Bar Diameter Is an Important Component of Knee-Spanning External Fixator Stiffness and Cost

HYUNCHUL KIM, MS; JOSEPH P. RUSSELL, BS; ADAM H. HSIEH, PHD; ROBERT V. O’TOOLE, MD

abstract

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The authors’ objective was to determine the effects of bar diameter on the stiffness and cost of a knee-spanning external fixator. The authors studied 2 versions of an external fixator with a difference in bar diameter (small bars, 8-mm diameter; large bars, 11-mm diameter). Fixators were tested using frame dimensions and a synthetic fracture model appropriate for a tibial plateau fracture. Five configurations of each fixator were tested: standard, cross-link, oblique pin, double stack, and super construct. The construct stiffness of each configuration (n=60) was measured in anterior-posterior bending, medial-lateral bending, axial torsion, and axial compression. Cost analysis allowed for calculation of the stiffness per unit cost. In the large bar group, an increase in construct stiffness was noted for all constructs and testing modes. Magnitude of stiffness increase ranged from 24% to 224% (P<.05 in all cases), depending on the configuration and loading mode. Increase in stiffness was so large that double-stack small bars performed similarly to standard construct large bars. Considering that the frame components have similar costs, the double-stack small bar fixator results in a 66% increase in cost for the same stiffness provided by the standard large bar. Bar diameter seems to have a large effect on knee-spanning external fixators. The authors observed an increase in stiffness of up to 191% under anterior-posterior bending despite an increase in bar size of only 37.5%. This finding might allow clinicians to use less expensive frames constructed of larger bars without sacrificing construct stiffness.

The authors are from the Orthopaedic Mechanobiology Laboratory (HK, JPR, AHH), Fischell Department of Bioengineering, University of Maryland, College Park; and the R Adams Cowley Shock Trauma Center (AHH, RVO), Department of Orthopaedics, University of Maryland School of Medicine, Baltimore, Maryland.

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Correspondence should be addressed to: Robert V. O’Toole, MD, R Adams Cowley Shock Trauma Center, Department of Orthopaedics, University of Maryland School of Medicine, 22 S Greene St, T3R62, Baltimore, MD 21201 (rvo3@yahoo.com).

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External fixation is commonly used for the temporary treatment of high-energy tibial plateau fractures and other fractures while awaiting resolution of soft tissue swelling before definitive plate and screw fracture fixation is applied. The external fixator must have sufficient construct stiffness to maintain fracture alignment. Achieving sufficient construct stiffness can be difficult because the frames commonly have long bar distances that accommodate pins being placed outside the zone of injury and the location of future plate fixation. Previous studies have shown that the stiffness of a monolateral external fixator can be increased based on the number and diameter of pins, the spread of the pins, and the distance of the connecting bar to bone.

Although it is theoretically known that increasing the diameter of the bar should increase stiffness, the magnitude of this effect on typical knee-spanning external fixator systems used in current clinical practice is unknown. Furthermore, the cost implications of increasing bar size have not been investigated, and the cost of temporary external fixators is substantial and might be of additional importance. The purpose of this study was to evaluate the mechanical and cost effects of increasing the bar diameter of a commercially available knee-spanning external fixator. The authors’ hypothesis was that even modest increases in bar diameter have large effects on construct stiffness for knee-spanning external fixation.

**Materials and Methods**

For the investigation, the authors chose a fixation system that had 8-mm diameter carbon fiber bars (small bars) (Hoffmann II MRI External Fixation System; Stryker Trauma AG, Selzach, Switzerland) and a system with 11-mm carbon fiber bars (large bars) (Hoffmann 3 Modular External Fixation; Stryker Trauma AG). The Hoffmann systems were chosen for investigation because the 2 versions are largely identical in all other manners, allowing the authors to study the single parameter of bar size in isolation.

The “standard” knee-spanning fixator configuration was modeled after what is commonly used in practice at the Level I trauma center at the authors’ institutions. To simulate what is typically needed in a clinical setting, the dimensions of the fixator were chosen to allow for room to span the knee joint and an area of the tibia for future plate and screw fixation and room for soft tissue (Figure 1). All fixators were placed using a technique that is identical to the technique that would be used in an operating room. All pins were inserted bicortically into the authors’ bone model using a guide. The pin clusters were oriented at a 45° angle relative to each other in the axial plane, as is typical at the authors’ center.

