Biomechanics of Two External Fixator Devices Used in Rat Femoral Fractures

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Abstract

The use of external fixators allows for the direct investigation of newly formed interfragmentary bone, and the radiographic evaluation of the fracture. We validated the results of a finite element model with the in vitro stiffness' of two widely used external fixator devices used for in vivo analysis of fracture healing in rat femoral fractures with differing construction (Ti alloy ExFix1 and PEEK ExFix2).

Rat femoral fracture fixation was modelled using two external fixators. For both constructs an osteotomy of 2.75 mm was used, and offset maintained at 5 mm. Tufnol, served as standardized substitutes for rat femora. Constructs were loaded under axial compression and torsion. Overall axial and torsional stiffness were compared between the in vitro models and FE results. FE models were also used to compare the fracture movement and overall pattern of von Mises stress across the external fixators.

In vitro axial stiffness of ExFix1 was 29.26 N/mm± 3.83 compared to ExFix2 6.31 N/mm± 0.67 ($p^*<0.05$). Torsional stiffness of ExFix1 was 47.5 Nmm/^{o} ± 2.71 compared to ExFix2 at 19.1 Nmm/^{o} ±1.18 ($p^*<0.05$). FE results predicted similar comparative ratios between the ExFix1 and 2 as the in vitro studies. FE results predicted considerably larger interfragmentary motion in the ExFix2 comparing to ExFix1.

We demonstrated significant differences in the stiffness' of the two external fixators as one would expect from such variable designs; yet, importantly we validated the utility of an FE model for the analysis and prediction of changes in fracture mechanics dependent on fixator choice.

Keywords: fracture fixation, finite element analysis, biomechanics

1. Introduction

Multiple physiological and mechanical factors govern the fracture healing process. Overall stiffness of the fracture fixation construct directly impacts the axial, torsional and shear interfragmentary movement at the fracture site (1-3). These subsequently impact the healing process and as with physiological healing, rigid fixation will lead to intramembranous ossification, while those that are less rigid, allow for the creation of cartilaginous callus and endochonrdral ossification (4, 5).

Rodents have been widely used to investigate the fracture fixation. They are an invaluable animal model used to understand the fracture healing process and to develop new technologies and treatments to address complications such as non-union. A number of external fixators have been used to fix femoral fractures in rodents. These fixators, typically result in a combination of intramembranous and endochondral ossification with studies illustrating healing by various biological scenarios in different models (6, 7).

The literature comparing the biomechanical differences of existing external fixators in rodents is limited. Harrison et al. (8) reported no significant difference in axial stiffness between aluminium and titanium fixator bar materials. However, pin material and thickness does have a large effect on torsional and axial stiffness. Mark et al. (9) reported a 50% decrease in axial stiffness and transverse stiffness of the fixator, when using a 1.0-mm compared to a 1.2-mm outer diameter pin. Willie et al. (10) demonstrated significantly reduced stiffness at the fracture site of titanium alloy pins versus stainless steel in fixators of the same design, with similar effects of body material and offset on stiffness as previous studies. Glatt et al. (11) reported the development of a variable stiffness PEEK fixator where fracture rigidity can be altered during healing. This PEEK fixator is gaining favour for use in the investigation of rodent fracture healing as the four pin construct is lighter than traditional titanium and stainless steel fixators and has been shown to be well tolerated in vivo (12). In contrast, the majority of studies utilise a more traditional unilateral fixator design such as the

Harrison et al. titanium alloy fixator. Recently reported variations of the Harrison fixator utilise 2 carbon fibre cross bars with four aluminium pins (13, 14); heavier than the Glatt fixator. Therefore, while there is a body of literature on the biomechanics of different external fixators on rodents (see also 15,16) and some variations of them e.g. their material properties and dimensions, to best of our knowledge, no study has compared the effects of a variable stiffness fixator and a static fixator on the in vitro stabilisation of a rat femoral fracture model. These are two different external fixator designs and a direct biomechanical comparison between them lacks in literature and is crucial to advance our understanding of the interplay between the biomechanical and biological factors in the context of fracture healing.

Studies investigating the effect of fixator construct on fracture stabilisation can be laborious, necessitating investigation of each design parameter-including crossbar number/size/ offset, pin size and each component material. Subsequently, the ability to utilise computational modelling to determine the mechanical characteristics of any fixator construct, is invaluable. So long as the models are validated using in vivo or in vitro experimental data finite element (FE) modelling provides a unique opportunity to model experimental scenarios computationally and accurately (17-19). As such, the creation of a validated design tool, that can replicate in vivo biomechanics, allow the augmentation and refinement of fixator characteristics to best suit experimental conditions, and yet does so in a timely and cost effective fashion-would be most beneficial to those working with fracture experimental models.

