Introduction

External skeletal fixation (ESF) is a popular method for the stabilization of long-bone fractures in small animals (1, 2). It offers several advantages over other fixation systems, such as minimally invasive fracture repair and the possibility of staged disassembly (1-3). A large variety of ESF devices are available, including the widely used IMEX SK™ system (4).

The treatment of long-bone fractures in patients weighing less than 5 kg has always been challenging for the veterinary surgeon. Complications such as nonunion, delayed union or malunion are more common in toy breed dogs (2, 5-9). An inherent biomechanical instability, impaired blood supply and complications concerning soft tissue coverage have been suggested to explain the higher prevalence of complications in very small animals and especially toy breed dogs (2, 5, 6, 10-13). Compared to external coaptation or intramedullary pinning, several studies suggest that ESF may be advantageous in the treatment of such fractures (5, 7, 9, 13, 14). Retrospective clinical studies found different bone healing rates of 89% to 100% for bone plates, 93% for ESF, 50% for intramedullary pinning, and 43% for external coaptation (2, 6, 7, 9, 15-20). Better results regarding pinning have been achieved with interlocking nails and intramedullary fully-threaded pins (21, 22). In comminuted fractures, ESF stabilization had fewer complications than plate fixation (14). In highly unstable fractures a type II configuration is recommended (23).

In order to prevent malunions and non-unions, adequate rigidity in early fracture healing is crucial (1, 3, 10, 11). On the other hand a certain amount of interfragmentary motion has been shown to enhance fracture healing (24, 25). However, the ideal rigidity of an external fixator remains unclear, and optimal values have so far only been described in a model of axial interfragmentary movement (25).

Toy breed dogs appear to be at a higher risk of stress protection (6, 7, 20). External skeletal fixation, especially in combination with techniques of staged disassembly, has been associated with increased bone stability and a lower degree of bone resorption compared to plate fixation (2, 3). The size and weight of ESF components are of great importance in very small animals (15, 16). Excessively large and heavy constructs may further impede the animal’s use of the affected leg, which may impair bone remodelling (8). Another challenge is the need for close pin placement in short fragments of juxta-articular fractures. Currently, this is a major limitation with most available ESF systems as well as bone plate systems (15).
A new universal micro external fixator system (UMEX™) has been introduced in human medicine to treat delicate orthopaedic conditions (26). The UMEX system is available in a smaller (alpha) and a larger (beta) clamp size and may prove to be a valuable alternative in the treatment of long-bone fractures in very small animals. The aim of our study was to biomechanically test three different UMEX alpha configurations. As a reference we wanted to compare the UMEX constructs to an established ESF system. However, due to the lack of a comparably sized ESF system, we compared the UMEX constructs to a widely used SK mini configuration in regard to stiffness, space needed for pin placement, and weight. We hypothesized that the biomechanical properties of the different UMEX configurations would show patterns comparable to the mini SK.

Materials and methods

External skeletal fixator constructs

An in vitro biomechanical experiment was conducted using solid, 10 mm diameter Delrin plastic rods\(^a\) as bone models in a 1 cm fracture gap simulation. Rods were cut to 170 mm lengths and predrilled at 20 mm increments in one to two planes, depending on the configuration of the construct. Pin separation across the fracture gap was 30 mm. Three different UMEX\(^b\) configurations and one SK\(^c\) configuration were assembled, accommodating 1.8 mm cortically threaded negative-profile half-pins and 1.8 mm non-threaded full-pins\(^d\) (Figure 1). Distance from cortex to connecting clamp (fixation pin working length) was set at 15 mm. The UMEX systems were constructed with alpha clamps and 316L knurled stainless steel rods (2.5 mm diameter). Alpha clamps are 8 mm long and have a width of 4.5 mm, weighing 1.7g. They accept pins with a non-threaded diameter of up to 2 mm. Clamping action is completed by simple clockwise tightening of a single insert screw. This set screw pushes the connecting rod against the pin, which is in turn trapped in the teardrop portion of the clamp hole. The special teardrop design of the pin insertion holes ensures a three-point fixation of pins (Figure 2). The UMEX connecting rods are manufactured with specially knurled surfaces to improve pull-out strength at the pin-rod interface. The SK ESF systems consisted of mini clamps and stainless steel rods (3.2 mm diameter). The 3.2 mm rod is the only size that fits into the SK mini clamps (4). The diameter of this rod was too large for the UMEX alpha clamps, therefore different rods had to be used in this experiment. Three different six-pin (three pins per fragment) UMEX constructs were investigated: Type Ia, type Ib, and type II modified. The type Ia frame was a standard unilateral, uniplanar frame. The type Ib frame was assembled as a unilateral, biplanar frame with three pins per segment of which the proximal and distal segmental pins were connected with the lateral rod. The middle segmental pins were connected with the cranial rod. The plane of the fixation frame was defined as the medio-lateral direction. In addition, a standard unilateral, uniplanar SK external fixator was assembled.

