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Clinical Biomechanics 13 (1998) 441–448

CLINICAL BIOMECHANICS

Stability of external circular fixation: a multi-variable biomechanical analysis

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Received 13 December 1996; accepted 14 January 1998

Abstract

Objective. To determine how the manipulation of the parameters of fixation and components of the circular external frame could improve and maintain optimal stability of bone fragments.

Design. We performed a multi-parametric biomechanical analysis of the extrinsic parameters effecting bone fragment stabilization. Results of testing are presented as a percent change in stiffness due to the manipulation of frame components and their interaction with other fixation parameters.

Background. Although there have been investigations of the biomechanical characteristics of circular external fixation, they have been limited to either individual frame components or full frame comparisons. Therefore, these studies did not provide a comprehensive understanding of how the manipulation of circular fixator components influences bone fragment stability.

Methods. Mechanical testing was performed in three phases examining the effect of numerous components including ring diameter, wire angle, ring separation, etc. on axial, torsional and bending stiffness.

Results. For phase I (single ring) and phase II (double-ring block), ring diameter was the most significant factor affecting axial and torsional stiffness, while wire angle, ring separation, and their interaction had the most influence on bending stiffness. Phase III (two double-ring blocks) showed that ring positioning with respect to the osteotomy site had the most affect on bending and torsional stiffness while axial stiffness was non-linear and dependent upon the applied load.

Conclusions. The stability of bone fragments within a circular external fixator is affected by manipulation of the parameters of fixation or individual components of the frame. The contribution of each component to overall bone fragment stability is dependent upon the mode of loading. The changes in overall stability of bone fragments are dependent not only on the individual frame components but also upon their interaction with other parameters of fixation.

Relevance

Understanding how the manipulation of individual frame components will affect overall bone fragment stabilization will allow the surgeon to better control the stability of bone fragments for each clinical situation.

Keywords: External circular fixation; Biomechanics; Stiffness; Bone fragment fixation; Orthopedics; Ilizarov

1. Introduction

Circular external fixation using thin tensioned wires has had increasing popularity over the past decade, gaining recognition for its advantages in fracture healing, limb lengthening and deformity correction. Those advantages include better tolerance of thin wires by bony and soft tissues, and stable, but not rigid, fixation of bone fragments promoting axial micromotion during weight bearing. Based on clinical experience and numerous biomechanical studies, the optimal stabilization of bone fragments within an external circular device is achieved through a two-level segmental fixation with three or four wires inserted at right angles [1,2]. In most clinical situations, however, wires can not be positioned at right angles due to anatomical limitations or complexity of bone deformity [3,4]. Reducing the number of wires or decreasing the

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PII: S0268-0033(98)00007-2

angle between wires may affect the bending stiffness of the bone fragment, therefore compromising the overall fixation. In these cases, bone stabilization can be improved by the manipulation of other parameters of fixation or frame components such as ring diameter, wire diameter, ring separation, etc.

Although there have been investigations of the biomechanical characteristics of circular external fixation [5–15], they have been limited to either individual frame components or full frame comparisons. The goal of this study was to approach the biomechanics of circular external fixation in a methodical fashion to determine the relative effect of not only individual frame components and specific parameters of fixation but their interactions on axial compression, torsional stiffness, and bending. The main hypothesis to be tested was that manipulation of individual components of the circular external frame could improve and maintain optimal stability of bone fragment fixation.

2. Methods

Mechanical testing was divided into three phases. Phase I investigated the effect of two cross-tensioned wires on stability of single-ring bone fragment fixation (Fig. 1 (A)). Components tested in this phase were ring diameter (RD): 120, 160, 200 mm; wire diameter (WD): 1.5, 1.8 mm; wire tension (WT): 90, 130 kg; and wire angle (WA): 30°, 45°, 90°. Phase II examined components of a double-ring block, which is a typical frame configuration for two-level segmental bone fragment fixation (Fig. 1 (B)). This phase tested ring diameter (RD): 120, 200 mm; ring separation (RS): 10, 200 mm; osteotomy site wire angle (OWA): 0°, 30°, 90°; joint site wire angle (JWA): 30°, 90°; and number of wires (WN): 3, 4. A 0° OWA represented a three-wire double-ring block with the single wire inserted in the M–L plane and bisecting the JWA. Loads were applied to the Delrin 100 mm from the ring with the OWA. Phase III studied the stability of two asymmetric-length bone fragments fixed to an external frame (Fig. 1 (C)). The fixator consisted of two double-ring blocks connected by four rods, which is a typical frame configuration used for fracture fixation, tibial lengthening, and deformity correction. This phase investigated the osteotomy site wire angle for both proximal (PWA) and distal (DWA) fragments: 0°, 30°, 45°, 90°; distal fragment ring separation (DRS): 10, 60, 110, 160 mm; and number of wires (WN): 6, 7, 8. In addition, two types of wire (olive, smooth), different ring materials (stainless steel, carbon composite, aluminum), and two types of connecting rods (threaded, telescopic) were examined in this phase.

