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A Biomechanical Comparison of Locking Versus Conventional Plate Fixation for Distal Fibula Fractures in Trimalleolar Ankle Injuries

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A B S T R A C T

Previous biomechanical studies have advocated the use of locking plates for isolated distal fibula fractures in osteoporotic bone. Complex rotational ankle injuries involve an increased number of fractures, which can result in instability, potentially requiring the same fixed angle properties afforded by locking plates. However, the mechanical indication for locking plate technology has not been tested in this fracture model. The purpose of the present study was to compare the biomechanical properties of locking and conventional plate fixation for distal fibula fractures in trimalleolar ankle injuries. Fourteen (7 matched pairs) fresh-frozen cadaver leg specimens were used. The bone mineral density of each was obtained using dual x-ray absorptiometry scans. The fracture model simulated an OTA 44-B3.3 fracture. The syndesmosis was not disrupted. Each fracture was fixated in the same fashion, except for the distal fibula plate construct: locking ($n = 7$) and one-third tubular ($n = 7$). The specimens underwent axial and torsional cyclic loading, followed by torsional loading to failure. No statistically significant differences were found between the locking and conventional plate constructs during both fatigue and torque to failure testing ($p > .05$). Our specimen bone mineral density averages did not represent poor bone quality. The clinical implication of the present study is that distal fibular locking plates do not provide a mechanical advantage for trimalleolar ankle injuries in individuals with normal bone density and in the absence of fracture comminution.

Ankle fractures account for 9% of all fractures, with approximately 187 fractures per 100,000 people annually (1,2). Unstable fractures will have improved outcomes with anatomic reduction and stabilization. Operative management of lateral malleolar fractures generally involves open reduction and internal fixation using standard AO techniques, with lateral plates the most commonly used fixation method. Variations of this technique have become popular, especially with the development of locking compression plate (LCP) technology. LCPs allow the opportunity to combine internal fixation and dynamic compression principles, depending on the fracture type and site. The LCP functions as a fixed-angle device, converting shear stress to compressive forces at the screw–bone interface. This force conversion improves fracture fixation, because cortical bone has a greater resistance to compressive stress than to shear stress (3–6). The LCP also provides angular and axial stability, which can obviate the need for exact plate contouring, thereby reducing the risk of primary loss of reduction (4). LCPs have been advocated for open reduction and internal fixation of distal fibular fractures, in particular, for patients with low bone density and/or comminution with bone loss (7–9). To our knowledge, no studies have evaluated the biomechanical utility of locking plate fixation in higher energy, multiple fracture, rotational ankle injuries. We wanted to test whether the stability of trimalleolar ankle injuries would benefit from the fixed angled properties afforded by locked plate fixation of the distal fibula fracture. Furthermore, we questioned whether this technology would be indicated in situations in which the other fractures in the injury model have been stabilized with traditional fixation implants. The purpose of the present study was to compare the biomechanical properties of locking compression and conventional plate fixation in the complex rotational ankle injury using a cadaveric model.

Materials and Methods

Specimens

Seven matched pairs of fresh-frozen human lower leg specimens (age 72.3 ± 5.4 years, no sex restriction) were used for the present study. Planar radiographs were

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taken before testing with a surgical C-arm (Philips BV Pulsera; Koninklijke Philips N.V., Amsterdam, The Netherlands). Specimens with anatomic deformities or evidence of previous fracture or surgery were excluded from the present study. Bone density scans (dual x-ray absorptiometry; Hologic QDR-2000; Hologic, Inc., Bedford, MA) were also performed on the diaphysis, with the skin and musculature intact. After bone density scanning, the specimens were prepared for mechanical testing. The soft tissue was removed from the sole of the foot and from the proximal 10 cm of the lower leg to improve fixation to the potting material. All ankle bony and ligamentous structures were left intact, including the syndesmosis. The specimens were then potted anatomically in a polyurethane casting agent (Smooth Cast 300; Smooth-On, Easton, PA) to securely fix them to the mechanical testing frame before and after surgical preparation.

Surgical Preparation

A supination–external rotation trimalleolar ankle fracture pattern was simulated in each, reproducing type 44-B3.3 fractures of the AO Muller classification system, with an oblique distal fibular osteotomy, an osteotomy of the posterior malleolus, and a transverse osteotomy of the medial malleolus (10). The syndesmosis was not disrupted, and the fractures were anatomically reduced. The distal fibular fracture was reduced using a 3.5-mm nonlocking interfragmentary lag screw. Seven specimens were fixated with a LCP (2.7/3.5-mm LCP Lateral Distal Fibula Plate; DePuy Synthes, West Chester, PA), secured with 2 proximal 3.5-mm bicortical locking screws and 3 distal 2.7-mm unicortical locking screws (Fig. 1*A*). The other ankle in each pair was fixated with a 5-hole one-third tubular plate (one-third tubular plate with collar 5 holes; DePuy Synthes) using 3.5-mm non-locking screws with 2 bicortical proximal screws and 2 unicortical distal screws (Fig. 1*B*). The medial and posterior malleolar fractures were fixated in the same fashion in all 14 specimens. Two 4.0-mm partially threaded cancellous screws for the transverse medial malleolar fracture and one 4.5-mm partially threaded screw oriented anterior to posterior across the posterior malleolar fracture were placed.

