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Biomechanical comparison of fixation stability using a Lisfranc plate versus transarticular screws☆

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ABSTRACT

Background: To obtain adequate fixation in treating Lisfranc soft tissue injuries, the joint is commonly stabilized using multiple transarticular screws; however iatrogenic injury is a concern. Alternatively, two parallel, longitudinally placed plates, can be used to stabilize the 1st and 2nd tarsometatarsal joints; however this may not provide adequate stability along the Lisfranc ligament. Several biomechanical studies have compared earlier methods of fixation using plates to the standard transarticular screw fixation method, highlighting the potential issue of transverse stability using plates. A novel dorsal plate is introduced, intended to provide transverse and longitudinal stability, without injury to the articular cartilage.
Methods: A biomechanical cadaver model was developed to compare the fixation stability of a novel Lisfranc plate to that of traditional fixation, using transarticular screws. Thirteen pairs of cadaveric specimens were tested intact, after a simulated Lisfranc injury, and then following implant fixation, using one method of fixation randomly assigned, on either side of each pair. Optical motion tracking was used to measure the motion between each of the following four bones: 1st metatarsal, 2nd metatarsal, 1st cuneiform, and 2nd cuneiform. Testing included both cyclic abduction loading and cyclic axial loading.
Results: Both the Lisfranc plate and screw fixation method provided stability such that the average 3D motions across the Lisfranc joint (between 2nd metatarsal and 1st cuneiform), were between 0.2 and 0.4 mm under cyclic abduction loading, and between 0.4 and 0.5 mm under cyclic axial loading. Comparing the stability of fixation between the Lisfranc plate and the screws, the differences in motion were all 0.3 mm or lower, with no clinically significant differences ($p > 0.16$).
Conclusions: Diastasis at the Lisfranc joint following fixation with a novel plate or transarticular screw fixation were comparable. Therefore, the Lisfranc plate may provide adequate support without risk of iatrogenic injury to the articular cartilage.

1. Introduction

Lisfranc injuries have been reported at a rate of approximately 1 per 55,000 people per year [1]. Injury to the Lisfranc joint can range from low energy falls to high energy crushes such as automobile accidents [1–4]. Depending on the type of impact, the injury may result in anything from small bone avulsion fractures to complete ligament ruptures and midfoot dislocations [5]. If misdiagnosed or left untreated, Lisfranc injuries may lead to a progressive flatfoot deformity or midfoot instability, along with symptomatic osteoarthritis in the future [5].

Successful treatment of Lisfranc injuries requires sufficient stabilization and fixation of the joint space. Open reduction and internal fixation (ORIF) or arthrodesis of the Lisfranc joint has been found to be essential in treating a Lisfranc injury [6,7]. To obtain adequate fixation for an ORIF, the joint is commonly stabilized using multiple transarticular screws, allowing for reduction and

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presumed ligament repair; however iatrogenic injury to the articular surfaces is a concern with this method. Alternatively, two parallel, longitudinally placed plates can be used to stabilize the 1st and 2nd tarsometatarsal joints [8]; however this method may not provide adequate stability in the transverse plane or along the Lisfranc ligament. Several biomechanical studies have compared earlier methods of fixation using plates to the standard transarticular screw fixation method, highlighting the potential issue of transverse instability using plates [8–10]. To address shortcomings of both plates and transarticular screws, a novel dorsal plate has been developed that is intended to provide transverse as well as longitudinal stability, without risk of injury to the articular cartilage.

For the present study, a biomechanical cadaver model was developed to compare the fixation stability of the novel Lisfranc dorsal plate to that of a transarticular screw fixation. The objective of this study was to quantify and compare the stability provided by the novel Lisfranc dorsal plate to the stability provided by transarticular screw fixation.

2. Methods

2.1. Specimen preparation

Fifteen matched pairs of fresh-frozen cadaveric lower legs were obtained to conduct a paired analysis of two methods of Lisfranc injury fixation. Each specimen was scanned using a Hologic 2000 bone densitometer (Hologic, Inc., Waltham, MA) to determine the bone mineral density (BMD) in g/cm^2 of the region of interest (1st metatarsal, 1st cuneiform, 2nd metatarsal, and 2nd cuneiform). Additionally, plain radiographs were taken to document the initial conditions of all specimens. After these initial scans, two pairs of legs were excluded from the study due to poor bone quality (98 year old) and severe pre-existing untreated musculoskeletal disorders (untreated talipes equinus) leaving thirteen pairs total for further testing (N = 26).

All specimens were fixed in 15° dorsiflexion, with respect to the tibia axis, to correspond with the position of the foot during maximum loading during gait. To achieve this, a 3.5″ screw was inserted through the calcaneus, talus, and tibia. To further stabilize the hindfoot a second screw was inserted through the navicular and calcaneus (Fig. 1).

For each specimen the Lisfranc joint space was exposed without disrupting the ligaments, muscles, or tendons, so that motion tracker flags could be secured to each of the four bones: 1st metatarsal, 2nd metatarsal, 1st cuneiform, and 2nd cuneiform.

