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**What Are the Biomechanical Properties of the Taylor Spatial Frame™?**

Running title: Taylor Spatial Frame™ Mechanical Stability

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**Abstract**

Background The Taylor Spatial Frame ™ (TSF) is a versatile variant on the traditional Ilizarov circular fixator. Although in widespread use, little comparative data exist to quantify the biomechanical effect of substituting the tried-and-tested Ilizarov construct for the TSF hexapod system.

Questions/purposes This study was designed to investigate the mechanical properties of the TSF system under physiologic loads, with and without the addition of a simulated bone model, with comparison to the standard Ilizarov frame.

Methods The mechanical behaviors of three identical four-ring TSF and Ilizarov constructs were tested under levels of axial compression, bending, and rotational torque to simulate loading during normal gait. An acrylic-pipe fracture model subsequently was mounted, using fine wires and 5 mm half pins, and the testing was repeated. Load-deformation curves, and so rigidity, for each construct were calculated, with statistical comparisons performed using paired t-tests.

Results Under axial loading, the TSF was found to be less rigid than the Ilizarov frame (645 ± 57 N/mm versus 1269 ± 256 N/mm; mean difference, 623 N/mm; 95% CI, 438.3-808.5 N/mm; $p < 0.001$), but more rigid under bending and torsional loads (bending: 42 ± 9 Nm/degree versus 78 ± 13 Nm/degree; mean difference, 37 Nm/degree; 95% CI, 25.0–47.9 Nm/degree; $p < 0.001$; torsion: 16 ±2 Nm/degree versus 5 ± 0.35 Nm/degree; mean difference, 11 Nm/degree; 95% CI, 9.5–12.2 Nm/degree; $p < 0.001$). On mounting the bone models, these relationships broadly remained in the half-pin and fine-wire groups, however the half-pin constructs were universally more rigid than those using fine wires. This effect resulted in the TSF, using half pins, showing no difference in axial rigidity to the fine-wire Ilizarov (107 ± 3 N/mm versus 107 ± 4 N/mm; mean difference, 0.05 N/mm; 95% CI, -6.99

to 7.1 N/mm; $p > 0.999$), while retaining greater bending and torsional rigidity. Throughout testing, a small amount of laxity was observed in the TSF construct on either side of neutral loading, amounting to 0.72 mm (± 0.37 mm) for change in loading between -10 N and 10 N axial load, and which persisted with the addition of the synthetic fracture model.

Conclusions This study broadly shows the TSF construct to generate lower axial rigidity, but greater bending and torsional rigidity, when compared with the Ilizarov frame, under physiologic loads. The anecdotally described laxity in the TSF hexapod strut system was shown in vitro, but only at low levels of loading around neutral. It also was shown that the increased stiffness generated by use of half pins produced a TSF construct replicating the axial rigidity of a fine-wire Ilizarov frame, for which much evidence of good clinical and radiologic outcomes exist, while providing greater rigidity, and so improved resistance, to potentially detrimental bending and rotational shear loads.

Clinical Relevance If replicated in the clinical setting, these findings suggest that when using the TSF, care should be taken to minimize the observed laxity around neutral with appropriate preloading of the construct, but that its use may produce constructs better able to resist bending and torsional loading, although with lower axial rigidity. Use of half pins in a TSF construct however may replicate the axial mechanical behavior of an Ilizarov construct, which is thought to be conducive to bone healing.

## Introduction

It is recognized that the biomechanical environment at a fracture or osteotomy site is one of the key factors in the process of bone healing [16]. One of the more powerful tools at the surgeon's disposal in influencing this environment is circular external fixation using the Ilizarov technique. The use of modular circumferential rings connected by longitudinal treaded rods, with an almost limitless combination of bone-fixation fine wires or half pins, allows each frame construct to be tailored to each patient's individual needs. When applied appropriately, there is much evidence that circular fixation generates beneficial levels of axial micromotion during weightbearing while limiting deleterious shear strain [14, 17, 23, 26]. This combination is thought to generate an osteogenic mechanical environment [8, 10, 11, 17, 36], and the biomechanical theory has been supported by clinical results [12, 33].

More recently, substitution of the threaded rods of the traditional Ilizarov frame for a hexapod system of six telescopic struts at the focal level, like in the Taylor Spatial Frame™ (TSF) (Smith & Nephew Inc, Memphis, TN, USA), has allowed more versatile application of circular fixation. This arrangement allows manipulation of a fracture or osteotomy site with six degrees of freedom by differentially altering the length of the six struts. This allows simultaneous correction of multiplane deformity and distraction and compression in a much simpler manner than would be possible using traditional Ilizarov equipment and with several reports of good clinical and functional outcomes after such treatment [1, 19, 35]. Numerous studies exist in the literature investigating and describing the potential effects of altering any one of the modular parts of a traditional Ilizarov circular frame on the biomechanical environment at a fracture site [2, 20, 23, 24, 27], however there are almost no studies examining the biomechanical behavior of a hexapod construct [21] and none, to our knowledge, directly investigating the potential effect that substituting the longitudinal rods of an Ilizarov fixator for the six struts of a hexapod may have on the overall behavior of a fixator construct, and therefore on the mechanical environment of a fracture site. Given the existing body of evidence comparing in vitro biomechanical behavior of circular frames with clinical outcomes, and the ever-expanding understanding of the complexities of bone healing biomechanics, a biomechanical comparison was planned to better understand the behavior of the TSF in isolation, but also to see whether any observed differences in the behavior of the frame components in isolation persisted once the effect of bone-fixation elements in the simulated fracture model were added, and therefore whether any clinically important effect was likely to result from substitution of a TSF for a traditional Ilizarov frame.

