**In vitro biomechanical comparison of the effects of cerclage wires, an intramedullary pin and the combination thereof on an oblique osteotomy of the canine tibia**

J. van der Zee
Valley Farm Animal Hospital, Faerie Glen, Pretoria, South Africa

**Keywords**
Cerclage wires, intramedullary pin, oblique tibia osteotomy, biomechanics

**Summary**
**Objective:** To compare the in vitro biomechanical effects of single loop cerclage wires, an intramedullary pin and the combination thereof as applied to an oblique mid-diaphyseal osteotomy of canine tibiae.

**Methods:** Three groups of nine bones with long oblique osteotomies were repaired with the following methods: 1) Three single loop cerclage wires and a transcortical skewer pin, 2) intramedullary pinning with a smooth Steinmann pin, and 3) a combination of both methods. The repaired constructs were tested in a single cycle four-point-bending test to failure. Load displacement curves were drawn and the following parameters were calculated or extrapolated: Stiffness, load at yield, and force resisted at 2 mm actuator displacement. The latter was determined to demonstrate the difference in the amount of energy absorbed between the different groups.

**Results:** The stiffness and force resisted at 2 mm displacement of the groups with cerclage wires were significantly higher than the group with an intramedullary pin alone (p ≤0.05). The differences in stiffness (p = 0.15) and force required at 2 mm displacement (p = 0.56) between cerclage wires and the combination of cerclage wires and intramedullary pins were not significant.

**Clinical relevance:** Cerclage wire repair results in higher stiffness than repair with an intramedullary pin. When cerclage wires are combined with an intramedullary pin, the intramedullary pin does not provide protection to the cerclage wire repair and the wires or the bone under the wires has to fail before the pin resists significant load.

**Introduction**
The use of intramedullary pins and cerclage wires in fracture repair is an inexpensive method which can give good results if correctly used (1–3). When not correctly applied however, this method does lead to frequent complications often resulting in nonunion of the fracture (4–7). Cerclage wires are often the first line of treatment for long oblique periprosthetic proximal femoral fractures as a complication of total hip arthroplasty in dogs and humans (8–11). Furthermore, cerclage wires are still frequently used in human orthopaedic surgery for spiral tibial fractures or oblique metacarpal fractures (12, 13). A lot of research has been done on the optimal application technique for the use of cerclage wires (14–18). Although there are many published guidelines for the use of intramedullary pins and cerclage wiring in small animals, the author could not find any reports which compare the biomechanical effects of the use of cerclage wires alone, with intramedullary pins alone, or the combination of these two methods as applied to a fracture simulation (1, 19–23). The lack of these data has been mentioned previously (24). The objective of this study was to determine if an intramedullary pin, when placed according to the current guidelines, protects the repair with cerclage wires of a long oblique osteotomy. The hypothesis was that the osteotomies stabilized with cerclage wires and an intramedullary pin combined would not have significantly higher stiffness than the group with cerclage wires alone and that the groups with cerclage wires would have higher stiffness than the group with an intramedullary pin alone.

**Materials and methods**
**Specimen preparation**
Twenty-seven tibiae were collected from cadavers of 14 dogs that were euthanatized for reasons other than this research project. The tibiae were selected from dogs older than one year of age, between 18 and 25 kg...
body weight, and free from clinical and radiographic signs of disease. Tibiae with a medullary diameter of 6.5 mm to 8 mm and with a length (proximal tip to distal tip) of 175 mm to 205 mm were selected by measurements made of radiographs of the bones. All soft tissues and the fibulae were removed after which they were wrapped in paper towels, soaked in 0.9% saline, identified and frozen at –15°C. The bones were allocated into size groups using a two-directional table of bone diameter and bone length, with three dimension groups for both parameters (Appendix Table 1 - available online at www.vcot-online.com). The random block test was used to allocate nine bones to each of the three experimental test groups. Prior to testing, each tibia was thawed for 12 hours at room temperature. A fracture model was created in the bones as follows: The bones were mounted in a special jig that was mounted on an electrical band saw. The jig ensured that the bones were sawn perpendicularly to the sagittal plane and precisely medio-lateral, from a proximal caudal point, exactly through the midpoint, to a distal cranial point. The osteotomy was made at an angle of 20 degrees to the long axis. The bone length, cranio-caudal external diameter at midpoint, cranio-caudal medullary diameter at midpoint, and length of the osteotomy were measured with a Vernier calliper and recorded. The repair procedures were performed by the same person (JvdZ).

