**In vitro** biomechanical evaluation and comparison of a new prototype locking plate and a limited-contact self compression plate for equine fracture repair

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**Keywords**
Locking plates, locking screws, equine, fracture repair

**Introduction**

There are a variety of postoperative complications that may contribute to an unsuccessful outcome in horses undergoing surgical management of long-bone fractures, such as laminitis of the contralateral limb, surgical site infection, implant loosening, implant failure, and catastrophic failure of the repaired bone (1, 2). In order to minimize the risk of development of any of these complications it is essential for the equine patient to be able to ambulate normally immediately following surgery and for the implants to be strong enough to withstand the high loads experienced during weight bearing (3). The current trend in AO internal fixation is biological osteosynthesis, which emphasizes preservation of the blood supply of the periosteum and the surrounding soft tissues. Reported benefits of biological osteosynthesis over the traditional techniques in human patients include lower infection rates, reduced blood loss, and suitability for minimally invasive percutaneous osteosynthesis, all of which could reduce the complication rate in horses undergoing fracture repair (4–6). One such method of biological repair includes the use of locking compression plates⁴ (LCP) in combination with locking screws to create an internal fixator, which does not rely on friction between the plate and the bone for implant stability. Because the plate is not compressed to the bone by the screws, the periosteum does not need to be removed (potentially preserving blood supply), and the plate does not need to be accurately contoured. Additionally, due to its design, the LCP can be used as a conventional compression plate, as a locked internal fixator, or as an internal fixation system providing absolute stability in reconstructable fractures or relative stability in non-reconstructable fractures, depending on the fracture location and configuration (4, 6,

**Methods:** The plates alone were tested in four-point bending single cycle to failure. The MC3-plate constructs were created with mid-diaphyseal osteotomies with a 1 cm gap. Constructs were tested in four-point bending single cycle to failure, four-point bending cyclic fatigue, and torsion single cycle to failure.

**Results:** There were no any significant differences in bending strength and stiffness found between the two implants. The MC3-NP-LP construct was significantly stiffer than the MC3-LC-SCP in bending. No other biomechanical differences were found in bending, yield load in torsion, or mean composite rigidity. Mean cycles to failure for bending fatigue testing were similar for both constructs.

**Clinical significance:** The NP-LP was comparable to the LC-SCP in intrinsic, as well as structural properties. The NP-LP construct was more rigid than the LC-SCP construct under four-point bending, and both constructs behaved similarly under four-point bending cyclic fatigue testing and torsion single cycle to failure. The new NP-LP implant fixation is biomechanically comparable to the LC-SCP in a simulated MC3 fracture.

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⁴ Synthes, West Chester, Pennsylvania, USA

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tested constructs in this study had a lower overall stiffness than an equine bone and implant construct would exhibit (9). The constructs in that study included the dynamic compression plate (DCP), the limited-contact dynamic compression plate (LC-DCP), the LCP, and the clamp rod internal fixator for repair of a simulated midshaft oblique fracture. Their results showed the LCP to be superior due to its high yield strength, stiffness and the least amount of motion at the fracture site (9). Another study compared the LC-DCP and LCP constructs in osteotomized canine femurs. In that study the LCP construct was stiffer at the level of the osteotomy gap than the LC-DCP construct in four-point bending (10). Likewise, Sod and colleagues also observed biomechanical superiority when comparing LCP and LC-DCP constructs in osteotomized equine third metacarpal bones (MC3) (11).

Recently a new prototype 4.5 mm locking plate (NP-LP) has been developed for use in equine fracture repair. Similar to the current 4.5 mm LCP design, this plate has a combination hole, which accepts either locking or traditional cortical screws. However, it differs in four main features including: stacked combination holes at both ends of the plate, the acceptance of larger diameter screws, the shape and design of the locking head, and the angulation of the locking portion of the combination hole, which diverges by seven degrees from the longitudinal axis of the plate (Fig. 1–3).

The plate holes of the NP-LP allow the placement of either 5.5 mm cortical screws or 5.5 mm locking screws in the combination hole rather than the 4.5 mm cortical or 5.0 mm locking screws that are used in the current 4.5 mm LCP design. Finally, although the thread angle and pitch configuration is similar to a traditional cortical screw, the locking screws have a threaded head that is cylindrical in shape rather than conical. This is intended to reduce the possibility of cold welding that can occur with the current titanium and stainless steel LCP system, thus eliminating the need for a torque limiting device during application (12, 13).