This biomechanical study therefore was designed to answer the following research questions: (1) What are the mechanical load-deformation characteristics of the TSF construct alone, and how does this compare with the standard Ilizarov frame under physiologic levels of axial compression, axial torsion, and bending loads? (2) How are these characteristics altered when tested with the addition of mounted bone models using fine wires or half pins?

**Materials and Methods**

An experimental biomechanical study was designed testing Ilizarov and TSF constructs of identical dimensions alone and with a bone model mounted using fine wires and half pins. An acrylic tube with an outer diameter of 32 mm and a 4-mm wall (Clear Plastic Supplies, Chesterfield, UK) was used as a mechanical substitute for bone. This symmetric, uniform model was chosen to minimize any variability between testing cycles that may be caused by minor differences in wire placement in more anatomic models or by bone density variation in cadaveric samples, a common practice in similar mechanical studies [20, 24, 25, 27, 28, 30, 31]. While not attempting to reproduce a directly clinically comparable environment, the reproducibility of such mechanical modeling allows greater confidence in reporting observed differences with small sample sizes, a finding confirmed by the low variance observed in preliminary testing and which was used to guide the sample size for this study.

Three identical four-ring frames for each of the six configurations tested were constructed according to manufacturers' specifications using 155-mm aluminum TSF rings (Smith & Nephew Inc). These were assembled with 50-mm spacing between common segment rings, spanned by four 6-mm threaded rods, and 175-mm spacing across the unstable focal segment. This was spanned with six medium TSF FAST-FX$^{TM}$ struts (Smith & Nephew Inc) for the TSF constructs or four 6-mm threaded rods for the Ilizarov constructs (Fig. 1).

After testing the frame elements alone, bone models were mounted using new clinical standard 1.8-mm smooth wires tensioned to 130 kg or predrilled self-tapping 5-mm half pins. To maximize reproducibility and isolate the effect of the frame construct, in keeping with previous studies [6, 20, 27, 31], bone models were mounted centrally in each ring with fixation elements placed at theoretically "ideal" crossing angles of 90°. To prevent bone apposition during loading, simulating an entirely unstable fracture or corticotomy, bone models were mounted with a 20-mm fracture gap.

Constructs were mounted in the testing apparatus (Tinius Olsen H25K-S UTM; Tinius Olsen Inc, Horsham, PA; and Uniaxial manual torsion testing machine; University of Leeds, Leeds, UK) with bespoke mounting jigs allowing rigid fixation of the frame constructs, and bone substitutes, to the apparatus (Fig. 2). Constructs were tested separately under axial loading, AP bending, and axial torsion up to 700 N, 20 Nm, and 20 Nm, respectively, at loading rates of 6 mm/minute/1.01°/minute and with data being collected only after three preconditioning cycles of full loading. Data were recorded for three testing cycles for each construct configuration. Physiologic loading was considered to consist of 500 N axial loading, 20 Nm bending, and 5 Nm torsional load, being analogous to loads shown to be supported by a tibial frame during normal gait at 30 days postoperatively [8, 32].

Data were collected and collated using Microsoft® Excel® (Microsoft Corporation, Redmond, WA, USA) with data sets transferred for graph plotting and statistical analysis to Graph Pad Prism® (Version 6; GraphPad Software, Inc, La Jolla, CA, USA). Load deformation curves were created for each construct configuration and loading regime for the full range of loading. Nonlinear regression analysis then was performed to determine the construct rigidity, defined as the mean slope of the linearly elastic portion of the load deformation curve. Rigidity in this context refers to the mechanical load-deformation properties of a complex structure such as a circular frame, as opposed to stiffness, being a property of a uniform material. Analyses were done to examine for statistically significant differences in rigidity between constructs. The data met assumptions for parametric testing using the D'Agostino and Pearson omnibus test and QQ-plot analysis and therefore Student's t-test and ANOVA testing with post hoc analysis using Tukey's method were performed. This methodology is validated in small sample-size studies and commonly applied to such mechanical modeling studies where conditions are almost entirely controlled and comparison between two samples, with low variance, is required [7, 20, 25]. While recognizing the potential for type II error with small sample sizes, a p value less than 0.05 was considered statistically significant throughout.