Biomechanical Testing

For fatigue testing, the specimens were loaded under multi-axial, physiologic conditions using an Instron® 8521 load frame (Instron® Inc., Norwood, MA) before and after surgical device implantation. Testing simulated the end of the single leg stance phase of gait immediately before the opposite heel strike. The foot was oriented with respect to the tibia in 15° of dorsiflexion, 10° of supination, and 0° of internal and external rotation (Fig. 2). These orientations, in particular, dorsiflexion and supination, have been shown to substantially increase the mechanical loading on the fibula (11,12). The specimens were cyclically loaded in axial compression to 5K cycles, a nondestructive limit determined from preliminary testing. Synchronized axial/torsional loading was applied between 0.5 body weight/0 Nm and 1.25 body weight/1 Nm (body weight = 850 N, axial loading/external torque) at 1 Hz. The axial compressive force, external rotation torque, axial displacement, and rotation were monitored, and the axial and rotational stiffness were calculated at 100 cycles, noted as precycling, and 4900 cycles, noted as postcycling.

For the torque to failure analysis, the specimens were again placed in the test frame in the same initial configuration as for the fatigue testing. An 800 N compressive load was maintained throughout failure testing, and the specimen was rotated at 60° per second until failure (8,13). Failure was determined present at the first local maximum when the torque started to decrease. Stiffness was calculated using the initial linear region of the torque versus angle graph.

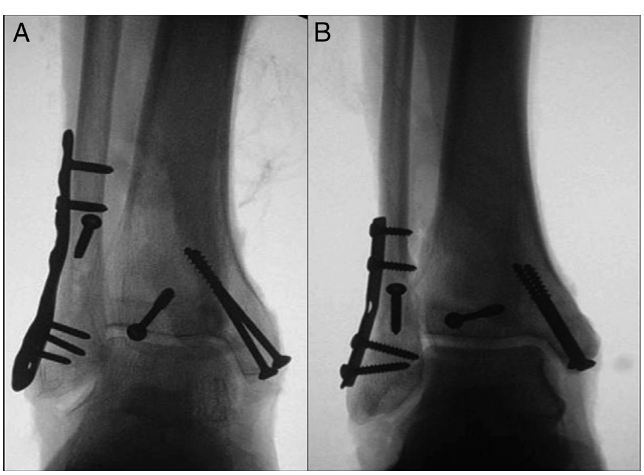

Fig. 1. Anterior to posterior plain radiographs of the (*A*) locking (left leg) and (*B*) nonlocking (right leg) distal fibular plate constructs. Both constructs incorporated an interfragmentary lag screw for the fibular fracture. The medial and posterior malleolar fractures were fixated in the same fashion in both groups.

Fig. 2. Prepared specimen (with an implanted locking plate) secured within the testing apparatus.

The outcome measures, including axial stiffness, rotational stiffness, torque at failure, rotational stiffness during failure testing, and angle at failure, were analyzed with matched pairs analysis ($\alpha = 0.5$) and correlated using a standard Pearson correlation procedure (JMP, version 7; SAS Institute, Cary, NC).

Results

No statistically significant differences were seen between the locking and nonlocking plate constructs during both the fatigue and the torque to failure testing ($p > .05$). In the fatigue testing, the axial and torsional stiffness values for the locking plates were 1785.6 ± 473.9 N/mm and 1.2 ± 0.5 Nm/degree after 4900 cycles, respectively. The corresponding values were decreased by 249.3 ± 259.9 N/mm and 0.47 ± 0.39 Nm/degree from those of the intact, nonfractured specimen. For the nonlocking plates, the axial and torsional stiffness values were 1592.6 ± 461.7 N/mm and 1.6 ± 0.4 Nm/degree, respectively. These axial and rotation values were 160.6 ± 225.3 N/mm and 0.45 ± 0.28 Nm/degree less than those for the corresponding intact state. The outcome measures for fatigue testing, including axial stiffness and rotational stiffness, for the intact specimen and after fracture and repair are summarized in Table 1. The change in axial and rotational stiffness before and after repair is also listed in Table 1, with calculated p values.

The data showed that for the torque to failure testing, the locking plates had an average failure torque and angle of 20.7 ± 5.8 Nm and 22.8° ± 8.0°, respectively. The nonlocking plates had similar values of