2.2. Biomechanical loading

Previous studies using cadaver models have fixed the hindfoot in 30° of plantarflexion, with the goal of maximizing the strain on the Lisfranc joint [8,11]. For the present study our goal was to create maximum shear forces between all the joints that comprise the Lisfranc complex. In plantarflexion, these joints may be compressed longitudinally, with little shear forces applied.

The maximum ground reaction force is applied during the stance phase of gait, on the plantar aspect of the medial phalanges [12]. The position of maximum ground reaction load is at approximately 75%–80% of the stance phase corresponding to a maximum dorsiflexion of close to 15°. With the foot positioned in about 15° dorsiflexion, 2° eversion, and 1° abduction [13,14], the forces acting on the foot are a vertical force equal to 1.2*Body weight (BW), a medial force equal to 0.5*BW, and an anterior force equal to 0.15*BW [13].

Therefore, in order to accurately recreate this loading scenario, we held the foot in about 15° dorsiflexion, 2° eversion, and 1° abduction, and applied the resultant force to the plantar aspect of the medial phalanges. Furthermore, as much as possible, the load was applied through the big toe so as to maximize shear forces at the Lisfranc joint.

2.2.1. Abduction loading

Based on the above estimations, during the toe-off position of gait, the medial forefoot experiences an abduction force equaling about 5% of BW; roughly equivalent to the component of the ground reaction force in the medial lateral direction [13,15]. Considering a weight of 200 lbs, typical of a moderately overweight

Fig. 1. Screw placement through (1) Calcaneus, talus, and tibia, and (2) Navicular and Calcaneus.

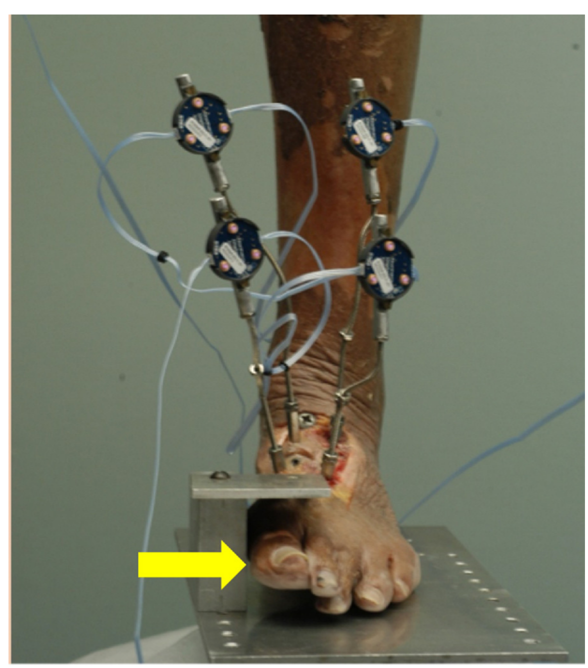

Fig. 2. Abduction loading set-up: Tibia attached to MTS actuator via ring and pointed screws, 1st metatarsal against metal bar on flat plate attached to load cell.

person, the medial lateral component of the ground reaction force acting on the toe is 44.5 N. Therefore, cyclic loading was applied to a peak load equal to a 44.5 N abduction load acting on the 1st metatarsal simulating the toe-off position (Fig. 2).

A flat plate was attached to the top of a load cell aligned with the actuator of the biaxial MTS 858 MiniBionix servohydraulic load frame (MTS, Eden-Prairie, MN). The proximal tibia was rigidly attached to the actuator using a custom made cylindrical ring with several pointed-tip screws inserted transversely into the bone for additional fixation. Holes at 1″ increments were created along the length of the plate that allowed varied placement of a metal bar. A small plate was attached to the top of the bar to prevent the foot from sliding up. With the bar secured next to the 1st metatarsal, the rotation of the leg via the actuator created an abduction force on the foot, which was measured by the load cell as a torsional moment. The maximum torque was adjusted based on the distance of the applied load to the center of the load cell, so that the resulting force on the 1st metatarsal would be 44.5 N.

For the intact state, each specimen was first preconditioned for 200 cycles at 0.5 Hz prior to the 10-cycle loading step at 0.1 Hz similar to previous biomechanical studies [2]. For the injured state, each specimen was again loaded for 10 cycles at 0.1 Hz. After fixation, each specimen was first preconditioned for 200 cycles at 0.5 Hz prior to the 10 cycle loading step at 0.1 Hz.

2.2.2. Axial loading

To replicate the loading experienced by the forefoot during the toe-off position of the gait cycle, an axial load was applied to the head of the first metatarsal of the foot. The tibia was rigidly attached to the actuator of the MTS load frame using a custom made cylindrical ring with pointed screws inserted into the bone. A raised metal bar was attached to the center of the load cell so the load would only be applied to the first metatarsal, creating a shear force in the fixation device. Additionally, to avoid the foot from slipping off the bar under higher loads, a small platform was attached to the top of the bar and zip-ties were used to secure the toe to the bar and plate. Finally, the six-degree-of freedom load cell, which was attached to gimbals, which controlled rotation in the coronal and sagittal planes, was rotated 15° to measure the forces and moments in the proper anatomical planes.