## Results

### Frame Elements Alone

Axial loading of the Ilizarov frame in isolation produced a mean rigidity of 1269 N/mm (± 256 N/mm) for the linear portion of its load-deformation curve (Fig. 3A). By contrast, the TSF showed a lower mean rigidity on axial loading of 645 N/mm (± 57 N/mm; mean difference, 623 N/mm; 95% CI, 438.3-808.5 N/mm; $p < 0.001$) (Fig. 4A). Additionally, while loading of the Ilizarov produced a largely linear trend in deformation, axial loading of the TSF produced a marked initial "toe" region of increasing rigidity with increased loading around neutral, representing a mean deformation of 0.72 mm (± 0.37 mm) for a change in load from 10 N distraction to 10 N compression, before a linear relationship returned.

Under bending loads, the Ilizarov frame produced a mean rigidity of 42 Nm/degree (± 9 Nm/degree), less rigid than the TSF at 78 Nm/degree (± 13 Nm/degree; mean difference, 37 Nm/degree; 95% CI, 25.0–47.9 Nm/degree; $p < 0.001$) (Fig. 4B). Once again, and again in contrast to the Ilizarov, the load-deformation curve of the TSF generated a marked toe region, representing 0.5° (± 0.16°) deformation from -5 to 5 Nm loading (Fig. 3B).

On torsional testing, the TSF construct had a rigidity of 16 Nm/degree (± 2 Nm/degree), higher than that of the Ilizarov at 5 Nm/degree (± 0.35 Nm/degree; mean difference, 11 Nm/degree; 95% CI, 9.5–12.2 Nm/degree; $p < 0.001$) (Fig. 4C). Again there was an initial nonlinear toe region to the load deformation curve with 10 Nm loading around neutral producing 2° (± -0.13°) deformation from -5 to 5 Nm (Fig. 3C).

Bone-frame Constructs

Axial loading of the all fine-wire constructs showed the fine-wire Ilizarov frame to be slightly more rigid than the fine-wire TSF (107 ± 4 N/mm versus 100 ± 1.7 N/mm; mean difference, 7 N/mm; 95% CI, 0.09–14.2 N/mm; $p = 0.047$). Both constructs produced nonlinear load-displacement curves, with initially increasing construct rigidity followed by more linear behavior (Fig. 5A). In both constructs a toe region of laxity in the load deformation curves was observed between -10 N and 10 N loading with no difference seen, with the numbers available, between the two designs (0.7 ± 0.03 mm versus 0.5 ± 0.04 mm; mean difference, 0.17 mm; 95% CI, -0.07 to 0.41 mm; $p = 0.178$) (Table 1). Axial loading of half-pin constructs produced similar results with the half-pin TSF less rigid at 107 N/mm (± 3 N/mm) than the half-pin Ilizarov at 120 N/mm (± 0.7 N/mm; mean difference, 13 N/mm; 95% CI, 6.1–19.7 N/mm; $p = 0.001$), although with more linear plots observed for both.

Notably, however, the half-pin TSF construct showed no difference in rigidity from the fine-wire Ilizarov, with the numbers available, in the linear phase of loading (107 ± 3 N/mm versus 107 ± 4 N/mm; mean difference, 0.05 N/mm; 95% CI, -6.99 to 7.1 N/mm; $p > 0.999$). Again there was a demonstrable initial toe phase of laxity in the half-pin TSF construct with a mean deformation of 0.5 mm (± 0.08 mm) for 10 N around neutral loading, greater than that observed in the half-pin Ilizarov at 0.25 mm (± 0.16 mm; mean difference, 0.25 mm; 95% CI, 0.01–0.48 mm; $p = 0.042$). In the fine-wire TSF and Ilizarov, however, there was no difference in the toe-phase observed with the numbers available (0.7 ± 0.03 mm versus 0.5 ± 0.04 mm; mean difference, 0.17 mm; 95% CI, -0.07 to 0.41 mm; $p = 0.178$).

Application of bending forces to the half-pin constructs produced similar results to the frames tested in isolation, with the TSF more rigid at 13 Nm/degree (± 0.2 Nm/degree) than the half-pin Ilizarov at 12 Nm/degree (± 0.1 Nm/degree; mean difference, 0.6 Nm/degree; 95% CI, 0.1–1.1 Nm/degree; $p = 0.013$). Again, a higher degree of toe-phase laxity was observed for 5-Nm loading around neutral (1.7° ± 0.01° versus 1.1° ± 0.03°; mean difference, 0.7°; 95% CI, 0.3°–1.0°; $p = 0.001$) (Fig. 5B). In contrast, bending loading of the fine-wire constructs reversed the relationship seen in the frames alone, with lower rigidity observed in the fine-wire TSF at 4 Nm/degree (± 0.2 Nm/degree) than the fine-wire Ilizarov construct at 5 Nm/degree (± 0.2 Nm/degree; mean difference, 1.2 Nm/degree; 95% CI, 0.7–1.6 Nm/degree; $p < 0.001$). This bending testing of the fine-wire constructs did show unexpectedly high levels of deformation under physiologic loads, particularly in the TSF construct, with an element of deformation occurring through sliding of the bone models along the smooth fine wires.