**Experimental design**

**Group I (C): Fixation with three single loop cerclage wires and a transcortical skewer pin**

The sawn bones were reduced with a large towel clamp that was placed perpendicularly to the osteotomy line, thereby creating compression on the osteotomy and preventing displacement thereof. A 1.5 mm hole was pre-drilled perpendicularly to the osteotomy plane, about 2 mm lateral from the sagittal midline and crossing the osteotomy in the middle of its length. A 1.6 mm Kirschner wire was placed through this hole, and the ends cut to approximately 5 mm in length. The Kirschner wire was placed away from the midline to avoid interference with the intramedullary pin in group III (C+IMP). A 1.25 mm stainless steel single loop cerclage wire with a pre-formed loop was placed around the skewer pin, with the knot on the medial side and tightened with a wire tightener according to the prescribed clinical use. Two more single loop cerclage wires were placed on either side of the oblique wire, 6 mm from the osteotomy ends (Figure 1 A, B).

**Group II (IMP): Fixation with intramedullary pin only**

A 4.8 mm Steinmann pin, corresponding to 60% to 74% (medullary diameter 8 mm to 6.5 mm) of the medullary canal diameters was introduced in a retrograde method into the medullary canal of the proximal half of the sawn bone. The fragments were reduced with a bone clamp. The intramedullary pin was seated into the distal fragment until the tip became visible in the subchondral bone of the distal segment. The proximal end was cut flush to the bone. The skewer pin was not included in this group (Figure 1 C).

**Group III (C+IMP): Fixation with three single loop cerclage wires, transcortical skewer pin and intramedullary pin**

A 4.8 mm Steinmann pin was inserted retrograde into the centre of the proximal segment. A transcortical skewer pin and single loop cerclage wires were applied after reduction as in group I (C). The reduced bone was held with a bone clamp between the distal two wires whereafter the pin was seated into the distal segment and cut flush to the bone as in group II. No interference from the skewer pin or skewer pin deflection was noted (Figure 1 D, E).

**Mechanical testing procedure**

After repair, the bones were mounted in purpose made moulds ensuring accurate midpoint positioning between the two moulds where the bone ends were embedded in methyl methacrylate. The space

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**Figure 1** Radiographs of the repair methods: Cerclage wire with skewer pin (A+B), intramedullary pin (C), intramedullary pin + cerclage wire and skewer pin (D+E).

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a Röth Medical Components, Cape Town, South Africa
b Synthes, Johannesburg, South Africa
in the mould ends allowed for different lengths of bone to be embedded with the same final length for the testing jig to ensure that the applied bending moments on all the specimens were equal. All the bones were then mounted in a four-point-bending testing fixture so that the caudal surfaces would contact the inner contact points of the jig when it was pulled upward by the testing machine (Figure 2). The jig was connected to a universal testing machine c. The inner loading points of the testing jig were fixed and were spaced 60 mm apart. The outer loading points were designed to accept the epoxy embedded bone-ends. They were mounted with swivelled steel cables on a horizontal bar, which was attached to the top machine connection. The design of the top part of the jig allowed any anisometry of the bone-epoxy complex to balance out before the four-point-load was applied. The effective loading point distance between the outer loading points was 180 mm. This resulted in a moment arm of 60 mm. The testing machine specifications were set for a general tensile test as used for a cylindrical geometric object d. The tensile forces were measured with a 10 kN load cell. A sampling rate of 10 Hz was used. The specimens were loaded at a rate of 0.8 mm/s until failure or until the cross-head of the testing machine had moved through 20 mm. Load and displacement data were captured by the integrated software. Stiffness of the linear elastic region and load at yield point (at 0.2% offset) were calculated by the software. For each test the force resisted at 2 mm actuator displacement was recorded. The mode of failure was documented.