Failure at the bone-pin interface with temporary spanning external fixators is not typically a problem in healthy bone. Because the behavior of bone is not relevant in this scenario, the authors chose to use ultra-high molecular weight polyethylene pipes with an outer diameter of 50.8 mm and an inner diameter of 25.4 mm to simulate the femur and tibia. A comminuted Orthopaedic Trauma Association type C complete tibial plateau fracture was simulated by creating a 50-mm gap between the tubes, representing the femur and tibia. The size of the gap between the
tubes also prohibited the ends of the bone model from touching during compression and bending tests. Straight posts were placed into the 10-hole pin clamps using the holes closest to the fracture site. The carbon rods were attached to the outriggers at the most lateral positions possible on the straight posts, following the clinical practice of establishing a wide frame for increased stability of the fixator.

The standard construct was modified to use 4 common strategies for increasing the fixator frame stiffness: use of a cross-link, use of an oblique pin, double stacking, and building a super construct that included all of these techniques (Figure 2). For the cross-link configuration, an additional 250-mm carbon rod was clamped to the 2 existing carbon rods, proximally on the medial rod and distally on the lateral rod. Both proximal and distal cross-link couplings were 140 mm from the straight posts for small bar constructs and 170 mm for large bar constructs. For the oblique pin configuration, a 5-mm stainless steel pin was added to the construct, extending from the lateral carbon fiber rod to the femur model. The oblique pin was inserted 45 mm distal and 10 mm lateral to the femoral pin cluster. For the double-stack configuration, a second set of 10-hole pin clamps was fixed 60 mm above the standard pin clamps and a second set of carbon rods was attached. For the super construct configuration, the standard configuration was modified with the 3 previous configurations (cross-link, oblique pin, and double-stack) in tandem, following the same protocol as the individual configurations.

Of note, one difference was found between the 2 systems in addition to bar diameter. Five hundred-millimeter bars were used for the small bar group and 550-mm bars for the large bar group, but the pin-to-pin distances and all other aspects of the 2 models were identical. The 10% difference in bar length was accommodated by placing the pin bar clamp in a slightly different configuration, which theoretically might have provided a slight mechanical advantage to the small bar group.

Six different samples each for the small and large bar groups were tested repeatedly using all 5 fixator configurations, yielding a total of 60 tests. Configurations were tested in a random order to ensure that no bias in stiffness was introduced based on the order of testing. For every specimen, each configuration was tested sequentially under 4 loading modes: anterior-posterior bending, medial-lateral bending, axial torsion, and axial compression (Figure 3). Pilot studies were conducted to define load limits within which all deformations were fully recoverable to ensure validity of sequential construct testing.

All testing was conducted with an 858 Mini-Bionix II axial-torsional materials testing machine.
testing system (MTS Corp, Eden Prairie, Minnesota). For anterior-posterior and medial-lateral bending, specimens were mounted on custom fixtures to maintain orientation while allowing free rotation in the appropriate plane during a 3-point bending test. For the bending tests, specimens were supported by aluminum rods 5 cm from the ends of the femoral and tibial tubes. During testing, a horizontally oriented aluminum rod was brought down to make contact with the tibial bone model 15 mm away from the gap between the pipes under a tare load of 1 N. A 5-mm displacement was then applied at a rate of 1 mm per second while force data were collected. Loads were directed posteriorly for anterior-posterior bending and medi- ally for medial-lateral bending.

For axial compression, specimens were mounted vertically with the tibial component constrained in the bottom fixture. On the femoral end, an aluminum hemisphere was brought into unconstrained contact with an aluminum surface under a tare load of 1 N. A 10-mm displacement was then applied at a rate of 1 mm per second while force data were collected.

For axial torsion, both the femoral and tibial ends of the specimens were instrumented with hexagonal ball joints to eliminate parasite constraint effects and enable testing in pure torsion. Specimens were loaded to a maximum torque of 4.5 N•m at a rate of 0.5 N•m/s.

The total cost for each configuration was calculated by tallying the price of each component based on US list prices provided in 2012 in $USD. Actual costs vary between hospitals, but the calculation provides a relative comparison of the cost of each of the configurations. Cost per stiffness for the small bar configuration was calculated and then normalized with a value of the large bar standard configuration.

Data were recorded and then plotted in Microsoft Excel 2010 (Microsoft Corp, Redmond, Washington). The stiffness was determined by calculating the slope of linear region in the load-displacement curve. The results were statistically analyzed using SPSS version 14.0 software (SPSS, Inc, Chicago, Illinois). A Mann-Whitney nonparametric test using the Monte Carlo exact method was performed for a comparison of the fixator system. A P value of .05 was defined as statistically significant.

Results

Measured stiffness values for the various configurations under different loading modes are shown in Table 1. The 2 systems were tested in 5 configurations with 4 testing modes, yielding 20 possible bivariate comparisons. The large bar fixator system was stiffer than the small bar fixator system (P<.05) for 18 of the 20 comparisons. The size of the increase in stiffness ranged from 24% to 224%. The only comparisons that did not reach statistical significance in medial-lateral bending were cross-link and super construct by 28% and 7%, respectively. Both constructs use a cross-link bar component (P=.07 and .18, respectively).