Under torsional loads, the fine-wire and half-pin constructs behaved similarly to frames tested in isolation. The fine-wire and half-pin TSF constructs were more rigid at 1.5

Nm/degree (± 0.02 Nm/degree) and 2 Nm/degree (± 0.02 Nm/degree), respectively, than the comparable Ilizarov constructs at 1.2 Nm/degree (± 0.06 Nm/degree; mean difference, 0.3 Nm/degree; 95% CI, 0.19–0.38 Nm/degree; $p < 0.001$) and 1.7 Nm/degree (± 0.04 Nm/degree; mean difference, 0.5 Nm/degree; 95% CI, 0.39–0.58 Nm/degree; $p < 0.001$)(Fig. 5C).

**Discussion**

The widespread uptake of hexapod systems such as the TSF with use of circular-frame constructs has simplified ring-fixation techniques in complex clinical scenarios, allowing successful management in cases that previously might have been considered highly technically challenging [4, 9, 35]. The majority of clinical and preclinical evidence regarding use and outcomes of circular external fixation is based on traditional Ilizarov equipment with some prior studies characterizing the effect of altering aspects of Ilizarov construct design on fracture mechanics [6, 11, 23, 28]. This knowledge, along with an understanding of the fracture-site mechanical environment and bone healing, allows a surgeon to tailor construction of a frame to suit the particular clinical scenario and therefore it is vital to understand what effect exchanging hexapod struts for threaded rods might have on the mechanical behavior of a circular frame construct so that this may be taken into account during preoperative planning. This mechanical study showed that the TSF hexapod frame is less rigid than the Ilizarov frame under axial loading, but more rigid under bending and torsional loading, however these differences became far less, if at all, apparent when tested as part of a frame construct with fine-wire or half-pin bone fixation elements included. The TSF showed an increased element of laxity around neutral loading in all planes compared with the Ilizarov, a difference which became far less apparent in the context of the fine-wire constructs, which were universally less rigid than those with half pins.

As an experimental biomechanical study, this study has certain limitations that must be taken into account in the interpretation and clinical translation of the results. As far as possible, the experimental design aimed to limit the potential variables between frames to focus specifically on the fundamental mechanical behavior of the TSF and Ilizarov frames and optimize reproducibility between testing cycles and constructs. Given this, it is not intended to consider the multitude of possible combinations in which a frame may be used, and therefore is not a clinical comparison of the Ilizarov versus the TSF. Each clinical case is different and the correct frame construct design remains up to the treating clinician, for example attempting to place a frame that seeks to minimize shear strain in an unstable transverse tibial fracture. The intention of this study is to help inform preoperative planning by providing a clearer understanding of what effect use of a TSF in place of an Ilizarov frame might have on the overall construct rigidity under physiologic loading conditions. Acrylic pipe as a bone model, while not replicating the geometry or anatomy of bone, was chosen as a symmetric, uniform and rigid analog, allowing simulation of bicortical fixation and stress distribution of wires and pins, while not influencing the comparative mechanical behavior of the constructs [5]. The advantage of this uniformity, when used with idealized fine-wire and half-pin crossing angles, lies in the reproducibility of the construction and loading of each frame so minimizing variability stemming from small differences in wire placement, contact area, or plane of loading between testing cycles and constructs. However, such simplified wire placements, particularly with smooth wires alone, may generate behavior that does not directly replicate the clinical scenario, as seen under bending loads. Nonetheless, this approach, with the resulting low levels of variability we observed between samples, allows greater confidence in analyzing and drawing conclusions from the small sample size used in this study, a commonplace practice in such mechanical studies where conditions are almost entirely controlled and variability is  low [20, 25, 27, 31]. Small sample sizes also may create difficulty in statistical analysis, particularly in the clinical scenario; however, in a comparative mechanical analysis such as this, with a highly controlled testing environment and almost identical samples, simple comparative statistics, and specifically the t test, which has been shown to be accurate in small sample analysis, allows comparison to be made between two different samples each with their own mean and distribution of results to determine if significant variability may exist [7]. Nonetheless, type II error is a concern, and p values not substantially lower than 0.05 should be regarded with circumspection.

Frame Elements Alone

Direct loading of the frame elements alone generated results broadly as might be expected from mechanical theory with cylindrical elements best resisting deformation when loaded along their long axis. In this way, it was seen that on axial loading, the Ilizarov threaded rods produced an extremely rigid construct, whereas axial loading of the TSF, with struts oblique to the direction of loading undergoing bending rather than purely axial loading, was less able to resist deformation, generating 0.8 mm (± -0.1 mm) and 1.7 mm (± 0.3 mm) displacement at 700-N loading, respectively [18]. Conversely, loading of the obliquely oriented TSF struts under bending and torsional loads resulted in superior resistance to deformation compared with the vertical Ilizarov rods. Throughout testing of the TSF, but most markedly during testing of the frame alone, an element of laxity or “play” was identifiable in the universal joints of the TSF struts. This is a factor briefly mentioned by Henderson et al. [21] but not investigated nor previously quantified elsewhere, to our knowledge. This resulted in motion at low levels of loading, just less than 1 mm of displacement for 10 N axial distraction to 10 N compression, or 0.5° angulation for -5 Nm to 5 Nm bending load [21].

