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A Comparative *In Vitro* Biomechanical Investigation of a Novel Bilateral Linear Fixator *vs*. Circular and Multiplanar Epoxy-Pin External Fixation Systems Using a Fracture Model in Buffalo Metacarpal Bone

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Abstract

Objectives: Comparative evaluation of in vitro biomechanical properties of a newly developed Bilateral Linear External Fixation Device (BLF), Circular External Fixator (CEF) and Multiplanar Epoxy-Pin Fixation System (EPF) in a fracture model.

Methods: A novel design of the BLF comprising of a pair of side bars (14 mm diameter) with threads carved in opposite directions (pitch 2 mm) on both sides from the centre, and multiple fixation bolts, which could be secured at desired levels, was developed using stainless steel alloy. Using buffalo metacarpal bones, different fixator constructs were prepared using a fracture model with 2 mm gap. The intact metacarpal bone, BLF-bone construct, CEF-bone construct and EPF-bone construct were subjected to axial compression (n=4 each) and craniocaudal 3-point bending (except CEF-bone construct, n=4 each) using a Universal Testing Machine. Load-deformation graphs were plotted and different mechanical parameters were recorded/calculated and compared between the constructs.

Results: The BLF was very versatile; it allowed the placement of fixation pins at desired levels and allowed bone reduction or compression by rotating the side bars in opposite directions even after bone fixation. EPF was the simplest. Under compression, BLF construct showed significantly higher mechanical strength than CEF and EPF constructs. When loaded under 3-point bending, no significant difference in the mechanical properties was seen between BLF and EPF constructs.

Clinical Significance: The newly designed BLF is adequately strong, and can be used to treat long bone fractures in animals weighing up to 350 kg. EPF, though the weakest among the fixator constructs tested, was found to provide sufficiently rigid bone fixation and may be effective in animals weighing up to 100 kg.

Keywords: External skeletal fixation; Biomechanical study; Circular external fixation; Bilateral linear fixator; Epoxy-pin fixation system

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Introduction

In large animals, management of fractures, especially open and infected, is difficult and pose a challenge [1]. In bovines, fractures of the metacarpus or metatarsus are commonly recorded and such fractures are frequently open [2-5]. Routinely used external and internal fixation techniques do not provide optimal environment for open fracture repair. External Skeletal Fixation (ESF), involving placement of a series of percutaneous pins that are connected externally to a rigid frame [6], has been described for stabilization of skeletal injury in man and animals. Among ESF systems, linear external fixation, and transfixation pinning and casting are most commonly practiced in small and large animals, respectively [7,8]. Though several ESF systems are being used in small animal fracture fixation, very few were developed for large animals and used in routine clinical practice [9-15].

Free-form external fixation is a type of ESF wherein the pins placed in the fracture fragments are connected to moldable polymers rather than clamps and steel connecting bars [16,17]. Free-

form ESF, using acrylic as a substitute for the connecting clamps and rods, has been used successfully for treatment of long bone fractures in small animals [17-20]. The epoxy putty has been shown to have similar strength, greater apparent modulus, and reduced toughness when compared with the methacrylates [21]. Epoxy-Pin Fixation (EPF), which was earlier used in birds and small animals [22,23], also proved useful in treating certain open long bone fractures in calves and foals [24].

The mechanical property of an ESF system depends on several factors, such as the size and number of fixation pins, the distance between the bone and external frame, and also the design and material used to construct the fixator frame [25-27]. Though there are several biomechanical studies involving ESF systems, there are only a few studies involving external fixators developed and used in large animals [9,28-30]. Keeping these in view, the present study was undertaken with the objective to evaluate and compare the *in vitro* biomechanical properties of a novel BLF (made of stainless steel side bars), CEF (made of aluminum rings) and multiplanar EPF (made of Epoxy + SS pins) systems developed for use in large animals (light to medium weight) using an *in vitro* fracture model.

Materials and Methods

Specimen collection and preparation of ESF constructs

Sixteen pairs of adult buffalo metacarpal bones from animals weighing about 250 Kg to 350 kg were collected from a slaughter house. Soft tissues were removed, and bones were wrapped in saline soaked towels and stored at - 20°C. The bones were thawed to room temperature (24°C to 27°C) before used for testing. Three different ESF constructs, namely Bilateral Linear Fixator (BLF), Circular External Fixator (CEF), and multiplanar Epoxy-Pin Fixator (EPF), were developed using a cadaveric bone model.

