Shiraz University

IJVR

ISSN: 1728-1997 (Print)
ISSN: 2252-0589 (Online)

Vol. 26 No. 3 Ser. No. 92 2025

IRANIAN JOURNAL OF VETERINARY RESEARCH

Original Article

A biomechanical study of a novel multi axial circular external fixation device for stabilization of a bovine cadaveric tibial fracture gap model

Kamalesh Kumar, K. S.[1]; Rohit Kumar,[2*]; Sai Kumar, T.[1]; Singh, B. P.[1]; Saxena, A. C.[2]; Amarpal,[2]; Pawde, A. M.[2] and Thakur, D.[3]**

[1]Ph.D. Student in Veterinary Surgery and Radiology, Division of Surgery, Indian Veterinary Research Institute, Izzathnagar, Bareilly, Uttar Pradesh, India; [2]Division of Veterinary Surgery and Radiology, Indian Veterinary Research Institute, Izzathnagar, Bareilly, Uttar Pradesh, India; [3]AMIE Mechanical Engineering, Design Lab, Indian Institute of Technology, School of Mechanical and Material Engineering, Mandi, Himachal Pradesh, India

[*]***Correspondence:*** *Rohit Kumar, Division of Veterinary Surgery and Radiology, Indian Veterinary Research Institute, Izathnagar, Bareilly, Uttar Pradesh, India. E-mail: drrohits.singh@gmail.com*

[**]***Co-correspondence:*** *K. S. Kamalesh Kumar, Ph.D. Student in Veterinary Surgery and Radiology, Division of Surgery, Indian Veterinary Research Institute, Izatnagar, Bareilly, Uttar Pradesh, India. E-mail: kskamalesh93@gmail.com*

10.22099/ijvr.2025.51532.7656

(Received 19 Nov 2024; revised version 25 Jun 2025; accepted 13 Jul 2025)

This is an open access article under the CC BY-NC-ND license (http://creativecommons.org/licenses/by-nc-nd/4.0/)

Abstract

Background: Circular external fixators are widely used in orthopaedic fracture management but often face limitations in complex fractures due to their rigid structure. The multi axial circular external fixator offers increased flexibility and adjustability, making it a promising alternative for cases involving anatomical variability. **Aims:** The present study aimed to compare the biomechanical properties, including stiffness and strength, of a novel multi-axial circular fixator (MCEF) with a traditional full-ring circular external fixator (CEF) using buffalo tibia fracture model. **Methods:** The study used eighteen buffalo tibiae divided into three groups: intact bones, CEF, and MCEF constructs. Each construct was subjected to axial compression and torsion tests. The constructs were subjected to loads until failure, and load-displacement curves were generated for each sample. Axial compression and torsion tests were performed to evaluate the biomechanical performance of both fixators. Mechanical parameters such as stress, stiffness, yield load, resilience energy, failure load, and maximum torque were measured and analysed. Appropriate statistical tests were performed to compare between groups. **Results:** CEF demonstrated higher stiffness in axial compression and bending as compared with MCEF, which showed greater flexibility under both axial and torsional loads. MCEF, however, provided enhanced multi-planar adaptability, making it better suited for fractures in anatomically variable regions. In torsion, CEF exhibited higher maximum torque and torsional strength. **Conclusion:** CEF provided greater axial and bending stiffness. The MCEF design allows flexibility and adaptability providing better optimization of fracture stabilization in complex fracture patterns, making it a viable alternative in large animal fracture management.

Key words: Biomechanical testing, Buffalo tibia fracture, External fixation systems, Ilizarov fixator, Multi-axial circular fixator

Introduction

External fixation devices play a pivotal role in managing complex fractures, particularly when internal fixation is not feasible or when additional stability is required to enhance bone healing. External fixation offers several advantages over intramedullary nailing, internal fixation, and open reduction. These include ease of application, greater construct flexibility, and improved access for wound care and monitoring after fixation, making it particularly useful in managing complex fractures and soft tissue injuries (Moss and Tejwani, 2007). Among these devices, circular external fixators, such as the Ilizarov device, have been widely used for the treatment of long bone fractures, deformity corrections, and limb lengthening (Lewis *et al*., 1998).

The Ilizarov external fixator or full ring circular external fixator (CEF) is a circular frame apparatus consisting of wires, developed by Gavriil Ilizarov in Kurgan, Siberia, during 1952 (Gasser *et al*., 1990). The technique is based on the biomechanical principle that axial compressive forces and slight movements promote biological bone healing across the fracture gap. This process is influenced by the tension applied to the wires (Mitousoudis *et al*., 2010). This design creates an optimal mechanical and biological environment for bone healing. These systems stabilize fractures by distributing forces through tensioned wires and rings, which promote healing via controlled micromotion (Paley, 1991). However, despite their effectiveness, traditional circular fixators come with certain limitations, including their fixed-axis design, which restricts the ability to adjust forces at the fracture site, thus complicating management in complex or multi-planar fractures. Furthermore, the unnecessary weight of the bigger sized rings can be reduced usingsmaller diameter rings in less muscular areas, improving patient comfort without compromising stability.

In response to these challenges, new designs are being developed to offer greater flexibility and adaptability in fracture management. One such innovation is the multi-axial circular fixator (MCEF), which enhances adjustability across multiple planes. This increased adaptability may allow for better optimization of fracture stabilization, particularly in more complex clinical conditions, and may result in improved outcomes for challenging fracture patterns. However, despite its potential, the biomechanical performance of MCEF remains underexplored, especially when compared with traditional circular fixators.

Biomechanical studies conducted in-vitro using animal models are essential for evaluating the stability and mechanical properties of fixation devices. Buffalo tibia bones, due to their structural similarity to human bones, provide a suitable model for simulating human fracture conditions (Singh *et al*., 2007). By comparing the performance of traditional circular fixators and the novel MCEF in a buffalo tibia fracture model, this study aimed to uncover valuable insights into the strengths and limitations of these systems, helping to guide future clinical applications.

Fracture management in large animals, especially those with open or infected fractures, presents unique challenges (Steiner and Anderson, 2009). Tibial fractures, common in cattle with incidence rates between 15% and 40%, are often compound and comminuted due to the animal's substantial weight (Adams, 1985; Aithal *et al*., 2004). Various external and internal fixation methods have been used, but compound fractures require stable fixation while allowing for wound management (Aithal *et al*., 2019). Recent studies have emphasized the emergence of modular, multi-axial external fixators that improve adjustability and clinical outcomes in both human and large animal orthopedics (Widanage *et al*., 2023; Bowers and Anderson, 2024). These designs aim to overcome limitations of traditional circular frames by accommodating complex anatomy and variable loading conditions.

The versatility of external skeletal fixators (ESF) has spurred renewed interest, especially for complex fractures resulting from traumatic injuries. ESFs provide advantages such as early limb function restoration, improved local blood flow, and effective management of comminuted fractures (Lewis *et al*., 1998). Circular and hybrid systems, in particular, have demonstrated efficacy in providing stable fixation for unstable fractures (Paley, 1991; Aithal *et al*., 2007). However, their application in large animals, especially ruminants with angular limbs, remains difficult (Aithal *et al*., 2004). Despite advances in external fixation technology, no ideal fixation device exists for managing tibial fractures in large animals, particularly when dealing with complex or unstable fractures. This gap highlights the need for further innovation in fracture management solutions.

