In vitro mechanical evaluation of torsional loading in simulated canine tibiae for a novel hourglass-shaped interlocking nail with a self-tapping tapered locking design

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**Objective**—To describe a novel interlocking nail (ILN) and locking system and compare the torsional properties of constructs implanted with the novel ILN or a standard 8-mm ILN (ILN8) by use of a gap-fracture model.

**Sample Population**—8 synthetic specimens modeled from canine tibiae.

**Procedures**—An hourglass-shaped ILN featuring a tapered locking mechanism was designed. A synthetic bone model was custom-made to represent canine tibiae with a 50-mm comminuted diaphyseal fracture. Specimens were repaired by use of a novel ILN or an ILN8 with screws. Specimens were loaded for torsional measurements. Construct compliance and angular deformation were compared.

**Results**—Compliance of the ILN8 was significantly smaller than that of the novel ILN. Mean ± SD maximum angular deformation of the ILN8 construct (23.12 ± 0.65°) was significantly greater, compared with that of the novel ILN construct (9.45 ± 0.22°). Mean construct slack for the ILN8 group was 15.15 ± 0.63°, whereas no slack was detected for the novel ILN construct. Mean angular deformation for the ILN8 construct once slack was overcome was significantly less, compared with that of the novel ILN construct.

**Conclusions and Clinical Relevance**—Analysis of results of this study suggests that engineering of the locking mechanism enabled the novel hourglass-shaped ILN system to eliminate torsional instability associated with the use of current ILNs. Considering the potential deleterious effect of torsional deformation on bone healing, the novel ILN may represent a biomechanically more effective fixation method, compared with current ILNs, for the treatment of comminuted diaphyseal fractures. (Am J Vet Res 2006;67:678–685)

As a result of biological and mechanical advantages, ILNs are considered the standard of care for use in the treatment of most diaphyseal fractures of the humerus, femur, and tibia in people. However, as the use of ILNs has increased, clinical limitations have been reported, which prompted several studies. An in vivo study revealed a significant delay in bone healing and functional recovery, which was attributed to torsional and bending instability, when results for an ILN were compared with those for an external fixator in experimentally induced tibial fractures in sheep. A similar in vitro study revealed that torsional compliance and deformation of tibial constructs implanted with ILNs were greater than those of constructs implanted with a PRC. In that study, ILN constructs under torsional loads had up to 28° of slack, whereas PRC constructs underwent continuous deformation throughout the testing period. Similarly, an in vitro study of 9 ILNs implanted in human tibiae revealed that there was torsional slack in all constructs, regardless of nail design. Analysis of results of these studies suggests that current human and veterinary ILNs do not provide torsional stability as much as initially anticipated, thus potentially contributing to complications, such as delayed healing or nonunion of fractures.

Contrary to ILNs used in humans, which often are implanted after the medullary cavity is reamed, ILNs are routinely used in veterinary medicine without reaming of the medullary cavity. Reaming allows for implantation of larger, stronger nails and increases contact between the nails and endocortices, thereby potentially improving stability of the repair. However, reaming severely impairs the medullary blood supply and has been associated with a higher incidence of infection and fat emboli. In contrast, the use of thinner ILNs without reaming of the medullary cavity has potential biological advantages, such as preservation of the endosteal and medullary blood supply; however, the procedure places the construct at a mechanical disadvantage by reducing the nail-bone contact area and increasing the working length of the locking device. This suggests that although nails that do not require reaming of the medullary cavity may be preferable from a biological standpoint, from a mechanical stand-

**ABBREVIATIONS**

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
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<tbody>
<tr>
<td>ILN</td>
<td>Interlocking nail</td>
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<tr>
<td>PRC</td>
<td>Plate-and-intramedullary rod combination</td>
</tr>
<tr>
<td>AMI</td>
<td>Area moment of inertia</td>
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<tr>
<td>br-DCP</td>
<td>Broad–dynamic compression plate</td>
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<tr>
<td>ILN8</td>
<td>8-mm ILN</td>
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<tr>
<td>SCP</td>
<td>Screw-cone peg</td>
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<tr>
<td>ASTM</td>
<td>American Society for Testing of Materials</td>
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</table>
point, construct stability of ILNs used without reaming of the medullary cavity relies primarily on the integrity and efficacy of the locking mechanism. Current ILN designs allow a mismatch between screws or bolts and the nail hole, which precludes rigid interaction between locking devices and nails. Indeed, it has been reported that in vitro torsional instability results from the discrepancy between the screw or bolt diameters and that of the nail hole. Furthermore, it was suggested that, under torsion, this discrepancy may be exacerbated by flattening of the screw threads and by structural damages to the edges of the nail holes induced by the screw. To increase construct stability, repairs have been augmented with various implants. Although improved stability has been subjectively reported for use of such techniques, repairs are potentially time-consuming and invasive, thereby offsetting the biological advantages of intramedullary nails.

