Effects of ring diameter and wire tension on the axial biomechanics of four-ring circular external skeletal fixator constructs

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**Objective**—To determine relative effects of ring diameter and wire tension on axial biomechanical properties of 4-ring circular external skeletal fixator constructs.

**Sample Population**—4-ring circular external skeletal fixator constructs and artificial bone models.

**Procedure**—4-ring constructs were assembled, using 50-, 66-, 84-, or 118-mm-diameter rings. Two 1.6-mm-diameter fixation wires were attached to opposing surfaces of each ring at intersection angles of 90° and placed through a gap-fracture bone model. Three examples of each construct were loaded in axial compression at 7 Ns to a maximum load of 400 N at each of 4 wire tensions (0, 30, 60, and 90 kg). Response variables were determined from resulting load-displacement curves (construct stiffness, load at 1 mm of displacement, displacement at 400 N).

**Results**—Ring diameter and wire tension had a significant effect on all response variables and had a significant interaction for construct stiffness and displacement at 400 N. Significant differences within all response variables were seen among all 4 ring diameters and all 4 wire tensions. As ring diameter increased, effect of increasing wire tension on gap stiffness and gap displacement at 400 N decreased. Ring diameter had a greater effect than wire tension on all response variables.

**Conclusions and Clinical Relevance**—Although effects of wire tension decrease as ring diameter increases, placing tension on wires in larger ring constructs is important because these constructs are inherently less stiff. The differential contribution of ring diameter, wire tension, and their interactions must be considered when using circular external skeletal fixators. (Am J Vet Res 2001;62:1025–1030)
used as bone models. Rings were fixed at intervals of 60 mm, using 3 threaded rods. Rod placement was staggered between each ring pair. To ensure consistent and accurate wire placement, holes for fixation wire were drilled prior to wire placement, using an indexed drill press. Fixation wires were placed on opposing surfaces of each ring and driven through the previously drilled holes in the centered bone models with an intersection angle of 90°. Fixation wires were secured to the rings, using the center hole of the fixation bolt (Fig 1). Tension was placed on fixation wires, using a calibrated dynamometric wire tension device. After assembly of each construct, a 1-cm segment of delrin rod (ie, bone model) was removed to create a gap-fracture model. This gap was centered between the 2 inner rings of the construct.

Constructs were nondestructively tested in axial compression, using a servo-hydraulic materials-testing machine under load control. Constructs were loaded at a rate of 7 N/s to a maximum load of 400 N. Constructs were secured to the testing machine at the proximal and distal ends of the delrin rods, using a custom-designed mounting apparatus.

Figure 1—Illustration of a 4-ring circular external skeletal fixator construct applied to a 19-cm solid rod, using a gap fracture model. Each ring is separated by an interval of 6 cm, and fixation wires are placed at intersection angles of 90°.

Figure 2—Representative load-displacement curves for each of the 4 diameters of rings at 0 kg of tension on fixation wires. Notice the increasingly nonlinear behavior as ring diameter increases.

Figure 3—Mean ± SD values for stiffness (top), displacement at 400 N (middle), and load at 1 mm of axial displacement (bottom) for 4-ring circular external skeletal fixator constructs as a function of ring diameter and tension of fixation wires.
Displacement was determined from the distance the actuator traveled. Load-displacement data were stored electronically. Applied loads did not exceed the elastic region of the load deformation curve to allow repeated testing of the same construct. Fixator components and bone models were examined after each test to evaluate plastic deformation, and they were discarded if damaged. Each construct configuration was loaded 5 times at each of 4 wire tensions (0, 30, 60, and 90 kg), resulting in 16 construct configurations. After each series of 5 trials, fixation wires were replaced and tightened to the appropriate tension. Three series of trials were performed for each configuration.

Data analysis—The initial load-displacement curve was excluded from analysis to remove effects of settling of fixator components. Data from the subsequent 4 trials were averaged. For each construct configuration, 3 response variables were determined (axial stiffness, defined as the slope of the load-displacement curve; displacement at 400 N; and load at 1 mm of displacement). For construct configurations that had nonlinear load-displacement behavior, stiffness was defined as the slope of the final 25% of the load-displacement curve. Response variables were analyzed, using a 2-way multivariate ANOVA with factors of ring diameter and wire tension. Means of interest were compared, using a Fisher protected least significant difference test. Values of $P < 0.05$ were considered significant.

Because of the potential clinical importance of a 1-mm axial displacement at a fracture gap, nomograms were constructed for each ring diameter tested in an effort to illustrate the wire tension required to achieve a 1-mm displacement for specific body weights of dogs based on our axial displacement data. Anticipated body weight of a dog for each ring diameter was based on our clinical experience. Axial load on the fixator was inferred from data recorded for the forelimb for clinically normal dogs during trotting on a force plate.

