Periarticular Locking Plate vs Intramedullary Nail for Tibiotalocalcaneal Arthrodesis: A Biomechanical Investigation

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Abstract

Augmented retrograde intramedullary (IM) nail fixation was compared with augmented periarticular locking-plate fixation for tibiotalocalcaneal arthrodesis. Specimens in 10 matched pairs were randomly assigned to a fixation construct and loaded cyclically in dorsiflexion. The groups did not differ in initial or final stiffness, load to failure, or construct deformation. No correlation was found between bone mineral density and construct deformation for either group. A humeral locking plate may be a viable alternative to an IM nail for tibiotalocalcaneal fixation in cases not amenable to IM nailing.

Achieving stable fixation in tibiotalocalcaneal (TTC) arthrodesis can be a surgical challenge. Potential complications include nonunion, delayed union, inadequate fixation, loss of fixation, implant failure, and infection.[1] Patients who undergo this procedure often have poor bone quality secondary to either disuse or inflammatory arthropathy, and bony fragmentation may be present with neuroarthropathy. It would be useful to identify a TTC fixation method that has optimal stability in poor bone stock.

Intramedullary (IM) nails have been shown to provide stable TTC fixation.[2-7] Disadvantages of this fixation method include difficulty of use in cases of poor-quality bone, posttraumatic tibial deformity, and past deep infection secondary to external fixation. A biomechanical analysis showed that a blade plate construct with TTC screw augmentation was superior in stiffness and fatigue strength to IM nail fixation with lateral-to-medial interlocking screws.[6] Placement of a blade plate, however, can be technically challenging because of the fixed position of the blade in 3 dimensions.

Periarticular locking plates offer the advantages of blade plates, but they also provide multiplanar screw fixation and can be technically easier to insert than a blade plate. We and others have used these locking plates in TTC arthrodesis, but clinical results have not been studied definitively. These plates performed better than blade plates in 2 fracture models.[8] A straight humeral locking-plate construct with single-plane screw holes was not different from an IM nail construct in initial stiffness, torsional load to failure, and construct deformation for TTC arthrodesis in a cadaver model.[9] The role of bone mineral density (BMD) was not evaluated in this study.

We hypothesized that the locking-plate construct with an augmentation screw would provide a more rigid construct for TTC fixation than IM fixation with an augmentation screw. Given the findings of a previous study,[10] we also hypothesized that stiffness of the locking-plate construct would be correlated with BMD. The purpose of this study was to compare a periarticular humeral locking plate and a retrograde IM nail, both augmented with a screw, in terms of construct rigidity, construct deformation, and final load to failure.

MATERIALS AND METHODS

Ten (5 male, 5 female) pairs of fresh-frozen cadaveric legs (mean age, 80 years; range, 62-91 years) were used in this study. Each specimen was stored at –20°C, thawed to room temperature before testing, and amputated 10 cm distal to the tibial tubercle and at the transverse tarsal joint. Dual-energy x-ray absorptiometry, or DXA (GE Lunar Scanner; GE Healthcare, Buckinghamshire, United Kingdom), was then performed to determine the BMD of the posterior calcaneus in a sample from each pair. One side of each matched pair was randomly assigned to receive IM fixation, and the contralateral side was assigned locking-plate fixation. The fibula was excised. This is often done clinically in TTC arthrodesis for exposure or to harvest bone graft for local use. Further, the fibula may be unsalvageable in these conditions and therefore cannot be used in the construct. All soft tissues were excised, except ligamentous structures, which maintained the relative bony positions