Bone-frame Constructs

The rigidity of a circular frame construct is a function of the rigidity of the frame and the stiffness of the bone fixation elements used. Numerous published studies describe the behavior of bone-fixation elements in circular frames under loading [2, 14, 25, 27, 37]. For example, Orbay et al. [27] described the rigidity of a two-wire, single-ring construct on axial loading at approximately 65 N/mm. Considering these findings, it is unsurprising that the bone-frame constructs we tested showed far lower resistance to deformation than the frame elements tested in isolation. The pattern of results observed was broadly similar to those obtained when testing the frames alone, although with less marked differences between the Ilizarov and TSF constructs. Likewise, the increased rigidity and more-linear load deformation properties of half-pin compared with fine-wire constructs, as seen throughout our testing, is well recognized [15, 20, 24]. Axial loading of bone-frame constructs followed these recognized trends, with half-pin constructs of the Ilizarov and TSF frames showing greater rigidity than with use of fine wires, and Ilizarov constructs proving more rigid than the TSF when comparing constructs using the same bone-fixation element. Interestingly, these two phenomena overlapped when comparing a half-pin TSF with a fine-wire Ilizarov frame, where the additional stiffness generated by use of 5-mm half pins increased the overall rigidity of the TSF construct to levels not different from those seen in a traditional fine-wire Ilizarov frame.

Under torsional loads, predictable mechanical behavior was observed with half-pin constructs more rigid than fine wire and the oblique struts of the TSF leading to less deformation than longitudinal Ilizarov rods when using the same bone-fixation elements. In contrast, bending load testing of the bone-frame constructs produced some unexpected results. Half-pin constructs were once again more rigid than fine wire and the TSF half-pin construct was, as expected, more rigid than the Ilizarov construct. However, the load deformation curve produced on applying a bending force to the fine-wire TSF was less rigid than that of the equivalent Ilizarov frame. This is the opposite result to that expected from testing of the frames alone. During these tests, shear was observed at the fracture site generated by movement of the bone model along the tensioned fine wires, producing increased deformation for a given bending load. Such wire slippage, although anecdotally occasionally reported clinically, may be exaggerated as a function of the idealized wire-crossing angles and simplified model. It may be hypothesized that the increased rigidity of the TSF frame compared with the Ilizarov may transfer greater load to the bone-wire interface, therefore producing more-exaggerated slip of the bone model on the tensioned wires, generating greater deformation. However, this phenomenon was described in biomechanical testing from this unit using purely Ilizarov materials, the effect being obliterated by addition of certain configurations of half pins to the constructs [20]. The limitations of the experimental setup must be considered when interpreting this finding and there may be a place for further investigation in future studies, particularly using more clinically relevant constructs and potentially the effect of using olive wires.

Throughout testing, an initial nonlinear toe region to the load deformation curves obtained when testing the TSF constructs was observed. This likely is caused by the slight laxity in the universal joints of the TSF struts, increasing deformation around neutral loading for the TSF beyond that observed with equivalent Ilizarov models. Fine-wire Ilizarov frames, however, also undergo nonlinear initial deformation as a result of self-stiffening of the wires on loading, and when comparing all fine-wire Ilizarov and TSF constructs, no difference was observed between overall behavior at these low loads, suggesting that self-stiffening of the wires is more important that the effect of TSF strut laxity, the effect of which is masked. This was the case under axial, bending, and torsional loading [6, 23].

## Clinical Relevance

This study was designed to highlight the mechanical differences in behavior under loading between the traditional Ilizarov ring fixator and the hexapod system of the TSF. To comment meaningfully how this knowledge may be applied to clinical practice, a current understanding of the biomechanics of bone healing must be considered. There is general consensus that a certain level of axial strain is desirable and necessary to stimulate bone healing with, among others, Kenwright and Goodship [22], as early as 1989, reporting increased callus mineralization and fracture stiffness in ovine tibial fractures with approximately 16% axial strain compared with more rigid fixation, although this was seen to deteriorate in quality somewhat with increased strains of up to 66% [10, 13, 17, 22, 34, 36]. Likewise, although there is less agreement on this, it generally is considered that shear strain, whether linear or rotational, is detrimental to bone healing and should be limited where possible [3, 8, 10, 29, 36]. Relating this to the current study, given the nonclinically representative nature of the models tested and infinite variability of possible fracture patterns, it is not possible to extrapolate the precise change in mechanical behavior that would be produced at a specific fracture site by use of a TSF. However, even if the magnitudes of differences observed in the current study are considered in light of the previously discussed studies, it may be questioned whether those statistical differences would be likely to represent clinically important ones. Differences in axial rigidity between the TSF and Ilizarov constructs translated only to, at most, 4% increased strain under 500 N loading, and less than 1% strain for bending and torsional loading in the current model. What has been broadly shown, however, is that the TSF system is less axially rigid than the Ilizarov frame, but more rigid under bending and torsional loads, and it is this general concept that may be applied to clinical planning. Again considering the previous discussion, in cases where the clinician has concerns regarding control of rotational or shear strain, but not about levels of axial strain, the TSF may confer an advantage over the Ilizarov while providing the benefits of a hexapod system. This would seem particularly true with use of a half-pin TSF over a fine-wire Ilizarov, where no difference in axial rigidity was observed, but with improved resistance to rotational and bending loads.