The aim of this study was to compare the biomechanics of two increasingly utilised rodent external fixators; a derivation of the Harrison et al titanium alloy fixator, and the Glatt/AO PEEK external fixator, but more importantly to validate an FEA model of design with the ex vivo data. These fixators where chosen specifically in order to attempt FEA validation with two very disparate fixator designs. We utilised a series of experimental in vitro testing and in silico computational models based on finite element method.

2. Materials and Methods

2.1 External fixator designs

The study compared two external fixator designs. The first (ExFix1) has two graphite cross bars of 2x40 mm, spaced 4 mm apart, fixed between two titanium alloy (Ti6Al-4v) blocks. These blocks measured 8mm in height, 10 mm in width and 7.2 mm in depth. This design used 4 titanium alloy threaded pins of 0.8/1.0 mm, fixed within the blocks with stainless steel grub screws. The second fixator (ExFix2) was comprised of a single PEEK crossbar and again four stainless steel threaded pins. The crossbar measured 16.5 mm long, 5 mm wide and 2 mm deep with four 1mm holes to locate the steel pins. A single 12.5 mm long, 1 mm wide rectangular opening runs parallel with the openings for the steel pins; again each pin measured 0.8/1.0 mm. The offset as measured from the free length of the pins beneath the crossbar to the upper surface of the bone, was kept constant at 5 mm throughout testing. ExFix1 weighed 6.23 g (range 6.22-6.31 g), and ExFix 2 3.11 g (range 3.08-3.65 g).

A hollowed homogenous rod of laminated Tufnol (Tufnol Composites, Birmingham, UK), of similar elastic modulus to adolescent rat femora (inner diameter 1.5 mm, outer diameter 4 mm, length 35 mm) served as standardised substitute for bone and fixed using ExFix1 (n=5) and 2 (n=5). Fixation was carried out using custom drill guides of 0.8 mm that allowed for the accurate predrilling of holes into the Tufnol, after which pins were manually screwed into position to breach both cortices by one thread. After the fixator was fixed to the Tufnol bone a fracture was created with a 2.75 mm fracture gap maintained.

2.2 In vitro testing

The Tufnol specimens were tested non-destructively using a Zwick (Zwick-Roell, Germany) materials testing machine to determine axial and torsional stiffness. In compression, a maximum load of 40 N was applied, with a preload of 0.5 N at a rate of 0.5 mm/min. Load was applied onto potted concave ends of the Tufnol via steel beads attached to the testing machine, and the loading-unloading process repeated three times for each sample.

In torsion both ends of the sample were fixed into titanium cylinders with grub screws to negate slipping during testing. One end of the Tufnol remained static, whilst a maximum vertical load of 40 N was applied to the other end with a lever arm of 75 mm, which led to torsion of 3000 Nmm. Loading was repeated three times per specimen and torsional stiffness was calculated by dividing the applied torque by the degrees of rotation of the proximal end of the Tufnol.

2.3 Finite element analysis

Computer-aided design models of the bone and two external fixators were developed in CATIA V5 (Dassault Systèmes, Paris FR - Figure 1). Dimensions exactly reflected those of the real-life fixator models and all parts assigned isotropic material properties; The Tufnol bone model has an elastic modulus of 6.5 GPa and Poisson's ratio 0.4 (20-22). Titanium alloy blocks in the ExFix1 have an elastic modulus of 96 GPa and a Poisson's ratio of 0.36. The Graphite rods have an elastic modulus of 4.1 GPa and a Poisson's ratio of 0.17. The PEEK crossbar of the ExFix2 has an elastic modulus of 3.6 GPa and a Poisson's ratio of 0.38. Finally, stainless steel pins in both fixators were given the same mechanical properties: an elastic modulus of 193 GPa and Poisson's ratio of 0.31. The effect of screw pull-out at the fixator-Tufnol interface was ameliorated by gluing these contacts during experimental testing; subsequently, the interface experienced minimal micro-motion upon loading in-vitro and allowed all pin-Tufnol interfaces to be modelled as "fully fixed".