In addition to providing adequate stability, implants must be sufficiently strong to withstand loads during the early postoperative period, particularly when cortical continuity is not achieved or local instability may result in a prolonged healing time. The AMI of an implant characterizes its ability to resist bending and is inversely proportional to the stress of a section under bending. Because a larger AMI results in small-of an implant characterizes its ability to resist bending and efficacy of the locking mechanism. Current prototypes of the novel ILN system were designed to be of a size comparable to the commercially available $8 \times 185$-mm ILN.

Prototypes for the locking device and novel nail were designed to address several primary constraints. The novel locking device had to provide rigid interaction with the nail. The shape of the novel nail had to be such that it would limit interference with endocortices and facilitate fracture reduction while limiting the risk of joint infraction attributable to perforation of the distal subchondral bone plate. Finally, shape and size of the novel locking device had to enhance successful insertion during surgery.

Secondary constraints that addressed theoretic stiffness and strength of the locking device and novel nail were also considered for the design. The AMI of the novel locking device had to be greater than that of a 4.5-mm bone screw and comparable to that of a commercially available 3.5-mm locking bolt. The AMI of the novel nail at the level of a locking hole had to be similar to that of an ILN8 with 3.5-mm bone screws in both the mediolateral and craniocaudal planes. Finally, AMI of the weakest part of the solid central section of the novel nail had to be similar to that of the solid section of a 3.5-mm br-DCP.

To address these constraints, an hourglass-shaped nail featuring an oblong bulletlike distal tip and a novel SCP locking device were manufactured from 316L stainless-steel certified to ASTM F138 standards (Figure 1). This material was similar in chemical composition and mechanical properties to that used to manufacture the ILN8 used in the study (ie, 316L ASTM F139). Stainless-steel 316L ASTM F138 was chosen for use in manufacture of the SCP locking device and novel ILN components because it is the current material of choice for medical manufacturers of similar implants.

The outside diameter of the nail ends was 8 mm, whereas the central portion featured a reverse entasis that reduced the middle portion of the nail to an outside diameter of 6 mm. Two tapered holes (diameter, 4.0 and 3.2 mm, respectively) were placed 11 mm apart in each end of the nail. The most proximal and distal nail holes were separated by 155.5 mm. The SCP was designed as a self-tapping, cortical-type screw (core diameter, 4 mm) with a central Morse taper that matched the nail hole and a solid distal tip (outside diameter, 3.2 mm). Prototype SCps were manufactured in 2 lengths (26 and 30 mm).

Measurement of AMI—Established methods were used to calculate AMIs. The AMI values were calculated for various sections of the SCP, at the level of the nail hole, and for the central section for the novel ILN.

Preparation of the bone model—in an attempt to limit specimen variability and circumvent the need to procure canine bones, we used a custom-made synthetic tibial model consisting of 30% glass-filled structural nylon. This material was chosen on the basis of its material properties, which are similar to those of cortical bone (Appendix). To mimic a gap-fracture model, specimens were manufactured in 2 symmetrical halves. Each half of the bone model featured a segment used for linkage to the holding fixtures, a tapering segment representing the metaphyseal region, and a final segment representing the diaphysis (Figure 2). Thickness of the wall was 2.5 mm throughout. Overall, length of the tibial model (between each holding fixture and including a 50-mm central gap) was 210 mm.

To allow accurate and consistent placement of the bone screws or SCps, a dedicated custom-designed drilling fixture was used to drill 2 pilot holes in all bone models. Centers of
the holes were exactly 11 mm apart to match the nail holes. The outside diameter of the pilot holes was 2.5 mm for the ILN8 and 4.0 mm (cis cortex) and 3.2 mm (trans cortex) for the novel ILN. This standardized procedure allowed all nails to be precisely and reliably centered longitudinally within the bone model. A matching alignment fixture was used during implantation to ensure that all of the nails were also axially aligned within the bone models (Figure 3). Custom-made polyurethane foam plugs were inserted at the ends of each bone model to maintain the ILNs in a centralized position during testing.