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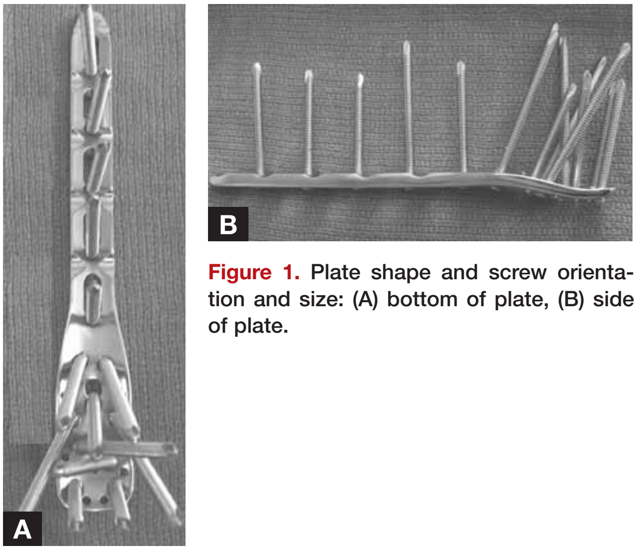

Figure 1. Plate shape and screw orientation and size: (A) bottom of plate, (B) side of plate.

while hardware was inserted. Ligamentous structures were removed before testing. All specimens were instrumented using a previously described protocol for in situ fusion.[6] The joint surface was left intact to avoid introducing an uncontrolled variable.[6,11,12] Each sample was placed in a position of neutral dorsiflexion, neutral rotation, and neutral varus/valgus alignment.

In the first group, TTC arthrodesis was performed with a 12×150-mm IM nail (Ankle Arthrodesis Nail; Biomet, Warsaw, Indiana). A guide wire was placed retrograde from a starting point just lateral to the medial border of the calcaneus in line with the center of the talus and tibia. Sequential flexible reaming followed in 0.5-mm increments up to 12.5 mm. The nail was placed retrograde through the calcaneus, talus, and tibia such that the distal tip of the nail protruded no more than 5 mm. Two proximal locking screws were then placed medial to lateral through the tibia. The articulating surfaces were then compressed with a threaded compression device using the manufacturer's instrumentation. A varied amount of compression based on the anatomy of each specimen was used to obtain the best possible compression without having the compression device impact the calcaneus. The goal was to achieve coaptation of the bony surfaces and then compression of 2 to 3 mm. Three distal locking screws were then placed. The 2 more proximal of these screws were placed lateral to medial through the talus and calcaneus, respectively. The third screw was placed posterior to anterior through the calcaneus. Finally, each sample received a partially threaded 6.5-mm (mean length, 110 mm; range, 100-120 mm) cannulated titanium screw (DePuy/ACE, Warsaw, Indiana) placed retrograde from the calcaneus through the talus and into the anterior tibial metaphysis to improve stability.[10,12]

In the second group, each sample received a 3.5-mm proximal humerus locking compression plate (LCP; Synthes, Paoli, Pennsylvania) with a 5-hole shaft and 14 holes total; this plate was 142 mm in length (Figure 1). Before the plate was applied, the sample received an augmentation screw. Plate contouring is often needed with nonlocking systems, but the relative stiffness of this locking plate and the nature of its locking mechanism obviated the need for contouring to maintain fixation. In some samples, however, a small portion of the Chopart tubercle or the lateral process of the talus was removed to accommodate the plate. An effort was made to position the plate inferiorly while maintaining flush contact with the bony surfaces and maximizing the number of screws in the calcaneus, as described previously.[10] The shaft portion of the plate was oriented proximally along the tibia. Plate apposition on bone was desirable, but in some areas this did not occur.

