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Biomechanical Comparison of Intramedullary Screw Versus Low-Profile Plate Fixation of a Jones Fracture

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Abstract

Background: Intramedullary screw fixation of fifth metatarsal Jones fractures often produces satisfactory results, however, nonunion and refracture rates are not negligible. The low-profile "hook" plate is an alternative fixation method that has been promoted to offer improved rotational control at the fracture site, but this remains to be proven. The purpose of this study was to document biomechanical performance differences between this type of plate and a contemporary solid, dual-pitch intramedullary screw in a cadaveric Jones fracture model.

Methods: Simulated Jones fractures were created in 8 matched pairs of fresh-frozen cadaveric fifth metatarsals. One bone from each pair was stabilized using an intramedullary TriMed Jones Screw and the other using a TriMed Jones Fracture Plate (TriMed, Inc, Santa Clarita, CA). Controlled bending and torsional loads were applied. Bending stiffness and fracture site angulation, as well as torsional stiffness, peak torque, and fracture site rotation were quantified and compared.

Results: Intramedullary screw fixation demonstrated greater bending stiffness and less fracture site angulation than plate fixation during plantar-to-dorsal and lateral-to-medial bending. Torsional stiffness of screw-fixed metatarsals exceeded that of plate-fixed bones at initial loading; however, as rotation progressed, the plate resisted torque better than the screw. No difference in peak torque was demonstrable between fixation methods, but it was reached earlier in specimens fixed with screws and later in those fixed with plates as rotation progressed.

Conclusion: In this cadaveric Jones fracture model, intramedullary screw fixation demonstrated bending stiffness and resistance to early torsional loading that was superior to that offered by plate fixation.

Clinical Relevance: Although low-profile "hook" plates offer an alternative for fixation of fifth metatarsal Jones fractures, intramedullary screw fixation may provide better resistance to bending and initiation of fracture site rotation. The influence of these mechanical characteristics on fracture healing is unknown, and further clinical investigation is warranted.

Keywords: Jones fracture, intramedullary screw, fifth metatarsal, plate fixation

Introduction

Proximal fifth metatarsal metadiaphyseal fractures, as first described by Jones in 1902,[10] are notorious for their high rates of delayed union, nonunion, and refracture when treated nonoperatively.[4,11] These poor outcomes are understood to be a function of both biological and mechanical factors. The precarious blood supply of the region of the fifth metatarsal in which Jones fractures occur has been well documented.[16] Additionally, biomechanical evidence suggests that these unstable fractures have a propensity for displacement due to forces exerted on the proximal fracture fragment by the peroneus brevis tendon.[15] Given the adverse healing conditions unique to Jones fractures, primary fixation has become the treatment of choice, particularly in the active individual.

Although a variety of fixation techniques have been described for the operative management of Jones fractures, the predominant method is via intramedullary screw. This mode of stabilization is the most studied and reported in the literature, with favorable outcomes in terms of rates of

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union, time to union, and return to activities compared with nonoperative management.[3,5,11] However, despite improved results, complications specific to intramedullary screw fixation, including penetration of the cortex, missing the medullary canal, screw head prominence, and soft tissue damage from reaming, are not infrequent, with an incidence as high as 45% in one series.[11] Additionally, treatment failures including nonunion, delayed union, and refracture still occur with notable frequency.[7,12,22] Early return to vigorous activities and suboptimal screw size are thought to be predictive of failure.[7,12,22] The limited ability of intramedullary screw fixation to control relative rotation between the fracture fragments during healing has also been implicated. In a model of midstance gait in which cadaver feet were subjected to concurrent axial load and tendon forces, it was found that the proximal fifth metatarsal was subjected not only to bending stresses, but also torsional stresses.[21] Thus, inadequate resistance to either bending or torsional loads can negatively influence outcome after intramedullary screw fixation.

Low-profile, fracture-specific plates are an alternative fixation method for the treatment of Jones fractures. They have been theorized to provide improved rotational resistance at the fracture site. Although short-term clinical results from Level IV studies using this fixation method are promising,[2,14] to our knowledge, there are no available data on the biomechanical profile of these plates, particularly when compared to intramedullary screws. The purpose of this study was to quantify the bending and torsion resistance of Jones fractures stabilized with this type of fracture-specific plate compared to that afforded by a an intramedullary screw of contemporary design. Our hypotheses were that the tested plate would provide improved torsional resistance compared to screw fixation, and that screw fixation would better resist bending loads.