Each specimen was loaded cyclically to a maximum of 230 N, similar to previous studies (Fig. 3) [11,16]. Finally, a load to failure test was performed with loading steps consisting of 15 cycles, increasing in amplitude by 5N increments until failure or axial displacement in excess of 80.0 mm.

2.3. Injury simulation

After initial loading, a Lisfranc injury was created in each specimen. The injury model was based on previously established model by Alberta et al. [8]; as follows: we transected the dorsal, plantar, interosseous, and plantar Lisfranc ligaments, as well as the Lisfranc and 1st and 2nd TMT joint capsules (Fig. 4). In order to ensure a complete transection of the Lisfranc ligaments, the intercuneiform ligament between the medial and middle cuneiform was only partially transected. Unlike Alberta et al, we did not transect all five TMT joints because we felt that transection of the first two TMT joints was sufficient in injuring this complex, without adding unnecessary instability in the entire keystone structure. This injury would be classified as a B1 by the Myerson classification system [17]. An orthopaedic surgeon with extensive clinical experience with these injuries and implants created all simulated injuries.

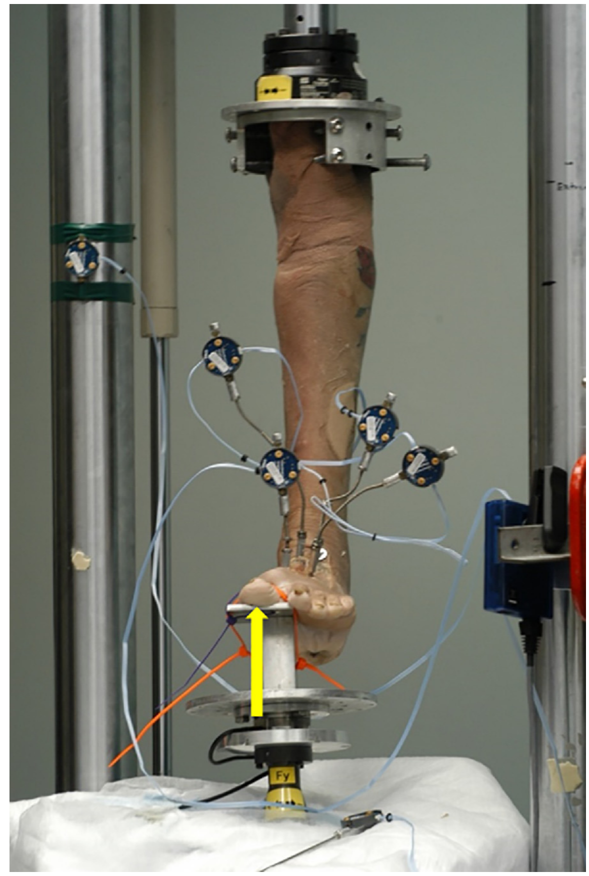

Fig. 3. Axial loading set-up: Superior end of tibia attached to MTS actuator via ring and pointed screws, 1st metatarsal positioned over and secured to metal bar and plate attached to load cell (blue circles are motion tracker flags rigidly attached to the 4 bones in the foot) (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.).

2.4. Fixation with Lisfranc plate or transarticular screws

Following intact and injured testing, each specimen was fixed with either transarticular screws or the novel Lisfranc plate. All surgeries were conducted using the surgical instrumentation and protocol described by the manufacturer. For Lisfranc plate fixation, one leg of each pair was randomly selected to have the Lisfranc Plate (Arthrex Inc., Naples, FL) implanted. One locking screw and three non-locking compression screws were used to attach the plate to the diaphyseal aspect of the bones. The screws were placed in the following order: Non-locking in medial cuneiform then an eccentrically placed non-locking in the 2nd MT, followed by a non-locking in the 1st MT, and finally a locking in the middle cuneiform. For transarticular screw fixation the contralateral leg was fixed using three 3.5 mm solid, fully threaded, cortical transarticular screws (one across the Lisfranc joint, one across the 1st TMT joint, and one across the 2nd TMT joint, Arthrex Inc., Naples, FL). No screws were placed from the medial to middle cuneiform, as this was not part of the injury pattern. (Fig. 5). All procedures were performed by an orthopaedic surgeon, experienced with both methods of fixation.

Following fixation, the previously described cyclic loading protocol was repeated. The specimens were then axially loaded until failure or until the axial displacement exceeded 80 mm.