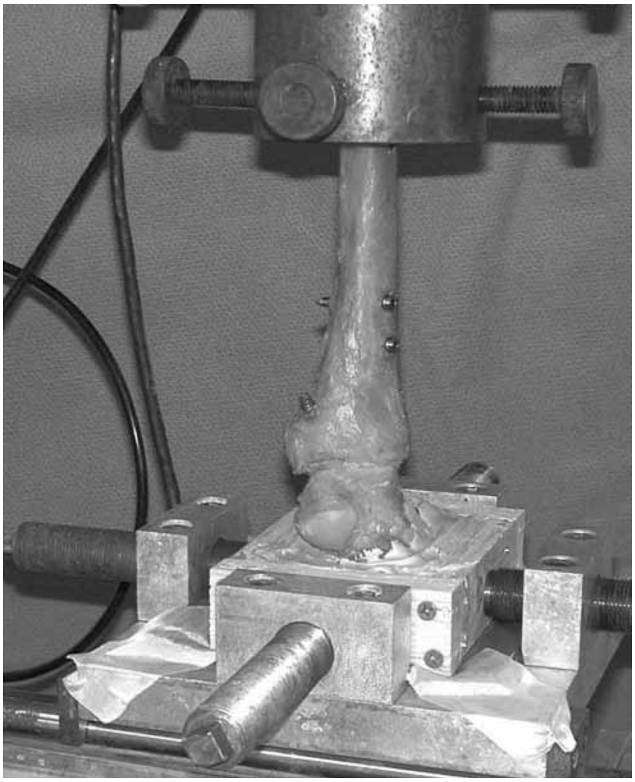

Figure 2. Test setup with intramedullary nail and augmenting screw in place.

Two 3.5-mm bicortical locking screws were placed across the calcaneus through the distal plate holes. A 3.5-mm nonlocking bicortical screw was then placed through the third most proximal shaft screw hole in compression. A unicortical 3.5-mm locking screw was placed in the most proximal shaft screw hole, and bicortical 3.5-mm locking screws were placed in the remaining proximal tibial shaft screw holes. The talar screw hole was filled with a 3.5-mm locking screw. When the position of the augmentation screw prevented use of a bicortical screw in this hole, a unicortical screw was used. The other holes in the periarticular portion of the plate were filled, when possible, with bicortical 3.5-mm locking screws. In approximately half the specimens, 1 hole was left open because of interference from the 6.5-

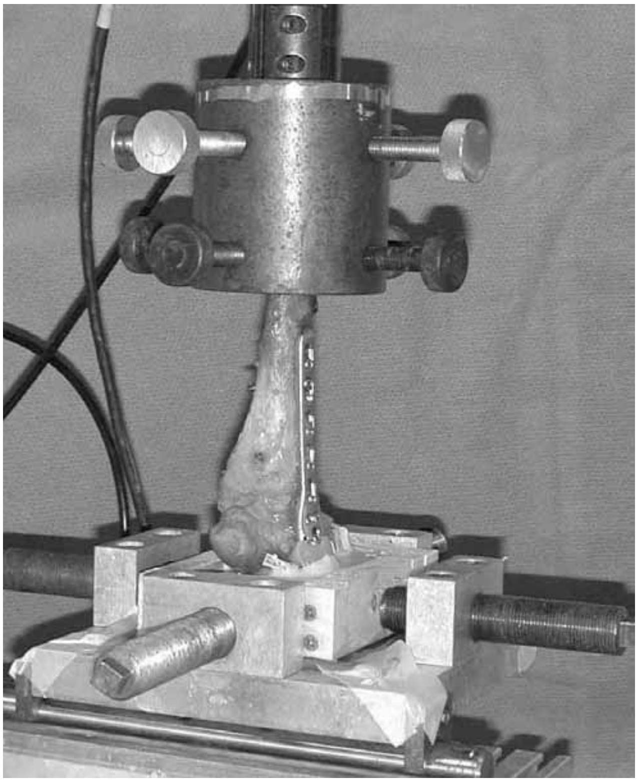

Figure 3. Test setup with plate in place.