Methods

Anatomic Specimens

Eight matched pairs of fifth metatarsals were dissected from fresh-frozen cadaveric feet and stripped of soft tissue. The average donor age was 56 years (±14 years; 6 male, 2 female). To account for confounding differences in bone quality, the bone mineral density of each specimen was quantified by dual-energy x-ray absorptiometry (DEXA) scanning of the posterior calcaneus using a Lunar PIXImus small-animal densitometer (Lunar Corporation, Madison, WI). Quality assurance testing was performed using a phantom with a known density. The mean bone mineral density in the intramedullary screw group was 0.495 (±0.096) g/cm^3 compared with 0.540 (±0.105) g/cm^3 in the plate group, and the difference was not found to be statistically significant (P = .134).

Specimen Preparation

The surface of each metatarsal was degreased using an acetone-soaked gauze sponge. A 1.6-mm Kirschner-wire was inserted from proximal to distal into the metatarsal along its longitudinal axis, and the projecting wire was grasped in the chuck of a drill press oriented perpendicular to the machine's table. A 2.5-cm-diameter by 2.5-cm-length stainless steel tube was positioned upright on the drill press table and the chuck lowered until the metatarsal head was precisely centered in the tube. The machine's depth stop was set at that point, the chuck and metatarsal were retracted, and polymethylmethacrylate cement (Fastray, H.J. Bosworth Co, Skokie, IL) was poured into the tube. The metatarsal was lowered back into the tube and the cement was allowed to harden, encasing the metatarsal head in a cylindrical block. The tube and Kirschner wire were then removed.

Jones Fracture Creation

The polymethylmethacrylate cylinder encasing the metatarsal head was clamped in a machinist's vise such that the bone's longitudinal axis was horizontal. The vise was placed against the guide fence of a band saw, and the metatarsal was passed through the saw blade to produce a planar cut perpendicular to the shaft, at the level of the metadiaphyseal junction, directed toward the 4-5 intermetatarsal facet. The saw cut was made up to, but not completely through, the medial cortex in order to maintain precise fragment alignment during subsequent fracture fixation. The simulated fracture created was such that there was no interdigitation between the fragments. This was intentionally done to create a worse-case scenario in terms of rotational stability. The distance between the fracture site and the center of each specimen's distal head was then measured with a machinist's caliper and recorded for use in the subsequent bending stiffness and angulation calculations.

Intramedullary Screw Fixation

One metatarsal from each matched pair was fixed using a TriMed Jones Screw (TriMed, Inc, Santa Clarita, CA)[20] (Figure 1). This implant is a stainless steel, noncannulated screw that achieves compression by means of a pitch differential between the leading (shank) and trailing (head) threads. The head end of the screw has a larger diameter, a smaller thread pitch, and a slightly tapered form that produces a wedging effect when seated in the bone. The screw tip has a projection that facilitates locating the drill hole. The selection of leading thread diameters (4.5, 5.5, and 6.5 mm) and lengths (40-60 mm) allows sizing to be individualized to the metatarsal.

Screw insertion was performed using the manufacturer's instrumentation and recommended technique.[19] The inside

Figure 1. TriMed Jones Screws.

diameter (ID) of each metatarsal's intramedullary canal was measured on an anteroposterior radiograph, and the screw diameter that best matched the size of the bone was selected as follows: metatarsals with ID ≤4.0 mm received a 4.5-mm screw; ID = 4.0 to 4.5 mm, a 5.0-mm screw; and ID ≥4.5 mm, a 6.5-mm screw. Each bone's outside diameter, as determined on anteroposterior and lateral radiographic views, was verified to be adequate to avoid "canal blowout" with the selected screw diameter. Two metatarsals were fixed with 4.5-mm screws, 4 with 5.5-mm screws, and 2 with 6.5-mm screws. Screw length was selected by measuring the straight portion of the metatarsal diaphysis, proximal to the distal lateral curvature, from the anteroposterior radiographs and selecting a length that placed the leading screw threads distal to the fracture site but no further distal than necessary.

