

Effect of bone diameter and eccentric loading on fatigue life of cortical screws used with interlocking nails

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Objective—To test the effects of bone diameter and eccentric loading on fatigue life of 2.7-mm-diameter cortical bone screws used for locking a 6-mm-diameter interlocking nail.

Sample Population—Eighteen 2.7-mm-diameter cortical bone screws.

Procedure—A simulated bone model with aluminum tubing and a 6-mm-diameter interlocking nail was used to load screws in cyclic 3-point bending. Group 1 included 6 screws that were centrally loaded within 19-mm-diameter aluminum tubing. Group 2 included 6 screws that were centrally loaded within 31.8-mm-diameter aluminum tubing. Group 3 included 6 screws that were eccentrically loaded (5.5 mm from center) within 31.8-mm-diameter aluminum tubing. The number of cycles until screw failure and the mode of failure were recorded.

Results—An increase in the diameter of the aluminum tubing from 19 to 31.8 mm resulted in a significant decrease in the number of cycles to failure (mean \pm SD, 761,215 \pm 239,853 to 16,941 \pm 2,829 cycles, respectively). Within 31.8-mm tubing, the number of cycles of failure of eccentrically loaded screws (43,068 \pm 14,073 cycles) was significantly greater than that of centrally loaded screws (16,941 \pm 2,829 cycles).

Conclusions and Clinical Relevance—Within a bone, locking screws are subjected to different loading conditions depending on location (diaphyseal vs metaphyseal). The fatigue life of a locking screw centrally loaded in the metaphyseal region of bone may be shorter than in the diaphysis. Eccentric loading of the locking screw in the metaphysis may help to improve its fatigue life. (*Am J Vet Res* 2003; 64:569–573)

In veterinary medicine, there has been a recent trend toward treatment of diaphyseal fractures by “biological osteosynthesis” with the use of interlocking nail

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fixation; anatomic alignment is achieved often without complete fracture reduction, especially in highly comminuted fractures. A limited surgical approach is used for application of the interlocking nail, which allows minimal disturbance of soft tissue attachments and vascular supply of fracture fragments. The interlocking nail can be used as a means of repair of diaphyseal fractures of the femur, tibia, and humerus. Results of clinical studies¹⁻⁷ indicate that 83 to 96% of dogs and cats with diaphyseal fractures heal without complications, and 80 to 90% of animals regained excellent or good limb function.

Depending on the conformation of a fracture, the interlocking nail acts in either a load-sharing or load-bearing manner. When the interlocking nail is used as a load-bearing construct in comminuted fractures, the mechanical properties of the bone-implant construct are of primary concern because of the potential over-activity of patients after surgery. In a multicenter prospective study, breakage of interlocking nails through a screw hole was reported to have occurred in 14% of 6-mm-diameter nails locked with 3.5-mm-diameter cortical screws.¹ Reducing the locking screw diameter from 3.5 to 2.7 mm improved the fatigue resistance of the 6-mm-diameter interlocking nail by 52-fold,⁸ but likely compromised the fatigue life of the locking screw. Subsequently, screw breakage has been reported,^{1,2,7} usually occurring at the nail-screw junction where the load is transmitted to the screw. Screw breakage is, in part, dependent on the bending stiffness of the screw, and small decreases in the core diameter of a screw result in large decreases in bending stiffness.⁹

Interlocking nail systems are subjected to loads of varying magnitudes depending on, among others things, the bone in which it is placed (femur, tibia, or humerus), the size and activity level of the individual animal, and the fracture configuration (simple or comminuted). Although not previously studied, locking screws are most likely subjected to different loading conditions depending upon their location within a bone (metaphyseal or diaphyseal). The unsupported length of a screw increases as the diameter of the bone increases from the diaphyseal to the metaphyseal region of bone. In addition, locking screws are often eccentrically loaded in the metaphyseal regions of bone because of the curvilinear nature of canine bone.

The purpose of the study reported here was to test the fatigue life of 2.7-mm-diameter cortical bone screws used in the 6-mm-diameter interlocking nail. The effects of diameter of the bone and eccentric loading of the screw on fatigue life were examined. We test-

ed the hypotheses that 1) screw fatigue life would decrease with an increase in bone diameter, and 2) screw fatigue life would increase with eccentric loading of the screw.