Stainless steel (unless otherwise noted) Ilizarov external fixation components (Smith & Nephew Richards, Memphis, TN) were used. Solid 38 mm diameter Delrin cylinders, centrally located within the ring(s), were used as a bone model. K-wires were tensioned with a dynamometric wire tensioner (Smith & Nephew Richards, Memphis, TN). To ensure clinical applicability of the results, parameters were restricted to those typically encountered during the utilization of the device. Variables not found to have significant affect on bone fragment stability in one phase were eliminated from subsequent phases.

Stability of fixation was measured in axial compression (AC), torsion (T), anterior–posterior (A–P) bending, and medial–lateral (M–L) bending using a universal testing system (MTS 858, Minneapolis, MN). Loads were applied through the Delrin at a distance of 100 mm from the wire–bone interface(s). For phases I and II, the rings were held rigidly to the test system and loaded at a rate of 15 N s^{-1}, 375 N mm s^{-1} and 1.2 N s^{-1} for axial compression, torsion and bending, respectively. For phase III, axial and torsional loads were applied at a rate of 20 N s^{-1} and 375 N mm s^{-1}, respectively through the distal fragment with the proximal fragment held rigidly to the test system. Transverse bending was applied at a rate of 3 N s^{-1} in a four-point bending method, which generated a constant moment across the frame. Axial displacement of the bone fragments was measured using an extensometer (MTS, model 632.11B-26) positioned across the fracture gap. Torsional rotation was determined using two transducers (Celesco, PT-1-1-10A) positioned at

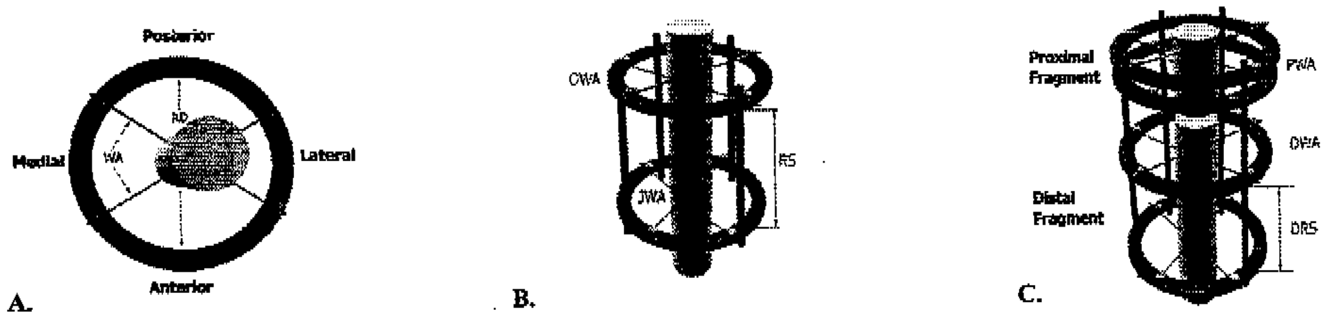

Fig. 1. Construct set-up for (A) phase I, (B) phase II, and (C) phase III.

the fracture gap. Transverse displacement of both fragments was analyzed with two linear potentiometers (Maurey Instr. Corp., M1326-1-6102) positioned at the fracture gap.

The load/displacement characteristics were plotted on an X–Y plotter (Soltec VP-6424S) and analyzed to determine the stiffness in all four loading modes. Resulting values had the units of Newton per millimeter (N mm^{-1}) for axial and bending modes, and Newton millimeter per degrees (N mm deg^{-1}) for torsion. Each configuration was tested three times per loading mode (AC, T, A–P, and M–L), and the averaged stiffness values were evaluated to determine the percent change in mechanical stiffness due to an individual frame component. Statistical analysis was performed with the PC/SAS software package, using a four-way analysis of variance (ANOVA), allowing for second-order interactions. The percent of total variance in stiffness that could be attributed to a component or their interactions, and the total variability that could be explained by the model, were calculated. Small percentage values for a component would suggest that no matter how that component was manipulated, the effect on bone fragment stabilization would be negligible. Conversely, large percentage values for a component indicated that manipulation of said component would have significant impact on bone fragment stabilization.