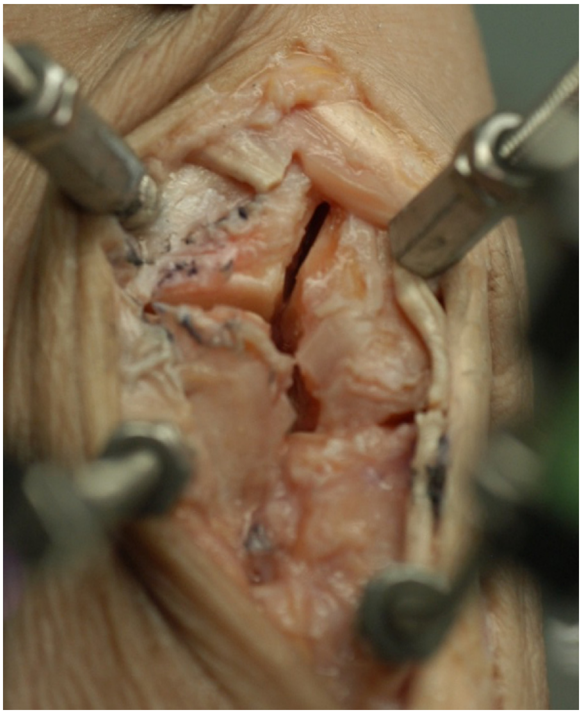

Fig. 4. Specimen with simulated Lisfranc injury: transected dorsal, plantar, interosseous, and plantar Lisfranc ligaments, as well as the Lisfranc and 1st and 2nd TMT joint capsules.

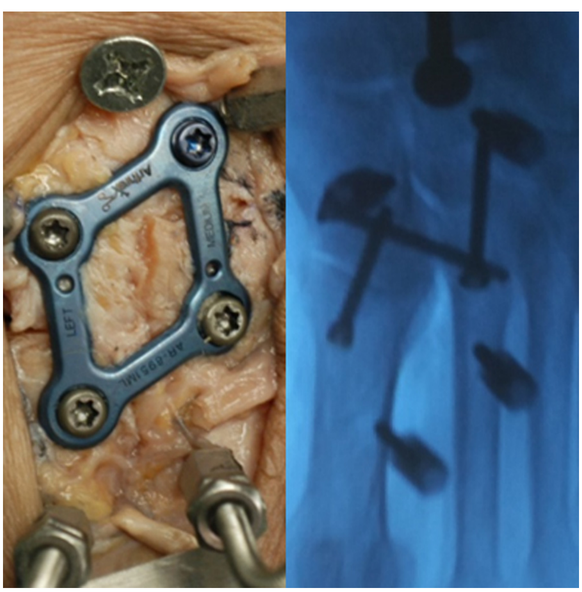

Fig. 5. Fixation methods: Left: Lisfranc Plate (Arthrex Inc.), Right: 3 transarticular srews.

2.5. Data acquisition and analysis

Six-degree-of-freedom motions of the bones were measured during cyclic loading. An Optotrak Certus Motion Capture System (Northern Digital Inc, Waterloo, Ontario, Canada) was used to record the 3D positions of the motion tracker flags during testing. The motions were normalized using values from the unloaded, or neutral position. From this, the motions of the bones relative to each other during the peak loads of the 10 cycle loading steps of both abduction and axial loading were determined. Specifically, for each state (intact, injured, and fixed) and loading scenario

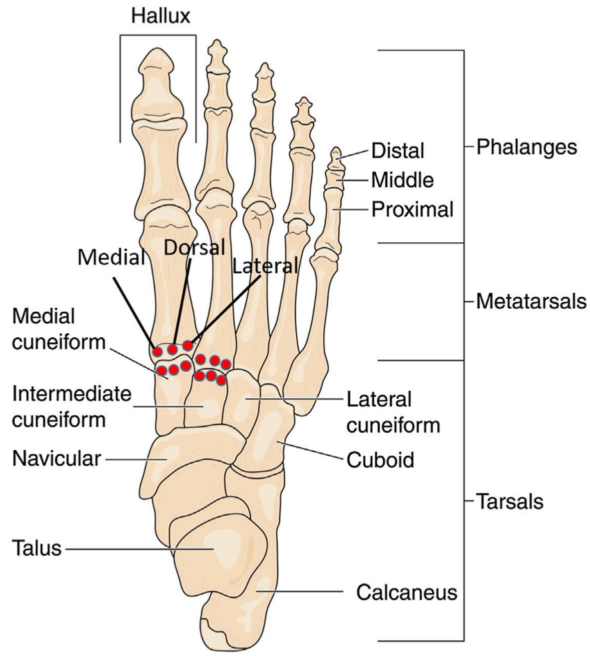

Fig. 6. Red dots indicate points that were digitized on each bone (medial, dorsal, and lateral) (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.).

(abduction and axial) we calculated the 3D distances and individual distances along the x, y, and z-axes between the following joints: (a) 1st metatarsal and 1st cuneiform (1M-1C), (b) 2nd metatarsal and 2nd cuneiform (2M-2C), (c) 1st metatarsal and 2nd metatarsal (1M–2 M), (d) 2nd metatarsal and 1st cuneiform (2M-1C) (Lisfranc joint).