The frames were constructed using established ESF application techniques, with all pins placed with low speed power insertion (300 rpm or less) after being pre-

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\(^a\) Delrin rod: Angst & Pfister, Zurich, Switzerland
\(^b\) Micro Universal Fixator alpha: Adler Mediequip Pvt. Ltd., Sangameshwar, India
\(^c\) External Skeletal Fixation System mini: IMEX Veterinary Inc., Longview, TX, USA
\(^d\) Negative profile pins: Adler Mediequip Pvt. Ltd., Sangameshwar, India
drilled with a 1.5 mm drill bit into the Delrin rods. The Type Ia frames had pin placement through the central longitudinal axis of the Delrin rods. The Ib frames had pin placement through the central longitudinal axis of the Delrin rod, with 90° pin divergence. In the type II modified frame, the most proximal and most distal pins were full-pins, the inner four half-pins were placed in an alternating manner. Full-pins were not threaded; half-pins had a negative-profile thread. All clamps were tightened with one Nm torque using a precision torque wrench.

**Construct testing**

The same ten constructs of each design were tested under different loads in the following order: axial compression, craniocaudal-bending, mediolateral-bending, and torsion. Tests were performed on a servo-hydraulic testing machine (Figure 3). For axial compression, the load was applied along the longitudinal axis of the Delrin rod. The craniocaudal-bending and mediolateral-bending tests were performed following the ASTM F382 - 99(2008) Standard Specification and Test Method for Metallic Bone Plates guidelines. The load was applied in the transverse plane, via four-point bending, with the proximal and distal ends of the Delrin rod placed on support rollers 180 mm apart, and the load applied via two loading rollers with a 60 mm gap, positioned symmetrically on both sides of the fracture gap. For torsional loading, the proximal and distal rod ends were clamped rigidly in cylindrical metal sleeves aligned with the rotational axis of the testing machine. Frames were loaded in the elastic range to 30 N for axial compression, 50 N for craniocaudal-bending and mediolateral-bending, and 0.2 Nm for torsion. The test values for very small animals (less than 5 kg) were extrapolated from biomechanical studies in larger dogs as no references could be found in the literature. The actuator speed was set at 1 mm/sec for axial compression and bending and 0.5°/sec for torsion. Each frame was repeatedly loaded four times, and stiffness was determined from the data of the fourth loading cycle by determining the slope of the linear portion of the load displacement curve. For axial compression and torsion, the slope of the curve provides a direct measure of compressive stiffness (N/mm) and torsional stiffness (Nm/°), respectively. For bending tests, the slope of the load/displacement curve is defined as the bending stiffness K (N/mm), according to the ASTM standard.

For each configuration rod, pins and clamps were weighed on a precision balance. Ten samples of each configuration were weighed and the average was calculated.

Furthermore, the closest possible pin placement of three UMEX alpha and SK mini systems was evaluated after the disassembly of the constructs. Alpha clamps can be placed in direct proximity to each other by using a hexagon key on the inside of the clamp. In SK mini clamps, the hex wrench must encompass a nut; therefore a small space is required between SK clamps to enable them to be tightened (Figure 2).

**Data analysis**

Data recording and statistical analysis were performed with spreadsheet software and statistical software. Group measurements were described using means, medians and 95% confidence intervals based on standard errors. Due to the small sample sizes (n = 10) per group, data were recorded as dot plots. A Kruskal Wallis one-way analysis of variance on ranks test was utilized to assess differences in constructs between the four groups of interest. If the Kruskal Wallis test was significant (p < 0.05) a series of Mann-Whitney U tests for the difference in medians was run for pairwise comparison. The threshold level for statistical significance was set to α = 0.05 / 6 = 0.0083 to adjust for the multiple comparison testing.

**Results**

Comparing the three UMEX constructs, we found that the type II modified was significantly stiffer in axial compression, torsion, and mediolateral-bending than type Ia and type Ib (α <0.0083). (Figure 4) In craniocaudal-bending, there was no significant difference between UMEX type II modified and UMEX type Ib, while both constructs were significantly stiffer than the UMEX type Ia (α <0.0083). The UMEX type Ib was significantly stiffer than UMEX type Ia also in all other loads (α <0.0083).