This study also shows that the TSF hexapod includes an inherent degree of laxity in the strut joints for approximately 10 N axial loading and 5 Nm bending or torsion around neutral. Although not reported in previous studies, this is widely accepted to be the case, and anecdotally, we have observed that patients sometimes describe discomfort and a feeling of instability on initial loading, which is frequently demonstrable on examination. In our experience replacing the hexapod with threaded rods will almost always eliminate this, again confirming the effect to be inherent to the hexapod struts of the TSF as suggested by this study. This effect also appears to decrease in many patients once the TSF correction begins, likely owing to increasing loading on the frame and compression or soft tissue tension. The effect of this laxity on bone healing is even less clear, but given the very low levels of strain this equated to on testing, it is unlikely to be of clinical importance to bone healing.

## Conclusions

Therefore, as far as it is possible to draw direct conclusions from such a biomechanical study, and within the limitations previously outlined, the findings presented here would indicate that with use of the TSF system, a half-pin construct most closely replicates the tried-and-tested beneficial axial rigidity of an Ilizarov fine-wire frame, while further limiting potentially deleterious shear strain. Future studies might be designed to investigate this effect in more clinically relevant models and configurations, to help identify the specific clinical scenarios where use of a hexapod system may confer a biomechanical advantage and technical and practical ones.

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## Legends

**Fig. 1A-F** The six frame constructs tested were the **(A)** Ilizarov alone, **(B)** Taylor Spatial Frame$^{TM}$ (TSF) alone, **(C)** fine-wire Ilizarov, **(D)** fine-wire TSF, **(E)** half-pin Ilizarov, and **(F)** half-pin TSF.

**Fig. 2** A fine-wire Taylor Spatial Frame$^{TM}$ mounted in the axial testing apparatus (Tinius Olsen H25K-S UTM; Tinius Olsen Inc) is shown.

**Fig. 3A-C** Load deformation plots for testing of frame elements alone under **(A)** axial, **(B)** bending, and **(C)** torsional loads are shown. TSF = Taylor Spatial Frame$^{TM}$

**Fig. 4A-C** Box and whisker plots of stiffness data for Ilizarov and TSF elements alone, fine-wire, and half-pin constructs under **(A)** axial, **(B)** bending, and **(C)** torsional loading are shown. The boxes represent the 25$^{th}$ to 75$^{th}$ percentiles with whiskers from minimum to maximum values. The line through the middle of the box represents the median and the + the mean. TSF = Taylor Spatial Frame$^{TM}$

**Fig. 5A-C** Load deformation plots for testing of bone-frame constructs under **(A)** axial, **(B)** bending, and **(C)** torsional loads are shown. TSF = Taylor Spatial Frame$^{TM}$

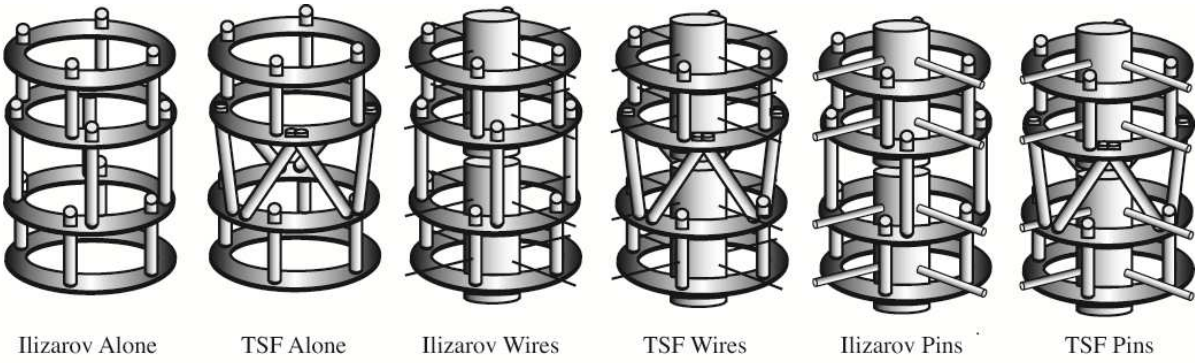


Figure 1. Frame constructs tested.