Materials and Methods

Implants—All locking screws and interlocking nails were manufactured from 316L stainless steel. Locking screws were standard 2.7-mm-diameter cortical bone screws^a that had a core diameter of 1.9 mm, 1 thread/mm of screw shaft, and thread lengths of 32 and 36 mm. Interlocking nails^b were 6 mm in diameter. At the end of each nail were 2 transverse 2.7-mm-diameter screw holes that were centered 11 mm

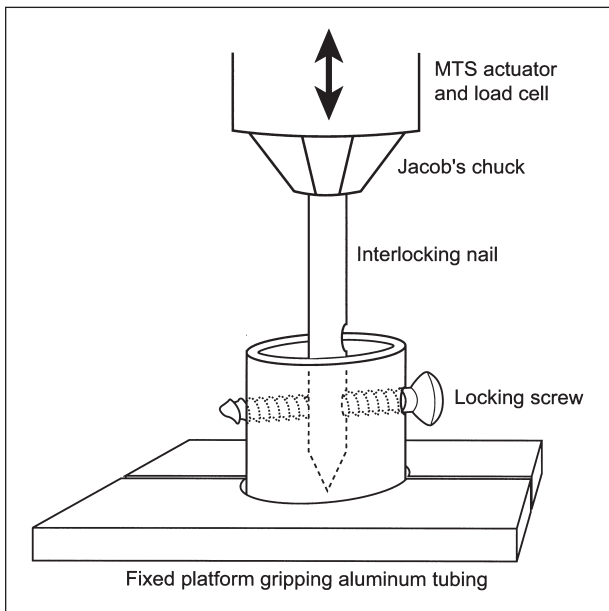


Figure 1—Schematic drawing of materials testing setup. Each screw was cyclically loaded in 3-point bending. MTS = Materials testing system.

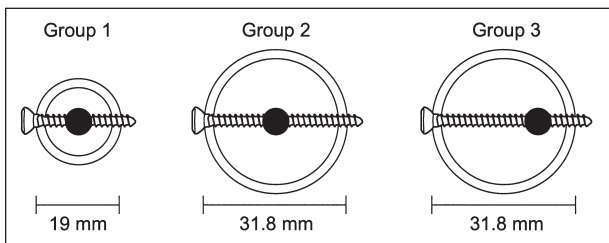


Figure 2—Schematic drawing of simulated bone model configurations for screw fatigue testing as follows: group-1 screws (n = 6) had the interlocking nail centrally located within 19-mm-diameter aluminum tubing, group-2 screws had the interlocking nail centrally located within 31.8-mm-diameter aluminum tubing, and group-3 screws had the interlocking nail eccentrically located within 31.8-mm-diameter aluminum tubing.

apart. Two sizes of aluminum tubing^c were used for the simulated bone model. The tubing had an outer diameter of 19 mm (small) or 31.8 mm (large) and a wall thickness of 2 mm. Radiographs of fracture repair in all dogs in which an interlocking nail was used at The Ohio State University from January 1999 to July 2001 were reviewed to select an appropriate size of aluminum tubing. From the radiographic views, maximal bone diameter and cortical thickness at the level of the proximal and distal metaphysis and at the diaphyseal isthmus were recorded by 1 investigator (RLA).

Testing system—Pilot holes were drilled (2-mm-diameter drill bit^d) and tapped (2.7-mm-diameter tap^e) across the diameter of the aluminum tubing. Screws that passed through a hole in the interlocking nail were then inserted. The aluminum tubing was secured to the base of the materials testing frame^d while the interlocking nail was connected to the actuator arm (Fig 1). Fatigue testing was performed under load control through the digital control program of the materials testing frame, with the screws cyclically loaded in 3-point bending. Each screw was cyclically loaded between 30 and 300 N at a rate of 10 Hz. The number of cycles until implant failure was recorded, with failure defined as either screw breakage or 1 mm of displacement at the point of maximum deflection. Testing stopped automatically when the implant failed. Each screw was tested in a new hole in the aluminum tubing to eliminate loosening of the screw within the hole by use of repeated cyclic loading. The interlocking nail was inspected after testing of each screw to ensure that there had not been any deformation of the nail hole.