One promising advancement in external fixation technology is the development of the Multi-Axial Circular External Fixator (MCEF), which offers enhanced flexibility and adaptability over traditional systems such as the Ilizarov fixator. Based on its modular and adjustable design, MCEF is hypothesized to provide superior adaptability in stabilizing anatomically complex fractures, particularly in regions with variable bone geometry or substantial muscular coverage. This improved adjustability is expected to result in better alignment, stress distribution, and clinical outcomes compared with conventional circular fixators, especially in large animal orthopedic applications.

To evaluate the performance of MCEF, *in vitro* biomechanical testing using buffalo tibia bones is employed. These bones offer a structurally relevant and cost-effective model for simulating fracture biomechanics in both human and veterinary contexts. Such testing is hypothesized to effectively capture critical mechanical parameters such as axial stiffness, torsional strength, and failure load that are essential for assessing the mechanical integrity of external fixation constructs. This approach enables direct comparisons with traditional full-ring circular fixators under standardized conditions.

It is further hypothesized that MCEF, owing to its multi-planar configuration and design innovation, can provide sufficient biomechanical stability under axial and torsional loads to serve as a viable alternative in the management of complex fractures in large animals. The potential to accommodate anatomical variability and address the limitations of conventional designs is a valuable advancement in external fixation systems. The present study was designed to explore these hypotheses through a comprehensive biomechanical evaluation.

Materials and Methods

Preparation of cadaveric tibia and ESF constructs

Eighteen bovine tibiae were collected from a local slaughterhouse and inspected for any signs of disease. Only bones that were physically and anatomically normal were selected for inclusion in the study. The bones were meticulously cleaned by removing all soft tissues. They were then wrapped in saline-soaked towels and stored at -20°C. Before testing, the bones were thawed to room temperature, which ranged between 24°C and 27°C.The collected bones were divided into three groups: Full Ring Circular External Fixator (CEF), Multi-Axial Circular External Fixator (MCEF), and

Table 1: Design of biomechanical study

Groups		Number of bones	Biomechanical test
Group A	Full ring CEF (CEF)	6 - Fractured bones 6 - Fractured bones	Axial compression Axial torsion
Group B	Multiaxial CEF (MCEF)	6 - Fractured bones 6 - Fractured bones	Axial compression Axial torsion
Group C	Intact bone (IT)	6 - Intact bones 6 - Intact bones	Axial compression Axial torsion

intact bone (IT). All constructed specimens underwent mechanical testing through destructive loading to evaluate their performance under both axial compression and axial torsion. The design of the study is summarized in Table 1.

The fixators were made from mild steel, a low-carbon steel containing 0.25% carbon, and were later nickel-plated. Mild steel was selected for prototype fabrication due to its low cost, machinability, and proven mechanical strength for in-vitro biomechanical studies. While stainless steel or titanium would be more appropriate for clinical applications, mild steel is widely accepted for preliminary mechanical evaluations in laboratory conditions (Aithal *et al*., 2010b). The transosseous pins and wires made of 316 L stainless steel were used to fix the ring with bone. Fixator rings with inner diameters of 20 cm and 15 cm were used to create two 4-ring circular external fixator (CEF) configurations. These rings were made of nickel-plated mild steel (low-carbon steel containing 0.25% carbon). The four-ring CEF (CEF) construct was developed using a full-ring model with an internal diameter of 20 cm. Another model, the novel multi-axial CEF (MCEF), was developed using two rings with diameters of 20 cm and 15 cm, respectively. The distance between the rings, connected either by threaded connecting bars or multi-axial linkages, was maintained at 4.5 cm.

Threaded connecting bars (10 mm diameter with 2 mm pitch) (Fig. 1b) and fixation bolts were used to secure the rings in the CEF model and also between the rings of the same diameter in the MCEF model. Two connecting bars were used on the cranial side of the two rings of different diameters. Two multi-axial linkages (Fig. 1c) were used as a connecting bar between the rings of different diameters on the caudal aspect of these two rings. The multi-axial linkages consisted of cylindrical components with adjustable lengths and hexagonal slots for tightening, which were adjusted using Allen keys. The two ends of each linkage were fixed to the extended connecting bars from proximal and distal rings, ensuring a stable construct.

The process of developing CEF and MCEF-bone constructs

The fixing of the rings to the bone involved the use of 3.5 mm K-wires made up of 316 stainless steel. These wires were driven through the bone with a low-speed electric drill and then secured to the fixator rings using slotted fixation bolts. The distance between the rings and cranial and caudal aspect of the bone were kept uniform in all the constructs considering anatomy of the bone and soft tissue coverage. Each ring was affixed with two 3.5 mm K-wires (Fig. 1a). One wire was inserted from caudomedial to craniolateral, while the other followed a craniomedial to caudolateral direction. Subsequently, the wires were secured by manually tightening the nuts on the fixation bolts with a wrench. The order of fixation for the rings was as follows: the most proximal ring was secured first, followed by the most distal ring, and then the two middle rings (Figs. 2a and b).

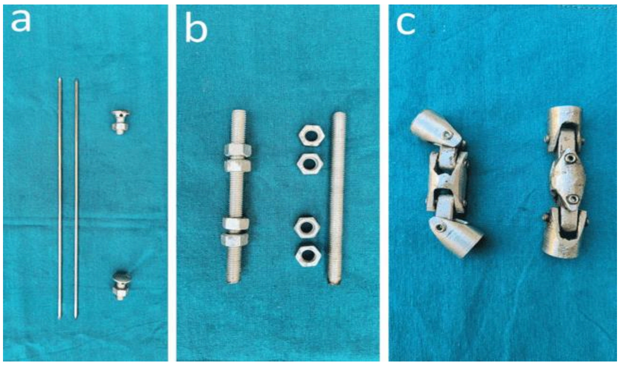

Fig. 1: (**a**) 3.5 mm K-wire with slotted bolts, (**b**) 10 mm connecting bars with nuts, and (**c**) Multi-axial linkages

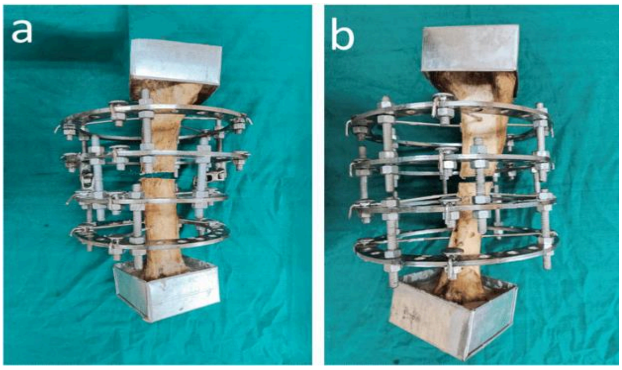

Fig. 2: Bovine tibia bone constructs prepared for biomechanical testing. (**a**) CEF, and (**b**) MCEF construct

In the bones affixed with fixators, a mid-shaft transverse fracture with a gap of 20 mm was intentionally created in all cases using an electric saw. This was done to replicate an unstable fracture configuration accurately. Eccentric placement of cadaveric bone was achieved in both the construct by keeping a distance of 1 inch between ring and cranium of the tibial tuberosity in the first ring and 2 inches cranial to the distal third tibial diaphysis in both the construct.