3. Results

3.1. Phase I — single ring (Fig. 2 (A)–(D), Table 1)

Ring diameter had the greatest effect in all four loading modes ($P = 0.0001$). However, each 40 mm enlargement in ring diameter decreased axial stiffness 30%, while reducing torsion, A–P and M–L bending stiffness only 10%. Axial and torsional stiffness were also reduced 10% ($P = 0.0001$) when the wire diameter was decreased from 1.8 to 1.5 mm. Wire angle had no substantial effect on axial or torsional stability, but was the most notable parameter affecting bending ($P = 0.0001$). A reduction in wire angle from 90° to 45°

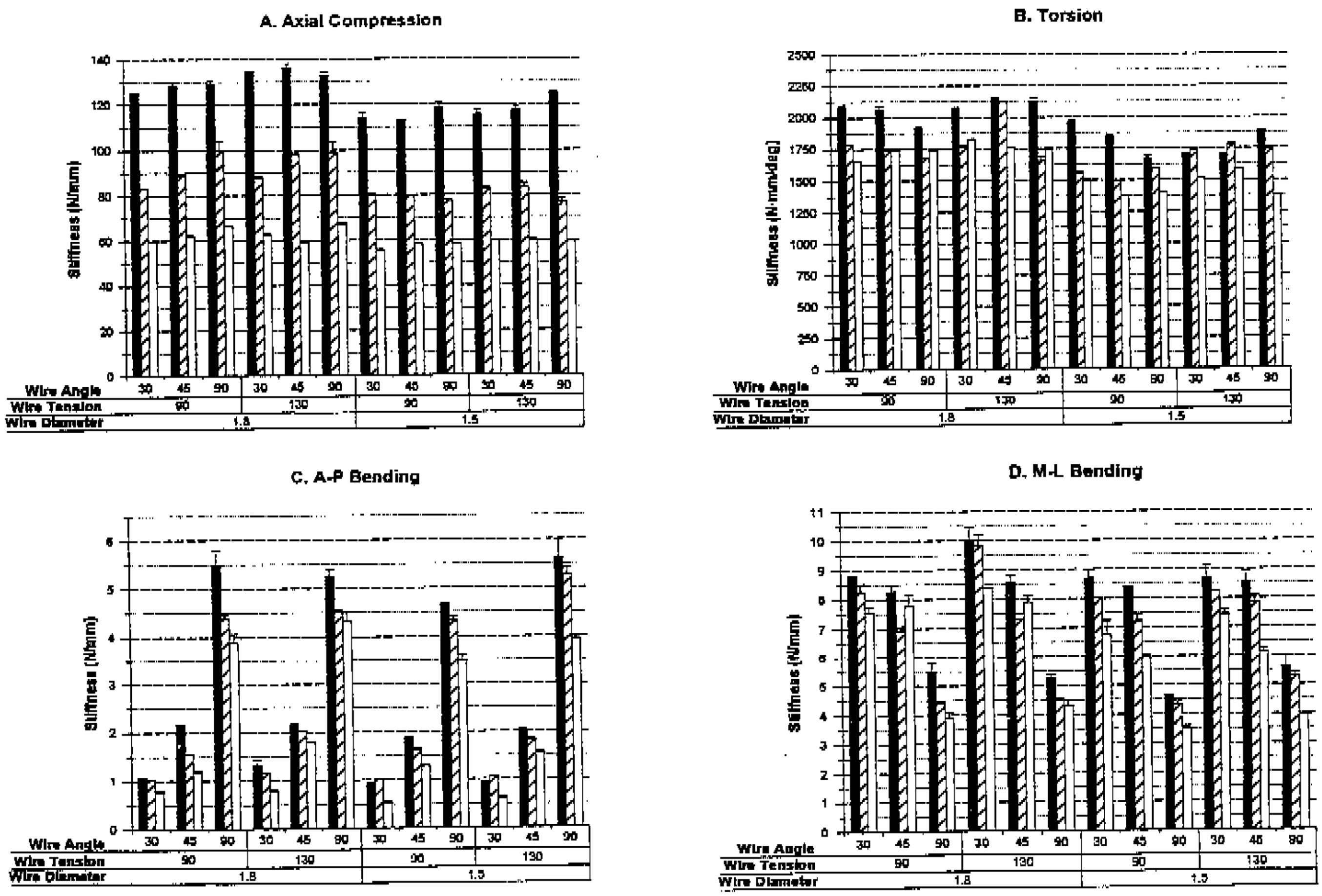

Fig. 2. Stiffness graphs for phase I (single-ring).

decreased A–P bending stiffness 160% and increased M–L bending stiffness 60%. However, a further reduction in wire angle (45–30°) decreased A–P bending stiffness an additional 90%, while improving M–L bending stiffness only 10%. Increasing wire tension, from 90 kg to 130 kg, had a negligible effect on all four loading modes.

3.2. Phase II — double-ring block (Fig. 3 (A)–(D), Table 1)

Due to a minimal contribution to stiffness during phase I, testing of wire diameter and wire tension were eliminated as variables. Therefore, 1.8 mm wires tensioned to 130 kg were used for phase II and III configurations. Because the effect of ring diameter on stability of fixation was linear, this parameter was reduced to two sizes (120 and 200 mm).

As with phase I, ring diameter had a substantial effect on axial compression, torsion, and bending ($P = 0.0001$). A ring diameter enlargement of 80 mm decreased axial stiffness 55%, torsional stiffness 10%, and bending stiffness 30%. The distance between the rings had the greatest effect on bending ($P = 0.0001$). Increasing ring separation from 10 mm to 200 mm improved A–P bending stiffness 340% and decreased M–L bending stiffness 55%. Ring separation had a nominal effect on torsional rigidity decreasing it 10% and had no substantial effect on axial stiffness.