Study design—The ILNs were applied to the synthetic tibial models (8 specimens/group). The ILN8 group used an 8 × 185-mm ILN with 4 bicortical 3.5-mm bone screws, whereas the novel ILN group used a 185-mm novel ILN and SCPs. Sample sizes were determined by use of a power analysis (power > 0.8) based on means and SDs obtained during a preliminary study.

Mechanical testing—Specimens were tested nondestructively in torsion in accordance with a protocol described in another study. All constructs were securely mounted in a custom-designed torsion fixture to ensure accurate axial alignment and consistent testing length for all specimens. The torsion fixture was linked to a servohydraulic testing machine via a pinion assembly. Tests were performed in load control with a torque of ± 5 Nm for 10 cycles.

Data acquisition—A 255-kg (2,500-N) load cell coupled with the hydraulic actuator recorded loads over time with simultaneous recording of actuator displacement at a sampling rate of 250 Hz. Because bone constructs were tested under load-control conditions, construct compliances (defined as the slope of the deformation-vs-load curve) were calculated in the 10th cycle between ± 1.5 Nm and ± 5 Nm by use of linear regression (r² > 0.99). Total construct compliance, calculated as the mean compliances during positive and negative loading, was used for statistical analysis.

Construct angular deformation was calculated from actuator displacement data and geometric dimensions of the torsion fixture. Construct slack was documented when the compliance curves appeared bimodal and was calculated as the difference between the y-axis intercept of the compliance slopes on the positive and negative loading curves.

Data analysis—Calculated AMIs of the SCP novel ILN, and ILN8 were qualitatively compared with AMIs of ILN8 and br-DCP reported in the literature. Construct compliance and angular deformation data were each compared by use of a 1-factor ANOVA. Student-Newman-Keuls post hoc tests were used whenever significant differences were identified. Significance was set at values of P < 0.05.

Results

Values for AMI—The AMI of the 3.5-mm screw, 4.5-mm screw, 3.5-mm bolt, and SCP was calculated. The AMI of various sections of the SCP (threaded section, 12.57 mm³ [core, 4.0 mm]; tapered midsection, 8.24 mm³ [core, 3.6 mm]; and smooth solid section, 5.15 mm³ [core, 3.2 mm]) were always larger than those of the 3.5-mm (1.63 mm³) or 4.5-mm (3.98 mm³) screws (core diameter of the 3.5- and 4.5-mm screws, 2.4 and 3.0 mm, respectively). In addition, the AMI for the threaded section of the SCP was 171% greater than that of a 3.5-mm bolt (7.36 mm³).
Table 1—Comparison of AMI for various sections of standard ILNs, a novel hourglass-shaped ILN, and a 3.5-mm br-DCP.

<table>
<thead>
<tr>
<th>Section</th>
<th>ILN8 (3.5 mm)*</th>
<th>Novel ILN*</th>
<th>ILN8 (4.5 mm)*</th>
<th>3.5-mm br-DCP</th>
</tr>
</thead>
<tbody>
<tr>
<td>Solid central section</td>
<td>201.06</td>
<td>63.62</td>
<td>201.06</td>
<td>Approx 551; approx 591</td>
</tr>
<tr>
<td>Nail hole</td>
<td>65.6</td>
<td>62.1</td>
<td>37.94</td>
<td>Approx 321</td>
</tr>
<tr>
<td>Mediolateral bending</td>
<td>174.18</td>
<td>171.43</td>
<td>146.45</td>
<td>Approx 5001</td>
</tr>
<tr>
<td>Cranio-caudal bending</td>
<td>11.95</td>
<td>12.65</td>
<td>11.95</td>
<td></td>
</tr>
</tbody>
</table>

Values reported for AMI are mm$^4$.
*Data computed by use of established methods.20 †Data obtained from another study.22 ※Values determined on the basis of a plate cross section of approximately 3.9 x 3.9 mm and a hole cross section of approximately 5.55 x 3.9 mm, assuming rectangular cross sections for the plate and screw hole.

Whereas the AMI of the smooth section of the SCP was 316% and 129% greater than that of a 3.5- or 4.5-mm screw, respectively, it was 30% less than that of a 3.5-mm bolt.

Values of AMIs for the novel ILN, ILN8 with 3.5- and 4.5-mm nail holes, and 3.5-mm br-DCP were calculated (Table 1). The AMI values for the novel ILN at the nail holes were similar to those of an ILN8 with 3.5-mm screws and larger than those of an ILN8 with 4.5-mm screws in both the mediolateral and cranio-caudal planes. The central section of the novel ILN had an AMI greater than that of a 3.5-mm br-DCP.