Fracture simulation

For the tibiae designated for mechanical testing, both the proximal and distal ends were embedded in methyl methacrylate (Pyrax® denture-based polymer resin from Pyrax Polymers, India). The aluminium pedestals of size 10 cm × 10 cm with a depth of 5 cm was used to fix the bones and methyl methacrylate for compression tests. An iron pedestal along with a holding bar of similar size was used for axial torsion tests to ensure stability of the bone constructs during torsional forces. Notably, each pedestal integrated a 5 cm segment of the bone (Figs. 2a and b). Pretesting cranio-caudal and mediolateral view radiographs (Figs. 3a-d) were taken to be compared with post testing radiographs.

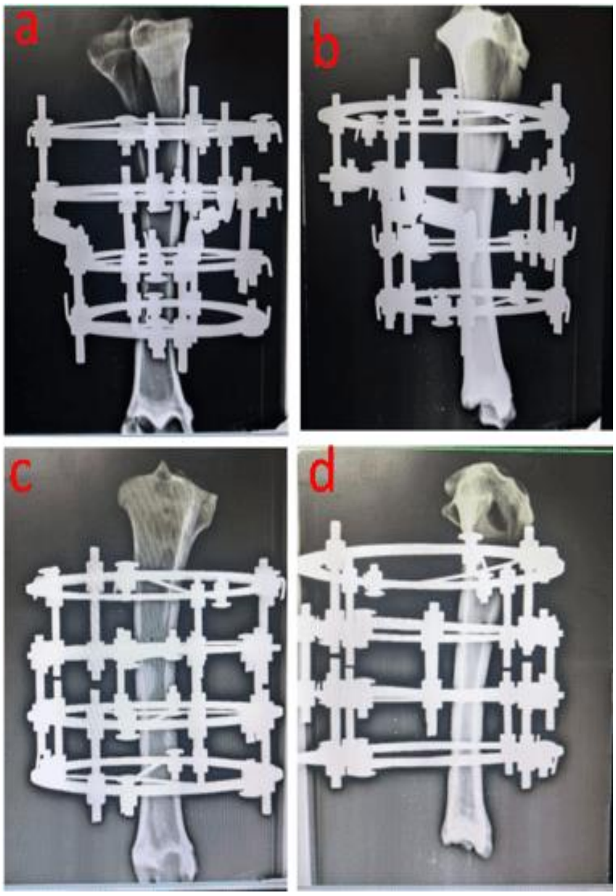

Fig. 3: Preoperative radiographs before test. (**a**) MCEF cranio-caudal, (**b**) Mediolateral view, (**c**) CEF cranio-caudal, and (**d**) Mediolateral view

Mechanical tests

The biomechanical tests were performed at Indian Institute of Technology, Mandi, Himachal Pradesh. All the tests were performed based on a load control protocol.

Axial compression

For the compression test, the specimens were loaded using a servo-hydraulic actuator mounted on a 50 kN Universal Testing Machine (Tinius Olsen India Pvt. Ltd.) equipped with a precision load cell with ±0.1% accuracy (Figs. 4a and b). Axial compression was applied along the longitudinal axis of the construct at a consistent displacement rate of 2 mm/s until failure (Steiner *et al*., 1993; Shi *et al*., 2000; Shah, 2019). The construct was considered as failed when either the implant or the bone failed. Failure was defined as the point of major yield in the load versus displacement curve (Figs. 5a-c). The mode of failure was recorded in each case. Load-deformation curves were recorded for each specimen, and the yield load and failure load were directly derived from these curves. The failure load (N) was calculated as the maximum load reached before failure, which was directly read from the load-deformation curve. To determine stress (N/mm²), the failure load was divided by the cross-sectional area of the specimen. Stiffness (kN/mm) was calculated as the slope of the initial linear portion of the load-deformation curve, representing the ratio of the load to deformation within the elastic region. The yield load (N) was defined as the load at which a specified permanent deformation occurred; typically determined using a 0.2% offset method on the curve. Yield energy (J) was calculated as the area under the load-deformation curve up to the yield point. The break force (N) was often synonymous with the failure load but was specifically noted as the load at which the material ultimately fractured. Displacement (mm) was measured as the change in length of the specimen at the point of failure or yield. Mean values ± standard error (SE) for different parameters were estimated for the various constructs and then compared.

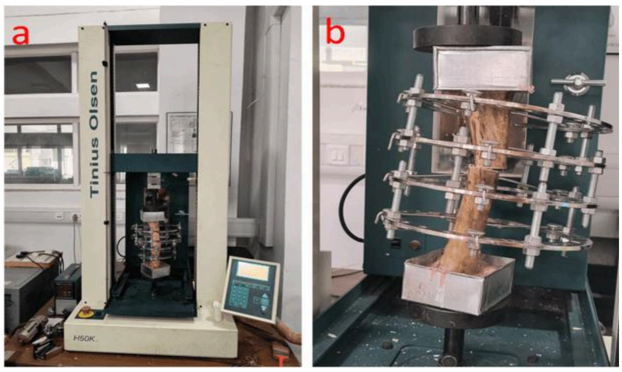

Fig. 4: 50 kN closed-loop compression testing machine showing (**a**) the CEF construct, and (**b**) the MCEF construct mounted for axial compression testing

Axial torsion

The constructs were subjected to torsional loading at a constant rotational speed of 12 degrees per min using a displacement-controlled protocol on a Torsion Testing Machine (Tinius Olsen India Pvt. Ltd.) with a maximum torque capacity of 1000 Nm (Figs. 6a and b) (Calhoun *et al*., 1992; Podolsky and Chao, 1993). The torque was axially applied on to the proximal end fixture. The constructs were twisted constantly along the clockwise direction (positive loading). The different parameters were calculated using the load-deformation (torque-