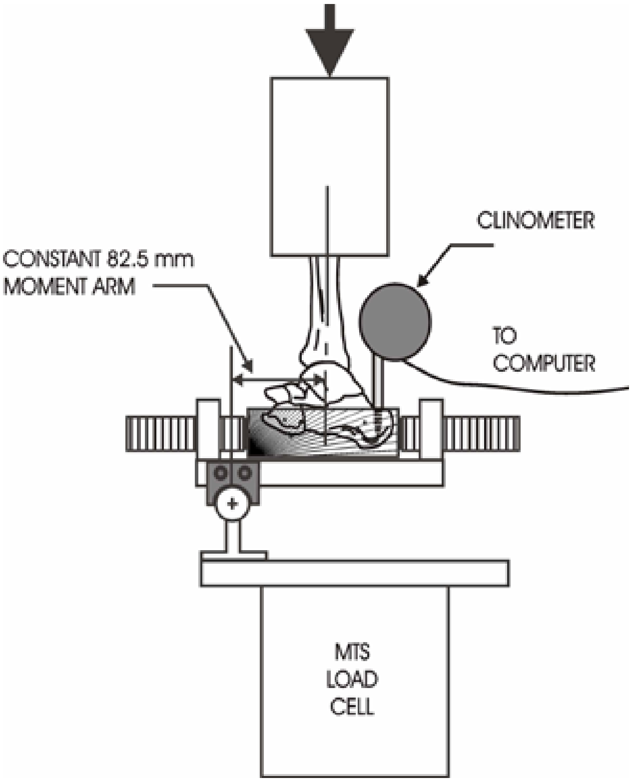

Figure 4. Angle measurement validation test setup.

mm augmentation screw. Eight or 9 locking screws were used in the calcaneus in all specimens. Locking screw holes were predrilled using a locking guide and a 2.8-mm bit. The nonlocking 3.5-mm screw was predrilled with a 2.5-mm bit.

Each sample was centered in a wooden box and secured with 4 2.0-mm Kirschner wires that were passed through 2 walls of the box and the calcaneus. Any exposed fixation hardware was covered in modeling clay to prevent adherence to the resin. The calcaneus of each sample was potted with a polyester resin while avoiding embedding the subtalar joint. Samples were then loaded into a servohydraulic frame (MTS Systems, Eden Prairie, Minnesota) for dorsiflexion loading (Figures 2, 3).[6,12]

The load cell had a maximum capacity of 2,500 N, and the resolution of the load cell was 0.1% of full scale (2.5 N). All samples were loaded cyclically using a sinusoidal waveform from a minimum of 26 N to a maximum of 260 N at 3 Hz for 250,000 cycles. This load, along with a constant moment arm of 82.5 mm (resulting in a torque of 21.5 N-m), was chosen to simulate 6 weeks of partial weight bearing.[13] The constant moment arm was established by precise positioning of the specimen box on the mounting plate. The bearings were self-lubricating bronze bushings. Failure was defined as 10° of dorsiflexion or fracture,[9] with the angle calculated from the actuator displacement. Fracture of the tibia or talus was monitored visually, and fracture of the calcaneus would have resulted in more than 10° of dorsiflexion. Load and deflection data were collected for the initial cycle and the last cycle. Construct stiffness was calculated from the load and deflection data collected at the first and last cycles based on the linear portion of the load deflection curve. Construct deformation was calculated as the difference in actuator piston position between the initial cycle and the last cycle. This measurement was obtained at a load of 150 N for all specimens. All samples that had not failed after 250,000 cycles were monotonically loaded to failure at a rate of 10 mm per minute. At the completion of each test, the wooden mounting box was disassembled and the transfixion wires removed. The resin was then carefully removed from the calcaneus. A visual analysis was performed on the fixation hardware.