Interfaces such as at the crossbar-pin interface had inherent micro-motion as they were either threaded into position or held with grub screws. Thus two simulations were created, one with all contacts "fully fixed" and a second with all grub screws and threaded contacts "relaxed" to account for this motion. The relaxed model used contact elements at the interfaces with a friction coefficient of 0.4 (17). The expectation being that the properties of each fixator would be between these two extreme models.

In order to replicate the boundary conditions of the test rigs, the constraints were applied within the concave housing of the Tufnol under axial loading conditions and along the outside face of the housing under torsional loading conditions. Additionally, the surface/node in which the load was applied was also constrained to translate in only the axis parallel to the line of loading.

Analyses were carried out in FE package ANSYS (Academic Research, Pennsylvania USA). Tetrahedral elements were used to mesh all components of the fixators and Tufnol. Convergence was tested on each fixator by increasing the number of elements from ca. 5,000 to 2,000,000 incrementally. The solution for ExFix1 converged to within 5% at approximately 135,000 elements when measuring axial stiffness and approximately 260,000 elements when measuring torsional stiffness. For ExFix2, the solution converged for both quantities of interest at approximately 322,000 elements. Results converged substantially faster with the use of mid-side nodes, and as such they were used throughout.

In addition to axial and torsional stiffness, FEA was also used to evaluate fracture gap displacement as measured by nodes either side of the osteotomy. Von Mises stresses were calculated for each fixator and the points of maximal stress also determined. It must be noted that since in this study no detail validation of the strain pattern was carried out the stress results were analysed qualitatively.

2.4 Statistical Analysis

Statistical analysis was performed on the experimental data. The ANOVA assumption of normality was tested using the Shapiro–Wilks normality test. If the assumption was met, an ANOVA was performed, if not, a Mann Whitney *U* test was used. The data was analysed using Prism 4.03 (GraphPad Software Inc., San Diego, USA) and a significance level when comparing data was set at $p<0.05$.

3. Results

3.1 Axial stiffness:

ExFix1 was 29.26 N/mm± 3.83 compared to ExFix2 6.31N/mm± 0.67 ($p^*<0.05$). The fully restricted FEA model predicted axial values of 79.95N/mm and 31.57N/mm for ExFix1 and 2 respectively. The model under secondary contact conditions produced axial values of 46.12 N/mm and 7.52 N/mm respectively (Figure 2A).

3.2 Torsional stiffness:

ExFix1 was 47.5 Nmm/º ± 2.71 compared to ExFix 2 at 19.1 Nmm/º ±1.18 ($p^*<0.05$). The fully restricted FEA model predicted torsional stiffness of 98 Nmm/º and 50 Nmm/º for ExFix1 and 2 respectively. The model under secondary contact conditions produced torsional stiffness of 89.8 Nmm/º and 27 Nmm/º respectively (Figure 2B).

3.3 Comparative ratios:

The ratio of ExFix1: ExFix2, axial and torsional stiffness based on the in vitro experimental data was 4.6 and 2.5 respectively. The same ratio based on the FEA with fully fixed interface conditions were 2.5 (46% lower than the experimental data) and 2 (20% lower than the experimental data) for the axial and torsional stiffness respectively. The same ratio based on the FEA with relaxed interface were 5.1 (11% greater than experimental data) and 3.3 (32% greater than experimental data) for the axial and torsional stiffness respectively (Figure 3).

3.4 Fracture movement:

Total fracture movement as measured in the FE models, was greater for ExFix2 in all planes versus ExFix1. Less than 1mm of movement occurred with ExFix1 at the maximal loading however, in the ExFix2 the fragments come into contact leading to a fracture movement of about 2.7 mm based on the relaxed interface model. Under axial loading ExFix1 was found

to have 0.54 and 0.91 mm of movement with the fully fixed and relaxed models. Whereas ExFix2 demonstrated 1.49 and 2.75 mm of movement respectively. Under torsional conditions, ExFix1 showed 0.52 and 0.64 mm of movement with the fully fixed and relaxed models. Versus ExFix2 with 2.20 and 2.74 mm of movement respectively (Figure 4A and b).

3.5 Stress pattern:

The stress contour plots of the equivalent von Mises stresses for each fixator component are shown in Figure 5. In all components of the fixator ExFix1 experienced lower overall stress than ExFix2, in both axial and torsional loading. For all FE analysis maximum stress occurred at the pin-Tufnol interface. In axial loading of both fixators, stress peaks in the pin closest to the point of loading was seen, whilst in torsion, maximum stress occurred in the pins either side of the fracture gap.