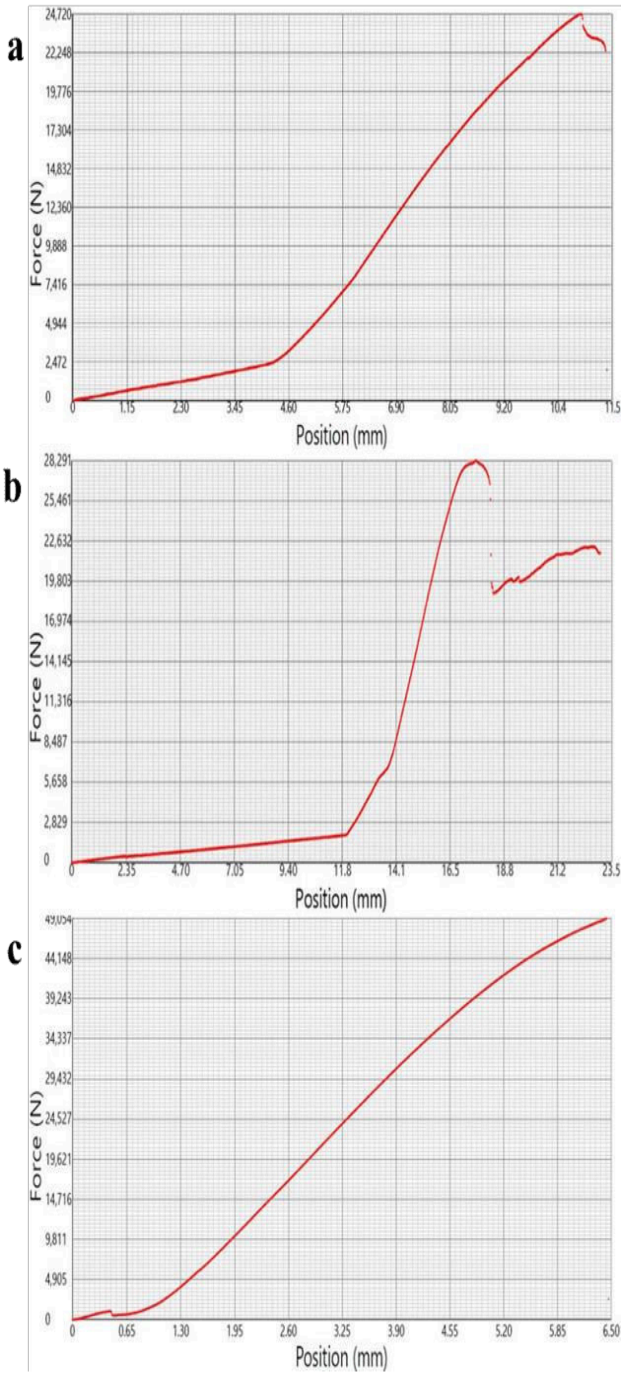

Fig. 5: Showing force/position curve under axial compression. (**a**) MCEF, (**b**) CEF, and (**c**) Intact bone

angle) curves (Figs. 7a-c). The maximum torque (Nm) was identified as the highest value of torque applied to the specimen before failure or plastic deformation occurred. The angle at maximum torque (T max) was determined as the angular displacement (in degrees) corresponding to the maximum torque. The maximum angle (degrees) was recorded as the greatest angular deformation experienced by the specimen prior to complete failure. Stiffness (k) was calculated from the slope of the linear elastic region of the torque-angle curve, representing the specimen's resistance to deformation under torsion. The torsion strength (Nm/mm²) was computed by dividing the maximum torque by the cross-sectional area of the specimen, providing a measure of the ability of the material to withstand torsional forces per unit area.

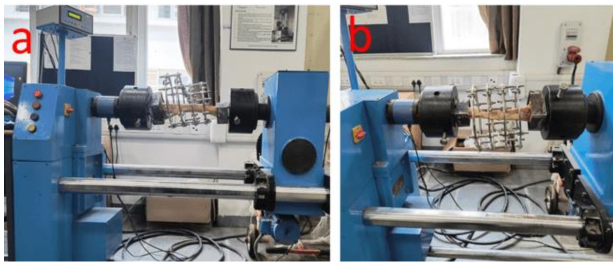

Fig. 6: Constructs mounted on the material testing machine for axial torsion testing. (**a**) MCEF construct, and (**b**) CEF construct

Post-testing construct photographs radiographs

After testing of the sample, a cranial photograph (Figs. 8a-f), a cranio-caudal and medio-lateral radiographs (Figs. 9a-j) of bone implant constructs were taken to assess the status of the constructs.

Statistical analysis

The data were subjected to analysis using commercially available SPSS software version 26.0 (SPSS, Inc., Chicago, IL). Single measurements of continuous variables between groups were compared using one way analysis of variance (ANOVA) and post hoc Tukey's HSD test was employed. The level of significance was set at $P<0.05$ (Snedecor and Cochran, 1994).

Results

None of the constructs in the axial compression and torsion tests exhibited complete failure during the initial testing phase. In the axial compression test, the failure load and stress parameters are detailed in Table 2. The intact bones had a significantly higher failure load (43031 ± 4233 N) compared with both MCEF (24863.50 ± 1018 N) and CEF (27469.50 ± 3246 N) constructs, indicating a greater ability to withstand axial forces ($P<0.0001$). However, both MCEF and CEF showed sufficient axial stiffness to maintain structural stability, though MCEF displayed slightly lower stiffness due to its multi-axial linkages, which provided flexibility but reduced rigidity.

In the torsion tests, CEF demonstrated the highest maximum torque (70.05 ± 18.82 Nm) compared with MCEF (65.96 ± 5.71 Nm), as shown in Table 3, though no statistically significant difference was observed between these two constructs ($P=0.832$). Intact bones, as expected, had the lowest flexibility under torsional loads, reflecting their natural resistance to torsion.

Failure modes were observed primarily in the form of plastic deformation and stress concentrations rather than complete fracture. In the MCEF constructs, plastic deformation occurred in the bones (n=6), while wire