Experimental design—Screws were tested in 3 simulated bone model configurations (Fig 2) with 6 screws/group. Group-1 screws had the interlocking nail centrally located within 19-mm-diameter aluminum tubing. Group-2 screws had the interlocking nail centrally located within 31.8-mm-diameter aluminum tubing. Group-3 screws had the interlocking nail offset by 5.5 mm from center (half the distance from the nail to the inner wall of the tubing) within 31.8-mm-diameter aluminum tubing.

Data analysis—Data were examined to determine the effect of bone diameter (group 1 vs 2) and eccentric loading (group 2 vs 3) by a Student *t*-test. A Bonferroni correction was applied, because there was multiple hypothesis testing with 3 possible comparisons; thus, values of *P* < 0.017 were considered significant.

Results

Bone diameter and cortical thickness—Radiographs of 12 femoral fractures and 1 humeral fracture that were repaired with 6- (n = 1) or 8- (12) mm-diameter interlocking nails were reviewed. Mean (± SD) bone diameters were 35.6 ± 7.1, 30.0 ± 6.1, and 19.4 ± 2.0 mm at the level of the proximal and distal metaphysis and isthmus, respectively. Mean cortical thickness at these 3 levels were 1.7 ± 0.7, 2.4 ± 0.6, and 3.0 ± 0.2 mm, respectively.

Table 1—Fatigue life of 2.7-mm-diameter cortical bone screws loaded in cyclic 3-point bending in a simulated bone model by use of aluminum tubing and a 6-mm-diameter interlocking nail

Measurements	Simulated bone models		
Diameter of bone model (mm)	19	31.8	31.8
Position of interlocking nail ^a	Central (6 screws)	Central (6 screws)	Eccentric (6 screws)
Mean (± SD) cycles to failure	761,215 ± 239,853 ^b	16,941 ± 2,829 ^c	43,068 ± 14,073 ^d

^aAs described in Figure 2.
^{b,c,d}Values with different superscript letters are significantly (*P* < 0.017) different.

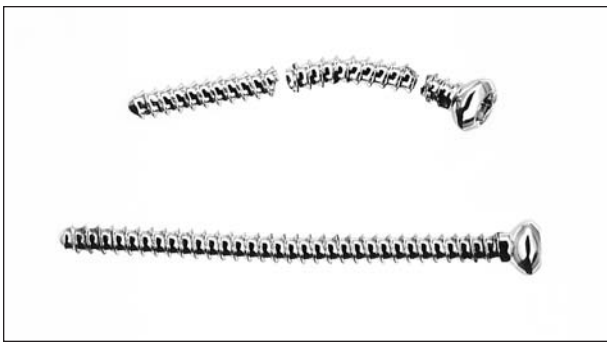


Figure 3—Photograph of cortical screws. All screws in group 1 failed by fracturing in 1 to 3 places. Group-2 and -3 screws failed by deformation without overt fracturing of the implant.

Fatigue testing—Fatigue life data for the 3 groups of screws are shown (Table 1). An increase in the diameter of the aluminum tubing from 19 to 31.8 mm resulted in a significant ($P < 0.001$) decrease in the number of cycles to failure. Eccentric loading of the locking screw resulted in a significant ($P = 0.006$) increase in the number of cycles to failure. All screws in group 1 failed by fracturing in 1 to 3 places (Fig 3). The location of these fractures was at the junction of the screw with the interlocking nail or inner wall of the tubing. Group-2 and -3 screws failed by deformation without overt fracturing of the implant.

Discussion

Bone is an anisotropic viscoelastic material that can modify its mechanical properties in response to *in vivo* loading. As such, bone lacks the consistent mineral density and strength that are essential for efficient mechanical testing.¹⁰ In addition, in fatigue strength studies, bone often breaks before the test material has signs of fatigue.¹¹ Because of the biologic variability of bone, manufactured materials are often used instead of bone to study the mechanical properties of orthopaedic implants. For these reasons, aluminum tubing was used instead of cadaver bone for the fatigue tests in our study. The diameter and wall thickness of the aluminum tubing used in our study were representative of the metaphyseal and diaphyseal bone diameter and cortical bone thickness, respectively.

There are no standards for testing interlocking nails and locking screws as there are for bone plates and screws. Three-point bending was selected for fatigue testing in our study on the basis of the clinical observations that screws failed by bending. Torsional forces acting on the bone-implant construct were not considered in our study and likely result in the locking screws being loaded in a bending mode that is more complex than simple 3-point bending.