4. Discussion

This study compared the mechanical characteristics of two commonly used external fixators in small animal fracture models. We used our in vitro findings to validate a series of finite element models based on axial and torsional stiffness data. Between the two fixators, we found significant differences in stiffness in both the axial and rotational planes, with ExFix1 markedly more rigid in both planes. Throughout the study we maintained a constant offset, pin material and pin diameter, thus allowing the fixator design and crossbar material (Ti alloy/carbon fibre vs. PEEK) to be the dominating factors on overall stiffness. Previous studies have determined that pin size and material are the greatest determinants of fixator stiffness and interfragmentary fracture movement, also demonstrating the importance of offset and pin number (10, 23, 24), our data also suggests the significant impact that the fixator material properties and bar configuration have on the overall stiffness.

In vitro axial stiffness of both ExFix constructs were significantly less than those found with locked nailing techniques (25). ExFix1 was a third as stiff, and ExFix2 just over half as stiff

as reported nailing data (25). Conversely rotational stiffness was greater for the external fixators than locked intramedullary nails, and indeed was greater than physiological numbers from intact bone (torsional stiffness 23 Nmm/º). This greater stiffness in rotation, if related in vivo, will lead to reduced interfragmentary movement in shear and as such will impact bone formation.

Our data suggests the FE model could predict the relative differences between the two external fixators. However, the FE models consistently predicted larger stiffness' then those found in vitro, this difference was considerably larger in the "fixed" model that did not account for any micro-motion at the pin-Tufnol or the pin-fixator interfaces. When relaxing the interfaces, the comparative ratios fell notably and were closer to the experimental in vitro data (see Figure 2). Again highlighting the fundamental role of micromotion at the interfaces in both the in silico and in vitro tests (17).

A relatively large body of work has evaluated the role of FE modelling in clinical fracture fixation scenarios. For example, Ramlee et al. (26) reviewed two external fixators with an FE model and their effects on subtalar dislocation reduction, similarly, Varga et al. (27) reviewed the use of compression screws in scaphoid fracture fixation and the effects these have on interfragmentary forces and again fracture reduction. Both of these studies amongst others (e.g. 3, 28) have validated FE models and underlined their utility in clinical fracture management. Our study uses the modelling technique in the preclinical setting; importantly allowing an understanding of the fracture mechanics without the need for lengthy in vivo experiments. Moreover, the validation of our relaxed FE model, that adjusts for interface micro motion, results in the creation of a tool that can allow the design and manipulation of a fixator to best suit different experimental parameters.

The difference in stiffness has a predictable effect on movement at the fracture gap, which has important implications on fracture healing. Interfragmentary motion of between 0.2-1 mm perpendicular to a diaphyseal fracture has been found to promote union; however, excessive

axial and shear motion will result in delayed healing (1-3). Under axial conditions ExFix2 experiences significant motion where bony fragments come into contact. ExFix1, however, restricts vertical motion under axial loading to less than 1 mm, within the desired envelope. Under torsion, this increases to a value equating to a rotation of up to 17 degrees. ExFix1 limits rotation to less than half this amount at the same levels of loading. Under axial loading, translation and rotation at the fracture gap in ExFix1 is also negligible. Additionally, our findings are particularly relevant when investigating biological and pharmacological interventions where variability in stress across the gap will directly influence the efficacy of these factors (29-31).

The specific pin where the maximum stress occurs changes between loading conditions. In axial loading, maximum stress is located on the most proximal pin in both ExFix1 and ExFix2 whereas under torsion, maximum stress occurred in the pin nearest the proximal end of the fracture. These changes are likely to be a function of the constraint of the Tufnol bone creating higher stresses in the pins adjacent to the fracture site.

While the FE model could not exactly represent the in vitro assembly boundary conditions, the two conditions that were investigated can accurately predict upper and lower limits for in vitro results. Ultimately, we demonstrated considerable differences in the overall stiffness between the two fixators, which should be considered when comparing experimental *in vivo* data on fracture healing. Given a consistent fracture gap fractures stabilised using ExFix2 are more likely to heal though endochondral ossification or go onto a delayed or non-union compared to ExFix1. The in silico model where the threads are not fully bonded, predicted the comparative stiffness between the two fixators, as evidenced by the similar ratios. This data suggests that a computational protocol that includes the micro-motion present at the pin-bone interface, results in a reproducible model of experimental conditions. Further in vivo and computational work is required to demonstrate the effect of gap distance and fixator stiffness on the rate, type and quality of ossification and healing.