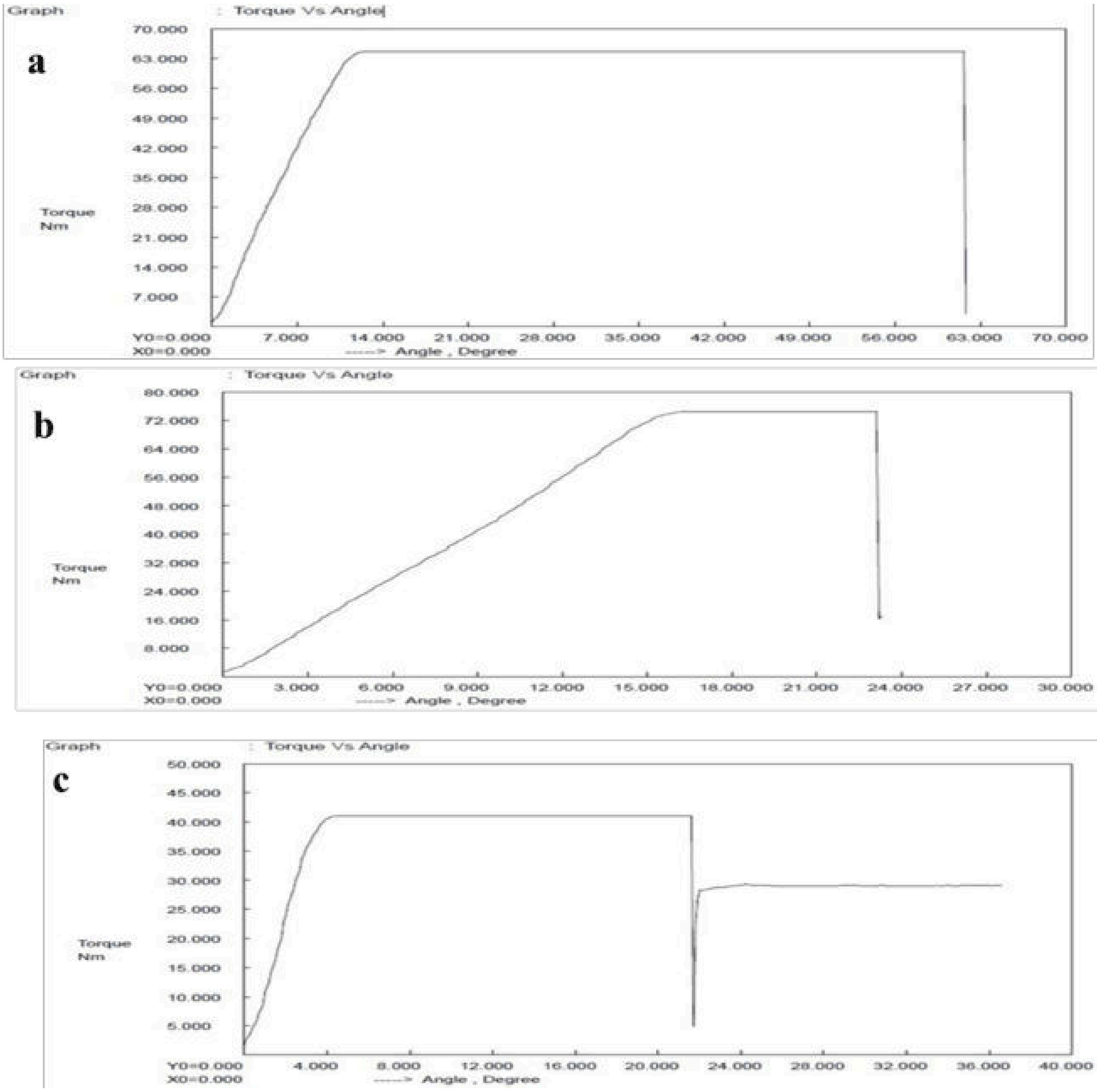

Fig. 7: Showing Torque Vs Angle graph under axial torsion. (**a**) MCEF, (**b**) CEF, and (**c**) Intact bone

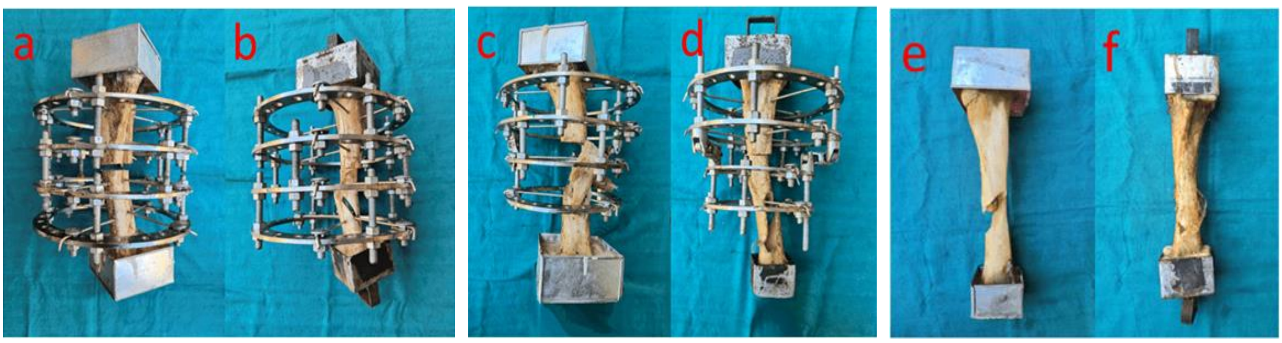

Fig. 8: Post-mechanical testing photographs of (**a**) CEF under compression, (**b**) CEF under torsion, (**c**) MCEF under compression, (**d**) MCEF under torsion, (**e**) intact bone under compression, and (**f**) intact bone under torsion

Table 2: Mean ± SD values of the mechanical testing variables measured for various specimens subjected to axial compression

S. No.	Parameters	MCEF (A)	CEF (B)	Intact (C)
1	Failure load (N)	24863.50±1018[b]	27469.50±3246[b]	43031±4233[a]
2	Stress (N/mm^2)	15.44±1.05[b]	17.05±1.50[b]	29.26±2.32[a]
3	Stiffness (N/mm)	17038±603[b]	18734±724[b]	31383±2689[a]
4	Yield load (N)	31780±1265[b]	33737±1947[b]	57350±3409[a]
5	Yield energy (J)	98.72±13.68[c]	136.72±29.46[b]	178.16±13.21[a]
6	Break force (N)	12951.33±2573[b]	15860.00±3832[ab]	19872±4142[b]
7	Displacement (mm)	16.32±1.23[a]	17.82±1.14[a]	17.85±1.05[a]
8	Weight (kg)	4.09±0.12[b]	4.14±0.09[b]	1.17±0.1[a]

Values with different superscript letters [a, b] and [c] in each rows represent significantly different mean±SD. The same letters indicate no significant difference between the groups

Table 3: Mean ± SD of the mechanical testing variables measured for various specimens subjected to axial torsion

S. No.	Parameters	MCEF (A)	CEF (B)	Intact (C)
1	Maximum torque (Nm)	65.96±5.71[b]	70.05±18.82[b]	47.45±4.30[a]
2	Angle at T. max (degree)	25.23±3.29[b]	29.87±19.19[b]	5.77±1.78[a]
3	Maximum angle (degree)	24.67±6.31[b]	32.65±17.78[b]	5.98±1.66[a]
4	Stiffness (k)	4.50±1.06[b]	3.13±1.51[b]	8.72±2.22[a]
5	Torsion strength (Nm/mm^2)	0.21±0.02[b]	0.22±0.06[b]	0.15±0.01[a]

Values with different superscript letters [a] and [b] in each rows represent significantly different mean±SD values. The same letters indicate no significant difference between the groups

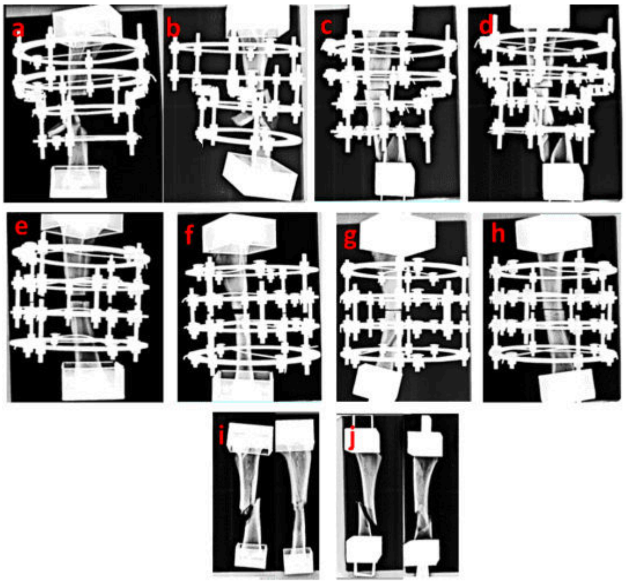

Fig. 9: Postoperative radiographs after test. (**a**) MCEF cranio-caudal, (**b**) Mediolateral view post axial compression tests, (**c**) MCEF cranio-caudal, (**d**) Mediolateral view post axial torsion tests, (**e**) CEF cranio-caudal, (**f**) Mediolateral view post axial compression tests, (**g**) CEF cranio-caudal, (**h**) Mediolateral view post axial torsion tests. Intact bone radiographs post (**i**) Axial compression, and (**j**) Axial torsion tests

bending was noted in three cases. The CEF constructs experienced wire bending (n=3) and plastic deformation of the bones (n=6), with two cases showing loosening of the fixation bolts under high stress. Both constructs demonstrated signs of fatigue under torsional loads, but no catastrophic failures were observed during the compression or torsion tests.

Discussion

Bone is a dynamic living tissue, and its healing is influenced by both mechanical and biological factors. Maintaining the bones blood supply creates an optimal biological environment for healing. Additionally, proper stability of the bone ends is crucial to prevent excessive movement, which could result in non-union. The biomechanical environment of a fracture is determined by the mechanical properties of a fracture fixation device (Fleming *et al*., 1989). Research suggests that interfragmentary strain below 10% is necessary for successful healing of fracture (Egol *et al*., 2004; Tan *et al*., 2014). CEFs serve two main purposes: they facilitate distraction to promote osteogenesis at the fracture interface during treatment and transfer stresses between the two bone segments (Widanage *et al*., 2023).