We tested another pair to determine the correlation between the angle calculated by the MTS actuator piston movement and the actual angle of the calcaneus. One specimen was instrumented with an IM nail and the other with a locking-plate construct. The angles measured during testing were validated by placing a clinometer (Accustar; Schaevitz Sensors, Hampton, Virginia) on the calcaneus with a vertically positioned 4-mm half-pin (Figure 4). The clinometer measured the angle in the sagittal plane of the calcaneus while an Exakt angle meter (Starrett, Athol, Massachusetts) measured the tibial angle in the sagittal plane. Calculation of the angle between the tibia and the calcaneus was based on the angles measured with the clinometer and the angle

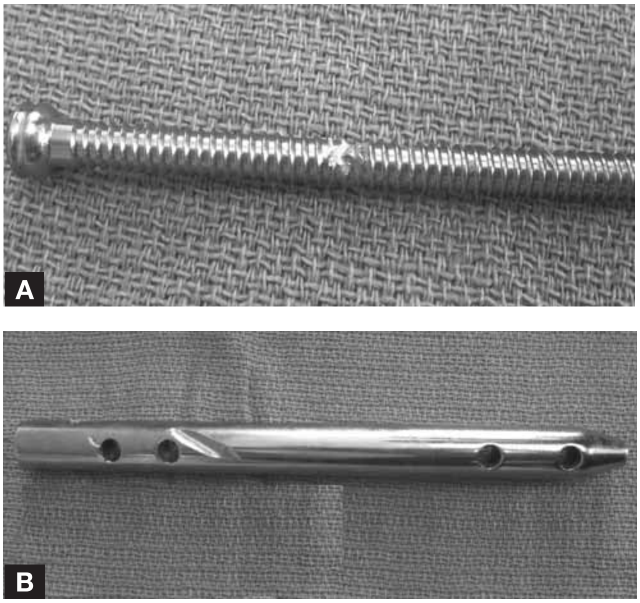

Figure 5. Damage to (A) distal transverse locking screw and (B) nail adjacent to more proximal distal transverse locking screw hole.

meter. Angles were measured before cyclic loading, after 250,000 cycles, and after load to failure.

In addition to recording stiffness in Newton-meters per degree, we calculated the same data in Newtons per millimeter to allow comparison with previous studies from our institution.

Statistical Considerations

Power. Power calculation was based on what was done in a similar study of ankle fusion.[6] Based on either initial or final stiffness, a sample size of 9 in each group had 80% power to detect a significant difference in stiffness at the .05 level where a difference exists.

Statistical Analysis. A paired 2-tailed *t* test was used to determine whether observed differences in final stiffness, load to failure, or construct deformation were significant ($P \leq .05$). The Mann-Whitney test was used to compare initial stiffness because the data were not normally distributed. Pearson correlation was used to determine whether there was a significant relationship between BMD and stiffness, failure load, or construct deformation in the 2 fixation groups. Pearson correlation was also used to determine whether there was a significant correlation between the angle calculated by MTS actuator piston movement and the actual angle of the calcaneus measured by clinometer.

Results

All samples completed testing of 250,000 cycles without failure. Median (quartile) initial stiffness was, in Newton-meters per degree, 15.9 (13.9, 19.0) for the nail group and 14.5 (11.0, 20.0) for the plate group ($P = .32$) and, in Newtons per millimeter, 134 (117, 160) for the nail group and 122 (93, 169) for the plate group ($P = .32$). There were no significant between-groups differences ($P = .38$) in final stiffness, either in Newton-meters per degree (nail mean, 29.6; standard error of the mean [SEM], 2.5; plate mean, 28.2; SEM, 2.4) or in Newtons per millimeter (nail mean, 248.9; SEM, 66.5; plate mean, 237.5; SEM, 64.1). For dorsiflexion load to failure, there was no significant difference ($P = .31$) between the nail group (mean, 1,108 N; SEM, 363.4 N) and the plate group (mean, 1,007 N; SEM, 198.4 N). All samples failed by excessive dorsiflexion movement of more than 10° during the monotonic load-to-failure phase. In the locking-plate constructs, failure mechanisms included the plate pivoting proximally on the tibia. Both ankle and subtalar motion could be detected visually, but this motion was not quantified. In the nail constructs, failure mechanisms were more difficult to detect but could be viewed after the samples were disassembled. The IM nail tended to open the cavity in the calcaneus, and the cavity was more oval than circular. No catastrophic failure occurred during either cyclic loading or monotonic loading to failure. The difference in construct deformation between the nail group (2.8 mm, 2.0°) and the plate group (2.3 mm, 1.6°) did not reach significance ($P = .07$).