The BLF developed using stainless steel alloy [31], comprising of a pair of side bars with threads carved in opposite directions (2 mm pitch) on both sides from the centre, and multiple fixation bolts, were used. The side bars measured 14 mm in diameter and 22 cm to 45 cm in length (three different sizes). The centre of the side bar had a hexagonal flat surface to facilitate easy turning with a wrench. The rectangular fixation bolts had an eccentrically placed threaded-hole to get secured along the side bars, and on the other side an 8 mm smooth-hole drilled perpendicular to the length of side bar to accept a 6 mm to 7 mm pin. The threads in the fixation bolts were also carved in opposite directions to secure along the corresponding side of the side bar. BLF construct was prepared in cadaver metacarpus by passing 2 fully threaded positive profile (6 mm) pins mediolaterally in proximal and distal portions of the diaphysis after predrilling each site with a 5.5 mm drill bit. The distance between the pins was kept at 40 mm. The pins were then fixed to the 22 cm long side bars using fixation bolts with the help of nuts at a fixed distance of 50 mm from the bone on either side (Figure 1).

The CEF developed using aluminum alloy [13] was used in the present study. The rings were of 14.5 cm inner diameter, 20 mm width and 5 mm thickness. Each full ring had 22 equidistant 10 mm holes for fixation of connecting rods and bolts. Connecting rods were 12.5 cm long and 8 mm in diameter, and made of stainless steel. The 8 mm bolts were allotted to accept 3 mm pins. The fixation pins (316 L SS, 3 mm dia.) were passed in craniomedial to caudolateral and caudomedial to craniolateral directions (crossed at 60 angle to 700 angle), two pairs each in proximal and distal bone segment, and fixed to the rings using slotted bolts (Figure 2). The pins were not tensioned before fixing to the rings. The distance between the successive pins was kept constant at 40 mm, and the distance between the bone and the rings was kept at 50 mm (same as in BLF).

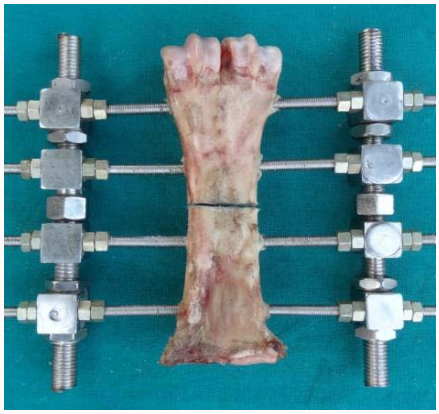

Figure 1: BLF-bone construct used for mechanical testing.

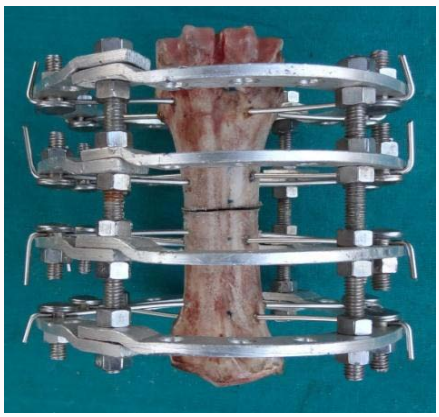

Figure 2: CEF-bone construct used for mechanical testing.

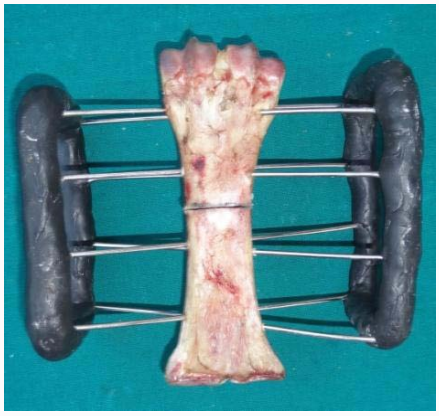

Figure 3: EPF-bone construct used for mechanical testing.