Data analysis

The mean values, standard deviations, maximum and minimum values were determined for all bone dimensions and tested parameters. To confirm even distribution of bone diameter and bone length between the three groups, the dimensions were compared by one-way analysis of variance (ANOVA). The stiffness, load at yield, and force resisted at 2 mm actuator displacement of the three groups were compared by ANOVA and any differences were confirmed by Bonferroni multiple comparisons. A p-value of less than 0.05 was accepted as significant.

Results

Standardization of bone specimens

There was no significant difference between the dimensions of the three groups of bones (Appendix Table 2 - available online at www.vcot-online.com). An osteotomy angle of 20° was used to give a theoretical value of osteotomy length: diameter of 2.93:1. The actual average value for osteotomy length to external diameter was 2.67:1. In all cases, the osteotomy length was more than 2.5 times the external diameter.

Mechanical parameters

Load-deformation curves were obtained for all specimens (Figure 3). The values of the tested parameters are presented in Table 1.

The stiffness of group I (C) was not significantly different from group III (C+IMP). The stiffness of both group I (C) and group III (C+IMP) was significantly higher than that of group II (IMP). There was no difference in yield load between the three groups (p ≥ 0.12).

The force resisted after 2 mm of actuator displacement for group II (IMP) was significantly lower than the other groups (p ≤ 0.05). There was no significant difference between groups I (C) and group III (C+IMP) (p = 0.56).

Mode of failure

The characteristic modes of failure of some of the samples are depicted in Figure 4 and are explained in the following text. Seven of the nine samples in group I (C) failed by fracture of the bone at or close to a wire. Four of these seven had partial unbending of wires with a fracture (Figure 4A) and the other three had no visible unbending of wires (Figure 4B). Two specimens of group I (C) failed by unbending of the wires (Figure 4C). In group II (IMP), all constructs failed by distraction of the two fragments due to bending of the intramedullary pin. Four of nine developed failures were associated with bending of the intramedullary pin. Two failures were seen on the tension side with minimal bending of the intramedullary pin.

Table 1 Descriptive statistics of the measurements recorded during testing to failure in four-point bending. Data are presented as a mean ± standard deviation for each configuration. Within each row of the table, mean values that share the same superscript lower case letter are not significantly different.

<table>
<thead>
<tr>
<th></th>
<th>Cerclage (n = 9)</th>
<th>IMP (n = 9)</th>
<th>Cerclage + IMP (n = 9)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stiffness (Nm/mm)</td>
<td>11.7 ± 1.7 a</td>
<td>7.9 ± 1.1 b</td>
<td>14.0 ± 1.6 a</td>
</tr>
<tr>
<td>Load at yield (N)</td>
<td>445.8 ± 120.8 b</td>
<td>475.4 ± 183.7 a</td>
<td>548.6 ± 143.7 a</td>
</tr>
<tr>
<td>Force at 2mm displacement (N)</td>
<td>233.2 ± 34.2 a</td>
<td>112.2 ± 18.0 b</td>
<td>251.5 ± 39.9 a</td>
</tr>
</tbody>
</table>

a  IMP = intramedullary pin.
fractures after a large amount of displacement (Figure 4D). Failure of specimens in group III (C+IMP) was very similar to group I (C), followed by bending of the intramedullary pin (Figure 4 E, F).

Discussion

This study revealed that bones stabilized with single loop cerclage wires were stiffer than those stabilized with intramedullary pins alone, and that in the constructs where intramedullary pins and cerclage were used together, there was no improvement in stiffness by adding the pin. While this study did not attempt to define all the mechanical properties of the implants as used, it was attempted to define the mechanical behaviour and failure mode of the repair methods tested, when subjected to a bending moment. The canine tibia was used because of its relative straight conformation and because of the frequent use of intramedullary pinning with or without cerclage wiring in clinical fractures (1, 27). When a repaired tibia is loaded in vivo, the bone-fixation complex is subjected to compressive, bending and torsional loads (28). In this study, constructs were only evaluated in caudal bending perpendicular to the plane of the osteotomy. Torsional loading was found to be the best technique of determining the strength of bone and bone-stabilization complexes, but as intramedullary pinning has no resistance against torsion, it was not considered as the best test for this study (29). Intramedullary pinning as used in dogs does not fill the entire medullary canal and will not fully resist the shear forces along an oblique fracture that result from compressive loads (1, 30).