At the time of screw insertion, the screw guide-wire was placed from proximal to distal along the desired trajectory and the canal was prepared using the appropriate cannulated drill and tap. The screw was inserted to approximately two-thirds of its ultimate depth, and the simulated fracture was completed through the 4-5 intermetatarsal articular surface using a handheld microsagittal saw. The screw was then advanced until fracture compression was seen and the trailing (head) end was flush with the cortex.

Plate Fixation

The contralateral metatarsal from each pair was fixed using a TriMed Jones Fracture Plate (Figure 2) applied to the lateral aspect of the bone using the manufacturer's instrumentation and recommended technique.[18] This stainless steel, low-profile, precontoured plate was available in 2 sizes (5-hole and 7-hole) and was fastened to the bone with 2.3-mm nonlocking cortical screws with lengths ranging from 7 to 18 mm. The manufacturer's technique did not specify plate sizing criteria or number of screws for each fracture segment. In order to standardize fixation, we chose the 7-hole plate size, allowing each segment of the fracture to have 3 points of fixation. Proximally, this consisted of 2 unicortical screws and the intraosseous tines. Distally, 3 bicortical screws were used, including 1 screw placed eccentrically in the distal oblong hole to provide compression. Before using the compression instrument and securing this screw, the Jones fracture was completed using a handheld microsagittal saw.

Figure 2. TriMed Jones Fracture Plates.

After fracture fixation, the proximal end of each bone was embedded in an aluminum cup using polymethylmethacrylate cement in a manner similar to the embedding of the distal head. To enhance anchoring of the proximal fragment, 6 small stainless steel screws were inserted into its cortex before immersing it in the polymethylmethacrylate. In plated specimens, the portion of the plate extending proximal to the fracture line was covered with modeling clay to isolate it from the embedding material, as was the exposed surface of the screw head in screw-fixed specimens. Extreme care was taken to ensure the anchoring screws did not interact with the tested implants. A fixture was used during the embedding of the proximal fragment to align the proximal aluminum cup coaxially with the distal polymethylmethacrylate cylinder. The proximal fragment was embedded to a point 2 to 3 mm proximal to the fracture line.

Bending Stiffness Quantification

Each prepared specimen was subjected to controlled bending in the plantar-to-dorsal and lateral-to-medial directions to determine bending stiffness and fracture site angulation under load in the sagittal and coronal planes, respectively. To accomplish this, the cup in which the base of the bone was

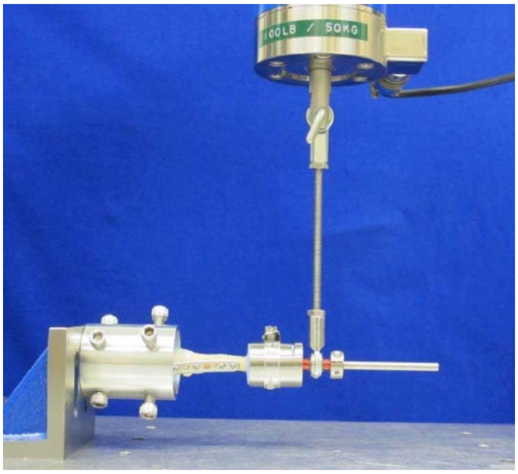

Figure 3. Bending test apparatus. Controlled bending load was applied 25 mm from the center of the metatarsal head.

embedded was rigidly fixed to a right-angle plate with the bone horizontal. The angle plate was secured to the horizontal table of a computer-controlled servohydraulic materials testing machine (Model 1321, Instron Corp, Norwood, MA) (Figure 3). A rod fitted with spherical bearings at each end linked the machine's actuator-mounted load cell to another rod attached to the distal polymethylmethacrylate cylinder and projecting distally. Linear displacement was applied at a rate of 0.1 mm/s at a point 25 mm from the center of the metatarsal head until the desired peak load at the metatarsal head was reached. The actuator was then returned to its starting position, and the displacement was repeated while recording load and displacement at a sampling rate of 250 Hz. A peak load of 18 N at the metatarsal head was used. A previous study of Jones fracture bending stiffness in which subfailure loading was applied used a peak load of 12 N,[16] or one-half of the load to which the fifth metatarsal head is subjected during the push-off phase of gait in normal ambulation.[8] We quantified stiffness and angulation at this load, but also at the higher 18 N load (a 50% increase) out of concern that 12 N might not adequately challenge the contemporary fixation devices evaluated in the current study.

