the ankle. The three locking screws were removed from the brace to allow flexion. The foot was in 20° plantar flexion, and the tibia was displaced medially to the ankle. In this configuration, the brace increased the inversion resistance of the ankle by 0.9404 N-m/deg (Fig. 7). This was a 14.2% increase in stiffness from the unbraced ankle.

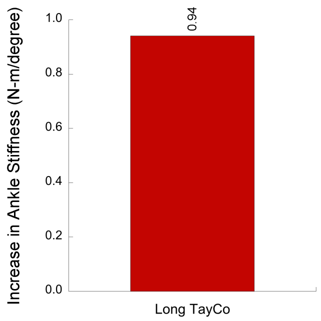


Fig. 7. The TayCo brace increased the inversion resistance of the ankle by 0.94 N-m/deg.

Discussion

The data indicate that the long TayCo brace is much more effective at resisting ankle version than the short brace. It is comparable to a walking boot. The advantage over the walking boot is the ability to allow flexion of the ankle by removing the three locking screws.

The results represent the stiffness of the construct under a reasonably high axial load, but lower than body weight and much lower than the force applied at heel strike during walking or running. The results are also consistent with low testing rates and loads. Higher loads could fracture the brace.

The TayCo brace outperformed the walking boot in eversion. However, eversion is an uncommon injury mechanism. After an ankle sprain, the lateral ligaments are likely to be injured, and additional support is needed to resist inversion of the ankle.

The injury limit of the ankle is most often defined by 30° to 40° of inversion, rather than applied moment (1). This reflects the limits on stretching of the ligaments. While different individuals may have ligaments of differing cross-sections and, therefore, stiffness, the maximum extension of ligaments is similar for all individuals. The moment resisted by the ligamentous structures at this point approaches an asymptote, reaching about 10 N-m (2). For all of the brace constructs tested, the resisting moment of the brace would exceed 10 N-m at approximately 10° of either inversion or eversion.

The use of the foam foot/ankle/shank model complicates direct interpretation of the mechanics of the braced ankle. Measurements of cadaver ankles suggest that the ankle has almost no resistance to inversion/eversion for up to 5° of motion (2). The moment at 10° of inversion is less than 25 kg-cm (2.45 N-m), and that in eversion is only slightly higher. This is predicted to be replicated *in vivo* due to the time required for the inversion/eversion muscles to fire during foot plants or landing from a jump (3). In contrast, the foam model had a resisting moment of approximate 20 N-m at 3.8° of version in either direction. However, given the stiffness of all of the braces, the resistance to version would increase much more rapidly than the unbraced ankles. In the case of the short TayCo brace, the moment borne by the brace increases at 1.115 N-m/deg, and would exceed the contribution of the ankle ligaments at only a few degrees of version in either direction (2). It is likely that the bending resistance increases more rapidly as the angle increases, but this was not tested to avoid damage to the braces and the artificial ankle.

Some error was unavoidable. The testing fixture had finite stiffness. An estimate suggests that in the worst-case scenario, about 10% of the deflection may come from the fixture (see the appendix). However, when the results are converted to moment/degree, and the stiffness of the unbraced leg is subtracted, the resulting stiffness should represent the incremental stiffness of the brace within this linear range. That is the stiffness of the foam foot-ankle-shank complex and the stiffness of the fixture are captured in the measurement with no brace. These can be subtracted to understand the additional contribution of the brace, because the load-displacement curves were linear. An additional source of error was potential motion between the shoe and foot plate. Visual observation indicated that this was minimized under the applied axial load, but it is an unmeasurable error. The walking boot was firmly bolted to the foot plate, but some motion may still occur due to flexion of the plastic on the sole of the boot.

References

1. Aydogan U, Glisson RR, Nunley JA., 2006, “Extensor Retinaculum Augmentation Reinforces Anterior Talofibular Ligament Repair,” *Clin Orthop Relat Res*., 442:210–215.

2. Chen, J., Siegler, S., Schneck, C.D., 1988, “The Three-Dimensional Kinematics and Flexibility Characteristics of the Human Ankle and Subtalar Joint—Part II: Flexibility Characteristics,” *J Biomech Eng*., 110(4): 374-385.

3. DeMers M.S., Hicks J.L., Delp S.L., 2017, “Preparatory co-activation of the ankle muscles may prevent ankle inversion injuries,” *J. Biomech.,* 52:17-23

Appendix: Estimate of fixture stiffness

The deflection of the fixture under the applied loads should be considered compared to the deflection of the ankle model.

First approximate the stiffness of the device using properties of aluminum and the cross-sectional geometry:

The modulus of aluminum is E=69 GPa = 69000 $\frac{N}{mm^2}$

The cross-sectional 2nd area moment is $I = \frac{bh^3}{12}$ where *h* is the thickness of the plate and *b* is the width. For this case, $b = 444.5\ mm, \quad h = 6.35\ mm$.
The bending stiffness is hence $EI = 654.4 \times 10^6\ N \cdot mm^2 = 654.4\, N \cdot m^2$.

The change in angle at the point of moment application and the deflection at the proximal tibia can be approximated in terms of the applied load. The change in angle due to the load is to negative contribution from the bending moment and a positive contribution from the applied load:

$$\phi = \frac{PL^2}{2EI} + \frac{ML}{EI} = \frac{PL^2}{2EI} + \frac{(-PL)L}{EI} = P\frac{(\tfrac{1}{2}m)^2}{2 \cdot 654.4 N \cdot m^2} - P\frac{(\tfrac{1}{2}m)^2}{654.4 N \cdot m^2} = -\frac{P}{5235N}\, radians$$

where *P* is the load, *M* is the moment, and the distance from the applied load to the fixation of the foot plate is approximately ½ *m*.

The change in deflection is the sum of the deflection due to the load, the deflection due to the moment, and a rigid body deflection of the tibia due to the imposed angle from the moment and the load:

$$\delta = \frac{PL^3}{3EI} + \frac{ML^2}{2EI} + \phi L = \frac{PL^3}{3EI} + \frac{(-PL)L^2}{2EI} + \frac{P}{5235N}(-L)$$

$$\delta = P\frac{(\tfrac{1}{2}m)^3}{3 \cdot 654.4 N \cdot m^2} - P\frac{(\tfrac{1}{2}m)^3}{654.4 N \cdot m^2} - P\frac{1}{5235N}(-\tfrac{1}{2}m)$$

$$\delta = P\left[\frac{1}{15705} - \frac{1}{5235} + \frac{1}{10470}\right] = -\frac{1}{31410} = \left(-31.84 \times 10^{-6}\ {}^{m}\!/\!_{N}\right)P$$

Hence, the combined effects of the moment and loading on the fixture account for 2.1 *mm* of deflection, or less than 10% of the applied 25 mm deflection when the load was 67 N, which was approximately the maximum. Moreover, the slope of the load-deflection curve for the fixture is 31410 N/m compared to approximately 3000 N/m stiffness measured, which is an order of magnitude different, and results in less than 10% error in the calculated stiffness of the construct.