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References

1. Goodship AE, Kenwright J. The influence of induced micromovement upon the healing of experimental tibial fractures. J Bone Joint Surg Br. 1985;67,650–655.

2. Claes LE, Heigele CA, Neidlinger-Wilke C, Kaspar D, Seidl W, Margevicius KJ, Augat P. Effects of mechanical factors on the fracture healing process. Clin Orthop Relat Res. 1998;355,S132–147.

3. Moazen M, Jones AC, Leonidou A, Jin Z, Wilcox RK, Tsiridis E. Rigid versus flexible plate fixation for periprosthetic femoral fracture - computer modelling of a clinical case. Medical Engineering & Physics 2012;34,1041-1048.

4. McKibbin B. The biology of fracture healing in long bones. J Bone Joint Surg. 1978;60-B,150–162.

5. Mark H, Nilsson A, Nannmark U, Rydevik B. Effects of fracture fixation stability on ossification in healing fractures. Clin Orthop Relat Res. 2004;419,245-50.

6. Histing T, Garcia P, Holstein JH, Klein M, Matthys R, Nuetzi R, Steck R, Laschke MW, Wehner T, Bindl R, Recknagel S, Stuermer EK, Vollmar B, Wildemann B, Lienau J, Willie B, Peters A, Ignatius A, Pohlemann T, Claes L, Menger MD. Small animal bone healing models: standards, tips, and pitfalls results of a consensus meeting. Bone. 2011;49,591-599.

309 7. Holstein JH, Garcia P, Histing T, Kristen A, Scheuer C, Menger MD, Pohlemann T.
310 Advances in the establishment of defined mouse models for the study of fracture healing and
311 bone regeneration. J Orthop Trauma. 2009;23,S31-8.

312 8. Harrison LJ, Cunningham JL, Strömberg L, Goodship AE. Controlled induction of a
313 pseudoarthrosis: a study using a rodent model. J Orthop Trauma. 2003;17,11-21.

314 9. Mark H, Bergholm J, Nilsson A. An external fixation method and device to study fracture
315 healing in rats. Acta Orthop Scand 2003;74,476–482.

316 10. Willie B, Adkins K, Zheng X. Mechanical characterization of external fixator stiffness for a
317 rat femoral fracture model. J Orthop Res 2009;27,687–693.

318 11. Glatt V, Evans C, Matthys R. Design, characterisation and in vivo testing of a new,
319 adjustable stiffness, external fixator for the rat femur. European Cells and Materials. 2012;
320 12,289-299.

321 12. Glatt V, Bartnikowski N, Quirk N, Schuetz M, Evans C. Reverse Dynamization: Influence
322 of Fixator Stiffness on the Mode and Efficiency of Large-Bone-Defect Healing at Different
323 Doses of rhBMP-2. J Bone Joint Surg Am. 2016;98,677-87.

324 13. Ho CY, Sanghani A, Hua J, Coathup M, Blunn G. Mesenchymal stem cells with
325 increased stromal derived factor 1 expression enhance fracture healing. Tissue Eng Part A.
326 2015;21, 594-602.

327 14. Lee O, Coathup M, Gooip A, Blunn G. Use of mesenchymal stem cells to facilitate bone
328 regeneration in normal and chemotherapy treated rats. Tissue Eng. 2005;11,1727-1735.

329 15. Kaspar K, Schell H, Toben D, Matziolis G, Bail HJ. An easily reproducible and
330 biomechanically standardized model to investigate bone healing in rats, using external
331 fixation. Biomed Tech. 2007;52,383-390.

16. Strube P, Mehta M, Putzier M, Matziolis G, Perka C, Duda GN. A new device to control mechanical environment in bone defect healing in rats. J Biomech. 2008;41,2696-2702.

17. Moazen M, Mak J, Jones AC et al. Evaluation of a new approach for modelling the screw-bone interface in a locking plate fixation: a corroboration study. J Eng in Med Part H. 2013; 227(7),746-756.

18. Macleod A, Panka P, Simpson A. Does screw-bone interface modelling matter in finite element analyses? J Biomech. 2012;45,1712-1716.