Torsional Stiffness Quantification

Torsion testing of each prepared specimen was performed after completion of the bending tests to determine torsional stiffness and fracture site rotation at the target torque. The aluminum cup containing the base of the metatarsal was rigidly attached to a custom-built rotary actuator mounted on the table of the servohydraulic testing machine and deriving its motion from the machine's linear actuator (Figure 4). The polymethylmethacrylate

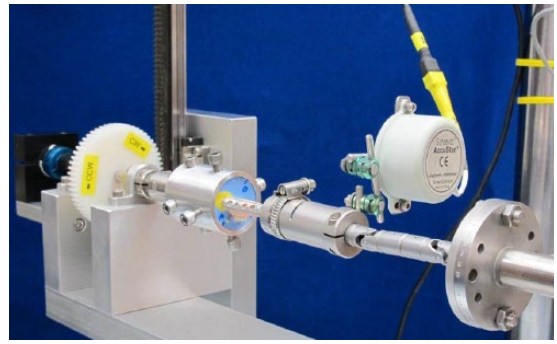

Figure 4. Torsion test apparatus. External rotation was applied to the distal metatarsal fragment. Both proximal fragment and metatarsal shaft rotation were monitored to accurately calculate motion at the fracture line.

cylinder containing the head of the metatarsal was attached to a double universal joint, which was in turn connected to an electronic reaction torque cell (Model TQ301, Omegadyne, Inc, Sunbury, OH) mounted on a linear bearing. The double universal joint and linear bearing ensured that the axis of fracture site rotation and the specimen length were not constrained during the tests. Rotation of the metatarsal base during testing was measured by an RVDT (Model R30D, Lucas Control Systems, Hampton, VA) incorporated into the actuator. To maximize the accuracy of measurement of rotation at the fracture site, an electronic clinometer was attached to the distal end of the bone to record any rotation occurring there as a result of backlash in the universal joints, allowing later subtraction from the rotation indicated by the RVDT.

Controlled internal rotation of the base of the metatarsal relative to the shaft (equivalent to external rotation of the shaft relative to the base) was then applied at a rate of 0.5 degrees/s until a peak torque of 1.0 N·m was attained or until the fracture site had rotated 10 degrees. Proximal fragment and distal fragment rotation and the resulting induced torque were recorded at a sampling rate of 250 Hz. The peak torsional load was selected through preliminary empirical testing to determine a torque that would challenge the fixation but not cause grossly observable damage to the fixed specimens.

Data Analysis

For bending tests, load and displacement data were converted to bending moment and angulation, respectively, based on the length of each specimen from the fracture line to the center of the metatarsal head. Bending stiffness was then calculated in the regions from 9 to 12 N of load and 15 to 18 N of load at the metatarsal head, and expressed in units of newton-meters of bending moment per degree of fracture site angulation. Fracture site angulation at 12

Table 1. Mean Bending Stiffness[a] in the Regions of 9 to 12 and 15 to 18 N of Load.

	Plantar to Dorsal		Lateral to Medial	
	9-12 N	15-18 N	9-12 N	15-18 N
Screw	1.16 ± 0.78	1.07 ± 0.66	2.23 ± 1.26	1.83 ± 0.81
Plate	0.40 ± 0.17	0.39 ± 0.18	1.16 ± 0.81	1.46 ± 0.87
P value	.012	.017	.017	.208

[a]Units are newton-meters per degree of angulation.

Table 2. Mean Fracture Site Bending Angulation.[a]

	Plantar to Dorsal		Lateral to Medial	
	Load = 12 N	Load = 18 N	Load = 12 N	Load = 18 N
Screw	0.90 ± 0.63	1.29 ± 0.87	0.43 ± 0.34	0.65 ± 0.51
Plate	2.29 ± 1.22	3.24 ± 1.74	1.64 ± 1.18	1.97 ± 1.45
P value	.012	.012	.012	.025

[a]Values are degrees ± standard deviation.

and 18 N was calculated mathematically from the head displacement at those loads, with the assumption that the fulcrum was located at the level of the fracture line.

Torsional stiffness was calculated from the torque-rotation relationship during the initial, linear portion of the rotation range (prior to 3 degrees of rotation in most cases) and expressed as newton-meters of torque per degree of fracture site rotation. Stiffness in a second linear region was also calculated for plate-fixed specimens. The highest torque recorded during rotation application and the amount of fracture site rotation associated with that point were documented.