> There are two broad categories of locking plates: fixed-angle and variable-angle locking plates. In the latter, the screw can be locked with a certain tolerance within a cone at an angle in the range of one to 15 degrees from perpendicular. The mechanism locking the screw in the plate also comes in two types; in the first, the screw head is locked in its chamber by a threaded locknut, whereas in the second, the screw head is itself threaded and screws into the plate or into an adapted lip (8).

Several in vitro studies comparing the mechanical properties of different well-established implants against the LCP have shown the later to be biomechanically superior (9–11). One such study compared the bending properties of four different constructs using a bone substitute. This bone surrogate has similar properties as equine long bone in bending tests. However, this material has half the Young’s modulus of immature equine bone, thus the

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* Canewasit tubes: Erhard Hippe KG, Spremberg, Germany
* Equilock®: Securos, Fiskdale, Massachusetts, USA
The objectives of our study were to determine and compare the bending stiffness, strength and structural stiffness of the NP-LP to the limited-contact self compression plate (LC-SCP) using a four-point bending test single cycle to failure. Additionally, we wanted to determine and compare the mean yield load, mean yield bending moment, mean failure load, mean failure bending moment, and mean composite rigidity of the NP-LP and LC-SCP when used to repair a simulated mid-diaphyseal comminuted fracture of the equine third metacarpal bone. We hypothesized that both implants would possess similar biomechanical characteristics when tested alone and that the NP-LP construct would be stronger than the LC-SCP under bending and torsional loads when applied in osteotomized equine third metacarpal bones.

Materials and methods

Single cycle testing of bone plates alone

An eight-hole 4.5-mm broad NP-LP was compared to an eight-hole 4.5 mm broad LC-SCP. Comparison of the characteristics of the broad LC-SCP and broad NP-LP is provided in Table 1. Three constructs of each plate design were tested to failure according to the American Society for Testing and Materials (ASTM) F382–99(2003) standards which provide a comprehensive reference test method for bone plates used in the surgical internal fixation of the skeletal system (14).

Mechanical testing

A jig specially designed for this study was connected to a materials testing frame to support the plate under loading, in accordance with the recommendations set forth in the ASTM Standard Specification and Test method for Metallic Bone Plates F382–99 (2003), Annex A1 and A2 (14). For all plates, the outer and resting roller support spans were located at equal distances (36 mm) away from the adjacent loading roller so that two screw holes were located between the adjacent loading and support rollers. The loading rollers were in contact with the surface of the bone plate intended to be in contact with the bone. The long axis of the plate was aligned perpendicular to the axis of the rollers.

The test was conducted at a constant displacement rate of 0.1 mm/s until obvious plastic deformation was reached (15). Force and displacement data were acquired at 10 Hz and recorded using commercially available software.

Data analysis

Single cycle bend testing of the bone plates was performed in accordance with the ASTM. The data collected were used to establish the bending stiffness of each bone plate defined by the ASTM standard as the maximum slope of the linear elastic portion of the load versus load-point displacement curve (N/mm). The 0.2% offset method was calculated using the formula: 

\[ q = 0.002 \cdot a \]

where 'a' equals the centre span roller distance (36 mm). The point where the bending moment deflection curves intersected the offset bending stiffness slope was the bending strength (Nm). The bending structural stiffness was determined using the expression:

\[ E_{1} = \frac{(2h+3a)K|b}{} \]

where 

\[ E = \text{modulus}, \quad I_{e} = \text{moment of inertia}, \quad K = \text{bending stiffness}, \] 

\[ a = \text{centre span distance}, \] 

\[ h = \text{loading span distance}. \]

Table 1  Comparison of the characteristics of the broad new prototype 4.5 mm locking plate and the broad limited-contact self compression plate.