Mean calcaneal bone density was 0.48 g/cm^2 (range, 0.31-0.79 g/cm^2). There was no significant correlation between BMD and initial or final stiffness, failure load, or construct deformation for either group ($P \geq .4$). Gouges representing metal damage from drill or screw insertion were found on the hardware in 3 specimens in the nail group. These markings were indicative of drill malposition during placement of the augmentation screw. In 2 of these cases, there was an approximate 1-mm mark on a single distal transverse locking screw (Figure 5A); the third case had a mark 10×1×0.5 mm at the distal end of the nail adjacent to the most proximal distal transverse locking screw (Figure 5B). No incidental damage from interference with the augmenting screw was found on the blade plates or associated screws.

Testing of the additional matched pair confirmed that TTC joint movement was closely related to the angle calculated by MTS actuator piston movement. The difference between the clinometer angle and the calculated angle before cyclic loading and after 250,000 cycles was less than 0.5° in both the nail group (before, clinometer 1.6° vs calculated 1.9°; after, clinometer 2.9° vs calculated 3.3°) and the plate group (before, clinometer 1.2° vs calculated 1.1°; after, clinometer 2.2° vs calculated 2.0°). At failure load, the maximum difference between the clinometer angle and the angle calculated by actuator piston movement was 0.7° in the nail group (clinometer, 9.3°; calculated, 10.0°) and 0.4° in the plate group (clinometer, 10.4°; calculated, 10.0°). The correlation coefficient was 0.995.

DISCUSSION

We found no significant differences between the periarticular locking-plate construct and the IM nail construct in initial or final stiffness or construct deformation, which are the tested factors of concern during partial weight bearing in the first 6 weeks after surgery. In addition, the 2 constructs did not differ in load to failure. The augmented periarticular locking-plate construct may provide an effective alternative to the IM nail and may be useful particularly in cases in which IM nailing is precluded. These findings are consistent with a finding from another TTC arthrodesis study—that a periarticular locking-plate construct was significantly stronger than a blade plate construct.[10] Our data are also consistent with findings of no difference in rigidity in 3 of 4 factors tested between a standard straight locking plate augmented with a TTC screw and a nail construct augmented with a TTC screw.[9]

In the present study, there was no relationship between BMD and stiffness, failure load, or construct deformation for either construct. Lack of positive correlation between BMD and these other parameters suggests that construct performance was not affected by BMD. Chiodo and colleagues[6] found an inverse correlation between BMD and the difference in construct deformation between the specimens of each pair fixed with an IM nail or blade plate plus augmentation screw. This finding suggests that the blade plate was more rigid than the IM nail in specimens with low BMD. Chodos and colleagues[10] found a positive correlation between BMD and both dorsiflexion load to failure and torsional load to failure with a proximal humeral locking plate, which suggests that this construct was less stable at failure loads in specimens with low BMD. As in the present study, the investigators did not find a correlation between BMD and stiffness in either construct tested. The difference in findings with respect to BMD correlation with failure load in the locking plate may be related to use of femoral rather than calcaneal bone to determine BMD in the previous study. More important, the variations in technique between the 2 surgeons in these studies may account for the difference in findings. Slight differences in plate position may have had an effect on fixation. Because poor-quality bone is common in patients who undergo this procedure clinically, it might be helpful to investigate construct performance further, specifically in poor-quality bone.