Construct compliance—Compliance curves for the ILN8 were bimodal, whereas compliance curves for the novel ILN were unimodal (Figure 4). In the bimodal curves for the ILN8, there was no quantifiable torque in the central region, which corresponds to the change in the direction of torque. For all practical purposes, this region represented slack in the construct and reflected an abrupt change in angular deformation without resistance to applied torques. Conversely, the unimodal shape of the compliance curves for the novel ILN reflects the lack of slack in the novel ILN implanted constructs. The ILN8 construct was significantly (P < 0.001) less compliant (mean ± SD, 0.821 ± 0.007/Nm), compared with compliance for the novel ILN construct (1.022 ± 0.011/Nm).

Angular deformation of constructs—Mean ± SD maximum angular deformation of the ILN8 construct (23.12 ± 0.65°) was significantly (P < 0.001) greater, compared with that of the novel ILN construct (9.45 ± 0.22°). Mean construct slack in the ILN8 group was 15.13 ± 0.63°. Mean angular deformation for the ILN8 construct once slack was overcome at high torques (7.97 ± 0.05°) was significantly (P < 0.001) less, compared with the mean value for the novel ILN construct (9.45 ± 0.22°).

Discussion

Although favorable clinical outcomes have been reported23 in small animals for use of currently available ILNs, clinical studies24 have also revealed intraoperative or postoperative instability of ILNs. In those studies, 12% to 14% of the animals had delayed healing or required supplementation of the initial ILN repair to provide adequate stability. Similarly, delayed union rates as high as 18% have been reported21 in humans treated by use of tibial nails without reaming of the medullary cavity. Although the optimal mechanical environment favorable to bone healing remains controversial, it is generally accepted that excessive interfragmentary motion delays bone healing, in contrast to results for controlled axial micromotion.35-40 In particular, the effect of shear motion on fracture healing continues to be contentious. Indeed, although numerous studies35-40 in sheep and dogs have indicated deleterious effects of torsional and shear motion on early bone healing and functional recovery, another study25 in rats revealed that the callus 2 and 4 weeks after surgery was significantly larger in tibial fractures subjected to local shear strains, compared with the callus in those with rigid stabilization. However, ex vivo motion at the fracture site 2 weeks after surgery was significantly greater in tibiae subjected to shear strains. Results of such studies, supported by experimental evidence of construct instability with current ILNs, emphasize the need for a more effective ILN design.7,9 In an attempt to improve construct stability and functional recovery, a study25 would be beneficial.
was conducted to evaluate an experimental nail featuring bolts threaded into the nail holes in vitro and in vivo by use of sheep with gap fractures of the tibiae. That study revealed that although experimental and conventional nails had similar stiffness in mediolateral bending and axial compression, the experimental nail was significantly stiffer in shear and craniocaudal bending. In addition, the experimental nail had higher (but not significantly so) stiffness in torsion. As a result, the experimental nail group had significantly smaller interfragmentary motion throughout the 9 weeks of the experiment, which translated to superior bone healing (as evaluated by use of histomorphometry, radiography, and ex vivo biomechanical testing) and faster and more complete functional recovery (as evaluated by use of gait analysis).39

On the basis that there was no slack for the novel ILN, we determined that the ILN described here had better torsional stability, compared with that for the currently available ILNs. Interestingly, the novel ILN had a deformation pattern similar to that for PRCs evaluated in another study22 (ie, lack of torsional slack and angular deformation of the construction of approx 9.5° and 12°, respectively, for identical loading conditions). This suggests that the novel ILN design may improve the biomechanical environment for fracture healing.

To increase construct stability, the SCP was designed with a central Morse taper, thereby eliminating motion at the interface between the nail and locking device. A Morse taper features a matching trunnion (male component) and bore (female component) and is commonly used to reliably join modular components during total hip arthroplasty.41 In addition, the tapered design facilitates the insertion of the SCP by providing a self-centering feature, even when there is slight misalignment between the cortical pilot hole and nail hole. This may help reduce the incidence of misaligned holes during treatment of fractures with ILNs.