<table>
<thead>
<tr>
<th>Measurement parameters</th>
<th>Broad 4.5 mm NP-LP</th>
<th>Broad 4.5 mm LC-SCP</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thickness (mm)</td>
<td>5.2</td>
<td>5.2</td>
</tr>
<tr>
<td>Width (mm)</td>
<td>17.5</td>
<td>17.5</td>
</tr>
<tr>
<td>Hole length (mm)</td>
<td>8.6</td>
<td>8.6</td>
</tr>
<tr>
<td>Hole spacing (mm)</td>
<td>18</td>
<td>18</td>
</tr>
<tr>
<td>Area moment of inertia*</td>
<td>0.153</td>
<td>0.147</td>
</tr>
<tr>
<td>Polar moment of inertia*</td>
<td>1.16</td>
<td>0.16</td>
</tr>
</tbody>
</table>

Key: NP-LP = new prototype locking plate; LC-SCP = limited-contact self compression plate. *Cross section between screw holes.

Bone plate and third metacarpal bone construct testing

Following owner consent, twelve pairs of equine MC3 bones were harvested from adult horses of different breeds, weighing more than 450 kg, which were euthanatized for reasons other than musculo-skeletal disease. The bones were stripped of soft tissue and the MC3 bones were wrapped in saline soaked towels followed by plastic to prevent dehydration. They were stored in pairs at ~20 °C until testing could be performed. The bones were thawed at room temperature (20–22 °C) and kept moist in saline solution-soaked towels until use.

Of each pair of limbs, one limb was randomly assigned to the experimental group in which a NP-LP was used in the creation of a bone and implant construct, while a LC-SCP was used to make a similar construct in the matched limb.

Following completion of the constructs, the 12 pairs of MC3 were divided into three test groups (4 pairs each) for (1) four-point bending single cycle to failure testing, (2) four-point bending cyclic fatigue testing and (3) torsional testing to failure.
Specimen preparation

For the MC3-NP-LP construct preparation, the plate was placed on the dorsal surface of each MC3 and secured in place with bone reduction forceps in the proximal and distal holes of the plate. A 5.5 mm cortical screw was inserted in the fourth and fifth holes in neutral position (all holes were numbered from the end of the plate closest to the proximal end of the bone). The plate-MC3 construct was secured in a positioning jig, and an oscillating saw\(^6\) with a 0.4 mm thick blade\(^6\) was used to make a 1 cm osteotomy gap in the mid-diaphysis between the fourth and fifth hole of each plate. The screws were tightened to compress the plate against the bone. The remaining holes were filled using 5.5 mm locking screws using a 4.0 mm drill bit through a threaded drill guide attached to the locking area of the combination hole of the plate, followed by the creation of the threads using a 5.5 mm tap guarded by a 5.5 mm sleeve. The MC3-LC-SCP constructs and osteotomy gap were created in the same fashion as the MC3-NP-LP constructs (Fig. 4). All holes were filled with 5.5 mm cortical screws placed in neutral fashion. For all constructs, the bone was cooled during drilling by irrigation with saline solution at 20 °C. All cortical and locking screws were tightened with a limited torque screw driver\(^7\) to a torque of 4.5 Nm.

Mechanical testing

For four-point bending, each MC3-NP-LP and MC3-SCP construct was positioned on a bending fixture within the materials testing device for both single cycle to failure and cyclic fatigue testing. For all constructs, the outer support span was 18 cm and the inner (palmar) support span was 4 cm (Fig. 4). This was done in order to avoid unequal contact of the support points due to the irregular surface of the bone (11, 16, 17). The constructs were tested at a constant displacement rate of 6 mm/s until failure, and the force (Newtons) applied at the time of failure was recorded. The force and displacement data were acquired at 0.1 second intervals. Failure was defined as the point at which the recorded load first dropped suddenly, indicating some form of structural damage.

For four-point bending in a cyclic fatigue testing, a load of 0–1000 N was applied at 4 Hz using a haversine wave for 136,800 cycles, which corresponds to a limb loading activity of a stalled horse during a period of one month (21). The test was stopped at the time of construct failure or when the number of cycles was completed. The number of cycles to failure was recorded, with failure determined by a sudden decrease in load and then assessing by visual, radiographic and videotape means, if there was plastic deformation of implants, fracture of implants, screw loosening, bending or breakage. During testing the specimens were moistened with physiological saline solution periodically.