As the main function of intramedullary pinning is the neutralization of bending forces, and because cerclage wires also resist bending forces, it was decided to test the repaired specimens in a bending test (31). A four-point-bending test has been demonstrated to apply an even bending moment over the entire test specimen without the localised concentration of force due to the fulcrum point of a three-point-bending test (31).

Many of the specimens repaired with the single loop cerclage wires failed by bone fracture without wire deformation, which means that the strength of the single loop cerclage wires was close to the strength where the bone would fail before the wire. Using double loop cerclage wiring could have made the difference between the groups with cerclage wires and the group without cerclage wires even bigger, as they resist more load before unbending.
and some specimens failed by unbending of the wires (32).

The decision to do a comparison of the load resisted with 2 mm actuator displacement of the machine head was made to demonstrate an important concept. This value relates to the elastic potential energy absorbed by the specimen at a specific displacement (33, 34). To include all the test specimens, a value had to be used where none of the specimens had undergone plastic deformation. Displacement of 2 mm was chosen as a convenient value close to the yield point of the weakest specimens.

Reduction with the cerclage wires prevents distraction of the bone fragments, imitating the geometry of solid bone and resulting in bending of the bone-implant complex until slipping of the bone under the wires, deformation of the wires, or bone fracture occurs (35). Adding an intramedullary pin to the bones repaired with wires, according to the current recommendations, did not add significant bending stiffness or increase load required to resist 2 mm displacement compared to bones stabilized with wires only. This is probably due to two factors: 1) The pin itself is not very stiff due to a small area moment of inertia relative to the construct and it has a long working length as applied here. 2) There is the potential for considerable bone movement relative to the pin because the pin does not fill the medullary cavity.

The area under the load-displacement curve represents the energy or ‘work’ done to create that amount of displacement, and for linear responses, is related to the stiffness of the structure (33). The difference in the force resisted after 2 mm of displacement was analyzed to highlight this difference between the groups. This value also represents the elastic potential energy that the structure is absorbing up to that point, as no plastic deformation has occurred. The constructs with wires only (C), resisted significantly more force than IMP at the same amount of actuator displacement in this initial phase. Adding an intramedullary pin to the constructs with wire as in C+IMP, did not show a significant difference from C. Therefore, in the elastic phase of loading, the wires and bone of the C+IMP group absorb most of the energy rather than the pin, and are therefore not effectively protected from failure by addition of an intramedullary pin.

The groups with cerclage wires failed by bone fractures at a cerclage wire, with or without cerclage unbending. Most of the fractures occurred underneath the middle wire, adjacent to the skewer pin. The main difference between the two cerclage groups came after the yield point. Constructs in Group C were completely unstable after failure, whereas the main fragments of constructs in Group C+IMP were held in some alignment, albeit with significant deformation of the wires, pin, and bone with resultant malalignment and instability. The mode of failure of IMP was very different from either C or C+IMP as there was fragment separation with bending of the pins and a lot more deformation. In the samples that did develop a fracture (4/9), the fracture type was very different from C and C+IMP as shown in Figure 2. This study confirms that the cerclage wires had a distinctly different effect on the constructs than the intramedullary pin, with higher stiffness, confirming the initial hypothesis and with a different mode of failure.

Conclusion and clinical recommendations

The stabilization of a long oblique fracture of a canine tibia with single loop cerclage wires as tested in a four-point-bending test was not significantly protected by the addition of an intramedullary pin. If cerclage wires are going to be used in combination with intramedullary pinning, the emphasis must be on the correct use of cerclage wiring with an intramedullary pin as support. The addition of supplementary fixation such as an external fixator should also be considered.

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Conflict of interests

No conflicts to declare. The results of this study were presented in a thesis as a requirement of the MMEDVET degree at the Medical University of South Africa.

References