In EPF constructs, 3 mm pins were passed in craniomedial to caudolateral and caudomedial to craniolateral directions (crossed at 60 to 700) as in CEF constructs (Figure 3). Four pins (two pairs) were passed in proximal and distal segments each. The transcortical pins in the same plane were bent and joined each other (using adhesive tapes) to make a temporary scaffold. The two side bars on both sides were connected at the proximal and distal ends using additional

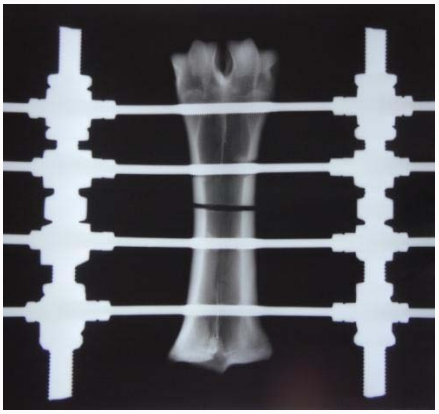

Figure 4: Dorsopalmar radiographic view of BLF-bone construct.

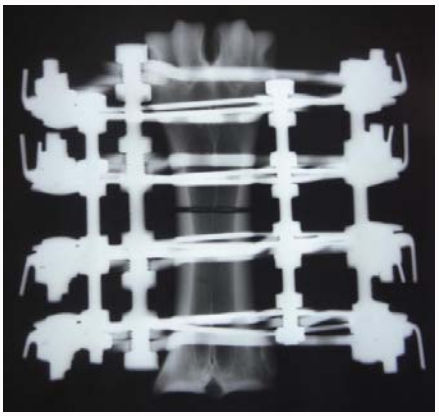

Figure 5: Dorsopalmar radiographic view of CEF-bone construct.

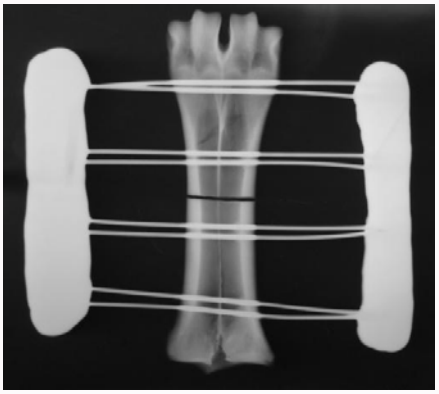

Figure 6: Dorsopalmar radiographic view of EPF-bone construct.

pieces of pins [32]. After thoroughly mixing the epoxy resin with hardener (750 g), the epoxy mold in dough stage was applied along the scaffold by incorporating the bent pins within to construct the side bars (dia. 20 mm to 22 mm). The distance between the bone and connecting side bars of EPF construct was kept same as that in BLF/CEF constructs. The epoxy was then allowed to dry.

In all the fixator-bone constructs, a 2 mm gap was created at the centre of the bone using an electric saw to simulate the fracture condition. Radiographs of all the test constructs were made before testing (Figures 4-6). The proximal and distal ends of the bones were embedded in methyl methacrylate1 pedestals (5 cm × 4.5 cm × 2 cm) to facilitate better anchoring of the construct during mechanical testing. The pedestals incorporated a 10 mm length of the bone on either end.

Table 1: Grouping of different fixator constructs for mechanical testing.

Group	Type of construct	No. of specimens	Type of test
A	BLF (stainless steel)	4	Axial compression
		4	Craniocaudal 3-point bending
B*	CEF (aluminum)	4	Axial compression
C	EPF (epoxy-SS pin)	4	Axial compression
		4	Craniocaudal 3-point bending
D	Intact bone	4	Axial compression
		4	Craniocaudal 3-point bending

*In group B, 3-point bending test could not be done due to lack of space between the rings for insertion of prongs

Mechanical tests

The fixator constructs were grouped as detailed in the Table 1.

Compression: The specimens were loaded under compression using a Universal Testing Machine [2]. Axial compression was directed along the longitudinal axis of the construct at a constant aquator displacement rate of 1 mm/sec until failure. The load-deformation curves were obtained for each individual specimen. The yield load and failure load were calculated directly from the load-deformation curve for each specimen. The structural stiffness represented by the Young's modulus of elasticity was calculated as the slope of linear portion (elastic region) of load-deformation curve. Mean ± SE values for different parameters were estimated for different constructs and were compared.

3-Point bending: The 3-point craniocaudal bending test was done in groups A, C and D. The central load support contacted the cranial aspect of the middle of the specimen and load was applied at 1 mm/sec per second until failure [2]. The load-deformation curves were obtained for each specimen. The yield load and ultimate failure load were calculated directly from the load-deformation curve for each specimen. Young's modulus of elasticity (stiffness) for each specimen was calculated as the slope of linear portion (elastic region) of load-deformation curve similar as in compression testing. Bending moment was calculated using the following equation [33].

$$M = \frac{PXL}{4}$$

Where M = Bending Moment, P = Peak load attained before failure and L = Span Length

Mean ± SE values for different parameters were estimated for different constructs and compared.