The small and large bar groups had similar changes in stiffness for the 5 fixator constructs across the 4 testing modes (Figure 4). Most notably, for anterior-posterior bending, medial-lateral bending, and axial compression, groups were distinctly clustered into a lower range (standard, cross-link, oblique pin) and a higher range (double-stack, super construct) of stiffness. The higher range groups were both significantly stiffer than each of the lower range groups (P<.005), and no statistical differences were found among the groups within each range. In axial torsion, a progressive increase in mean stiffness was observed from standard to cross-link to oblique pin to double-stack. The super construct was significantly stiffer than all the other construct configurations (P<.005).

The cost for each configuration is listed in Table 2. The cost relative to the standard frame increased 25% for cross-link, 12% for oblique pin, 87% for double-stack, and 123% for super constructs. Costs were almost the same for small and large bars. Although the costs were almost identical for a given configuration in both systems, the cost-efficiency per stiffness was not similar for both constructs because the stiffness varied. The normalized cost per unit stiffness is shown in Table 3. All small bar configurations had less cost-efficiency compared with the large bar standard configuration. The small bar cross-link configuration indicated the highest cost per unit stiffness in all loadings. The double-stack and super construct had relatively low cost per unit stiffness among the small bar groups but still showed 111% to 182% greater cost per unit stiffness than the large bar standard fixator.

Discussion

Although numerous investigators have examined the mechanical properties of external fixator systems, it is unknown what effect increasing bar diameter has on modern knee-spanning external fixator systems. The current study shows that even modest increases in bar diameter seem to have a large impact on construct stiffness. This finding might have both a clinical and financial impact because it might allow clinicians to avoid more expensive solutions, such as double stacking, in favor of less expensive ways of achieving the same stiffness by simply using systems with larger bars.

When comparing the small and large bar frames, all configurations in the large bar group were considerably stiffer than those in the small bar group in each mode of anterior-posterior bending, medial-lateral bending, axial torsion, and axial compression. It is notable that the standard configuration in the large bar group improved the stability under anterior-posterior bending and compression by 191% and 216%, respectively, compared with the standard configuration in the small bar group. This study found that the
large bar fixator had a great advantage in stiffness with only a 37.5% increase in the diameter of the connecting bar. In terms of percentage increase in stiffness, the small bar group benefited more from double stacking than did the large bar group, but this is likely because of the already high stiffness of the large bar system.

A temporary knee-spanning external fixator is commonly used for treating high-energy proximal tibial and distal femoral fractures. Ideally, external fixator pins are placed outside the location of future plate placement, making the spanning distance between pins longer than it would otherwise be and the construct less stiff. To hold the fracture in an adequately reduced position, the frame must resist the deforming forces inherent with all fractures. A frame that is not adequately stiff can lack the ability to hold the fracture in the reduced position, even at the time of initial surgery, which negates one of the benefits of placing the patient in the frame to begin with. Hence, adequate stiffness of the knee-spanning fixator is critical to the effectiveness of temporary stabilization before permanent fixation.

Relatively little recent work has been done on the mechanics of knee-spanning fixators despite their common use and the clinical importance of achieving adequate stiffness at a reasonable cost. A recent study by Mercer et al examined the relative stiffness of 4 configurations of knee-spanning external fixation: anterior femoral pins with monotube, anterolateral femoral pins with monotube, anterolateral femoral pins with 2 connecting rods, and a hinged ring fixator. The authors found that the hinged ring fixator was the stiffest configuration compared with the other configurations. Also, the dual-bar frame was more stable than the monolateral bar frame in varus, valgus, torsion, and anterior-posterior shear. However, comparisons of cost increase relative to increase in stiffness were not conducted in that study.

The current authors investigated the frame stiffness of 5 external fixator frame configurations using a commercially available external fixator system and the newly developed Hoffmann 3 external fixator system. The major modification for the Hoffmann 3 system was an increased carbon bar diameter (from 8 to 11 mm), which provided the opportunity to study the effect of the single parameter of bar size on frame mechanics.