Results

Gross plastic deformation was not observed in any construct. Appearance of the load-displacement curve varied with ring diameter and wire tension. The 30-mm rings had linear load-displacement behavior at all wire tensions. As ring diameter increased, load-displacement behavior became more exponential with the curves approaching linearity at higher loads (Fig 2). As wire tension increased, load-displacement behavior became more linear in the rings with a larger diameter. The 118-mm rings had linear load-displacement behavior only at a wire tension of 90 kg. Structural failure was not observed in any trial.

Ring diameter and wire tension each significantly ($P < 0.001$) affected construct stiffness, displacement at 400 N, and load at 1 mm of displacement. Significant differences in construct stiffness, displacement at 400 N, and load at 1 mm of displacement were seen among

![Figure 4](image-url)
all 4 of the ring diameters and all 4 of the wire tensions. Wire tension was directly related to construct stiffness and load at 1 mm of displacement. For these variables, as wire tension increased, construct stiffness and load at 1 mm of displacement increased. Wire tension was inversely related to displacement at 400 N (ie, as wire tension increased, displacement at 400 N decreased). Ring diameter was inversely related to construct stiffness and load at 1 mm of displacement. Ring diameter was directly related to displacement at 400 N (Fig 3). Ring diameter had a greater effect on construct stiffness, displacement at 400 N, and load at 1 mm of displacement, compared with the effect of wire tension.

Ring diameter and wire tension had a significant ($P < 0.001$) interaction for construct stiffness and displacement at 400 N. As ring diameter increased, the effect of increasing wire tension on stiffness decreased. Similarly, as ring diameter increased, the effect of increasing wire tension on displacement at 400 N decreased.

Nomograms illustrating the wire tension required to achieve 1 mm of displacement for each ring diameter revealed poor association with anticipated body weights of dogs (Fig 4). The nomogram curve of wire tension required to achieve 1 mm of displacement for 66-mm rings was within the anticipated body weight for this ring diameter. The nomogram curves of wire tension required to achieve 1 mm of displacement for 50-, 84-, and 118-mm rings were outside the anticipated body weights for each of these ring diameters.

**Discussion**

The nonlinear load-deformation behavior seen in the study reported here is consistent with results of other studies in which investigators examined axial stiffness characteristics of circular external fixation devices. A self-tensioning effect during axial loading is observed when small-diameter wires are used as fixation elements; thus, there is an initial exponential increase in stiffness that yields to a linear increase in stiffness at higher loads. It can be difficult to appropriately determine construct stiffness for nonlinear behavior. Other investigators have addressed this problem by performing logarithmic transformation of the data, determining stiffness at multiple points along the load-displacement curve with an assumption of linearity at each point, or determining stiffness over the terminal linear portion of the load-displacement curve. The validity of each method can be argued. Logarithmic transformation of our data did not result in consistent linear behavior for all constructs, making it impossible to conduct comparisons. Although determining stiffness over the final 25% of the load-displacement curve may bias the data toward higher stiffness values, we believe this method provides the most valid means of comparison among constructs.

The transition from nonlinear to linear load-displacement behavior for the constructs that had nonlinear behavior was less obvious than the transition observed in single-ring circular fixator constructs. We believe this difference can be attributed to the additive effect of loading multiple rings in the same construct. This has the effect of distributing the loading behavior over a greater load-and-displacement range. Because of this additive effect, we were observing a smaller window of behavior during the physiologic loading range used in the study reported here, compared with that in another study; therefore, the transition of load-displacement behavior from nonlinear to linear may have been less obvious.

Construct stiffness decreased significantly with increases in ring diameter. This effect has been observed by others and it is probably attributable to the fact that deflection of a wire subjected to a specific load is dependent, in part, on functional length of the wire. Accordingly, displacement at 400 N increased significantly with increasing ring diameter, and load at 1 mm of displacement decreased significantly with increasing ring diameter. Ring diameter had a greater effect on all response variables than did wire tension, corroborating the results of other studies in which investigators examined wire tension and ring diameter. Unfortunately, ring diameter is dictated by size of the patient, so surgeons are limited in their ability to adjust construct biomechanics by varying the ring diameter. Surgeons should select the smallest-diameter ring that can be placed on a limb while maintaining approximately 2 cm of distance between the inner surface of the ring and the soft tissues.