Figure 2. fine-wire TSF axial loading.

| | **Ilizarov wires** | | **Ilizarov Pins** | | **TSF wires** | | **TSF Pins** | |
|---|---|---|---|---|---|---|---|---|
| Axial loading (N) | Axial deformation (mm) | | Axial deformation (mm) | | Axial deformation (mm) | | Axial deformation (mm) | |
| -10 | -0.3 | +/-0.1 | -0.1 | +/-0.0 | -0.5 | +/-0.3 | -0.3 | +/-0.1 |
| 0 | 0.0 | +/-0.0 | 0.0 | +/-0.0 | 0.0 | +/-0.0 | 0.0 | +/-0.0 |
| 10 | 0.2 | +/-0.1 | 0.2 | +/-0.1 | 0.1 | +/-0.3 | 0.2 | +/-0.0 |
| 500 | 5.9 | +/-0.7 | 4.2 | +/-0.1 | 6.3 | +/-0.1 | 5.0 | +/-0.1 |
| Ridgidity (N/mm) | 106.9 | 3.8 | 119.8 | +/-0.7 | 99.8 | +/-1.7 | 106.9 | 3.3 |
| Bending loading (Nm) | Bending deformation (Deg) | | Bending deformation (Deg) | | Bending deformation (Deg) | | Bending deformation (Deg) | |
| -5 | -0.5 | +/-0.0 | -0.6 | +/-0.0 | -0.7 | +/-0.0 | -0.7 | +/-0.0 |
| 0 | 0.0 | +/-0.0 | 0.0 | +/-0.0 | 0.0 | +/-0.0 | 0.0 | +/-0.0 |
| 5 | 0.8 | +/-0.1 | 0.5 | +/-0.0 | 1.0 | +/-0.3 | 1.1 | +/-0.0 |
| 20 | 3.9 | +/-0.1 | 1.7 | +/-0.0 | 4.9 | +/-0.1 | 2.3 | +/-0.0 |
| Ridgidity (Nm/deg) | 4.95 | +/-0.2 | 12.2 | +/-0.1 | 3.79 | +/-0.2 | 12.8 | +/-0.2 |
| Torsional loading (Nm) | Torsional deformation (Deg) | | Torsional deformation (Deg) | | Torsional deformation (Deg) | | Torsional deformation (Deg) | |
| -5 | -5.8 | +/-1.3 | -3.0 | +/-0.1 | -5.3 | +/-0.3 | -3.2 | +/-0.1 |
| 0 | 0.0 | +/-0.0 | 0.0 | +/-0.0 | 0.0 | +/-0.0 | 0.0 | +/-0.0 |
| 5 | 4.7 | +/-0.7 | 4.0 | +/-0.1 | 5.3 | +/-0.2 | 4.0 | +/-0.1 |
| Ridgidity (Nm/deg) | 1.2 | +/-0.1 | 1.7 | +/-0.0 | 1.5 | +/-0.0 | 2.2 | +/-0.0 |

Table 1.

Deformation and rigidity results for testing of frame - bone model constructs.

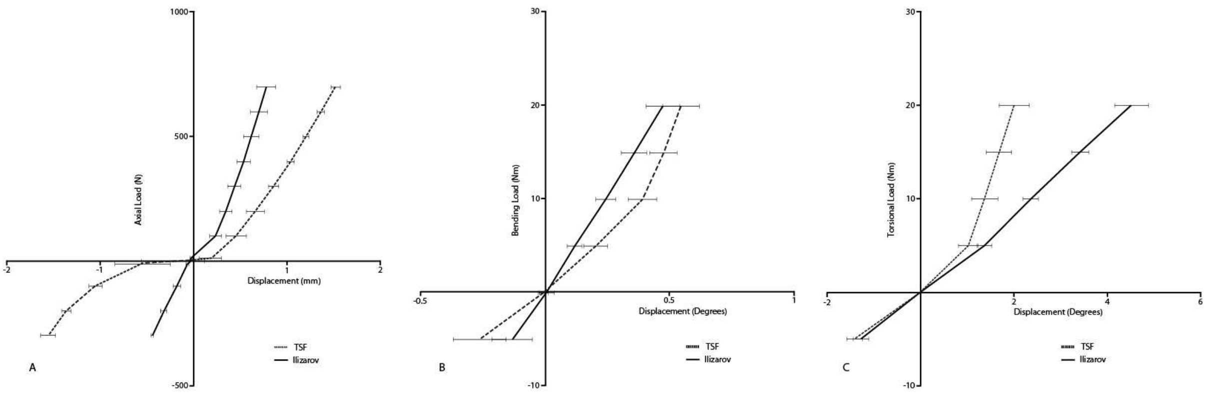


Figure 3. Load deformation plots for testing of frame elements alone under axial, bending and torsional loads.