Each of the described measures was compared between groups using 2-tailed Wilcoxon matched-pair signed rank tests. Significance was determined at the P = .05 level. The degree of correlation between the mechanical measures and bone mineral density was established by calculating Pearson product-moment correlation coefficients and their significance.

Results

Bending Stiffness

The mean sagittal and coronal bending stiffness of each fixation construct is reported in Table 1. During plantar-to-dorsal bending, intramedullary screw fixation demonstrated greater stiffness than plate fixation at both evaluated peak loads of 12 N (P = .012) and 18 N (P = .017). During lateral-to-medial bending, intramedullary screw fixation again was a stiffer construct than plate fixation, but this was found to be statistically significant only at the lesser of the 2 assessed loads (P = .017).

Fracture Site Angulation

The mean angulation at the fracture site for each bending direction and each applied peak bending load is summarized in Table 2. In both plantar-to-dorsal and lateral-to-medial bending, intramedullary screw fixation resulted in significantly less fracture site angulation than plate fixation at both tested loads (P range = .012-.025).

Torsional Stiffness

At the initiation of rotation application, the torsional stiffness (resistance to rotation) of intramedullary screw-fixed specimens was approximately 3-fold greater than that of plate-fixed specimens (P = .025) (Table 3). In 7 of the screw-fixed specimens, stiffness then decreased precipitously when friction at the fracture line was overcome and slippage occurred between the proximal and distal fragments (Figure 5). This fracture site "slip point" took place at 0.93 ± 0.28 degrees of rotation and 0.47 ± 0.28 N·m of torque, on average. After that point, torque plateaued, increased slightly, or decreased slightly as the remainder of the rotation was applied. In the remaining screw-fixed specimen, there was no abrupt slip and stiffness remained nearly linear through the rotation range. The target peak torque of 1.0 N was attained only in that screw-fixed specimen.

In the plated specimens, stiffness consistently decreased markedly in the latter half of the 10-degree rotation range. Consequently, Table 3 reports a second stiffness value for this region. Based on visual observation, the stiffness decrease at more extreme amounts of rotation was secondary to movement of the screws and retrograde prongs in the proximal fragment. Figure 6 depicts a representative plate fixation torque-rotation curve.

Table 3. Torsional Stiffness and Fracture Site Rotation.

	Peak Torque (N·m)	Rotation at Peak Torque (Degrees)	Torsional Stiffness 1st Linear Region (N·m/Degree of Rotation)	Torsional Stiffness 2nd Linear Region (N·m/Degree of Rotation)
Screw	0.62 ± 0.30	5.61 ± 3.87	0.60 ± 0.30	N/A
Plate	0.78 ± 0.12	9.73 ± 0.78	0.22 ± 0.17	0.06 ± 0.02
P value	.272	.028	.025	

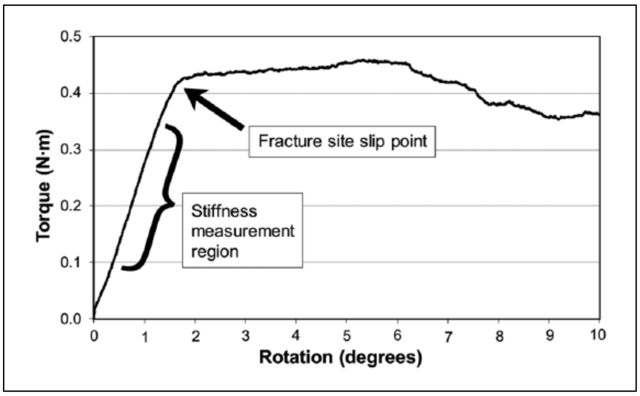

Figure 5. Example of a torsion test torque-rotation curve from a screw-fixed specimen illustrating the abrupt stiffness decrease coinciding with initiation of slippage at the fracture site.