Utilizing a materials testing device\(^7\), the torsional testing was set up and performed by drilling a 6.2 mm hole through the bone perpendicular to its longitudinal axis at the proximal and distal metaphyses 3.5 cm away from the plate ends. The construct was then placed in jigs and secured in place by placing 6.4 mm diameter smooth pins\(^8\) which were tightly inserted through holes in the jig and the bone in the proximal and distal portion of the construct. The constructs were attached to the testing device with the longitudinal axis of the mid-shaft of the MC3 aligned along the axis of loading (Fig. 3). The displacement was recorded at a constant rate of 0.17 rad/s until rotation of 0.873 rad was attained. Torque and angle data were recorded at 0.1 second intervals using a computer software system\(^9\).

Radiographs were made before and after testing to examine the manner in which the constructs failed using a storage phosphor screen\(^10\) system. Latero-medial and dorso-palmar radiographic views\(^11\) of MC3 at KVP 90, 250 mA, 17 mAs, were performed to visually assess repair techniques and site of implant failure. All tests were videotaped\(^12\).

Data analysis

Load deformation curves were generated for all tests. For four-point bending and torsion tests, load was represented as the bending moment (Nm) and torque (Nm) respectively. The mean (± SD) was calculated for composite rigidity, yield bending moment, failure bending moment, yield load and failure load in four-point bending, and composite rigidity and yield load for torsion. The composite rigidity was defined as the slope of the linear portion of the bending moment/angular displacement or torque/angular rotation curves. The yield points were determined by the offset method of two percent according to ASTM, which represents the amount of deviation from the load-deformation curve at which point permanent deformation of the sample occurs. For four-point bending, the ultimate failure point was

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\(^{6}\) Compact Air Drive\(^6\): Synthes, West Chester, Pennsylvania, USA

\(^{7}\) Standard Tooth Sawblade, model 532.067S: Synthes, West Chester, Pennsylvania, USA

\(^{8}\) Limited torque screw driver: CDI Torque Products, City of Industry, California, USA

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**Fig. 4**

A) Line drawing of a new prototype 4.5 mm locking plate construct, dorso-palmar and lateral views. B) Illustration of a construct in a four-point bending jig (left) and in a torsion jig (right).

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determined as the point of maximum load past which failure of the sample occurs, determined by a sudden drop of the plastic region of the curve. The data, trend lines and intersection points were plotted with a software programme.

In four-point bending, the loads and displacements that were recorded were converted to applied bending moments and bending angles as previously described using the following formulas (9):

\[ \alpha = 2\arcsin\left(\frac{x}{h}\right) \]

Where \( F \) = force (yield or ultimate failure load) and \( h = 70 \text{ mm} \) (distance of 70 mm between the inner and outer supports).

\[ \alpha = \text{bending angle}, \ x = \text{displacement} \ (\text{mm}) \text{ and } h = \text{lever arm} \]

Data were analyzed using a commercially available software programme.

Constructs curves for each test were analyzed with the aid of the videotapes to help with the identification of yield and failure points. Radiographs were used to assess the sites of bone and implant failure. The presence of screw loosening was determined by tightening each screw with the torque limiting screwdriver and noting if the screw turned, both by visual and tactile assessment. The location and fracture type were described, and the bone, plate and deformation of screws were recorded for all constructs.

Statistical analysis

Plates

The mean and standard deviation of bending stiffness, strength and structural stiffness were calculated. Confidence intervals were determined around the differences between means. Paired t-tests were used to compare the mean values between groups. Statistical significance was set at \( p < 0.05 \).

Constructs

The mean and standard deviation of yield load, yield bending moment, failure load, composite rigidity and failure bending moment for four-point bending in single cycle to failure, cycles to failure in four-point bending, and the yield load and composite rigidity for torsion for each construct configuration (MC3-NP-LP, MC3-LC-SCP) were calculated. Paired t-tests were used to compare the mean values between groups, since the bones assigned to each pair of samples came from the same horse. Statistical significance was set at \( p < 0.05 \).

Results

Mechanical Testing

Plates

Plastic deformation was achieved in all of the bone-plate constructs by applying a maximum load ranging from 1732.8 to 2593.5 N (mean = 2214.1 N). None of the plates fractured during this mechanical testing. Both the NP-LP and LC-SCP plates bent at the screws holes. The differences between groups for bending strength (\( p = 0.196 \)), bending stiffness (\( p = 0.11 \)) and bending structural stiffness (\( p = 0.11 \)) were not significant (Table 2).