The strength and failure pattern of ILNs depend on the design and size of the nail hole and locking device.42,43 In comminuted fractures, implants are mainly subjected to bending stresses.20 Therefore, the AMIs used in the study reported here were calculated with respect to theoretical bending conditions. This choice allowed for comparison with results of ILNs described in the literature. The AMI of an implant is a structural property that characterizes the geometric distribution of the material with respect to the axis of loading. All things being equal, an implant with a larger AMI will sustain lower stresses during cyclic loading, which in turn will extend its fatigue life.44 One limitation of early ILNs designed for veterinary use was the weakness of the nail holes, which, because of the sharp local decrease in AMI, acted as stress concentrators that led to nail failure.21,22 On the basis of the AMIs, ILNs are weakest at the nail hole in mediolateral bending. Reducing the diameter of the screw hole from 4.5 mm (early ILN8 design) to 3.5 mm (current ILN8 design) results in a 5.7-fold increase in local AMI, which translates into an 8-fold increase in fatigue life of an 8-mm nail.32 This design change improves structural properties of a nail to the detriment of a thinner or weaker screw as indicated by approximately a one-third decrease in AMI.32 Because of the failure patterns of 6-mm nails and screws and increased strength of ILNs, the novel ILN reported here was devised so that the AMI of the locking device was equal to or greater than that of the 4.5-mm screw and the AMI of the nail at the holes was similar to that of the ILN8 with 3.5-mm screws. Design of the SCP was such that its smallest AMI (ie, AMI for the peg section) had to be greater than that of a 4.5-mm bone screw; AMI for the peg section was 3.15 mm², whereas AMI for the 4.5-mm bone screw was 3.98 mm², which represented an increase of approximately 30%. At the same time, the smallest AMI for the SCP could be smaller than that of a 3.5-mm bolt (7.36 mm²). This design constraint was selected on the basis of the fact that failure of 4.5-mm screws has not been reported for the first generation of ILNs. Furthermore, although the corresponding AMI of the novel ILN (62.1 mm²) was slightly smaller than that of the ILN8 with a 3.5-mm screw (65.6 mm²), it is 64% greater than that of an ILN8 with a 4.5-mm screw, thus suggesting that the novel ILN should have an estimated fatigue life similar to that of the currently available ILN8s with 3.5-mm screws. Moreover, the improved stability of the novel ILN (no slack) could potentially result in shorter healing time, thereby rendering the slight theoretic decrease in fatigue life of the novel ILN clinically irrelevant.

The hourglass design of the nail provides several potential benefits. First, this design will contribute to preserving the endosteum and improve restoration of the medullary blood supply after implantation,13,16 This shape may also facilitate implantation of the nail in curvilinear bones by limiting contact between the nail and endocortices. The impetus for use of a relatively large nail is often guided by the necessity to have a strong locking mechanism.46 Consequently, it is recommended that the largest possible nail be used, principally on the basis that the fatigue life of an 8-mm nail is 10 times that of a 6-mm nail.47 Alternatively, a comparable fracture could be treated successfully by use of a 3.5-mm br-DCP.48 Because the AMI of the solid section of a 6-mm nail (approx 64 mm³) is similar to that of a 3.5-mm br-DCP (approx 55 mm³), it can be argued that the strength of an ILN8, which has an AMI for the solid section of 201 mm³, is not warranted. Furthermore, the use of larger medullary implants can substantially impede the cortical blood supply.49

Finally, the bullet-shaped distal tip of the nail was designed to facilitate reduction of the fracture, particularly with regard to restoring length of the bone without increasing the risk of penetration of the distal joint associated with the use of trocar points. The oblong tip of the novel ILN could facilitate insertion through the proximal metaphysis and permit deep anchorage in the distal metaphysis, compared with results for the flat truncated tip in currently available ILNs; this may allow for treatment of more types of fractures.

Shape and material properties of bones used in a study50 in humans have SDs that exceed 100%. These SDs are likely to be even greater in dogs on the basis of...
the large number of breeds and conformations. This amount of variation implies that several hundred specimens may be required to statistically detect significant differences. This is unsuitable for evaluation of implant biomechanics, particularly during the development phase. A model of a canine tibia was created for the study reported here to limit interspecimen variability and thereby allow for a better evaluation of the implant. Simple tubes of various materials have been used repeatedly for fatigue testing; however, complex shapes may be more appropriate to model physiological loads when devices for fracture fixation are evaluated. In an attempt to create a more realistic tibial model, tapered ends created on the basis of measurements obtained by our laboratory group were incorporated into the model to mimic the larger metaphysis. In another experimental study that used a bone model, ILNs were more susceptible to failure when implanted in the center of larger bones. Accordingly, to subject the novel ILN to the most stringent conditions and allow meaningful comparison between groups, all nails were locked near the ends of the model (outside diameter, 26 and 23 mm) and were maintained in a central location by use of a polyurethane plug.