Prior to loading the specimen 3 points of interest were digitized on each bone (medial, lateral, dorsal) (Fig. 6). To analyze the 3D motion, the magnitude of the vector between the dorsal point on each bone comprising the joint of interest, was calculated.

2.6. Data reduction and statistical analysis

Matched pair experimental design was used in the current study. A total of 13 pairs of lower legs were used (N = 26). Motion between all points of interest were calculated in the axial direction during axial loading, and in the mediolateral direction during abduction loading. In addition, the 3-Dimensional motion between all points of interest were calculated under both axial loading and abduction loading. These motion calculations were repeated for the same specimens under intact, injured, and implanted conditions. Differences were calculated for each motion as follows: (Intact – Injured) and (Injured – Implanted). In addition, differences were calculated for each motion between contralateral sides of each pair as follows: (Plate – Screw), that is motion of each leg fixed with the Lisfranc plate minus corresponding motion of each leg fixed with the transarticular screws. The means and standard deviations of all paired differences are presented in tables. Paired t tests were used to compare differences in means of each motion for all of the pairs listed above. All data analysis was performed using SPSS Statistics Version 17.0 (IBM, Houston Texas). A power analysis was conducted prior to commencement of the study, based on the results of previous similar cadaver studies [8]. Given 80% statistical power and alpha of 0.05, this initial power analysis indicated that approximately 16 pairs of cadaver legs would be sufficient. In-house pilot studies indicated that with 13 pairs, it would be possible to detect differences of 0.29 mm or larger with statistical significance.

3. Results

3.1. Overview of 3D motions under abduction and axial loading

The 3D motion between the bones in each of the following joints were calculated: 1M-1C, 2M-2C, 1M-2M, and 2M-1C (Lisfranc joint). In all, 13 cadaveric legs were used; one side of each cadaver was assigned to the Lisfranc plate group, the other, to the screw group.

In the Lisfranc plate group, 3D motion of the Lisfranc joint under abduction load ranged from −0.6 to 0.6 mm, (mean 0.1 mm ± 0.4) (Positive values indicate an increase in distance under load suggesting diastasis, whereas negative values indicate a decrease in distance under load suggesting compression). After creation of simulated injury, the motion ranged from −0.4 to 2.0 mm (0.6 mm ± 0.6). After fixation with the Lisfranc plate the 3D motion ranged from −0.6 to 1.4 mm (0.4 mm ± 0.5). Among the feet assigned to the screw group, 3D motion of the Lisfranc joint under abduction load ranged from −0.6 to 0.7 mm (0.4 mm ± 0.3). After creation of simulated injury, the motion ranged from −0.7 to 0.95 mm (0.4 mm ± 0.4). After fixation with the screw the 3D motion ranged from −0.8 to 0.8 mm (0.15 mm ± 0.6).

In the plate group, 3D motion of the Lisfranc joint under axial load ranged from −0.4 to 0.8 mm (0.5 mm ± 0.3). After creation of simulated injury, the motion ranged from −0.3 to 1.7 mm (0.7 mm ± 0.5). After fixation with the Lisfranc plate the 3D motion ranged from −0.4 to 1.0 mm (0.4 mm ± 0.4). Among the feet assigned to the screw group, 3D motion of the Lisfranc joint under axial load ranged from −1.0 to 2.3 mm (0.5 mm ± 0.8). After creation of simulated injury, the motion ranged from −0.4 to 1.4 mm (0.7 mm ± 0.3). After fixation with the screw the 3D motion ranged from −0.6 to 1.2 mm (0.5 mm ± 0.4).

3.2. Differences in 3D motions under abduction loading

Under abduction loading, in the Lisfranc plate group, following creation of a simulated injury, the 3D motion of the Lisfranc joint increased significantly by 0.5 mm ± 0.8 ($p = 0.05$). The motions of all of the other joints (1M-1C, 2M-2C, 1M-2M), did not change by more than 0.1 mm ($p > 0.22$) (Table 1). After fixation with the Lisfranc plate, the 3D motion of the Lisfranc joint decreased, but not significantly. Likewise, motions of the 1M-1C and 1M-2M joints did not change significantly, whereas motion of the 2M-2C decreased by 0.3 ± 0.5 ($p = 0.05$).

In the screw fixation group, following creation of a simulated injury, the 3D motion of the Lisfranc joint did not change ($p = 0.87$) (Table 1). Among the other four joints, the motion of the 2M-2C joint increased by 0.2 mm ($p = 0.04$), whereas the remaining joints did not change significantly. Following fixation with screws, 3D motion of the Lisfranc joint and all of the other joints decreased slightly; however, the decrease was only significant for the 2M-2C joint (0.5 ± 0.8 mm) ($p = 0.03$).