Constructs

The differences between the MC3-NP-LP and MC3-LC-SCP constructs in the mean yield load, mean yield bending moment and failure load (\( p = 0.3 \)) under four-point bending in single cycle to failure were not significant. However, the MC3-NP-LP construct had a significantly greater composite rigidity than the MC3-LC-SCP (\( p = 0.02 \)) (Table 2). Under fatigue testing, the mean number of cycles to failure under four-point bending for the MC3-NP-LP constructs (85,094.5 ± 39,559) and the MC3-LC-SCP constructs (84,192.25 ± 30,680) were not significantly different (\( p = 0.97 \)).

Moreover, the differences between groups in yield load (\( p = 0.085 \)) or mean composite rigidity (\( p = 0.056 \)) in single cycle to failure torsional testing were not significant (Table 4).

Failure mode analysis

Single cycle to failure, four-point bending

MC3-NP-LP failure mode: Failures occurred by fracture(s) at the level of the screws adjacent to the osteotomy (n = 3) or by fracture between the first and second screw (n = 1).

MC3-LC-SCP failure mode: Failures occurred by fissures at the cis cortex through the fourth and fifth screws (n = 1), by complete fracture through the first and second screw (n = 2), and by a fracture between the fifth and sixth screw (n = 1).

In all constructs, permanent deformation of the plate at the gap level occurred.

Cyclic fatigue testing, four-point bending

MC3-NP-LP failure mode: Failure occurred in three of the four constructs by fracture of the plate through the fifth hole.

Table 2

<table>
<thead>
<tr>
<th></th>
<th>Broad 4.5 mm NP-LP</th>
<th>Broad 4.5 mm LC-SCP</th>
<th>Difference (95% CI)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bending stiffness (N/mm)</td>
<td>817.83 ± 7.40</td>
<td>906.42 ± 61.98</td>
<td>88.59</td>
</tr>
<tr>
<td>Bending structural stiffness (Nm²)</td>
<td>15.90 ± 0.14</td>
<td>17.62 ± 1.20</td>
<td>1.72</td>
</tr>
<tr>
<td>Bending strength (Nm)</td>
<td>33.37 ± 7.76</td>
<td>44.55 ± 2.83</td>
<td>11.18</td>
</tr>
</tbody>
</table>

Key: NP-LP = new prototype locking plate; LC-SCP = limited-contact self compression plate; CI = confidence interval.
Table 3  Mechanical testing parameters (Mean ± standard deviation) for the MC3-NP-LP and the MC3-LC-SCP constructs under four-point bending in single cycle to failure. MC3 = third metacarpal bones, NP-LP = new prototype 4.5 mm locking plate, LC-SCP = limited-contact self compression plate.

<table>
<thead>
<tr>
<th></th>
<th>MC3 – NP-LP</th>
<th>MC3 – LC-SCP</th>
<th>Difference (95% CI)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Yield load (N)</td>
<td>4370.32 ± 608.25</td>
<td>3349.34 ± 1107.40</td>
<td>1020.99 [-854.41 – 2896.39]</td>
</tr>
<tr>
<td>Yield bending moment (Nm)</td>
<td>153.33 ± 21.51</td>
<td>118.02 ± 39.24</td>
<td>35.31 [31.11 – 101.73]</td>
</tr>
<tr>
<td>Composite rigidity (Nm/rad)</td>
<td>1166.34 ± 167.32*</td>
<td>734.22 ± 356.20*</td>
<td>432.11 [152.03 – 1016.27]</td>
</tr>
<tr>
<td>Failure load (kN)</td>
<td>11744.41 ± 1036.64</td>
<td>105009.78 ± 2027.68</td>
<td>1234.63 [2145.67 – 4614.92]</td>
</tr>
<tr>
<td>Failure bending moment (Nm)</td>
<td>421.45 ± 36.49</td>
<td>376.46 ± 70.95</td>
<td>44.97 [73.45 – 163.39]</td>
</tr>
</tbody>
</table>

Key: MC3 = third metacarpal bones; NP-LP = new prototype locking plate; LC-SCP = limited-contact self compression plate; CI = confidence interval.
* Indicates that differences are significant (p = 0.02).