Post-testing construct photographs and radiographs: After material testing, gross photographs and dorsopalmar and mediolateral radiographs of the constructs were made to observe the status of the fixator components and the bone at an exposure of 55 kVp and 20 mA with a 130 cm FFD.

Statistical analysis

The data were analyzed using commercially available software programs, Microsoft Excel (Part of Microsoft Office Professional Edition, Microsoft-2007) to calculate Young's modulus, bending moment, yield load and ultimate failure load from load-deformation curves and SPSS 15.0 (SPSS Inc., Chicago, IL) for Analysis of

Table 2: Mean ± SD of mechanical testing variables for different specimens in axial compression.

Groups	Stiffness (kN/mm)	Yield load (kN)	Failure load (kN)
A (BLF)	8.75 ± 0.10[c]	41.33 ± 0.53[c]	53.78 ± 2.67[b]
B (CEF)	8.10 ± 0.19[b]	38.08 ± 0.48[b]	49.03 ± 0.37[b]
C (EPF)	4.70 ± 0.32[a]	27.15 ± 0.52[a]	32.58 ± 1.49[a]
D (Intact bone)	11.05 ± 0.09[d]	46.03 ± 0.35[d]	61.75 ± 0.78[c]

Values with different superscript alphabets differ significantly (P<0.05)

Table 3: Mean ± SD of mechanical testing variables for different specimens in 3-point bending.

	Stiffness (kN/mm)	Yield load (kN)	Failure load (kN)	Bending moment (Nm)
A (BLF)	0.13 ± 0.01[a]	1.56 ± 0.05[a]	2.65 ± 0.07[a]	132.25 ± 3.42[a]
C (EPF)	0.07 ± 0.00[a]	1.15 ± 0.06[a]	2.55 ± 0.17[a]	127.50 ± 8.54[a]
D (Intact bone)	1.68 ± 0.07[b]	5.50 ± 0.48[b]	8.95 ± 0.22[b]	402.75 ± 9.98[b]

Values with different superscript alphabets differ significantly (P<0.05)

Variance (ANOVA), and Duncan's Multiple Range Test (DMRT) to compare the values between different groups. Paired '*t*' test was used to compare the mean values at different intervals with their base values in each group. The differences were considered significant at P<0.05 [34].

Results

Compression

The mean ± SE values of compressive stiffness, yield load and ultimate failure load recorded in different groups are presented in Table 2.

Stiffness: The axial stiffness of group D was significantly greater (11.05 kN/mm) than all other groups. The axial stiffness of groups A (8.75 kN/mm), B (8.10 kN/mm) and C (4.70 kN/mm) were significantly (P<0.05) different from each other. The axial stiffness was highest in group A and least in group C.

Yield load: The mean yield load of group D was 46.03 kN. Among the different constructs, yield load of group A (41.33 kN) was highest, followed by group B (38.08 kN) and group C (27.15 kN). The yield load of groups A, B and C were significantly (P<0.05) different from each other.

Ultimate failure load: The ultimate failure load of group D was 61.75 kN, which was significantly (P<0.05) greater than construct groups, A (53.78 kN), B (49.03 kN) and C (32.58 kN). No significant difference was seen in the values of ultimate failure load between groups A and B, and the values were lowest for group C.

3-Point bending

The craniocaudal 3-point bending test was performed in groups A, C and D, bending test could not be done in group B. The mean ± SE values for bending moment, stiffness, yield load and ultimate failure load calculated from load-deformation curves are shown in Table 3.

Bending moment: The bending moment of group D (402.75 Nm) was significantly (P<0.05) greater than that of group A (132.25 Nm) and group C (127.5 Nm). The bending moment of group A was non-significantly higher (P>0.05) than that of group C.

Bending stiffness: Mean value of bending stiffness was also significantly higher (P<0.05) for group D (1.68 kN/mm) as compared to group A (0.13 kN/mm) or group C (0.07 kN/mm). Comparison between groups A and C showed insignificant difference (P>0.05).

Yield load: Yield load was also highest in group D (5.50 kN) and lowest in group C (1.15 kN). No significant difference was seen between groups A and C.