19. Wieding J, Souffrant R, Fritsche A. Finite element analysis of osteosynthesis screw fixation in the bone stock: An appropriate method for automatic screw modelling. PLoS One. 2012;7,1371.

20. Tufnol Composites Limited. 2016. Carp Brand Tufnol [online]: http://www.tufnol.com/materials-full/fabric_laminates/carp-brand.aspx

21. ANSYS® Academic Research, Release 16.2. Engineering data, material properties, ANSYS, Inc.

22. AZO Materials. No date available. Supplier Data - Polyetheretherketone (PEEK) (Goodfellow) and Carbon - Graphite Materials [online].

23. Mark H, Nilsson A, Nannmark U. Effects of fracture fixation stability on ossification in healing fractures. Clin Orthop Relat Res 2004 419:245–250.

24. Mark H, Rydevik B. Torsional stiffness in healing fractures: influence of ossification: an experimental study in rats. Acta Orthop 2005;76,428–433.

25. Schoen M, Rotter R, Schattner S. Introduction of a new interlocked intramedullary nailing device for stabilization of critically sized femoral defects in the rat. J Orthop Res 2008; 26,184–189.

26. Ramlee MH, Kadir MR, Murali MR, Kamarul T. Biomechanical evaluation of two commonly used external fixators in the treatment of open subtalar dislocation-a finite element analysis. Med Eng Phys. 2014;36,1358-1366.

27. Varga P, Zysset PK, Schefzig P, Unger E, Mayr W, Erhart J. A finite element analysis of two novel screw designs for scaphoid waist fractures. Med Eng Phys. 2016;38,131-139.

28. Wang Y, Wong DW, Zhang M. Computational models of the foot and ankle for pathomechanics and clinical applications: a review. Ann Biomed Eng. 2016 Jan;44(1),213-221.

29. Sato M, Yasui N, Nakase T, Kawahata H, Sugimoto M, Hirota S, Kitamura Y, Nomura S, Ochi T. Expression of bone matrix proteins mRNA during distraction osteogenesis. J Bone Miner Res 1998;13,1221-1231.

30. Sato M, Ochi T, Nakase T, Hirota S, Kitamura Y, Nomura S, Yasui N. Mechanical tension-stress induces expression of bone morphogenetic protein (BMP)- 2 and BMP-4, but not BMP-6, BMP-7, and GDF-5 mRNA, during distraction osteogenesis. J Bone Miner Res 1999;14,1084-1095.

31. Seebach C, Skripitz R, Andreassen TT, Aspenberg P. Intermittent parathyroid hormone (1-34) enhances mechanical strength and density of new bone after distraction osteogenesis in rats. J Orthop Res 2004;22,472-478.

Figure Legends

Figure 1. Computer aided designs of both external fixator models, with arrows demonstrating load constraint conditions.

Figure 2A and B. Demonstrating the torsional and axial stiffness' of both external fixators in vitro and in silico.

Figure 3. Demonstrating the comparative stiffness ratios in torsion and compression for in vitro and in silico testing.

Figure 4A and B. Demonstrating total fracture movement as found in silico under compression (A) and torsion (B).

Figure 5. Equivalent von-Mises stress contour plots on the crossbars of both fixator models.