Previous studies have shown that the diameter of the connecting bar is a crucial contributor to increase the stiffness

### Table 1

<table>
<thead>
<tr>
<th>Construct</th>
<th>A-P Bending, N/mm</th>
<th>M-L Bending, N/mm</th>
<th>Axial Torsion, N•m/deg</th>
<th>Axial Compression, N/mm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Small bars</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Standard</td>
<td>2.17±0.45</td>
<td>4.55±1.06</td>
<td>0.28±0.03</td>
<td>1.05±0.11</td>
</tr>
<tr>
<td>Cross-link</td>
<td>2.43±0.22</td>
<td>5.27±1.43</td>
<td>0.33±0.02</td>
<td>1.22±0.08</td>
</tr>
<tr>
<td>Oblique pin</td>
<td>2.29±0.79</td>
<td>5.10±0.87</td>
<td>0.31±0.01</td>
<td>1.11±0.10</td>
</tr>
<tr>
<td>Double-stack</td>
<td>6.79±1.04</td>
<td>6.53±0.69</td>
<td>0.47±0.02</td>
<td>3.31±0.70</td>
</tr>
<tr>
<td>Super construct</td>
<td>8.11±1.30</td>
<td>8.36±0.68</td>
<td>0.58±0.05</td>
<td>5.94±0.50</td>
</tr>
<tr>
<td>Large bars</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Standard</td>
<td>6.33±1.49</td>
<td>6.42±0.58</td>
<td>0.48±0.04</td>
<td>3.32±0.25</td>
</tr>
<tr>
<td>Cross-link</td>
<td>6.86±1.01</td>
<td>6.70±0.56</td>
<td>0.54±0.05</td>
<td>3.47±0.29</td>
</tr>
<tr>
<td>Oblique pin</td>
<td>5.78±0.75</td>
<td>6.31±0.65</td>
<td>0.57±0.05</td>
<td>3.60±0.25</td>
</tr>
<tr>
<td>Double-stack</td>
<td>11.47±1.48</td>
<td>8.93±1.12</td>
<td>0.62±0.05</td>
<td>7.59±0.96</td>
</tr>
<tr>
<td>Super construct</td>
<td>12.50±1.02</td>
<td>8.97±0.63</td>
<td>0.80±0.08</td>
<td>9.13±0.52</td>
</tr>
</tbody>
</table>

Abbreviations: A-P, anterior-posterior; M-L, medial-lateral.
of external fixation for the treatment of tibial fractures. However, it was previously unknown how this would apply for knee-spanning frames or what the cost implications of this effect are. The current authors’ findings are generally consistent with the findings of the importance of bar diameter to fixator stiffness for tibial shaft fractures but are the first to show the magnitude of this effect on modern frames.

Comparisons of the effects of commonly used fixator construct configurations on stiffness were also examined in different loading modes. Consistent with the authors’ previous work, the addition of the cross-link and the oblique pin resulted in a stiffer than standard construct in all planes of loading. This is because an additional component provides more structural support for the standard frame to prevent it from bending. In addition, the double-stack and super construct configurations were significantly stiffer than were the other configurations in both the small and the large bar groups.

Although the super construct had a higher measured bending stiffness than did the double-stack configuration in anterior-posterior bending, medial-lateral bending, and axial compression tests, no statistically significant improvement was shown. In contrast, the torsional stiffness of the super construct was significantly greater than that of the double-stack frame in the large bar group. This is likely because of the addition of the cross-link to the frame. The cross-link configuration had a significant increase in torsion and compression test compared with the standard configuration. An additional bar across 2 long connecting rods in the frame produced more resistance to compression and torsion, contributing to an increased structural stability of the knee-spanning external fixators.

The increase in stiffness of the large bar system allows for a decrease in cost over small bar double-stack frames by using a simpler standard frame to provide the same construct mechanical performance. Ultimately, decisions regarding which frame type and configuration to use depend not only on cost-efficiency, but also on the desired construct stiffness. The authors’ stiffness and cost-efficiency data might be useful for guiding surgeons in their decisions for selecting an appropriate frame for fracture care based on the plane where additional stiffness is needed.

One limitation of the current study is that only a large gap was tested, such as would be present with a comminuted proximal tibial fracture. This study represents a worst-case scenario for frame stiffness. Furthermore, the authors studied only 1 company’s product. However, that company had recently made available a version of the fixator that essentially changed only 1 parameter, allowing the authors to study 1 effect in isolation without other important changes to the frame design. Another limitation is that the authors tested only 1 external fixator configuration. Although the authors would argue that this is a common configuration in clinical use today, there are other configurations commonly used that are either more stiff or less stiff and likely more or less expensive at baseline. As such, the details of the current study’s results might have varied if one of the other configurations had been used as the baseline frame, and these results might not apply to other frame configurations.

### Conclusion

Even a modest increase in bar diameter (from 8 to 11 mm) greatly improved bending stiffness in the knee-spanning external fixator. The effect was large enough that the standard large bar frame was similar to the double-stack small bar frame, despite the large increase in cost to create the double-stack small bar frame. Clinicians can use these data to help select the most efficacious and cost-friendly solutions when placing knee-spanning external fixators.
REFERENCES