Locking plates are thought to be advantageous in that they provide a fixed angled construct, much like that of an external fixator. Although it has been theorized that they may offer an advantage over standard plating techniques in osteopenic bone, this has not been clearly demonstrated in clinical studies. Potential pitfalls include additional exposure, prominent hardware, and limited compression at the arthrodesis site. Although the subtalar and ankle joints are more clearly visualized, devitalization of soft tissues may place healing at risk. Locking plates, however, do have several potential advantages. They may be easier to place than an IM device or blade plate. In cases in which posttraumatic deformity may preclude use of an IM device, a locking plate may be better suited. Finally, because an augmentation screw is placed before plate application, the risk for hardware impingement is lessened.

In 3 of our samples, evidence of hardware impingement with violation of the integrity of the IM implant was found with insertion of the TTC augmentation screw. This incidental observation may suggest the presence of a potential stress riser. Fixation in these specimens did not appear to be adversely affected.

In both groups, bending stiffness increased approximately twofold as cyclic loading progressed. This finding suggests that the construct settled during the first few loads.[10] We retained the comparison of first-cycle stiffness to determine whether one construct had relatively high stiffness from the start of cyclic loading.

Removal of the fibula with the plate construct is a concern with this method. Most internal locking-plate and nonlocking-plate systems for TTC arthrodesis recommend removal of the fibula and direct plating of the tibia. Ideally, in any fusion construct, it is preferable to maintain maximal bone stock to serve as hardware fixation points, to provide more surface area for healing, and to assist in load bearing. Especially in a 2-bone system, keeping both bones should, in principle, maximize stability. However, TTC fusion is often done as a salvage procedure after trauma with prior fibular fracture, talus avascular necrosis, or presence of multiple risk factors. If the fibula can be salvaged and is considered important because of concerns about the tibia or soft tissue, another method such as nail fixation or multiple screws may be preferable to plate fixation in TTC arthrodesis. It is possible (though technically challenging) to preserve the fibula in a plate construct provided that the width and depth of the plate allow for placement of the plate and screws. Most current plate designs, such as the one used in this study, are not adaptable to fibula preservation.

This study had several limitations. Testing in this model was limited to dorsiflexion and cannot be directly extrapolated to the clinical setting. Although testing in other directions would be clinically applicable, loading in multiple directions can alter the joint and soft tissues. We chose to focus on dorsiflexion because we believe that it is the most common motion in weight-bearing, standing, and walking. Cadaveric specimens used in biomechanical studies often have little to no deformity and as such do not accurately represent clinical pathology. Soft-tissue restraints were removed from all specimens and did not allow musculotendinous and ligamentous forces to act on the arthrodesis site. Articular surfaces were left intact in all samples—which allowed an uncontrolled variable to be avoided but does not reflect clinical practice. Although we used actuator angular displacement rather than direct measurement of joint movement, our additional test showed that angles calculated from actuator displacement closely paralleled subtalar and ankle joint movement as measured with a clinometer. This study did not assess fusion-site compression, which can differ between the 2 techniques. Further, our plating and nailing devices have features or jigs that allow for enhanced compression. The conclusions drawn regarding the specific devices used in this study may not be directly applicable to other locking plates or other TTC nailing systems. Finally, DXA scanning of the calcaneus has not been validated. However, the measurements used in this study allowed relative comparison among specimens.

In the present study involving TTC arthrodesis with dorsiflexion loading, construct rigidity with a periarticular locking plate did not differ from that with an IM nail, which suggests that the plate may be a viable option for this application.

Authors' Disclosure Statement and Acknowledgments

Dr. Schon wishes to report paid speaking presentations for DePuy/ACE, paid consulting for DePuy/ACE and Biomet, and past research support unrelated to the current study from Biomet. Dr. Schon and Mr. Parks also note that they hold a patent for a tibiotalocalcaneal (TTC) nail, but the nail has not yet been marketed, and there is no manufacturer. Dr. Ohlson, Dr. Shatby, and Ms. White report no actual or potential conflict of interest in relation to this article.

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