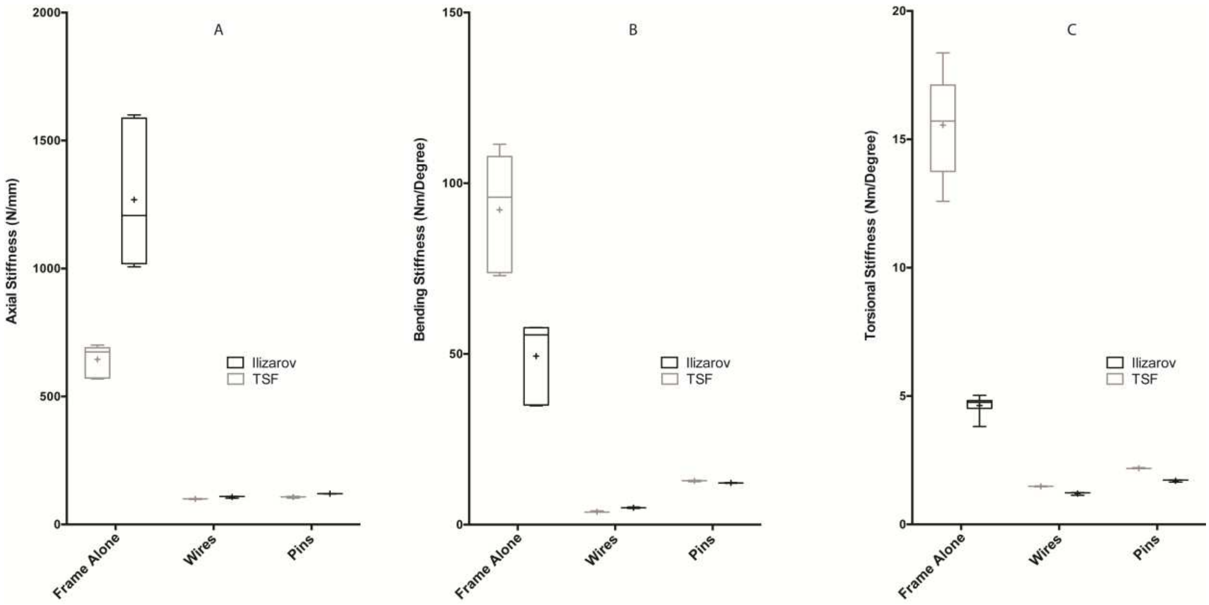


**Figure 4.** Box and whisker plot of stiffness data for Ilizarov and TSF; frame elements alone, fine-wire & half-pin constructs; under axial, bending and torsional loading. Boxes represent $25^{th}$ – $75^{th}$ percentiles with whiskers from minimum to maximum values. The line through the middle of the box represents the median and the +, the mean.

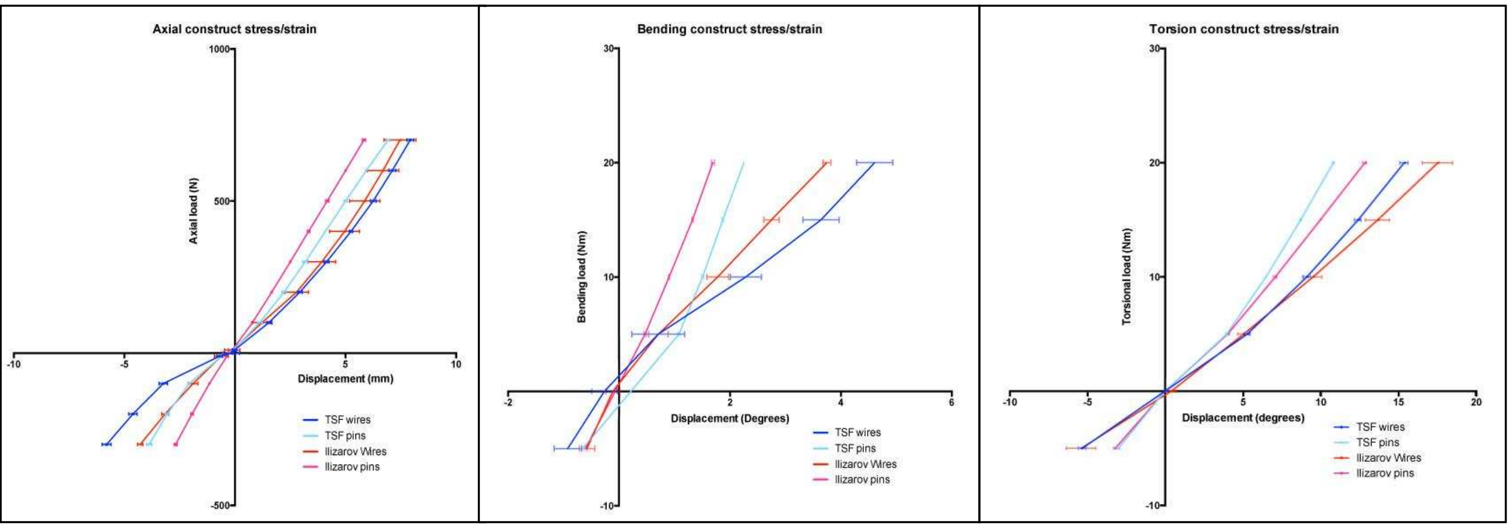


Figure 5. Load-deformation plots for testing of bone-frame constructs under axial, bending and torsional loads.