Ultimate failure load: The ultimate failure load of group D was significantly (P<0.05) greater (8.95 kN) than all other groups. Ultimate failure load did not differ significantly between groups A (2.65 kN) and C (2.55 kN).

Photographic and radiographic evaluation of post-testing constructs

Compression: The gross photographs and radiographs of ESF-bone constructs taken after compression testing showed cracks and splintering of bone fragments in all groups. Splintering was more at the site of osteotomy, especially at the proximal end of the distal fragment (Figures 7-9). In groups B and C, slight bending of the fixation pins was noticed. In group A, cracks occurred at the site of pin tracts, with splintering and loss of bone pieces at the proximal end of the distal fragment. In group D, splintering occurred at the distal end of the bone. In all the construct groups, the side bars/rings were intact and no deformation noticed.

Bending: In group A, when 3-point bending was applied at the middle of the bone at the site of osteotomy, deviation of bone ends occurred along with bending of fully threaded pins. The pins close to the osteotomy site showed more deformation (Figure 10a, 10b). In group C, epoxy columns got cracks/fractured at 2 sites to 3 sites along with bending/bowing of fixation pins (Figure 11a, 11b). The bone ends deviated away from the direction of applied force. In group D, when bending force was applied at the middle of the bone, the bone broke into two pieces with oblique fracture.

Discussion

External skeletal fixation devices have been widely used for treatment of long bone fractures and limb lengthening procedures in human beings and small animals [26,27,35,36]. ESF allows complete weight-bearing immediately after bone fixation, maintains normal joint mobility, and provides an optimal environment for osteosynthesis and wound-healing without the need for an implant at the fracture site [37]. ESFs are not commonly used in large animal fracture repair due to lack of suitable fixation devices that can sustain a heavy load and the forces exerted on angularly oriented bones. Use of ESF in ruminants has also been impeded by economic constraints and difficulties with postoperative care and management. For clinical acceptance, an ESF must be sufficiently rigid, well tolerated, easily applied, and inexpensive.

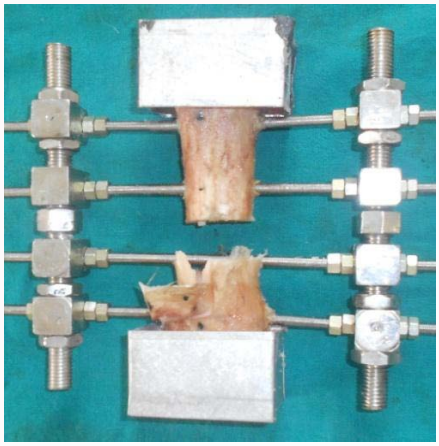

Figure 7a: Photograph of BLF-bone construct made after compression test.

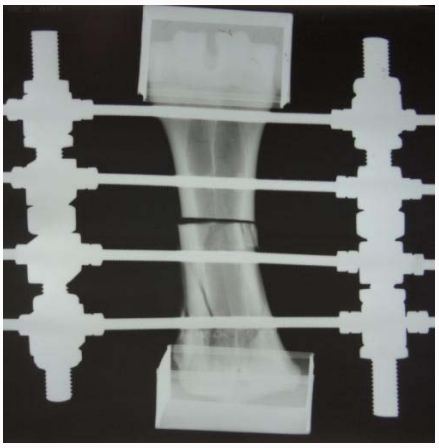

Figure 7b: Dorsopalmar radiographic view of BLF-bone construct made after compression test.

Figure 8a: Photograph of CEF-bone construct made after compression test.

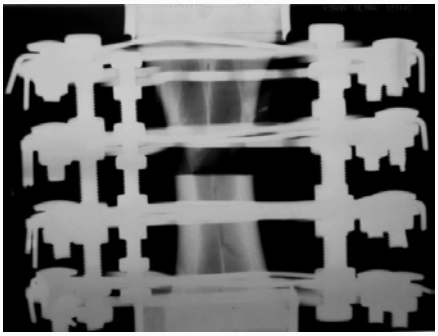

Figure 8b: Dorsopalmar radiographic view of CEF-bone construct made after compression test.