3.3. Differences in 3D motions under axial loading

Under axial loading, in the Lisfranc plate group, following creation of a simulated injury, the 3D motion of the Lisfranc joint increased 0.2 mm ± 0.3 ($p = 0.02$) (Table 2). 3D motions of the remaining four joints (1M-1C, 2M-2C, 1M-2 M), did not change significantly ($p > 0.15$) (Table 2). After fixation with the Lisfranc plate, the 3D motion of the Lisfranc joint decreased by 0.3 mm ± 0.5 ($p = 0.06$), though this was not statistically significant at the $p \leq 0.05$ level. Likewise, 3D motions of the 2M-2C and 1M-2M joints decreased by approximately 0.5 mm, whereas 3D motions did not significantly change.

In the screw fixation group, following creation of a simulated injury, the 3D motion of the Lisfranc joint decreased by 0.2 mm ± 0.4 ($p = 0.06$) (Table 2), though this was not statistically significant at the $p \leq 0.05$ level. 3D motions of the other four joints did not significantly change ($p > 0.09$) (Table 2).

3.4. Comparisons of motions between plate and screw fixation groups

Under abduction loading, differences in mediolateral motion between the Lisfranc plate and the screw fixation groups ranged from 0.1 to 0.2 mm, with no significant difference for any of the joints ($p > 0.30$) (Table 3). Similarly, the differences in 3D motion were not significantly different between the two groups ($p > 0.16$) (Table 3).

Under axial loading, differences in shear motion between the Lisfranc plate and the screw fixation groups ranged from 0.0 to 0.34 mm, with no significant difference for any of the joints ($p > 0.16$) (Table 3). Similarly, the differences in 3D motion were not significantly different between the two groups ($p > 0.21$) (Table 3).

3.5. Cyclic loading to failure

Under cyclic loading to failure of specimens fixed with both the Lisfranc plate and screws, all fixed joints maintained stability while the amplitude of load was increased. In the plate fixation group amplitude of cyclic loading was increased up to an average of 459 N (range 340–589 N). Similarly in the screw fixation group, the amplitude of loading increased up to an average of 463 N (range 275–757 N). There was no significant difference between the two groups. All of the specimens reached more than 80 mm of axial actuator displacement at these maximum loads. As indicated in the Methods section, the experiments were stopped at this large displacement since it exceeded clinically relevant range of joint motion.

4. Discussion

Proper management techniques of a Lisfranc soft tissue injury is vital to a successful outcome. It has been reported that successful

Table 1
The first column shows the average differences in 3D motion between Intact and Injured conditions, that is, Intact motion minus Injured motion, along with the standard deviation in parenthesis. The second column indicates the p value for the paired t-test comparing those two conditions. Similarly, the following columns show Injured minus Fixed conditions, along with associated p values. The same pattern is then repeated for the screw fixation group. Positive values represent diastasis, whereas negative values represent compression.

3D motion under abduction loading (mm)								
	Plate				Screw			
	Intact–Injured	*p value*	*Injured–Fixed*	*p value*	*Intact–Injured*	*p value*	*Injured–Fixed*	*p value*
1M-1C	0.0 (0.5)	0.92	0.0 (0.3)	0.64	0.0 (0.3)	0.99	0.0 (0.2)	0.56
2M-2C	0.0 (0.4)	0.66	0.3 (0.5)	0.05	−0.2 (0.4)	0.04	0.5 (0.8)	0.03
1M-2M	0.1 (0.3)	0.22	0.3 (0.6)	0.12	−0.2 (0.3)	0.08	0.3 (0.7)	0.11
Lisfranc joint: 2M-1C	−0.5 (0.8)	0.05	0.1 (0.8)	0.57	0.0 (0.6)	0.87	0.3 (0.7)	0.19

Table 2
The first column shows the average differences in 3D motion between Intact and Injured conditions, that is, Intact motion minus Injured motion, along with the standard deviation in parenthesis. The second column indicates the p value for the paired t-test comparing those two conditions. Similarly, the following columns show Injured minus Fixed conditions, along with associated p values. The same pattern is then repeated for the screw fixation group. Positive values represent diastasis, whereas negative values represent compression.

3D Motion Under Axial Loading (mm)								
	Plate				Screw			
	Intact–Injured	*p value*	*Injured–Fixation*	*p value*	*Intact–Injured*	*p value*	*Injured–Fixation*	*p value*
1M-1C	−0.2 (0.3)	0.04	−0.1 (0.7)	0.77	−0.2 (0.3)	0.09	0.0 (0.4)	0.92
2M-2C	−0.2 (0.7)	0.47	0.5 (0.8)	0.06	−0.3 (1.1)	0.40	0.7 (2.7)	0.35
1M-2M	0.1 (0.4)	0.64	0.6 (0.8)	0.02	−0.1 (0.4)	0.49	0.8 (1.5)	0.09
Lisfranc joint: 2M-1C	−0.2 (0.3)	0.02	0.3 (0.5)	0.06	−0.3 (0.9)	0.34	0.2 (0.4)	0.06

Table 3
The first column shows the average differences in Mediaolateral motion between Plate and Screw conditions, that is, Plate motion minus Screw motion, along with the standard deviation in parenthesis. The second column indicates the p value for the paired t-test comparing those two conditions. Similarly, the following columns show 3D Motion between Plate and Screw conditions, along with associated p values. For Axial loading, differences in Shear motion and 3D motion are presented, again following the format of Plate motion minus Screw motion. Positive values represent diastasis, whereas negative values represent compression.