Four-wire configurations were more stable than three-wire configurations for all modes of testing ($P = 0.0001$). Axial and torsional stiffness was improved 30% in four-wire configurations. However, neither joint nor osteotomy wire angles influenced axial compression and torsion. Furthermore, joint wire angle had no influence on bending stability, while osteotomy wire angle had a substantial effect on bending stiffness. Both A–P and M–L bending stiffness were improved 25–30% in four-wire configurations with osteotomy wire angle 30° and ring separation 10 mm.

Further increase in osteotomy wire angle (30–90°) improved A–P bending stiffness 30% and M–L bending stiffness 20%. In four-wire configurations with a 30° osteotomy wire angle and 200 mm ring separation, A–P and M–L bending stiffness were improved 10% and 60%, respectively.

Further increase in osteotomy wire angle to 90° improved A–P bending stiffness only 10%, while increasing M–L bending stiffness 70%.

Table 1
Percent contribution of parameters of fixation, components of external fixator, and their interactions to overall stability of the bone fragment(s) fixation

Component	Axial compression	Torsion	A–P bending	M–L bending
Phase I (single-ring)				
Ring diameter	93.7 (0.0001)	42.7 (0.0001)	3.6 (0.001)	10.2 (0.0001)
Wire diameter	3.6 (0.0001)	31.4 (0.0001)	0.1 (0.13)	1.5 (0.006)
Wire tension	0.4 (0.015)	3.6 (0.046)	0.6 (0.002)	2.0 (0.002)
Wire angle	0.4 (0.043)	2.1 (0.29)	93.3 (0.0001)	81.0 (0.0001)
% Total	98.1	79.8	97.6	94.7
Phase II (double-ring block)				
Ring diameter	88.9 (0.0001)	27.2 (0.0001)	5.3 (0.0001)	7.6 (0.0001)
Osteotomy wire angle	5.7 (0.0001)	39.0 (0.0001)	7.8 (0.0001)	34.7 (0.0001)
Ring separation	1.1 (0.0005)	13.2 (0.0001)	80.7 (0.0001)	8.9 (0.0001)
Ring diameter and ring separation	0.2 (0.054)	5.7 (0.003)	1.6 (0.0003)	0.03 (0.55)
Osteotomy wire angle and ring separation	0.6 (0.033)	2.8 (0.15)	1.2 (0.009)	46.2 (0.0001)
% Total	96.5	87.9	96.6	97.4
Phase III (two double-ring blocks)				
Proximal wire angle	3.9 (0.0001)	29.8 (0.0001)	10.3 (0.0001)	14.0 (0.002)
Distal wire angle	0.6 (0.013)	5.9 (0.05)	1.0 (0.045)	9.1 (0.005)
Distal ring separation	0.3 (0.08)	48.2 (0.0001)	84.0 (0.0001)	52.0 (0.0001)
Applied load	89.5 (0.0001)	NA	NA	NA
Distal wire angle and distal ring separation	0.6 (0.37)	2.7 (0.67)	0.01 (0.7)	13.1 (0.026)
% Total	94.9	86.6	95.3	88.2

3.3. Phase III — two double-ring blocks (Fig. 4 (A)–(D), Table 1)

Because ring diameter's effect on stiffness was linear for phase I (single ring) and phase II (double-ring block), this parameter was eliminated as a variable. Joint wire angle was also eliminated as variable due to its minimal contribution to stiffness during phase II testing. Therefore, 150 mm diameter rings with a joint wire angle of 90° were used for phase III configurations.

The load/displacement plots for axial compression demonstrated a non-linear relationship that was not seen in previous phases of testing. For analysis, load/displacement curves were divided into three load ranges: 0–215 N, 215–415 N, and 415–620 N as described by Podosky *et al.* [14]. There was an average 35–40% increase in axial stiffness between ranges. As with phase II, wire number had a substantial effect on axial stability ($P = 0.0001$) improving it 10% with the insertion of each additional wire. Distal fragment ring separation and both osteotomy wire angles had no effect on the axial stiffness.

Torsional stability was highly influenced by distal fragment ring separation ($P = 0.0001$). A 110 mm distal fragment ring separation was 5%, 10%, and 30% more stable than ring separations 160, 60, and 10 mm, respectively. The number of wires also had a substantial effect on torsional stiffness with a 10% improvement for each additional wire.

Distal fragment ring separation greatly influenced both A–P and M–L bending ($P = 0.0001$). A distance of 160 mm provided the most stable configurations regardless of wire angle. A–P bending stiffness demonstrated a 40% decrease, whereas M–L bending showed a 30% decrease in stiffness for each 50 mm reduction of distal fragment ring separation. Number of wires had a substantial effect on bending of the proximal fragment and only nominal influence on bending of the distal fragment ($P < 0.05$). This effect was best emphasized at a distal fragment ring separation of 60 mm demonstrating 30% stiffness increase with four-wire fragment fixation.