Table 4  Mechanical testing parameters (Mean ± standard deviation) for MC3-NP-LP and MC3-LC-SCP constructs under torsion in single cycle to failure. * Indicates significance (p = 0.02). MC3 = third metacarpal bones, NP-LP = new prototype 4.5 mm locking plate, LC-SCP = limited-contact self compression plate.

<table>
<thead>
<tr>
<th>1 cm Gap</th>
<th>MC3 – NP-LP</th>
<th>MC3 – LC-SCP</th>
<th>Difference (95% CI)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Yield load (N)</td>
<td>60.90 ± 8.36</td>
<td>67.24 ± 6.28</td>
<td>6.34 [8.58 – 21.26]</td>
</tr>
<tr>
<td>Composite rigidity (Nm/rad)</td>
<td>201.35 ± 25.41</td>
<td>247.08 ± 10.69</td>
<td>45.73 [4.81 – 86.65]</td>
</tr>
</tbody>
</table>

Key: MC3 = third metacarpal bones; NP-LP = new prototype locking plate; LC-SCP = limited-contact self compression plate; CI = confidence interval.

at the locking portion of the combination hole. One construct did not fail after completing the total number of cycles.

MC3-LC-SCP failure mode: All constructs failed by fracture of the plate either at the fifth (n = 3) or fourth (n = 1) hole.

Single cycle to failure, torsion

There was plastic deformation of the plate in all MC3-NP-LP and MC3-LC-SCP constructs in the region of the osteotomy gap.

Discussion

We assessed the mechanical properties of two orthopaedic implants by evaluating the traditional LC-SCP plate, and comparing these to the properties of a NP-LP prototype for stabilization of MC3 fractures in adult horses.

We did not find any significant differences in bending stiffness, strength and bending structural stiffness between the LC-SCP and new NP-LP implant. The NP-LP and LC-SCP have similar cross sectional dimensions through their screw holes, similar bending stiffness and area moment of inertia which give to both plates similar mechanical properties. These findings were also validated in previous studies (10, 15). Consequently, the selection of the NP-LP over LC-SCP should not be based on implant strength. Despite the similarities between the NP-LP and LC-SCP implants, a major advantage of the locking system is its ability to minimize disruption of the blood supply in highly comminuted diaphyseal fractures and provide greater screw fixation strength in human osteoporotic bone (3). The locking system between the locking screws and the plate creates a rigid apparatus, a single beam construct between the plate, screws and bone, where there is reduced motion between its components (11).

In general, the in vitro mechanical properties of gap-osteotomized MC3 stabilized with the new designed locking plate tested in four-point bending in single cycle to failure and torsion were not significantly different than those from the MC3-LC-SCP constructs. Both constructs resisted physiological tensile forces on the dorsal aspect of MC3, and exceeded the yield bending moment and mean failure bending moment calculated by Rybicki et al. (18). However, the composite rigidity in four-point bending was significantly greater for the MC3-NP-LP construct. Composite rigidity is a factor, which determines the effect of the internal fixation on long bone remodelling. Cortical bone remodels according to functional stress demands, and if an implant is too rigid, it can limit the stress experienced by the bone, thus leading to decreased bone deposition and generalized weakening of the bone in the later stages of healing (16, 19). On the other hand, an implant that provides a greater composite rigidity would allow a smaller amount of strain at the fracture site (19, 20). The increased composite rigidity offered by the MC3-NP-LP construct suggests that this new-prototype LP may result in a higher stress shielding, taking away the load from the bone. Recent research has shown that early temporary osteoporosis of the bone in contact with implants does not depend upon the degree of unloading (stress shielding) but rather on the amount of vascular damage caused by the implant (21). These characteristics could be beneficial in early stages of recovery for a horse that has to withstand its own weight immediately after surgery.

There are several factors regarding the experimental design of this particular study that are important to mention such as the one centimetre gap osteotomy, the
loading rate and the distance between supports for the force application in the four-point bending setting. The one centimetre gap was chosen to simulate a comminuted fracture model as previously described (16, 22, 23). A high loading rate was chosen to represent a high- and high-energy loading event as determined in earlier studies (11, 17, 22). Additionally, although ASTM standards recommend a testing configuration that locates the loading rollers at approximately the one-third points between the supporting rollers, in our study the distance between the two inner support points could not be more widely spaced as this would have caused uneven loading due to the conical shape of the bone (14, 16).