Differences in Motion (Plate – Screw) (mm)								
	Abduction loading				Axial loading			
	Mediolateral motion	*p value*	*3D motion*	*p value*	*Shear motion*	*p value*	*3D motion*	*p value*
1M-1C	0.2 (0.6)	0.32	0.0 (0.4)	0.83	0.2 (0.5)	0.16	−0.1 (0.9)	0.82
2M-2C	0.0 (0.2)	0.89	−0.1 (0.4)	0.42	−0.1 (0.4)	0.63	−0.7 (1.8)	0.21
1M-2M	−0.1 (0.3)	0.30	0.0 (0.1)	0.35	0.0 (0.4)	0.97	0.0 (0.2)	0.99
Lisfranc joint: 2M-1C	0.2 (0.8)	0.45	0.3 (0.7)	0.16	0.3 (1.0)	0.21	−0.0 (0.6)	0.80

outcomes strongly correlate with an anatomic and stable reduction and is less dependent on severity and initial diastasis of the injury [17]. Conventionally, Lisfranc injuries have been repaired using transarticular screw fixation; however, there are potential disadvantages to this method of fixation [8,18,19]. Specifically, concern has been raised regarding the potential risk of iatrogenic injury to the articular cartilage and gradual development of arthritis. The novel Lisfranc plate evaluated in the present study was designed to provide dorsal fixation of the 1st metatarsal, 2nd metatarsal, 1st cuneiform, and 2nd cuneiform using multiple screws, away from the joint. By attaching the plate to the epiphyseal or metaphyseal aspect of the bones and avoiding any transarticular fixation, risk of injury to the joint is avoided. The present study is the first to evaluate the potential advantages of this novel plate using a biomechanical model under simulated physiological loading.

Specimens fixed with the Lisfranc plate had similar 3D motions across all 5 joints compared to specimens fixed with the transarticular screws during both abduction and axial loading (Table 3). During both axial and abduction loading, the largest differences in 3D motion between the two fixation groups were about 0.3 mm under abduction loading and 0.5 mm under axial loading; however, none of the differences were statistically significant. Our power analysis suggested that, given the standard deviations and sample size used, the present study was able to detect differences of 0.29 mm or larger with 80% statistical power and alpha of 0.05. Since 0.29 mm is a reasonable threshold for clinically significant amount of motion, the results suggest, with sufficient statistical certainty that the two fixation methods are comparable from a clinical standpoint [1].

Due to transarticular screws ability to limit diastasis, they are currently the most accepted method in stabilizing soft tissue Lisfranc injuries. Similar to Alberta et al, we found screw displacement allowed more movement than a plate fixation method with displacement slightly above 0.5 mm, under axial loading [8]. To stabilize the Lisfranc injury, Alberta et al implemented two 3.5-mm screws through each TMT joint for the screw fixation method, and two 5-hole $^{1}/_{4}$ tubular plates with four 2-mm screws through each plate across each TMT joint for the plate method. They reported their plate method to allow 0.15 mm of diastasis with a standard deviation of 1.2 mm. Although this movement was less than our reported movement of 0.4 mm, the difference could be attributed to the designation of specific points of interest (digitized points) for calculation of motion. While we used the “dorsal” point (dorsal point bisects medial and lateral points) of each bone, Alberta et al measured movement between the medial aspect of the second metatarsal base and the lateral aspect of the medial cuneiform. Due to the miniscule distance between these two points, measured movement may inherently be very small.

Another group led by Panchbhavi et al., compared screw fixation to suture-button fixation and found the screw fixation to allow a median displacement at the Lisfranc ligament of just over 0.2 mm [20]. The reported movement is on the same magnitude as our study, but approximately 0.3 mm less than our findings. The reason may be due to differences in the way artificial Lisfranc injury was created. Panchbhavi et al. described their injury creation process as “A number-11 blade was inserted deep between the bases of the first and second metatarsals, and then all intervening structures were incised proximally up to the middle of the intercuneiform joint between the medial and middle cuneiform bones [20].” Although this incision path does disrupt the main Lisfranc ligament, to accurately replicate both transverse and longitudinal instability from a Lisfranc injury, the 1st and 2nd tarsometatarsal joint must also be injured [16]. In contrast to Panchbvhavi et al., we disrupted both the 1st and 2nd TMT joints in addition to the plantar ligaments between the Lisfranc joint, likely resulting in a slightly higher overall 3D movement for screw fixation.