In comparison to the established SK system, the UMEX type II modified system was stiffer than the SK in axial compression, torsion, and mediolateral-bending (α <0.0083). In craniocaudal-bending the SK was significantly stiffer than all UMEX constructs (α <0.0083). In addition, the SK was significantly stiffer than the UMEX type Ia and type Ib in all loads (α <0.0083) (Figure 4, Appendix Table 1 – Available online at www.vcot-online.com).

After testing was performed, the tightness of all clamps was confirmed.

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**References**

i. Precision balance Mettler Toledo Classic AB205-S; Mettler GmbH, Greifensee, Switzerland

j. MS Excel 2007; Microsoft, Redmond, WA, USA

k. NCSS 2007: NCSS, Kaysville, Utah, USA (available at: www.ncss.com)
The average weight of the SK system components (52.1g) was greater than any one of the other three constructs.

Differences between the average weight of components of the three UMEX constructs were relatively minimal (type Ia: 18.5g, type Ib: 21.2g, type II modified: 26.7g).

The closest possible placement of three pins in the same plane (type Ia configuration) with the UMEX device required 1.4 cm of fracture segment length. In contrast, the SK device required 2 cm of fracture segment length for the placement of three closely spaced pins (Figure 5).

**Discussion**

We performed in vitro biomechanical testing for different configurations of a new micro external fixation system (UMEX) as we thought it might prove an alternative in the treatment of long-bone fractures in very small animals. A mini SK type Ia system was tested as a reference.

We could not find any biomechanical studies on miniature ESF in the literature. Furthermore, there was no data available regarding loading forces, such as axial compression, mediolateral-bending and craniocaudal-bending, as well as torsion, for very small animals weighing less than 5 kg. Therefore we extrapolated the loads used in this study from historical force plate analyses of dogs with experimentally-induced chronic hindlimb lameness and clinically normal dogs (27, 28). These studies reported a peak vertical force at trot.

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**Figure 4** Dot plots showing construct stiffness in different loadings. All groups are significantly different from each other (α <0.0083), except the pair indicated by a bracket (*). A = SK type Ia; B = UMEX type II modified; C = UMEX type Ia; D = UMEX type Ib; i = axial compression; ii = torsion; iii = mediolateral-bending; iv = craniocaudal-bending.
of 18.9%, 44%, and 61.3% of presurgical peak values at two, six and 12 weeks, respectively and a peak vertical force in the hindlimb of 65.11% body weight (27, 28). Peak craniocaudal force in propulsion of a hindlimb was 7.69% body weight, in a forelimb 6.37% (28). Assuming a value of 9.81 m/s² for standard gravity, a 5 kg dog would subject the hindlimb to loads of approximately 6.04 N, 14.05 N and 19.58 N at two, six and 12 weeks postoperatively. A 2 kg dog’s hindlimb would be subjected to only 7.83 N 12 weeks postoperatively. Peak craniocaudal force of a normal mesomorphic dog of 5 kg would be 3.77 N in a hindlimb and 3.12 N in a forelimb. In this context, 30 N for axial compression and 50 N for craniocaudal bending were deemed sufficient loads for biomechanical testing, in order to cover a range exceeding the physiological compressive force and bending moments in these limbs. For mediolateral bending and torsion there were no force plate data available. Therefore, by scaling from previous studies conducted in larger dogs, values of 50 N for mediolateral bending and 0.2 Nm for torsion were defined.

As for comparison of different UMEX frames, results supported previous reports concerning the superiority of bilateral over unilateral and of biplanar over uniplanar constructs (1, 29-32). Our findings supported previous studies in that UMEX type Ia frames were the least stiff in all loadings (29-32). The UMEX type II modified frame was significantly stiffer than type Ia in all loads with a stiffness increase up to 1652% in axial compression. In craniocaudal bending, the UMEX type Ib equalled type II modified in terms of stiffness, whereas in all other loads type II modified was significantly stiffer than type Ib with an increase of up to 527% in axial compression. A regular type II construct using exclusively full-pins may have been even stiffer, nevertheless we assumed a type II modified would have more clinical relevance (29-31).

Stiffness obviously correlates positively with rod diameter and it has already been shown that a larger connecting rod, while maintaining the same pin diameter, has a significant effect on overall frame stiffness (29, 33). Due to fixed clamp sizes of the two systems, two different sized rods, a 3.2 mm SK rod and 2.5 mm UMEX rod had to be used. As expected, the larger and heavier SK constructs were considerably stiffer than the equivalent uniplanar, unilateral UMEX constructs. It can be assumed that although not tested in this study – the mini SK frame configurations will be stiffer than the UMEX counterparts. Nevertheless the UMEX type II modified was stiffer than a SK type Ia configuration. The only load where the unilateral and uniplanar SK was significantly stiffer than the UMEX type II modified was in craniocaudal bending. Due to insertion of fixation elements in the mediolateral direction, craniocaudal stiffness is typically the weakest in bilateral configurations (31). The knurled design of the UMEX rods may hypothetically increase stability of the constructs, as the pins are in direct contact with the rod. Further research is required to evaluate possible advantages.