The goal of the cyclic test was to determine the fatigue limit of the implants and to compare the mean total number of cycles between the two different fixation methods. We found that both construct types behaved similarly for cyclic loading. In order to compare the mechanical fatigue properties of both implants we chose to load each construct up to 25% of the mean yield load of the MC3-LC-SCP construct (~1000N) from the single cycle to failure tests results. Arbitrarily, we selected to limit the total number of cycles to 136,800, which is an equivalent of one month of limb loading activity of an adult horse confined in a stall (24). All but one of our constructs failed before reaching the maximum amount of cycles and the mean number of cycles to failure was equivalent to 18 days of limb loading activity in an adult horse, respectively. In our study, in all of the constructs that suffered fatigue failure, the plate fractured without evidence of the screws loosening or of bone fracture, which was assessed with the limited torque screwdriver and radiographs respectively. The fatigue mode of failure found in studies performed by Sod and colleagues using a different locking plate system is different from the one encountered in this study. However in those studies the constructs were assembled by compressing the osteotomy ends rather than leaving an osteotomy gap and using screws that had different sizes and patterns of placement, factors that ultimately can have an effect on the number of cycles to failure (11, 17, 25, 26). In our model, the failure occurred at the level of the screw holes even though all screw holes were filled and therefore more protected. This failure mode might have been the result of our gap model, in which the edge of the bone was quite close to the screw, and therefore the protection provided by the screw was reduced. Additionally, the gap distance between the inner and support loads were different in our study than those used previously. The larger screws were placed in areas of increased stress, whereas the greater holding stress is most useful at the end of the plate and at the level of the osteotomy, which could explain the differences observed in our study versus those by Sod and colleagues (11, 17, 25, 26).

In torsion testing, there were no significant differences between the construct types. The mean yield load and the composite rigidity were also not significantly different for the NP-LCP and LC-DCP fixation. This can be explained by the equivalent polar moment of inertia values for the NPD-LCP and LC-DCP. Plastic deformation was observed at the osteotomy site in torsion in all of the constructs, however, no failures were recorded in any of the fixation models presumably because the 5.5 mm screw size used in this experiment provided higher ultimate failure loads when compared with previous studies (11, 16, 17).

One limitation of our study was the variability of the MC3 collected for the experiment and the small number of specimens used. The MC3 were collected from euthanatized patients and since our hospital population Hospital name is mostly adult horses with a wide range of ages and a wide variety of uses, we likely obtained bones of different sizes, densities, ash contents, and failure stresses (27). These factors could have added more variability to our results. Many of the pair-wise comparisons we made in this study were not significantly different, which could be attributed to the relatively low power of the comparison tests. Post hoc power calculations indicated that approximately 38 pairs of cannon bones would be needed to detect a significant difference between the two methods of fixation. Due to financial constraints we could not add more samples to our study.

Both types of constructs resulted in failure of the plate at the level of the screw holes. In order to avoid failure of the implant at the screw holes, we suggest that the limited contact grooves be shifted from their current alignment to a proposed alignment between the screw holes. By shifting the limited contact grooves so that they are centred between the screw holes, we anticipate that the plate will fail between screw holes rather than at them. In addition, the current plate design results in an area moment of inertia that is not constant along the plate as the screw holes are located at the same level than the undercuts that provide the limited contact, thus reducing the cross sectional area of the plate.

In conclusion, we did not find any significant differences in the structural properties of the NP-LP and the LC-SCP implants in acute testing. Therefore the selection of the NP-LP over the LC-SCP should not be based on the strength of the fixation. Construct testing revealed that the MC3-NP-LP was more rigid than the MC3-LC-SCP in four-point bending single cycle to failure, but there were not any significant differences observed in four-point bending fatigue to failure or torsion to failure. Additional testing after implementation of our recommendations regarding the design of the underside of this plate should be performed. In addition, further investigations should be performed comparing axial compressive stability and cyclic torsion before performing in vivo studies.

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Conflict of interest

This study was partially supported by Securos, Inc. as they provided the plates and screws. Securos, Inc. had no role in the collection, analysis and interpretation of the data, in the writing of the manuscript, or in the decision to submit the manuscript for publication.
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