Fig 1

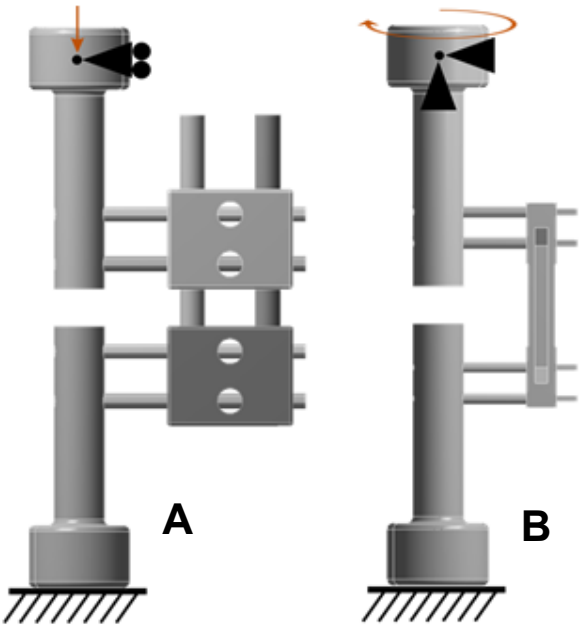
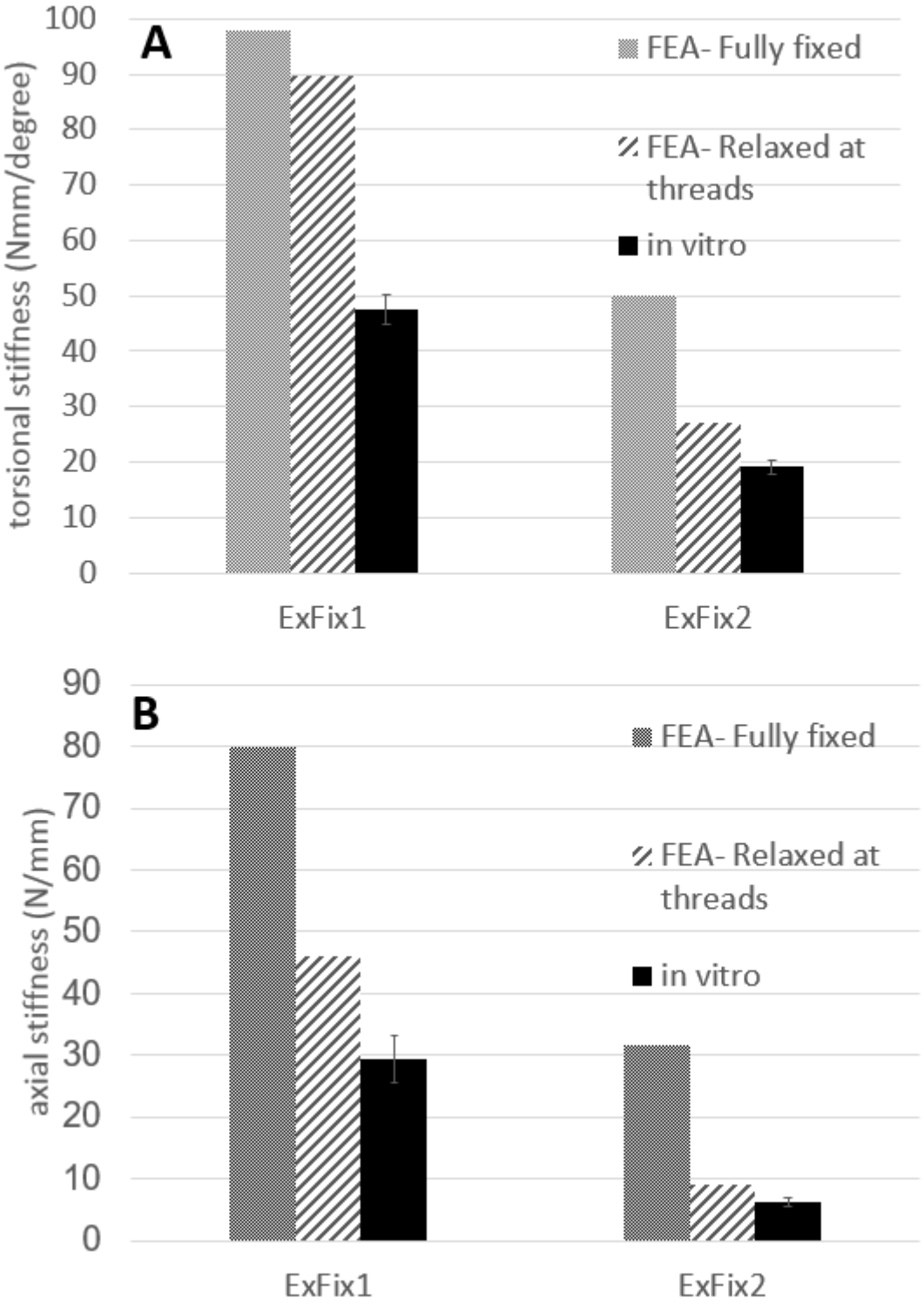