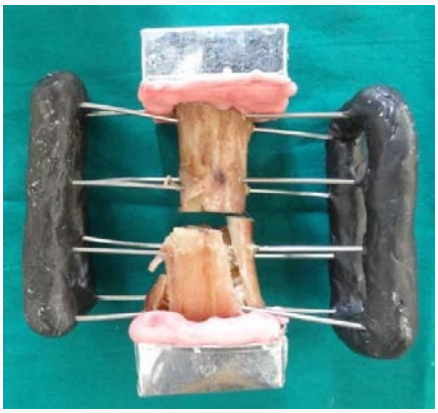

Figure 9a: Photograph of EPF-bone construct made after compression test.

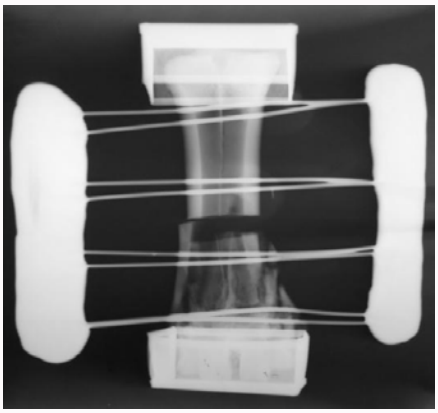

Figure 9b: Dorsopalmar radiographic view of EPF-bone construct made after compression test.

Fixation devices used for fracture stabilization must provide an optimal mechanical environment at the fracture site that mainly depends on the stability and rigidity of the fixation device. Ideally, fixators should provide adequate stability to maintain bone fragment alignment while allowing micromotion at the fracture site to favor bone healing [38-40]. Though several biomechanical studies evaluating ESF systems designed for use in small animal and human patients have been reported [25,38,41], there are only a few biomechanical studies of large animal fixators. Cervantes and associates [29] studied the biomechanical characteristics of a 4-ring CEF applied to equine third metacarpal bones and found that a 4-ring fixator using 2.25 in. pins/ring was not stiff enough for repair of an unstable third metacarpal bone fracture in a 450 kg horse. Rapoff et al. [28] designed an ESF for repair of long bone fractures in horses and cattle by use of wire ropes (6.4 mm) as the transosseous component and found that wire rope had a greater stiffness than a single pin in axial compression. A study comparing a bilateral linear, circular and hybrid fixators developed for large ruminants showed that the hybrid fixator had the greatest strength [9]. In this study, a novel bilateral

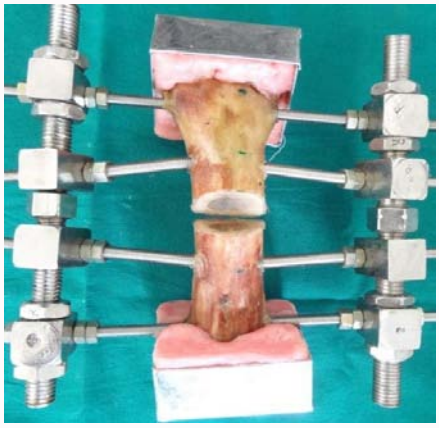

Figure 10a: Photograph of BLF-bone construct made after bending test.

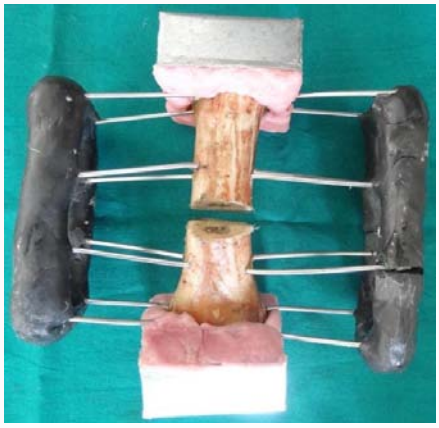

Figure 11a: Photograph of EPF-bone construct made after bending test.

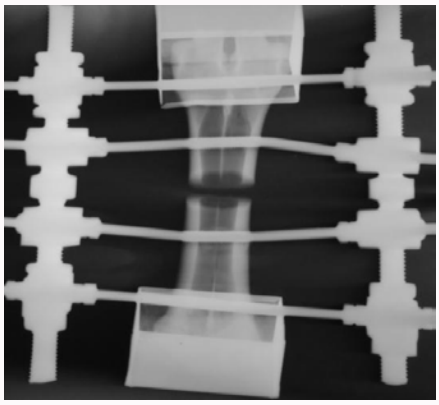

Figure 10b: Dorsopalmar radiographic view of BLF-bone construct made after bending test.