Wire angle in general had only a modest effect on both A–P and M–L bending. Therefore, this effect was noticeable only when the wire angle, either proximal or distal, was 90° producing 10% stiffness increase. Proximal wire angle was a more influential factor on A–P bending ($P < 0.05$), while distal wire angle had more effect on M–L bending stiffness ($P < 0.05$).

Transverse shear motion was determined by the relative displacement between the proximal and distal fragments. For the load range applied (up to 90 N) both A–P and M–L bending modes demonstrated less than 1 mm transverse shear between fragments for a distal ring block separation of 160 mm. At a ring separation of 110 mm, a force of 70 and 45 N was required to generate a displacement of 1 mm for A–P and M–L bending, respectively. Decreasing the distance between the distal rings to less than 60 mm required less than 30 N to generate the same amount of shear.

Telescopic rods improved bending and torsional stiffness of two double-ring blocks configurations 10% and 25%, respectively compared to 6 mm threaded rods. Although olive wires had no influence on axial, torsional, and A–P bending stiffness, they improved

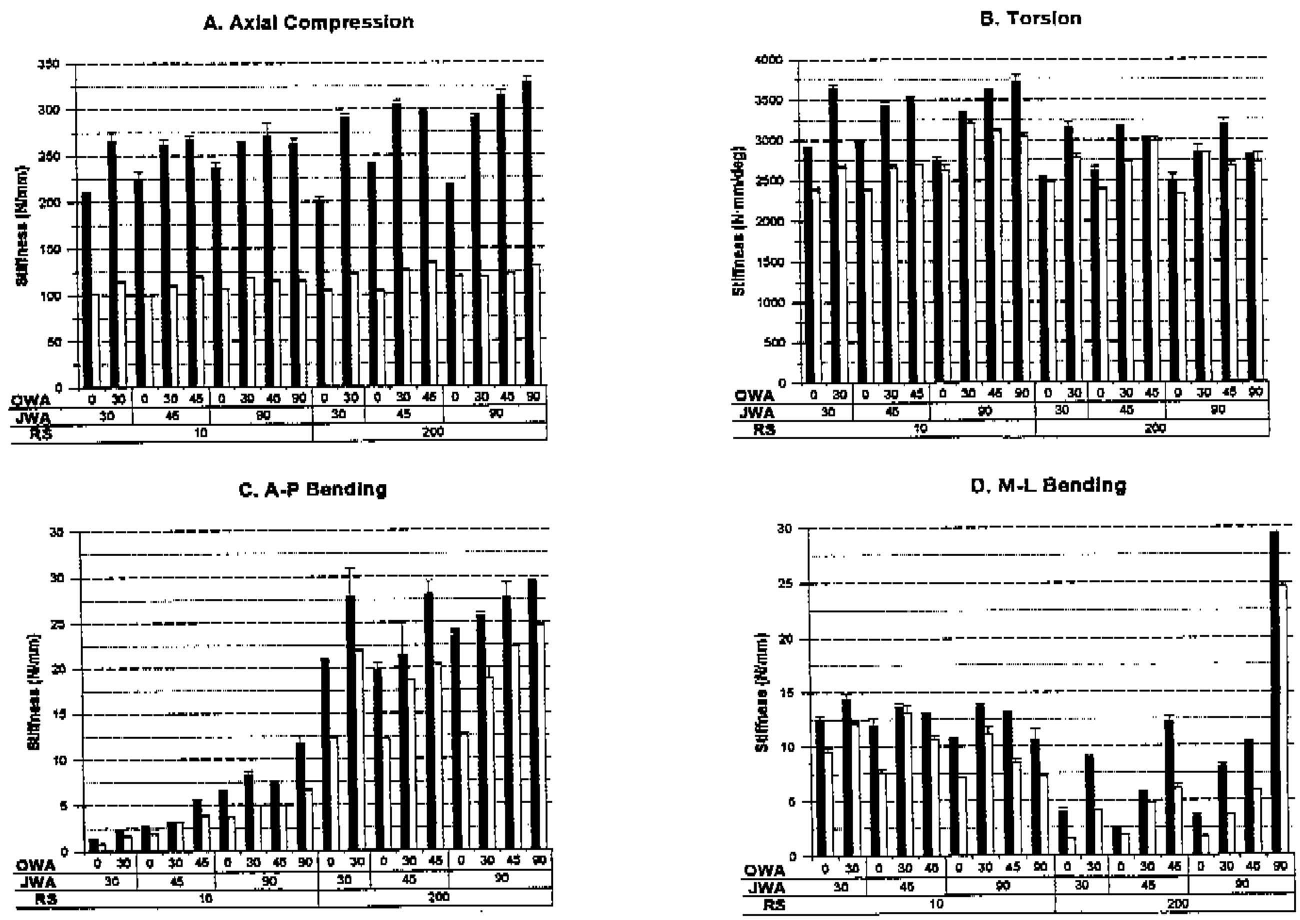

Fig. 3. Stiffness graphs for phase II (double-ring block).