Construct stiffness increased significantly with increases in wire tension. This effect also has been observed by others. Placing tension on fixation wires mitigates the nonlinear behavior of constructs that use rings of larger diameters, effectively bypassing the self-tensioning effect on the tension of the wires. The effect of increasing construct stiffness by placing tension on fixation wires is limited. Kummer found little effect of placing > 130 kg of tension on 1.8-mm-diameter wires and > 90 kg of tension on 1.5-mm-diameter wires. Bronson et al also found a diminishing effect on axial stiffness at higher values for wire tension. Wire tension is limited by yield strength of the wires and by stiffness of the rings to which the wires are attached. In the study reported here, displacement at 400 N and load at 1 mm of displacement were significantly affected by wire tension. Wire tension is not necessarily dictated by size of a patient; therefore, surgeons are able to vary wire tension to achieve the desired biomechanical behavior.

Controlled axial micromotion has a beneficial effect on fracture healing. One millimeter of axial motion is used in most models. To assess the ability of the fixation constructs examined here to achieve 1 mm of axial displacement under physiologic loading, nomograms were constructed for each ring diameter, based on our 4-ring constructs (Fig 4). Analysis of our results indicated that placing tension on wires in 50-mm constructs is unnecessary, because animals in which this ring diameter would be applicable do not load their constructs to the point that they produce 1 mm of axial motion. Thus, decreasing the stiffness of these constructs would be necessary to allow 1 mm of axial motion. Decreasing the diameter of the wire is 1 way to achieve this effect. One millimeter of axial motion can be achieved in 4-ring 66-mm constructs by simply varying wire tension in animals of the appropriate body...
weights. Placing tension of 90 kg on wires did not limit axial motion to 1 mm when 84- and 118-mm 4-ring constructs were used. This finding emphasizes the importance of placing maximal tension on wires of typical constructs that use 84- and 118-mm-diameter rings. Additional steps should probably be taken to increase axial stiffness of these constructs, such as use of additional rings, half pins, or drop wires.\(^2\) Analysis of our data revealed a significant decrease in the effect of wire tension on construct stiffness and displacement at 400 N as ring diameter increased. This finding was attributed to the functionally longer wires in large-diameter rings, and it has been observed by others.\(^2\) It is important to mention that although the percentage change in displacement at 400 N achieved with increases in wire tension decreased as ring diameter increased, the actual magnitude of the change increased. These results support our clinical impression that placing maximal tension on wires of large-diameter rings is necessary to approach ideal biomechanical behavior.

Several limitations exist that limit the application of results from this study to clinical settings. We chose to evaluate constructs of identical dimensions and varied only the ring diameter to allow direct comparisons among constructs. A 19-mm-diameter solid rod was used in all constructs. Bones of the distal limbs of cats or small dogs in which 50- or 66-mm-diameter rings would be used would be much smaller than 19 mm in diameter, and bones of the distal limbs of dogs in which 118-mm-diameter rings would be used often be larger than 19 mm in diameter. The diameter of the rod or bone containing the fixation wire has an effect on stiffness of the construct. The relatively large diameter of the solid rod may have artifically increased stiffness of our 50-mm constructs and decreased stiffness of our 118-mm constructs. Although delrin rods have been used as bone models in other studies\(^16,20\), their appropriateness is not known. Specifically, the behavior of the wire-bone interface may differ considerably; however, this potential difference would be of little consequence in axial loading. Total fixator length was not altered in the constructs evaluated; however, length of fixators was not a major contributor to axial stiffness in another study.\(^7\) The 50- and 66-mm-diameter constructs evaluated here would not have length dimensions appropriate for clinical use. Additionally, the circular external fixator constructs were idealized in that we used wire intersection angles of 90° and uniform spacing of rings. Anatomic considerations make these attributes difficult to achieve in clinical practice.\(^19\)

Recommendations regarding wire tension that were developed on the basis of the nomograms in this study should be interpreted cautiously. As mentioned previously, we used idealized constructs that may not accurately represent clinical situations. Although axial micromotion can stimulate healing of fractures, the precise amount of motion necessary to optimize healing of fractures is not known and is likely to be dependent on many factors, including size of the animal, size of the fracture gap, and fracture configuration.\(^16,28,30,31\) It is reasonable to assume that 1 mm of axial micromotion at the fracture site is not the optimal amount of micromotion for animals with the wide range of body weights and fracture types that would be treated with circular external skeletal fixators. Additional studies are necessary to better characterize factors that influence the effects of axial micromotion. Because of the assumptions necessary for this type of analysis and lack of information regarding optimal amount of fracture micromotion, results from the study reported here should only be used as a rough estimate of the appropriate amount of wire tension. Because of the inherent versatility in design of circular external skeletal fixators and the number and complexity of factors affecting healing of fractures, establishing specific simple guidelines regarding appropriate wire tension will probably not be possible.

\(^1\)IMEX circular external skeletal fixation system, IMEX Veterinary Inc, Longview, Tex.
\(^2\)MSC Industrial, Melville, NY.
\(^3\)Hofmann SAS, Monza, Italy.
\(^4\)Instron Model 8521, Instron Corp, Canton, Mass.

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