External skeletal fixation (ESF) devices are widely used for managing long bone fractures and performing limb lengthening procedures in both humans and animals (Ferretti, 1991; Cervantes *et al*., 1996; Lewis *et al*., 1999; Marcellin-Little, 1999). ESF offers several key advantages, including allowing immediate full weight-bearing after bone fixation, promoting joint mobility, and creating an optimal environment for osteosynthesis and wound healing, all without the need to place an implant at the fracture site (Anderson and St Jean, 1996). However, their use in large animal fracture repair has been limited due to a lack of appropriate fixation devices capable of withstanding the heavy loads and forces exerted on the angularly oriented bones of larger species (Shah, 2019). Additionally, economic constraints and challenges associated with postoperative care have further restricted the application of ESF in ruminants.

To ensure clinical acceptance of ESF in large animals, the device must be rigid, well-tolerated, easy to apply, and cost-effective (Singh *et al*., 2007). The stability and rigidity of the device are crucial in providing an optimal mechanical environment at the fracture site. An ideal external fixator must maintain bone fragment alignment while permitting micromotion to encourage bone healing (Rubin and Lanyon, 1984; Wu *et al*., 1984; Calhoun *et al*., 1992). While numerous biomechanical studies have been conducted on ESF systems for small animals and humans (Gasser *et al*., 1989; Calhoun *et al*., 1992; Lewis *et al*., 1998; Lewis *et al*., 2001), research on ESF systems for large animals is limited (Singh *et al*., 2007).

The present study compared the biomechanical characteristics of a novel MCEF construct with a traditional CEF using buffalo tibia bones as a model (Singh *et al*., 2007; Shah, 2019). The dimensions of pins, wires, rings, and connecting rods were decided based on the results of previous studies (Aithal *et al*., 2004, 2007; Singh *et al*., 2007). The design and dimensions of side bars and rings were standardized during a pilot study. Factors such as ring diameter, wire configuration, and wire tension were examined to determine their impact on stability and rigidity (Gasser *et al*., 1989; Calhoun *et al*., 1992; Lewis *et al*., 1999). Additionally, challenges related to tibia fixation, where eccentric positioning of rings is necessary due to anatomical constraints, were considered (Cervantes *et al*., 1996). The development of MCEF addressed these limitations by offering improved flexibility and ease of use, making it a promising candidate for more effective fracture management.

An important innovation in the design of the Multi-Axial Circular External Fixator (MCEF) examined in this study lies in its hybrid configuration, which combines rigid and flexible elements to optimize biomechanical adaptability (Bowers and Anderson, 2024). On the cranial (anterior) side of the construct, two standard threaded connecting bars provided fixed-length stabilization between the rings of differing diameters. These components ensured baseline structural rigidity, especially under axial loads. Conversely, the caudal (posterior) side featured two custom-designed multi-axial linkages. These linkages consisted of telescoping cylindrical elements with hexagonal end slots, which allowed for angular adjustment and secure fixation using Allen keys. The two ends of each linkage were anchored to extended bolts from the proximal and distal rings, forming a closed, yet semi-flexible frame.

Biomechanical aspects such as stability, load distribution, and rigidity must be evaluated when selecting or assessing ESF devices for tibia fractures. The biomechanical properties of fixators significantly influence their effectiveness in stabilizing fractures and facilitating bone healing (Rubin and Lanyon, 1984; Wu *et al*., 1984; Calhoun *et al*., 1992). In particular, the operating performance of CEFs is highly dependent on the overall stiffness of the device (Widanage *et al*., 2023). These factors are critical in determining which fixator provides the most secure bone hold, ensuring optimal healing and minimizing complications. Previous studies have highlighted these challenges. Cervantes *et al*. (1996) evaluated the biomechanical properties of a 4-ring CEF applied to equine third metacarpal bones and found that 2.25-inch pins per ring did not provide sufficient stiffness for unstable third metacarpal fractures in a 450 kg horse. Similarly, Rapoff *et al*. (1995) designed an ESF for horses and cattle using 6.4 mm wire ropes as transosseous components and observed that wire ropes exhibited greater stiffness under axial compression than a single pin. Clinical and biomechanical studies have demonstrated the effectiveness of CEF in large ruminants (Aithal *et al*., 2004; Aithal *et al*, 2010a, b). Additionally, a comparative study of bilateral linear, circular, and hybrid fixators for large ruminants indicated that hybrid fixators provided the highest strength (Dubey *et al*., 2021). However, it is important to note that comparing results across different biomechanical studies can be challenging, as these experiments are often conducted under varying conditions, using different machinery, fixation techniques, and metal components. These differences can introduce variability that complicates direct comparisons between devices and their mechanical performance.

The present study subjected CEF and MCEF systems to axial compression and axial torsion tests to assess their potential for providing stable fixation in large animal fractures. In order to replicate a more typical clinical fracture pattern, we developed a gap model by mid-diaphyseal osteotomy with a 2 cm gap, similar to a few previous models used for *in vitro* research (Gasser *et al*., 1989; Cervantes *et al*., 1996; Lewis *et al*., 1998). Axial compression is a critical factor due to the multiple stresses long bones endure under normal physiological conditions (Calhoun *et al*., 1992). Schneider *et al*. (1982) emphasized that torsional strain plays a crucial role in equine tibia biomechanics, as it significantly affects fracture healing and stability. This observation suggests that similar biomechanical forces, particularly torsional strain, could also be important in ruminants, given their comparable limb structure and weight-bearing dynamics (Singh *et al*., 2007). Therefore, managing torsional forces may be critical in treating tibial fractures in ruminants, as it is in equines. Bending forces, particularly in animals with angularly positioned long bones, are equally significant due to the constant bending stresses. Evidence from both human and animal studies indicates that craniocaudal disruptive forces during physiological loading at the fracture site are greater than lateromedial forces (Paul and Barbenel, 1974).

The biomechanical properties of an ESF system depend on several factors, including the size and number of fixation pins, angle between the pins, the distance between the bone and external frame, and the design and materials used in the fixator frame. In this study, variability was minimized, except for the materials used in constructing the fixator components. Earlier studies used pin placement at nearly 90° angles, but in this study, the angles ranged from 60° to 70°, reflecting clinical constraints. In conventional CEF, fixation pins are tensioned from one end to increase fixation strength, but in this study, the pins were not tensioned to reduce variability across groups. Although tensioning small-diameter pins enhances fixation stability, the effect diminishes with larger pins. As 3.5 mm pins were used in this study, the effect of tensioning was expected to be minimal.

When comparing the two ESF constructs for use in bovine tibial fractures, key biomechanical aspects such as stability, load distribution, and overall rigidity were focused on. These factors are critical in determining which fixator provides the most secure hold on the bone, promoting optimal healing and reducing complications. By assessing these variables across different designs, this study provides valuable insights into the strengths and limitations of each fixator for large animal orthopedic surgery.