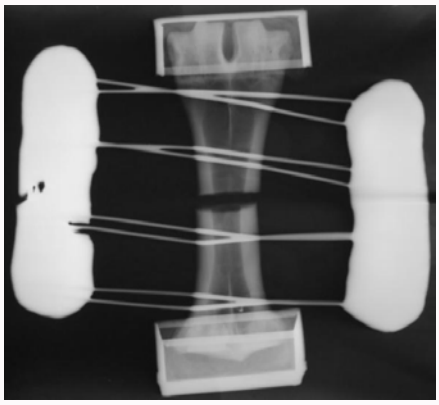

Figure 11b: Dorsopalmar radiographic view of EPF-bone construct made after bending test.

linear fixator, 4-ring circular fixator and multiplanar epoxy-pin fixator were subjected to axial compression and craniocaudal bending to determine if these ESF systems could provide stable fixation in large animal applications.

Under normal physiological conditions, long bones are subjected to different stresses, and axial compression is one of the important disruptive forces and hence an important loading mode for assessing fixator suitability [38]. Bending is equally important in animals, where long bones are placed angularly and are under constant bending stress. Evidence from human and animal studies suggests that disruptive forces in the craniocaudal plane during physiological loading of a fracture site are considerably greater than those in the lateromedial plane [42]. Hence, in the present study, different ESF designs were evaluated under compressive and craniocaudal bending loads.

In earlier studies, different models have been used for biomechanical properties of ESF [43-45]. In the present study, a mid-diaphyseal osteotomy with a 2 mm gap was used to simulate a clinical fracture. The larger gap model was not considered as the metacarpal bone is short. Similar models have also been used earlier by several researchers [44-46]. In most of the earlier studies, the distance between the fixation pins was kept at nearly 90°. In the present study, however, the distance kept between the pins was 60° to 70°. This was done to simulate the clinical situation, wherein it may not be possible to pass the cross pins at 90° angle in the metacarpus as it is flattened dorsopalmarly, and the extensors and flexors occupy the dorsal and palmar surfaces, along with vessels and nerves.

Several factors influence the biomechanical property of an ESF system, namely the size and number of fixation pins, the distance between the bone and external frame, and also the design and material used to construct the fixator frame, etc. In the present study, while preparing different fixator constructs, as far as possible the variables were kept to minimum, except for the materials used to construct the external fixator components. Generally, in CEF, the fixation pins are tensioned from one end to provide more strength to fixation. However, in the present study, the pins were not tensioned in all the construct groups to minimize the variation. Tensioning of small diameter pins has been shown to provide greater stability to fixation, as compared to un-tensioned pins [25]. It is also true that the effect of tensioning is more in small diameter pins, and as the size of pin increases the effect of tensioning reduces. In the present study, as relatively large diameter pins (3 mm) were used, the effect of tensioning would be less.

Compression test

Axial stiffness is the ability of the fixator to resist axial motion at the fracture site, i.e. motion along the central axis. BLF had greatest axial stiffness followed by CEF and EPF. The stiffness of fixator design depends on the strength of side bars/rings, and the orientation and size of fixation pins used [47]. The greater axial stiffness with BLF was probably attributed to the fixator design [9]. The difference in axial

stiffness between BLF and CEF was less, probably due to the presence of cross transfixation pins of 3 mm diameter in CEF, almost equating to 6 mm pins of BLF at corresponding levels. The compressive stiffness of BLF and CEF was about 80% of intact bone, indicating that both BLF and CEF could provide rigid stability under compression load in a transverses fracture model. The axial stiffness of EPF was the least, which was about 50% of BLF/CEF. As the number and orientation of pins used in CEF and EPF were the same, decreased stiffness with EPF was probably attributed to the weak epoxy-pin side bars. Unlike in CEF, where fixation pins were secured to the rings using fixation bolts, in EPF the fixation pins were just bent and joined using adhesive tape and epoxy dough. Nevertheless, the compressive stiffness of EPF was 4.70 ± 0.32 kN/mm, indicating that EPF can provide reasonably good stability under compression loading. The inherent elasticity associated with the use of small diameter fixation pins in CEF and EPF might allow axial motion of bone fragments at the osteotomy site, which could favor bone healing [43,48].

Yield load, the minimum load necessary for the material to start deforming, was also highest for BLF, followed by CEF. This indicates that BLF can withstand maximum load before deformation, which may probably be attributed to the large size fixation pins and strong side bars used in BLF construct. Further, the yield load for BLF and CEF was about 80% to 90% of that for intact bone, indicating that both these fixators are very rigid and can withstand maximum compressive load. Yield load was the least for EPF, suggesting that it can undergo deformation relatively early with less compressive load. Similar results were recorded with respect to failure load.