The interlocking nail and tubing construct used in our study can be represented in a theoretical analysis by a beam fixed at both ends with a concentrated load either at center or offset from center (Fig 4). When the load is concentrated at the center of the beam, the maximum deflection is located at the point of load in the center of the beam and can be calculated from the following equation:

$$\Delta(\max) = P \times L^3 / 192E \times I$$

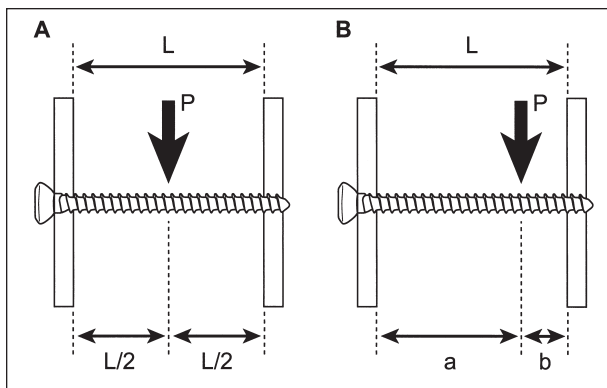


Figure 4—Schematic drawing of simulated bone model. Beam fixed at both ends with load concentrated at center (A) and offset from center (B). L = Length of the beam. P = Concentrated load on the beam. a = Measured distance along the beam to the point of load. b = Measured distance along the beam to the point of load such that $b = L - a$.

where $\Delta(\max)$ is the maximum deflection of the beam, P is the concentrated load on the beam, L is the length of the beam, E is the elastic modulus of the beam, and I is the moment of inertia of the beam.¹² When the load is concentrated at the center of the beam, the maximum moment is located at the point of load in the center of the beam and at both ends of the beam and can be calculated from the following equation:

$$M(\max) = P \times L / 8$$

where M(max) is the maximum moment in the beam, P is the concentrated load on the beam, and L is the length of the beam.¹² Increasing the length of the beam (ie, increasing the diameter of the tubing from 19 to 31.8 mm) increased the maximum deflection by 537% and the maximum moment by 85%. Under repeated loading, microcracks form in regions of high localized stress, and as the load fluctuates, crack propagation occurs as the crack opens and closes and progresses across the section.¹³ Increasing the maximum deflection and moment caused crack initiation to occur sooner and crack propagation to progress more rapidly, resulting in a shorter fatigue life of the locking screw. In addition, increasing the maximum deflection of the screw allowed failure by deformation without fracture of the implant in group-2 and -3 screws.

When the load is concentrated at a distance offset from the center of the beam (Fig 4), the maximum deflection and moment are no longer located at the point of load. Deflections and moments can be calculated from the following equations:

$$\begin{aligned} \Delta(a) &= P \times a^3 \times b^3 / 3E \times I \times L^3 \\ \Delta(\max) &= 2P \times a^3 \times b^3 / 3E \times I(3a + b)^2, \text{ at } x = 2a \times L / (3a + b) \\ M(a) &= 2P \times a^2 \times b^2 / L^2 \\ M(1) &= P \times a \times b^2 / L^2 \\ M(2) &= P \times a^2 \times b / L^2 \end{aligned}$$

where $\Delta(a)$ is the deflection at the point of load, $\Delta(\max)$ is the maximum deflection in the beam, M(a) is the moment at the point of load, M(1) is the moment in the left section of the beam, M(2) is the moment in the right section of the beam, P is the concentrated load on the beam, L is the length of the beam, a is the

measured distance along the beam to the point of load, b is the measured distance along the beam to the point of load such that $b = L - a$, E is the elastic modulus of the beam, I is the moment of inertia of the beam, and x is the point along the beam where the maximum deflection is located (as measured from the left).¹² Applying a load concentrated at a distance offset from the center of the beam (ie, eccentrically loading the screw such that $a = 3b$) decreased the maximum deflection by 31% and the deflection at the point of the load by 40%. While offsetting the load from the center increased the maximum moment (located at the right end of the beam) by 18%, the moment at the point of load was decreased by 29%, and the moment at the left end of the beam was decreased by 49%, resulting in a decreased total moment in the beam. Eccentric loading of the screw resulted in decreased deflection and total moment in the screw, resulting in a longer fatigue life.