In the present study, the axial compression test revealed that CEF demonstrated the greatest axial stiffness, followed closely by MCEF. This aligns with findings by Fleming *et al*. (1989), who emphasized that the axial stiffness of fixators depends heavily on the strength of the sidebars, rings, and the size and orientation of fixation pins. Higher axial stiffness was similarly noted by Fleming *et al*. (1989) and Podolsky and Chao (1993) when the bone was eccentrically positioned within the rings. The compressive stiffness of both MCEF and CEF was approximately 80% of that of intact bone, as previously demonstrated by Singh *et al*. (2007), suggesting that both systems can provide rigid stability under compressive loads in transverse fracture models. The slightly lower stiffness observed in the MCEF construct is primarily due to the design and configuration of the rings and the use of linkages instead of connecting bars.

When assessing the yield and failure loads, CEF and MCEF again showed superior performance. The yield load, which reflects the load at which plastic deformation begins, was highest for CEF and MCEF. This implies that these systems can withstand higher loads before deformation. Additionally, the failure load, defined as the maximum load the construct can sustain before structural failure, also demonstrated a similar trend. These findings are significant because both yield and failure loads are critical parameters that indicate the ability of a fixation system to maintain structural integrity under physiological stresses.

The failure load was significantly higher in intact bones, which maintained their natural structural integrity, as compared with bones with external fixation systems (MCEF and CEF). The introduction of fixators in these groups likely resulted in points of stress concentration, reducing the overall failure load of the bone. Furthermore, the stress results followed the same trend, where intact bones exhibited the highest stress resistance due to their homogeneous structure, while the external fixation systems, particularly MCEF, distributed loads less efficiently, resulting in lower stress resistance.

In terms of stiffness, intact bones once again outperformed the MCEF and CEF systems, as demonstrated by the higher stiffness values in group C. The lower stiffness in bones treated with fixators can be attributed to the mechanical properties of the fixator materials, which are more flexible compared with the natural rigidity of the bone. However, the flexibility introduced by these fixators could be beneficial in clinical cases where some degree of controlled movement is desirable to promote fracture healing.

During axial torsion testing, the CEF system showed superior performance in terms of maximum torque and torsional strength. This is likely due to the circular design of CEF, which distributes torsional loads more evenly across the construct. The MCEF constructs, while offering reasonable torsional resistance, performed less optimally compared with CEF but still better than intact bones. This suggests that external fixation systems, particularly CEF, provide biomechanical advantages in situations where fractures are subjected to significant torsional stress. The advantage of the conventional CEF over other fixator designs lies in its ability to permit axial movement while effectively resisting torsional and translational motions at the bone ends (Sarpel *et al*., 2005). The configuration of CEF, with four uniform rings arranged in a circular pattern, makes the fixator highly rigid on a horizontal plane, providing resistance to loading from any direction. In a biomechanical study by Bronson *et al*. (1998) each 40 mm increase in ring diameter resulted in a 30% decrease in axial stiffness. Additionally, a reduction in wire diameter from 1.8 mm to 1.5 mm led to a 10% decrease in both axial and torsional stiffness (P=0.0001) indicating reduced stiffness due to increased variations in ring diameters.

The flexibility provided by the CEF system is further highlighted by the greater angle at maximum torque and maximum angle values, indicating its ability to allow for controlled movement before failure. This design feature may be advantageous in cases where gradual bone deformation or alignment correction is required. However, it is important to note that the lower stiffness and yield loads in both MCEF and CEF systems indicate that while they enhance flexibility and torsional strength, they do so at the cost of reduced axial stiffness. Yilmaz *et al*. (2003) also reported that a hybrid external fixator is mechanically inferior to a four-ring circular Ilizarov frame. While hybrid external fixators have minimal impact on torsional stability, they significantly reduce axial stiffness and bending resistance. Despite these mechanical drawbacks, hybrid Ilizarov external fixation models may still be preferred due to their ease of use and increased patient comfort.

Clinical implications

Both MCEF and CEF systems offer specific biomechanical benefits depending on the forces involved in the fracture. The MCEF system provides a good balance of flexibility and support under both axial and torsional loads, whereas the CEF system excels in handling torsional forces, making it ideal for fractures with significant rotational stress. The choice between these systems should be based on the biomechanical needs of the fracture being treated in clinical settings. The findings show that every fixation device we created with mild steel alloy was robust enough to withstand a 300 kg force applied under compression and torsion. However, MCEF is also effective for managing fractures in areas where the bone's anatomy and surrounding muscle mass create challenges. This is particularly relevant for large ruminants, such as bovines, where tibial fractures present significant variations in muscle mass and bone contours. The MCEF configuration examined in this study aligns with recent biomechanical trends favouring hybrid and multi-axial fixators. Such systems offer customizable stiffness and alignment, which are crucial in managing anatomically challenging fractures in ruminants (Dubey *et al*., 2021; Bowers and Anderson, 2024). The adaptability of MCEF allows for tailored fixation, improving clinical outcomes. The biomechanical load-bearing capacity of MCEF is sufficient for use in clinical settings with large ruminants. Further research, including comparative studies in large ruminants, is necessary to confirm these findings and evaluate the long-term clinical efficacy of these systems in treating complex fractures.

The biomechanical comparison of MCEF and CEF systems for buffalo tibial fractures reveals that while intact bones provide superior axial strength and stiffness, the CEF system excels in torsional performance due to its circular ring design. MCEF offers a balance between axial support and flexibility, making it suitable for fractures in areas with anatomical variations, such as the bovine tibia. To enhance the understanding of these systems, further research should focus on evaluating their bending properties, as well as conducting clinical studies to validate their effectiveness in real-world cases, particularly for bovine tibial fractures.

Acknowledgements

The authors acknowledge ICAR-IVRI, Izatnagar, and IIT Mandi, Himachal Pradesh, for providing the facilities necessary for this study.

Conflict of interest

The authors declare that they have no competing interest.

References

Adams, SB (1985). The role of external fixation and emergency fracture management in bovine orthopaedics. Vet. Clin. North Am. Food Anim. Pract., 1: 109-129.

Aithal, HP; Amarpal and Kinjavdekar, P (2007). Transarticular fixation using a circular external fixator for management of fractures near the carpal joint in two calves. Vet. Rec., 161: 193-198.

Aithal, HP; Kinjavdekar, P; Amarpal, M; Pawde, AM; Singh, GR and Setia, HC (2010a). Management of tibial fractures using a circular external fixator in two calves. Vet. Surg., 39: 621-626.

Aithal, HP; Kinjavdekar, P; Pawde, AM; Dubey, P; Kumar, R; Tyagi, SK and Madhu, DN (2019). Epoxy-Pin external skeletal fixation for management of open bone fractures in calves and foals: A review of 32 cases. Vet. Comp. Orthop. Traumatol., 32: 257-268.

Aithal, HP; Singh, GR; Hoque, M; Maiti, SK; Kinjavdekar, P; Amarpal; Pawde, AM and Setia, HC (2004). The use of circular external skeletal fixation device for the management of long bone osteotomies in large ruminants: An experimental study. J. Vet. Med., 51: 284-293.

Aithal, HP; Singh, GR; Saxena, RK; Kinjavdekar, P; Amarpal, A; Pawde, AM; Hoque, M and Maiti, SK (2010b). Comparative mechanical, chemical, and microstructural properties of mild steel and aluminum rings used in large animal circular external fixators. Indian J. Anim. Sci., 80: 428-430.