M–L bending stiffness 20% but only for distal fragment ring separations of 60 mm, 110 mm, and 160 mm. The ring materials did not affect any mode of testing.

4. Discussion

Successful application of external circular fixation for limb lengthening, deformity correction, and fracture healing is dependent upon numerous biological and biomechanical factors. Biological factors include the type and level of osteotomy, the preservation and maintenance of adequate blood supply, and the quality of bony and surrounding soft tissues. The biomechanical factors determined to be important are fixator rigidity and stability of bone fragment fixation within the circular frame. The fundamental difference between circular and monolateral external fixators is the use of thin pretensioned wires as fixation elements. These wires are better tolerated by the bone and soft tissue, thereby significantly reducing the rate of complications. Small diameter fixation wires also provide elastic fixation at low loads allowing micromotion of the bone fragments during weight bearing, which has been deemed beneficial to fracture healing and the formation of distraction regenerate [16]. The stability of bone fragment fixation is highly dependent upon the number and orientation of the wires. Clinical experience and biomechanical studies have demonstrated that optimal stabilization of a bone fragment within rings can be achieved by two-level segmental fixation with three or four wires inserted at right angle to each other [1,2]. However, the anatomy of the limb segment may limit the number of fixation wires and the angle of their intersection, which is often less than 90°. Reducing the number of wires and decreasing the angle between them may affect the stability of bone fixation, especially in the sagittal plane. In these cases, compromised fixation can be improved by changing the other parameters of fixation wires such as length, diameter, or tension, as well as by the manipulation of other components of the circular fixator.

Several prior biomechanical investigations have analyzed the effect of some individual components of the external circular fixator on the bone fixation [5–15]. These studies, however, did not demonstrate the