According to Wolff’s law, bone modelling is dependent on the applied load. Although the optimal level of fracture stability has not been established, it is understood that inadequate stabilization can lead to excessive bone motion at the fracture site and, as a consequence, the development of delayed unions and nonunions (1, 3, 10). At the same time, excessively rigid fixation may shield the fracture site and predispose the bone to stress protection (24, 34). Stress protection can occur in both plate and external fixation, although a higher risk of bone resorption has been described in plate fixation (1, 7). Staged disassembly of ESF constructs has been shown to be beneficial in decreasing healing times in fracture repair (3). Very small animals appear to be at higher risk of stress protection (6, 7, 20, 35). The mechanical properties of the ESF are of great importance in achieving the optimal fracture repair (8, 15). Heavy or bulky constructs may impede weight bearing and mobility in very small animals. Decreased use of the limb may result in impaired vascularity, which may delay bone union and diminish the positive effects of ESF (8). Too rigid immobilization may even lead to severe bone atrophy (35). All of the different UMEX configurations were considerably lighter, weighing between 35% and 51% of the mini SK type Ia configuration. In this context, the use of a light UMEX construct could help to improve a small patient’s comfort.

Apart from the advantage in weight, pin placement was much closer in the UMEX compared to the SK system. This was due to clamp size and manner of application. A fragment length of 1.4 cm was needed to place three pins in a type Ia configuration in the UMEX systems, whereas a minimum fragment length of 2.0 cm was needed in the SK systems. This may be of great benefit in the fixation of very short juxta-articular fragments (15). Another approach to closer pin placement would be the use of biplanar, circular or hybrid configurations (16, 36).

Pins could be placed even closer using acrylic frames, an established method with successful results in the treatment of very
small animals (37–40). Other advantages of free-form external fixation are the increased versatility in the location and angle of pin placement, and its low cost. One disadvantage of acrylic constructs is the large rod diameter, which would need to be increased by about three to four times to reach similar stiffness as stainless steel bars (23, 41). Another limitation of acrylic frames is the difficulty of staged disassembly (37, 40). Regarding weight, a minimum of 19 g of resin acrylic for each clamp on ESF for animals weighing greater than 20 kg has been recommended (37, 42). To evaluate the advantages for animals weighing less than 5 kg, further research is required.

A possible disadvantage of the UMEF system compared to the SK system is that the pins may not be angled proximally or distally. Furthermore, there are no UMEF mini circular elements available at the moment, thus preventing an assembly of hybrid constructs.

A limitation of this study was that we were not able to test two comparably sized systems. This was due to the fact that the mini SK only allows the use of a 3.2 mm rod, which was too large for the UMEF alpha clamps (4). A better comparison of the UMEF and SK system would have been achieved if the same rod diameters had been used. In the same context, the torque used was a compromise between the different sized constructs. We decided on one Nm, whereas in the original description of the SK mini clamps, a mean torque of 3.85 Nm led to the feeling of ‘adequate clinical tightness’ in SK mini clamps (4). The use of less torque in our study did not lead to clamp loosening.

Threaded pins are reported to have significantly greater stability in axial and rotational loading (43, 44). However, conventional negative-profile pins, which have a reduction of the core diameter in the threaded region, are more susceptible to bending and breaking compared to smooth pins (43). Both constructs may have been improved in clinical application by using positive-profile threaded pins instead of the negative-profile threaded pins used in this study (44). Nevertheless this was not an issue in the in vitro situation, as the same pin thread was used in all constructs.

As there was no alternative available by the producer, UMEF type II modified full-pins were unthreaded, which may have increased the performance of this construct.

The principal goal of the present study was to compare the elastic stiffness of the UMEF systems to established ESF systems. Loading levels corresponded to what can be reasonably expected in a controlled postoperative situation, and therefore provide some insight into the promotion of fracture healing by sufficient immobilization. However, construct strength was not determined, therefore we cannot exclude that exceptionally high loading levels could compromise fixation in the UMEF or conventional ESF systems.

This in vitro study does not reflect in vivo conditions. Further in vivo studies are required to test the application of the UMEF ESF in a clinical setting.

In conclusion, the micro external fixator system due to its weight, size, rigidity and close placement of the pins, may provide an acceptable alternative for the treatment of long-bone fractures in very small animals.

Conflict of interest

None declared.

References