Anderson, DE and St Jean, G (1996). External skeletal fixation in ruminants. Vet. Clin. North Am. Food Anim. Pract., 12: 117-152.

Bowers, KM and Anderson, DE (2024). Delayed union and nonunion: current concepts, prevention, and correction: a review. Bioengineering. 11: 1-19.

Bronson, DG; Samchukov, ML; Birch, JG; Browne, RH and Ashman, RB (1998). Stability of external circular fixation: A multi-variable biomechanical analysis. Clin. Biomech., 13: 441-448.

Calhoun, JH; Li, F; Ledbetter, BR and Gill, CA (1992). Biomechanics of Ilizarov fixator for fracture fixation. Clin. Orthop. Relat. Res., 280: 15-22.

Cervantes, C; Madison, JB; Miller, GJ and Casar, RS (1996). An in-vitro biomechanical study of a multiplanar circular external fixator applied to equine third metacarpal bones. Vet. Surg., 25: 1-5.

Dubey, P; Aithal, HP; Kinjavdekar, P; Amarpal, GP and Nanjappa, MD (2021). 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. World J. Surg. Surg. Res., 4: 1-9.

Egol, KA; Kubiak, EN; Fulkerson, E; Kummer, FJ and Koval, KJ (2004). Biomechanics of locked plates and screws. J. Orthop. Trauma. 18: 488-493.

Ferretti, A (1991). The application of the Ilizarov technique to veterinary medicine. In: Bianchi-Maiocchi, A and Aronson, J (Eds.), *Operative principles of Ilizarov*. (1st Edn.), Milan, Italy: Medi. Surgical Vido., PP: 551-570.

Fleming, B; Paley, D; Kristiansen, T and Pope, M (1989). A biomechanical analysis of the Ilizarov external fixator. Clin. Orthop. Relat. Res., 241: 95-105.

Gasser, B; Boman, B; Wyder, D and Schneider, E (1989). Stiffness characteristics of the circular Ilizarov device as opposed to conventional external fixation. J. Biomech. Eng., 112: 15-21.

Gasser, B; Boman, B; Wyder, D and Schneider, E (1990). Stiffness characteristics of the circular Ilizarov device as opposed to conventional external fixation. J. Biomech. Eng., 112: 15-21.

Lewis, DD; Bronson, DG; Samchukov, ML; Welch, RD and Stallings, JT (1998). Biomechanics of circular external skeletal fixation. Vet. Surg., 27: 454-464.

Lewis, DD; Cross, AR; Carmichael, S and Anderson, MA (2001). Recent advances in external skeletal fixation. J. Small Anim. Pract., 42: 103-112.

Lewis, DD; Radasch, RM; Beale, BS; Stallings, JT; Lanz, OI; Welch, RD and Samchukov, ML (1999). Initial clinical experience with the IMEX TM circular external skeletal fixation system. Part I: Use in fractures and arthrodeses. Vet. Comp. Orthop. Traumatol., 12: 108-117.

Marcellin-Little, DJ (1999). Fracture treatment with circular external fixation. Vet. Clin. North Am. Small Anim. Pract., 29: 1153-1170.

Mitousoudis, AS; Magnissalis, EA and Kourkoulis, SK (2010). A biomechanical analysis of the Ilizarov external fixator. EPJ Web Conf., 6: 21002.

Moss, DP and Tejwani, NC (2007). Biomechanics of external fixation. Bull. NYU Hosp. Jt. Dis., 65: 294-299.

Paley, D (1991). Biomechanics of the Ilizarov external fixator. In: Bianchi-Maiocchi, A and Aronson, J (Eds.), *Operative principles of Ilizarov*. (1st Edn.), Milan, Italy: Medi Surgical Vido., PP: 33-41.

Paul, JP and Barbenel, JC (1974). Biomechanics. In: Ray, CD (Ed.), *Medical engineering*. (1st Edn.), Chicago, IL: Year Book Medical Publishers. PP: 199-215.

Podolsky, A and Chao, EYS (1993). Mechanical performance of Ilizarov circular external fixators in comparison with other external fixators. Clin. Orthop. Relat. Res., 293: 61-70.

Rapoff, AJ; Markel, MD and Vanderby, R (1995). Mechanical evaluation of transosseous wire rope configurations in a large animal external fixator. Am. J. Vet. Res., 56: 694-699.

Rubin, CT and Lanyon, LE (1984). Regulation of bone formation by applied dynamic loads. J. Bone Joint Surg. Am., 66: 397-402.

Sarpel, Y; Gulsen, M; Togrul, E; Capa, M and Herdem, M (2005). Comparison of mechanical performance among different frame configurations of the Ilizarov external fixator: experimental study. J. Trauma Acute Care Surg., 58: 546-552.

Schneider, RK; Milne, DW; Gabel, AA; Groom, JJ and Bramlage, LR (1982). Multidirectional *in vitro* strain analysis of the equine radius and tibia during dynamic loading with and without a cast. Am. J. Vet. Res., 43: 1541-1550.

Shah, MA (2019). Circular and hybrid external skeletal fixation systems for tibial fracture repair in large ruminants: Biomechanical and clinical study. Doctoral Dissertation, IVRI, Izatnagar. PP: 1-166.

Shi, D; Liu, K; Zhang, H; Wang, X; Li, G and Zheng, L (2020). Investigating the biomechanical function of the plate-type external fixator in the treatment of tibial fractures: a biomechanical study. BMC Musculoskelet. Disord., 21: 1-9.

Singh, GR; Aithal, HP; Saxena, RK; Kinjavdekar, P; Amarpal, M; Hoque, M; Maiti, SK; Pawde, AM and Joshi, HC (2007). In-vitro biomechanical properties of linear, circular, and hybrid external skeletal fixation devices developed for use in large ruminants. Vet. Surg., 36: 80-87.

Snedecor, GW and Cochran, WG (1994). *Statistical methods*. (8th Edn.), Ames, IA: Iowa State University Press. PP: 1-503.

Steiner, A and Anderson, DE (2009). Fracture management in cattle. In: Anderson, DE (Ed.), *Fracture management in cattle*. (2nd Edn.), St. Louis, Missouri, USA, Saunders Elsevier. PP: 253-258.

Steiner, A; Iselin, U; Auer, J and Lischer, CJ (1993). Shaft fractures of the metacarpus and metatarsus in cattle. Vet. Comp. Orthop. Traumatol., 6: 138-145.

Tan, BB; Shanmugam, R; Gunalan, R; Chua, YP; Hossain, G and Saw, A (2014). A biomechanical comparison between Taylor's spatial frame and Ilizarov external fixator. Malays. Orthop. J., 8: 35-39.

Widanage, KN; De Silva, MJ; Lalitharatne, TD; Bull, AM and Gopura, RARC (2023). Developments in circular external fixators: A review. Injury. 54: 1-16.

Wu, JJ; Shyr, HS; Chao, EYS and Kelly, PJ (1984). Comparison of osteotomy healing under external fixation devices with different stiffness characteristics. J. Bone Joint Surg. Am., 66: 1258-1264.

Yilmaz, E; Belhan, O; Karakurt, L; Arslan, N and Serin, E (2003). Mechanical performance of hybrid Ilizarov external fixator in comparison with Ilizarov circular external fixator. Clin. Biomech., 18: 518-522.