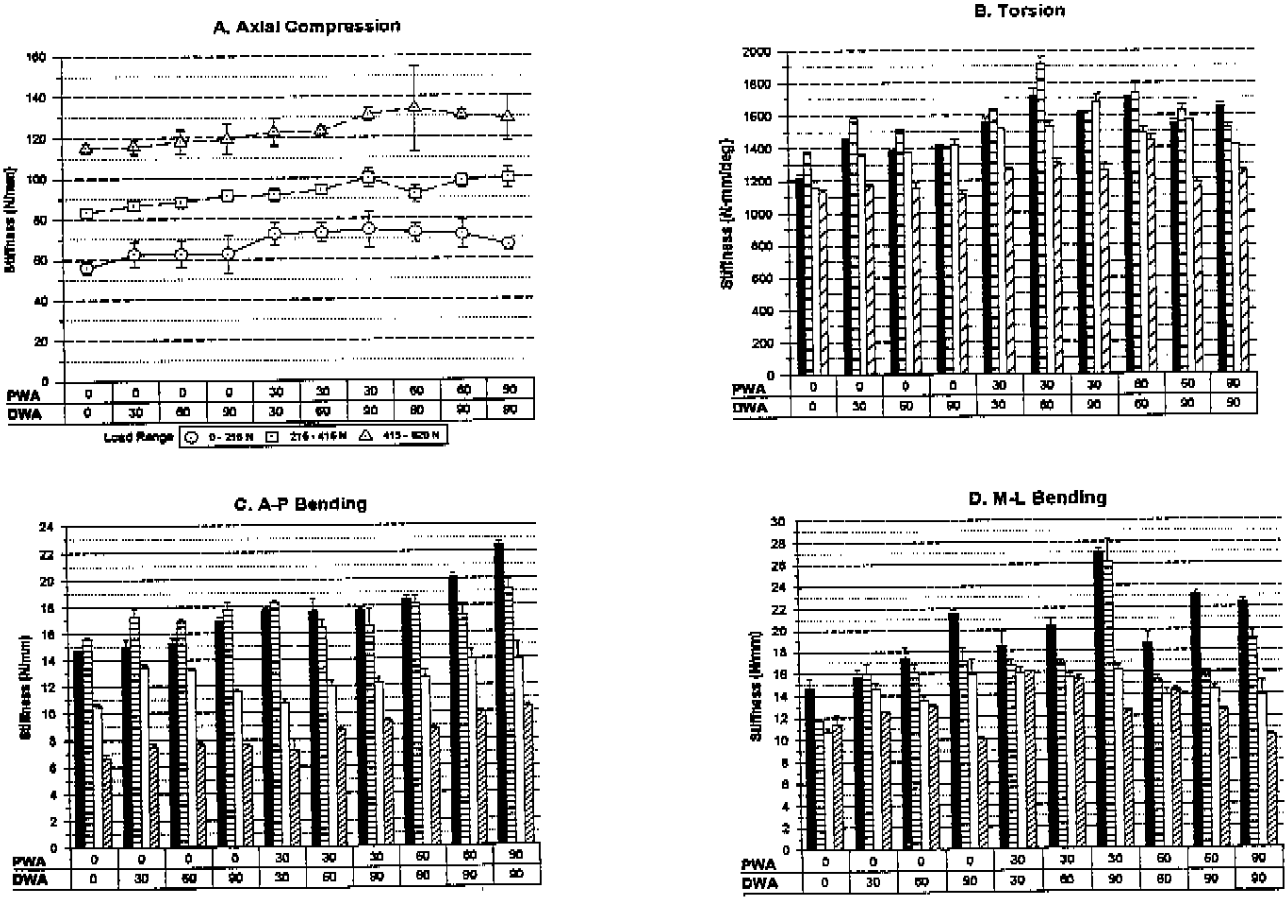

Fig. 4. Stiffness graphs for phase III (two double-ring blocks).

contribution of each individual component to the overall stiffness and how the interaction between different components influences the stability of bone fragment fixation. In the present study each individual component was biomechanically evaluated in three phases (single ring, double-ring block, and complete two double-ring block frame) to demonstrate the effect of said component on overall stiffness as levels of bone fragment fixation increased. Therefore, the utilization of a four-way analysis of variance allowing for second-order interactions permitted the quantification of the relative effect those different components and their interactions had on the variability seen in stability of fixation.

The present study confirmed the importance of ring diameter, wire angle, wire tension, and wire diameter on bone fragment stabilization with circular external support. Ring diameter had the most influence on overall stability, while wire angle had the most effect on bending stiffness. While the effect of wire diameter was significant for all modes of testing, except A–P bending (Table 1), it only substantially accounted for the variability seen in torsional stiffness. The effect of wire tension was also significant for all four loading modes, but had the least effect on construct variability, except M–L bending.

Ring diameter was the most influential component in all four loading modes of the double-ring construct (Table 1). Axial and torsional stiffness were improved by the insertion of additional wires while ring separation and wire angle were the most important components affecting bending. Individually, ring separation and wire angle could not adequately explain the variability in bending stiffness between different configurations. Therefore, bending stability was dependent upon both variables. A small wire angle substantially improved stiffness for a small ring separation whereas a large wire angle improves bone stabilization for a larger ring separation. The most prominent effect of this interaction was seen for M–L bending of the double-ring block configurations. The effect of ring separation on the stability of the three-wire double-ring frame configuration was previously studied by Orbay *et al.* [13]. Their results demonstrated that as the ring separation increases, M–L bending stiffness remains constant, whereas A–P bending stiffness increases but only when the ring separation is greater than 3 cm. Results of our testing were similar for both A–P and M–L bending stiffness but only for the small distances between two rings. With the larger ring separation (200 mm) used in the current study, M–L bending stiffness decreases due to bone slippage along the third wire. Distance between the rings in the double-ring block configurations also affected torsional rigidity reducing it when ring separation was increased.

Previous studies that have examined full frame configurations [5,7–11,14,15] have typically tested two equal length bone segments attached to symmetric fixation blocks. Usually two wire angles (i.e., 90 and 45°) were evaluated and compared to the standard reference construct. In clinical practice, however, an osteotomy for limb lengthening or deformity correction is often performed at the level of metaphysis generating a short proximal and long distal bone fragments. This surgical protocol was used in our study to assemble phase III frame configurations consisting of a short proximal and variable distal double-ring blocks.

During axial compression, bone fragment stability demonstrated a non-linear relationship with respect to the applied load. This non-linear relationship has been demonstrated by previous investigators [10,14] and was attributed to a stress stiffening effect of the wires, which are more resistant to deflection as loads increase. Further improvement of axial stiffness in our study was contributed to increasing number of fixation wires.

Torsional stability was substantially affected by the location of the proximal ring on the distal fragment (Table 1). The most stable frame configuration with respect to torsional stiffness occurred when the distal block ring separation was equal to the distance between the two double-ring blocks. Translating the proximal ring of the distal fragment either proximally or distally increases the length of either distal block ring separation or the distance between the two double-ring blocks, therefore reducing the torsional stability.

Analysis of the bending stability showed that there was a substantial reduction in bending stiffness in both A–P and M–L planes, with a reduction in distal block ring separation. Translating the proximal ring of the distal fragment distally creates a longer unsupported bone segment that is readily displaced, promoting shear between the bone fragments. Although transverse shear motion has not been extensively studied, this type of displacement between bone fragments is thought to be detrimental to fracture healing [17].

There are several limitations to the current study. Although the intrinsic factors (i.e., soft tissue tension and bone regenerate stiffness) provides substantial contribution to overall bone fragment stabilization [18], this study specifically focused on the extrinsic factors such as ring diameter, wire diameter, wire angle, etc. In clinical situations, variations in bone geometry and mechanical properties of bone tissues will inherently affect the overall stability of fixation. The use of a uniform plastic cylinder as a bone model eliminated this variability and allowed an investigation of only those factors that can be directly controlled during frame application.