Accurately simulating the strength of bone is most important when evaluating attachment devices such as screws; however, in nonfailure tests, mechanical properties that directly affect any alteration of the screw-bone interface during loading have not been determined. Therefore, general mechanical properties of human and canine cortical bone were used as guidelines for choosing an acceptable material. Although models of human tibiae are commercially available, they are too large to represent tibiae of giant breeds (405 to 375 mm in length) and are cost prohibitive for most veterinary studies. The 30% glass-filled structural nylon composite chosen for use in the study reported here was commercially available and relatively inexpensive. The current bone model provided comparable results to those of another study in which investigators evaluated ILNs by use of canine tibiae.

The larger compliance of the novel ILN construct, compared with compliance for the ILN8 construct, was attributed to the difference in the core diameter of the solid section of the nails (6 and 8 mm for the novel ILN and ILN8, respectively), which resulted in a 3-fold increase in AMI. Interestingly, compliance of the novel ILN construct group (1.022°/Nm) was comparable to that of a PRC construct (1.230°/Nm) reported in a similar study conducted by our laboratory group.

Overall angular deformation of the ILN8 construct was greater than that for the novel ILN construct primarily because of slack in the ILN8 system. Slack, which corresponds to a lack of resistance to applied torque, has been associated with looseness between the screw and nail. There was rigid interaction between the ILN8 and locking screws with increasing torque, which enabled the construct to resist applied loads. Resulting deformation for the ILN8 construct at high torque (approx 8°) was similar to the overall deformation for the novel ILN construct (9.5°). Similar to results for compliance, this difference was attributed to disparity in diameter of the core of the nails.

Evaluation of a PRC construct in another study conducted by our laboratory group revealed a similar overall angular deformation (approx 12°) to that for the novel ILN in the study reported here. In addition, there was no slack in the PRC construct.

Analysis of the study reported here suggests that reengineering of the locking mechanism in a novel hourglass-shaped ILN has eliminated the instability associated with the use of current ILNs inserted without reaming of the medullary cavity. Furthermore, the improved torsional stability of the construct matched that of a comparable PRC construct, a device routinely used to treat comminuted diaphyseal fractures. In contrast to standard plate osteosynthesis, an ILN can be applied remotely from the fracture site and be used to reduce and stabilize fractures by use of a closed technique. This less-invasive approach to fracture repair improves early bone healing. Because of the potential combined mechanical and biological benefits of this novel hourglass shape, the novel ILN could represent an effective and safe alternative to plate osteosynthesis while preserving the advantages of nails that do not require intramedullary reaming. Additional biomechanical studies to evaluate fatigue strength of the novel ILN are warranted. In vivo studies of the novel ILN should also be conducted to fully assess the potential use of the new hourglass-shaped ILN.

References

7. Schandelmaier P, Krettek C, Tischer H. Biomechanical...
## Appendix

Values for material properties of human and canine cortical bone and a composite synthetic nylon material.

<table>
<thead>
<tr>
<th>Material property</th>
<th>Human cortical bone</th>
<th>Canine cortical bone</th>
<th>30% glass-filled structural nylon*</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ultimate tensile strength (MPa)</td>
<td>106.8 (53–135)†</td>
<td>NA</td>
<td>140 (65–195)</td>
</tr>
<tr>
<td>Ultimate compressive strength (MPa)</td>
<td>158.8 (145.1–166.7)†</td>
<td>112.8‡</td>
<td>140</td>
</tr>
<tr>
<td>Ultimate shear strength (MPa)</td>
<td>68§</td>
<td>NA</td>
<td>7 (59–85)</td>
</tr>
<tr>
<td>Young modulus: E (GPa)</td>
<td>14.9 (8.2–17)II</td>
<td>12.26‡</td>
<td>7.2</td>
</tr>
<tr>
<td>Poisson ratio</td>
<td>0.49 (0.46–0.58)II</td>
<td>NA</td>
<td>0.35</td>
</tr>
<tr>
<td>Density (g/cm³)</td>
<td>1.9¶</td>
<td>0.84#</td>
<td>1.35</td>
</tr>
</tbody>
</table>

Values reported are mean or mean (range).

*Information reported elsewhere.† Wet femoral compact bone obtained from humans between 10 and 79 years of age.‡ Mean of wet long bones (femur and humerus).§ Humans ranged from 19 to 80 years of age.ǁ Results reported in another study.¶ Results reported elsewhere.ǁ Results reported in another study.\* NA = Not available.