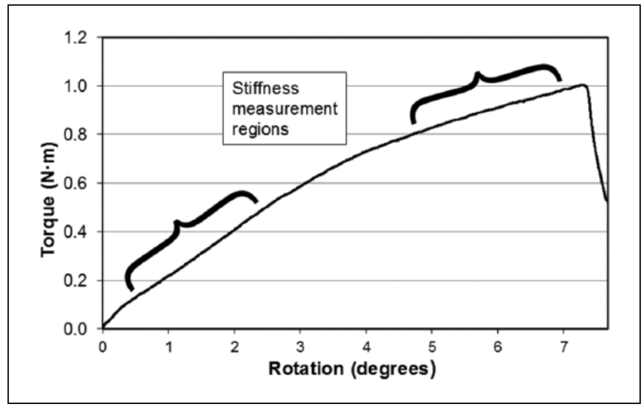

Figure 6. Representative torsion test torque-rotation curve from a plate-fixed specimen. Two distinct linear regions were commonly seen, with increasing torque but lower stiffness (slope) as rotation progressed.

Peak Torque and Fracture Site Rotation

No significant difference was demonstrable in peak torque during the 10 degrees of rotation between the fixation methods (P = .272) (Table 3). However, the amount of rotation at the peak torque was significantly greater with plate fixation (9.7 degrees) than with intramedullary screw fixation (5.6 degrees) (P = .028).

Correlation With Bone Mineral Density

In both screw- and plate-fixed specimens, there was a weak negative correlation between amount of bending angulation and bone mineral density, with r ranging from −0.212 to −0.291. There was a weak to moderate positive correlation between bending stiffness and bone density (r range 0.237 to 0.549). None of the correlations were significant with the available group sizes.

Greater degrees of correlation were observed with the torsional performance measures. Specifically, in plate-fixed specimens, there was a significant strong positive correlation between bone density and both peak torque (r = 0.826, P = .012) and torsional stiffness in the later portion of the rotation (r = 0.819, P = .013). There was a moderate negative correlation between bone density and the amount of fracture site rotation at the peak torque (r = −0.625 and −0.642 for screw-fixed and plate-fixed bones, respectively), but these were not statistically significant.

Discussion

Stabilizing fifth metatarsal Jones fractures can be difficult because of the potential for relative motion between the fragments, as the metatarsal is subjected to both bending and torsion by local deforming forces during weightbearing.[15,21] Standard fixation involves placement of an intramedullary screw, which can provide compression but has limited ability to resist relative rotation at the fracture site.[9] The intraosseous tine "hook" configuration of the TriMed Jones Fracture Plate was developed to address this shortcoming.[17] Although the biomechanical performance of various intramedullary screw designs for Jones fracture stabilization has been studied,[8-13] there have been no studies on the bending and torsion resistance of plate fixation in this same context. Further, to our knowledge there has been no previously reported laboratory study of Jones fracture fixation with the specific screw evaluated in the present study, which has a novel head design that may provide greater resistance to rotation within the proximal fragment than does a smooth, conventional screw head.

We found that intramedullary screw fixation performed better than plate fixation during plantar to dorsal bending and at smaller loads during lateral to medial bending. Of the tested bending conditions, the 1 condition in which no difference between plate and screw performance was demonstrable was lateral to medial bending at the higher of the 2 assessed loads. This was not unexpected, as the plate was placed laterally and therefore came under tension as the fracture line compressed during lateral-to medial loading.

During rotational loading, screw fixation demonstrated greater initial torsional stiffness than did plate fixation, but as rotation progressed beyond a few degrees, the plate resisted torque better than the screw. These somewhat mixed results are unsurprising given the 2 very different design approaches. In their study on the torsional resistance of Jones fractures fixed with 2 types of conventional partially threaded screws, Horst et al[9] observed that failure was accompanied by rotation of the proximal fragment about the unthreaded screw shank, rather than loss of purchase of the screw threads in the distal shaft. They reported a typical torque-rotation relationship paralleling that observed with screw fixation in the present study, characterized by relatively high initial stiffness followed by an abrupt drop. However, the stiffness magnitudes for the tested conventional 5.0- and 6.5-mm screws, respectively, were smaller than the magnitude measured with plate fixation in the present study and much smaller than that measured with the dual-pitch screw fixation that we evaluated. It is important to note that although suggestive, other variables could account for these differences, and may warrant further direct testing. The apparent performance advantage of the tested dual-pitch screw may stem from its tapered, threaded head design, which may achieve additional compression and endosteal purchase in the proximal fragment compared to conventional partially threaded screws.