5. Conclusion

The stability of bone fragments within circular external fixator is affected by manipulation of the parameters of fixation or individual components of the frame. The contribution of each component to overall bone fragment stability is dependent upon the mode of loading. Ring diameter, determining the length of the wires, has the greatest effect on stability of fixation. Minimization of the ring diameter substantially improves axial stability, torsional stiffness and bending rigidity of the bone fragments. Increasing the number of wires improves both axial and torsional stiffness, while torsional stability may be further improved by reducing the distance of ring separation. Bending stability in anterior–posterior or medial–lateral directions can be improved by reducing the relative wire angle while sacrificing stability in the opposing direction. However, a small wire angle permits bone slippage and special wires with stoppers should be used to increase the stability of the bone fragment.

The changes in overall stability of bone fragments are dependent not only on the individual frame components but also upon their interaction with other parameters of fixation. Torsional and bending stiffness are the most affected by component interactions. Ring diameter has substantial effect on torsional stability only for small (about 10 mm) distances of the ring separation. This effect is reduced as ring separation increases until the influence of ring separation dominates in torsional stiffness. The contribution of the interaction between wire angle and ring separation is similar. For the small ring separation, a minimization of the angle between crossing wires is the most effective way to increase bending stiffness. For the larger ring separation, either increasing the wire angle or the use of the special wires with stoppers will substantially improve bending stiffness.

Acknowledgements

This work was supported by the Texas Scottish Rite Hospital for Children Research Fund. The authors thank Dr. Michael Wattenbarger and Dr. Kelly Baker for their contribution to the research and editorial process.

References

[1] Paley D, Biomechanics of the Ilizarov external fixator. In Maiocchi AB and Aronson J (eds.), *Operative Principles of Ilizarov Fracture Treatment*, Medi Surgical Video. Milan, pp. 31–41.

[2] Catagni M. Fractures of the leg (tibia). In Maiocchi AB and Aronson J (eds.), *Operative Principles of Ilizarov Fracture Treatment*, Medi Surgical Video, Milan. pp. 91–124.

[3] Stitgne SH. Cairns ER, Ebraheim NA. Niemann JM, Jackson WT. Anatomical considerations of pin placement in the proximal tibia and its relationship to the peroneal nerve. Clin. Orthop. 1992;278:134

[4] Behrens F. General theory and principles of external fixation. Clin. Orthop. 1989;241:15

[5] Paley D, Flemming B, Catagni M, Kristiansen T, Pope M. Mechanical evaluation of external fixators used in limb lengthening. Clin. Orthop. 1990;250:50

[6] Aronson J, Harp JH. Mechanical consideration in using tensioned wires in a transosseous external fixation system. Clin. Orthop. 1992;280:23

[7] Calhoun JH, Li F, Ledbetter BR, Gill CA. Biomechanics of the Ilizarov fixator for fracture fixation. Clin. Orthop. 1992;280:15

[8] Chao EY, Kasman RA, An KN. Rigidity and stress analysis of external fracture fixation devices — a theoretical approach. J. Biomechanics 1982;15:971

[9] Fleming B, Paley D, Kristiansen T, Pope M. A biomechanical analysis of the Ilizarov external fixator. Clin. Orthop. 1989;241:95

[10] Gasser B, Boman B, Wyder D, Schneider E. Stiffness characteristics of the circular Ilizarov device as opposed to conventional external fixation. J. Bio. Eng. 1990;112:15

[11] Juan JA, Vera PP, Sanchez-Lacuesta HJ. Biomechanical consequence of callus development in Hoffman, Wagner, Orthofix and Ilizarov external fixators. J. Biomechanics 1992;25:995

[12] Kummer F. Biomechanics of the Ilizarov external fixator. Clin. Orthop. 1992;280:11

[13] Orbay GL, Frankel VH, Kummer FJ. The effect of wire configuration on the stiffness of the Ilizarov external fixator. Clin. Orthop. 1992;279:299

[14] Podolsky A, Chao EYS. Mechanical performance of Ilizarov circular external fixators in comparison with other external fixators. Clin. Orthop. 1993;293:61

[15] McCoy MT, Chao EYS, Kasman RA. Comparison of mechanical performance in four types of external fixators. Clin. Orthop. 1983;180:23

[16] Goodship AE, Kenwright J. The influence of induced micromovement upon the healing of experimental tibial fractures. J. Bone Joint Surg. 1985;67B:650

[17] Watson, JT, and Sauer P, The effects of fracture healing under cyclic shear micromotion, *Proceedings of the 4th Annual ASAMI Scientific Meeting*, Paper 2, New Orleans. LA, Feb 23, 1994.

[18] Hannu TA, ETS. Chao. Bone-healing patterns affected by loading, fracture fragment stability, fracture type, and fracture site compression. Clin. Orthop. 1993;293:8