Cyclic loading of an interlocking nail and bone construct is dependent upon the number of steps that a dog takes while ambulating. This number may vary as a result of the influence of factors such as size, age, and degree of discomfort. Among animals undergoing surgical repair of fractures, there is much variation in the use of the affected leg after surgery. Immediately after surgery, cycling (weight bearing) may be minimal as a result of pain. In addition, exercise restriction is usually imposed after surgery. Mean forefoot and hind foot stride frequencies at a walk are about 1.24 strides/s.¹⁴ Therefore, if a dog is walked 4 times daily for 5 to 10 minutes, then the number of cycles in an individual limb is estimated to be 1,500 to 3,000/d. Assuming normal weight bearing after surgery, our data predicts that fatigue failure of a centrally loaded 2.7-mm-diameter cortical bone screw within a large diameter bone would occur at about 17,000 cycles or within 6 to 11 days. These predictions do not take into account the use of 2 screws in most clinical situations, nor do they account for the increase in resistance of the interlocking nail and bone construct to fatigue failure that occurs as a fracture callus develops. In addition, a force approximately equal to 1 body weight (300 N) was chosen to cyclically load each screw, whereas peak vertical forces of only 60 to 70% of body weight are reached in the forelimbs and 40 to 50% of body weight in the hind limbs in clinically normal dogs during walking.^{15,16} The 6-mm-diameter interlocking nail is used in dogs that weigh approximately 20 to 30 kg, resulting in loads of 196 to 294 N.

Mechanically, an interlocking nail is weakest at the screw hole. The locking screws do not interact rigidly with the nail, and unlike a dynamic compression plate, they do not help to reduce the stresses in the nail at the screw holes. The area moment of inertia (ie, calculated bending stiffness) of a 6-mm-diameter interlocking nail with a 3.5-mm-diameter hole is 31% of a solid nail of equal diameter when the neutral axis is perpendicular to the screw hole.¹⁷ Failure of the interlocking nail through a screw hole has been reported¹ for the 6-mm-diameter nail with 3.5-mm-diameter screw holes in 8 of 56 dogs between 3 and 8 weeks after surgery, with 7 of those dogs requiring a second surgery. While reducing the screw hole size in the 6-mm-diameter inter-

locking nail from 3.5 to 2.7 mm increases the fatigue life of the nail, reducing the locking screw size compromises the fatigue life of the locking screw. Results of studies¹⁸⁻²⁰ that have analyzed the fatigue life of screws indicate that core diameter is the principal factor that determines fatigue life; screws with a larger core diameter have a longer fatigue life. Comparison of yield strength in 2.7- versus 3.5-mm-diameter screws in cantilevered bending found a reduction of 44% with the smaller diameter screw.²¹ The fatigue life of the 2.7-mm-diameter screw currently used with the 6-mm-diameter interlocking nail may be improved by increasing the core diameter of the screw to 2.7 mm. This could be accomplished by eliminating threads along the load-bearing portion of the screw.

Within a bone, locking screws are subjected to different loading conditions, depending on their location within the diaphysis or metaphysis. The results of our study indicate that the fatigue life of a 2.7-mm-diameter locking screw centrally loaded in the metaphyseal region of bone may be significantly shorter than in the diaphysis. Eccentric loading of the locking screw in the metaphysis may help to improve its fatigue life. Because of the configuration of a fracture, as well as the inherent size and curvature of the medullary cavity of canine bone, the surgeon does not necessarily have control of the eccentric placement of the interlocking nail within the metaphysis. The results of our study do not dictate a change in surgical technique or choice of interlocking nail. Rather, our data are important in providing improved understanding of the reasons for locking screw failure and greater insight into the likely fatigue life of an interlocking nail construct when the immediate postoperative radiographs are being evaluated. The surgeon may then use this knowledge to modify postoperative rehabilitation suggestions.

^aSynthes USA, Paoli, Pa.

^bInnovative Animal Products, Rochester, Minn.

^cMarmon/Keystone Corp, Butler, Pa.

^d858 Bionix servohydraulic materials testing frame, MTS Systems Corp, Eden Prairie, Minn.

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