It is not clear why metatarsals fixed with the "hook" plate evaluated in the present study showed lower torsional stiffness than did the screws during initiation of torsion. There are 2 possible explanations for this. First, no grossly observable plate motion or deformation occurred at the visible, distal fragment during testing, suggesting that fixation in the proximal fragment may have been suboptimal, and indeed motion was visible between the plate and proximal fragment at greater degrees of rotation. The lower density of the metaphyseal bone of the proximal fragment may contribute to this phenomenon and, based on our experience in this study, we recommend that when using this plate particular attention be paid to maximization of fixation integrity in the proximal fragment.

Second, we did not measure fracture site compression, and therefore cannot conclude whether differences in interfragmentary compression between the plate-fixed and screw-fixed fractures contributed to the documented difference in resistance to torsional loading. It should be noted that the simulated fractures created in this study intentionally included no interdigitation between the proximal and distal fragments in order to create a worst-case scenario in terms of rotational stability. Clinically, some degree of interdigitation and non-planarity is common, and would likely increase the torsion resistance of both of the fixation methods that were evaluated. Based on our observation of abrupt stiffness loss on initiation of fracture site rotation in screw-fixed metatarsals, this may be particularly relevant when intramedullary screws are used.

"Hook" plate fixation of proximal fifth metatarsal fractures was first described in 2003 by Carpenter and Garrett as an alternative fixation method, specifically in the presence of reduced bone quality and fracture comminution.[1] Since their description, 2 limited level IV studies have been published, suggesting acceptable early clinical performance using this fixation option, with radiographic union at an average of 7 to 8 weeks.[2,14] In one study, plate fixation avoided the need for postoperative immobilization and allowed initiation of early rehabilitation at an average of 3 to 4 days after surgery.[14]

Disadvantages unique to proximal fifth metatarsal plate fixation include the need for a larger incision, potential for soft tissue irritation from hardware prominence, and risk of disrupting the extraosseous blood supply from soft tissue dissection. However, there are scenarios in which plate fixation may be preferred, including fracture comminution, osteoporotic bone, and revision after nonunion or refracture. Plate fixation may be particularly useful in cases where intramedullary screw fixation is not an option, such as when the metatarsal canal diameter is too small or too large to accommodate the standard available screw sizes, or when there is loss of cortical integrity from cortical "blowout" or multiple previous revisions.

We acknowledge several limitations in the current investigation. First, inherent to cadaveric studies, our model could not assess fracture healing or clinical outcomes. Second, the controlled biomechanical tests used in this study allowed only isolated measurements of bending and torsion, clearly a simplification of what actually occurs during physiologic loading, where these forces act simultaneously in a more complex manner. We deliberately limited the bending tests to the most clinically relevant directions and to modest loads to prevent damage to the specimens before performing the torsion tests. Although plantar-to-dorsal bending of the fifth metatarsal can be assumed to occur during normal ambulation, lateral-to-medial bending is not as obvious. This loading direction was included because the mechanism of Jones fractures has long been understood to involve forces acting on the lateral border of the forefoot. [4,6,10,13] This was confirmed in a dynamic cadaveric gait study by Donahue and Sharkey,[6] who found both average and peak strains on the lateral cortex of the fifth metatarsal to exceed those measured on the dorsum of the metatarsal. Additionally, Jones fracture nonunions and fixation failures are most often manifested by lateral gapping, indicating medial angulation of the distal fragment. For our torsional testing model, only external rotation of the distal fragment relative to the base was evaluated. This direction was chosen based on knowledge of in vivo muscle forces and a previous study performed in a Jones fracture model where torsional loading was applied.[9] Finally, the use of monotonic loading, as opposed to cyclic loading, is an important limitation in this study because under physiologic conditions, fixation constructs for Jones fractures likely fail from repetitive, rather than catastrophic, loading. Because multiple directions of both bending and torsional loading were performed on each specimen, cyclic loading was not feasible in this study. Although each loading test was performed at subfailure loads, this may have resulted in a cumulative effect on the specimens. However, because all specimens in both fixation groups underwent the same sequence of testing, any potential effect on performance was systematic.

In conclusion, plate fixation was not found to provide better resistance to bending or initiation of torsion when compared to intramedullary screw fixation with the specific screw tested. Although we cannot infer actual differences in healing potential or functional outcomes based on these results, the current investigation does provide empirical evidence that should be considered when using plate fixation as an alternative for Jones fracture stabilization. Further prospective comparative studies are warranted to determine if clinical outcome differences exist between these fixation options.

Declaration of Conflicting Interests

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