Gross photographs and radiographs of the fixator constructs subjected to compression loading showed cracks and splintering of bone fragments at the site of osteotomy, especially at the proximal end of the distal fragment in all groups. In BLF construct, cracks also occurred at the site of pin-bone interfaces/along the pin tracts. Though there was bending/bowing of fixation pins, the side bars/rings showed no deformation. These findings support the biomechanical parameters studied, and suggest that all the fixator constructs are relatively strong against compression loading, and the failure of the construct occurred mostly due to development of cracks/fractures in the bone. This also indicate that in a ESF construct with a transverse fracture model having minimum gap between the bone fragments, the compressive load applied along the long axis is mostly transmitted through and absorbed by the bone itself with sparing effect on the external fixator components.

Bending test

The whole bone bending tests are generally conducted as 3-point or 4-point bending experiments. In the present study, 3-point bending was performed, since the length of bone was short and the position of ESF components made it difficult to have two prongs of the bending aquator of UTM to contact the bone in 4-point bending. The craniocaudal 3-point bending test could not be performed in CEF constructs as the prongs could not be placed over the bone due to obstruction from the rings.

Results of craniocaudal bending indicated that the bending stiffness of BLF construct was slightly higher than EPF construct, suggesting that BLF could better withstand the bending load as well than the EPF. However, there was no significant difference ($P>0.05$) between the BLF and EPF constructs, which could be attributed to the fixator design, and the use of more number of pins and multiplanar orientation of pins in EPF constructs. Yield load, failure load and bending moment also showed similar trend.

Post-test gross photographs and radiographs of fixator constructs subjected to 3-point bending revealed bending of bone fragments opposite to the loading direction in all groups. There was bending of fixation pins in all groups, which was more than that seen with compression loading. In EPF, the epoxy-pin side bars developed cracks, and bending of bone as well as K-wires occurred. These findings indicate that the fixator constructs are more vulnerable to bending load; and unlike during compression, during bending fixator components develop deformation. When compared between BLF and EPF constructs, the side bars of BLF are stronger and resistant to bending, whereas epoxy columns are relatively more brittle, and hence likely to develop cracks when subjected to bending.

When compared between compressions and bending loads, both BLF and EPF constructs could resist compression stress better than bending stress. Similar results were reported in earlier studies also [9,32,49]. This indicates that these fixators could be more useful for immobilization of fractures in long bones subjected to mostly compression loading; such as metacarpal and metatarsal bones and radius-ulna, which are straighter and parallel to the long axis of the limb, than those subjected to more of bending loads, such as tibia.

In the present study, *in vitro* biomechanical properties of different constructs were studied under static compression and craniocaudal bending loads; however, clinically these fixators will also experience other stresses such as torsion and fatigue. Similarly, the *in vitro* test results are obtained under ideal conditions like mid-diaphyseal transverse fracture situation. Further, clinically, based on the fracture location, more number of fixation pins and longer side bars/rings may be needed, especially for transarticular fixation. The biomechanical properties may also depend on the extent of bone loss at the fracture site. Hence, true testing of the fixator systems is possible only by clinical application. Nevertheless, the results of this study indicate that BLF is strong enough to sustain adequate compression and bending loads for use in fracture fixation in straight long bones of medium sized large animals. By increasing the size and number of pins the fixation rigidity can be further increased. Results also indicate that epoxy-pin fixation, which is simple, economical and less cumbersome could also provide sufficient strength to fixation in animals weighing at least up to 100 kg.

Conclusions

The novel bilateral linear fixation system developed for large animal use, which is least cumbersome to apply with minimal parts, proved strongest under both compression and bending loads. Hence, it may be useful to treat long bone fractures of straight bones (metacarpus, metatarsus and radius/ulna), especially open fractures in large animal practice. Epoxy-pin fixation system is biomechanically the weakest; nevertheless, it is easy to apply with minimal facilities and economical too, hence may be suitable for fracture fixation in light weight animals weighing up to about 100 kg, especially under field conditions.

[1]Pyrax®- denture based polymer resin, Pyrax polymers- Roorke, India.

[2]100 KN Servo-hydraulic Computerized Universal Testing Machine, ADMET, USA.

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