Fig 2



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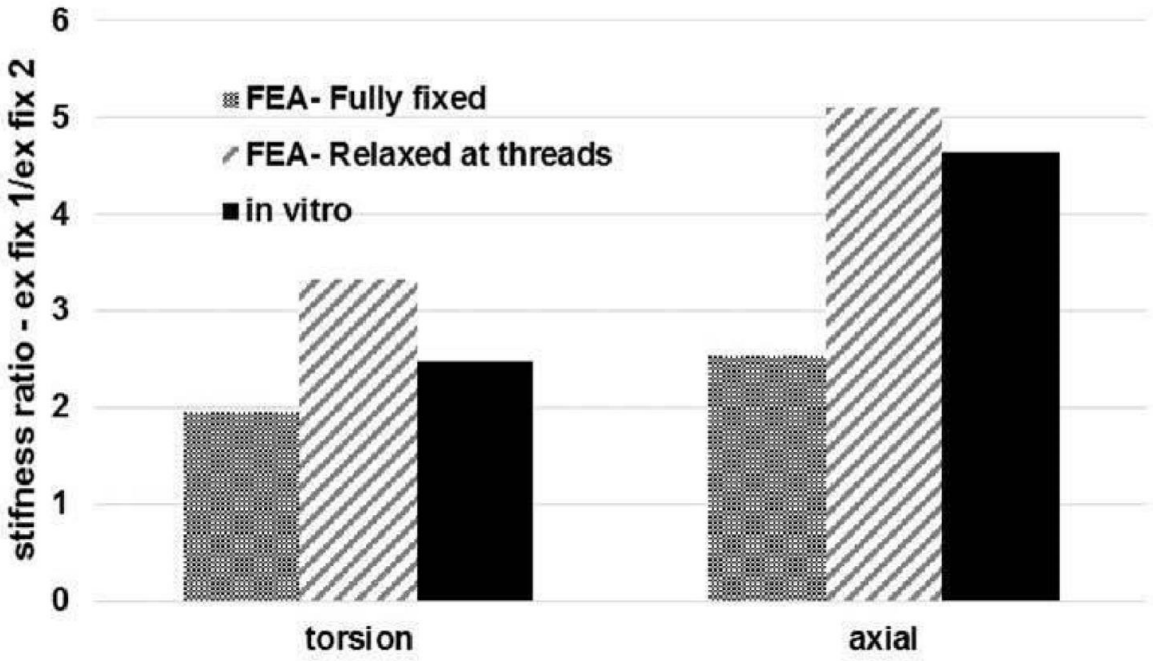

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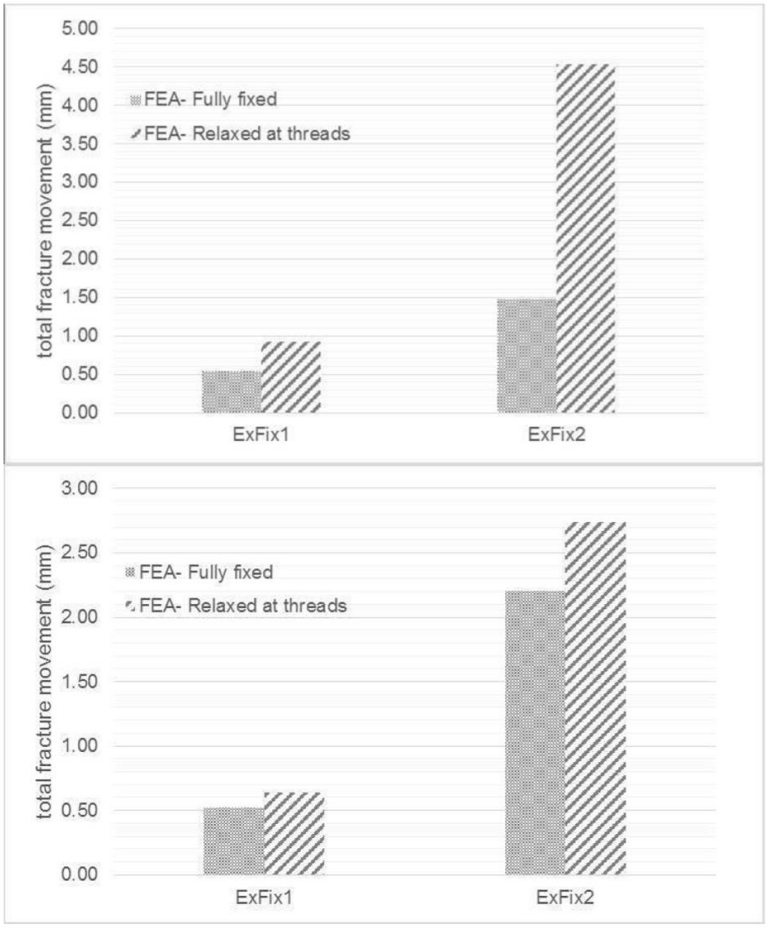

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