Pelt et al found average displacement of 1.0 mm which is higher as the motions measured in the present study. The differences in motion between the two studies may be easily explained by the different loading mechanisms. Specifically the loading in the present study was based on measurements during gait and in particular simulating the toe off motion, as explained in the methods section [13,14,21,22]. In contrast, Pelt et al. inserted a Kirschner wire, dorsal to plantar through the third metatarsal head, and used this wire to apply a "lateral pull" through the intermetatarsal ligaments, using a pneumatic actuator. Therefore, the loading in the present study was perhaps a more accurate simulation of physiological loading experienced during gait.

The present study compared two methods of ORIF for the fixation of ligamentous Lisfranc injuries: a dorsal Lisfranc plate versus transarticular screw fixation. The injury pattern that we used did not involve multiple columns that might be seen with high energy Lisfranc injuries. It should be noted that, as summarized in a recent systematic review of the literature, many studies have shown similar clinical results between ORIF and primary arthrodesis [23]. In this review, primary arthrodesis showed a reduction in the rate of surgical procedures for hardware removal as compared to ORIF. Primary arthrodesis also has the advantage of lower surgical cost. In fact, for higher energy Lisfranc injuries with multi-column involvement, a primary arthrodesis has shown superior clinical outcomes to ORIF [24].

Considering options for ORIF, our results are generally consistent with the clinical findings of Hu et al. They followed 60 patients on an average of 31 months and radiographically found 90.6% anatomic reduction for patients treated with dorsal plate fixation. On the contrary, they stated only 82.1% of patients treated with screw fixation showed successful anatomical reduction after 2 years. Along with successful reduction, Hu et al found 46.4% of patients treated with screw fixation suffered from degenerative joint disease, while only 25% of patients treated with dorsal plate fixation suffered from this [2]. Clinically this group has arrived at a similar conclusion as our study that the plate fixation may be biomechanically comparable to screw fixation while reducing the amount of future damage to the Lisfranc joint.

In addition to the previous clinical study, another study by Lau et al. found no significant difference when comparing plate or screw fixation outcome measurements. This group studied a total of 62 patients over a 6-year period and classified each fracture using the Hardcastle classification system. To measure a successful outcome, the Kellgren-Lawrence and Wilppula classification scales were used to report osteoarthritis and anatomic reduction. Although no specific benefits were found when comparing the two fixation methods this study showed fixation with dorsal plates or transarticular screws to be radiographically the same [10]. While we have demonstrated the similarities of the two fixation methods biomechanically, Lau et al. has clinically confirmed our findings.

The results of the present study are generally consistent with some clinical outcome reports. For example, Hu et al. compared ORIF and PA fixation to treat low energy Lisfranc injuries. They radiographically observed 25% of patients treated with ORIF dorsal plating to have degenerative joint disease while 46% of patients treated with PA to have degenerative joint disease. In addition to this analysis, they found successful anatomic reduction in 90% of patients treated with dorsal plating while only 82% of patients achieved sufficient reduction with PA[2]. Similar to Hu et al., Lau et al. used radiographic analysis to perform their study. Despite the similarities in the methods, Hu et al. found that "treatment with either transarticular screw/screws or dorsal bridge plating resulted in no significant difference." [10].

Contrary to the previously mentioned studies, Ly et al. showed that primary fusion produced a better outcome score than ORIF. However in this study Ly et al. compared ORIF to primary fusion for high energy Lisfranc injuries that involved all the columns of the foot, which is not the type of injury that pertained to this study [24]. Considering all of these previous clinical reports, as well as the results of biomechanical testing, the following factors should be taken into account in deciding on the treatment of choice for a given injury: Implant cost, severity of injury, bone quality, and general surgeon comfort with each method.

The present study had several limitations. First, the study implemented the use of cadaveric foot models. Another limitation was the inability to physically test the implants to failure. When testing to failure, either the bottom heel of the foot would be obstructed by the load cell or the load actuator would surpass its physical limitation of 80 mm displacement. Although we were not able to test to failure the amount of displacement and force seen prior to halting the test, surpassed realistic angular displacement and force. Finally, the present study did not consider other commonly used fixation methods, such as primary arthrodesis and suture-button fixation. Arthrodesis has become a viable route to healing soft tissue Lisfranc injuries and has been found to provide similar results to ORIF [25]. Suture-button fixation has also been shown to have similar fixation capabilities to transarticular screws [20]; however, we were able to compare two popular methods of a Lisfranc ORIF. By considering these limitations for future studies, an even more comprehensive profile of the novel Lisfranc plate's capabilities can be produced.

In summary, the results of the present study indicated that the novel Lisfranc plate method provided comparable stability of the joint compared to the current conventional transarticular screw method. The novel Lisfranc plate design may provide longitudinal stability that is comparable to the two plate design, while providing additional Lisfranc joint (1C-2M) stability. Future studies may address the biomechanical differences between the novel plate design and other unique methods in fixing Lisfranc injuries.

Conflicts of interest

None.

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