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**Article Title:** Comparison of Four Reconstructive Methods for Diaphyseal Defects of the Humerus after Tumor Resection

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Comparison of four reconstructive methods for diaphyseal defects of the humerus after tumor resection

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Abstract

The objective of the current study was to compare quantitative data on the biomechanical analysis of different techniques for fixation of intercalary bone defects of the humerus, by means of consistently applied methodology on composite models.

A total of twenty-five humeral specimens of composite models were used. An intercalary defect was created and reconstructed using plates, intramedullary nails, external fixators and segmental prosthetic implants. The specimens were loaded under axial compression, four-point-bending and torsion within the linear elastic region.

Modular segmental implants and intramedullary nails were able to compensate significantly greater amounts of compressive loads comparing to locking plates and external fixators. However, in flexion and torsion, the modular segmental implants and the external fixators were significantly better load-bearing devices comparing to the intramedullary nails and plates. Early mobilization of upper limb in patients with diaphyseal bone defects of the humerus could probably be more safe and tolerable when reconstructed with modular segmental implants.

Key words: intercalary, diaphyseal, defects, biomechanics, endoprosthesis
Introduction

Reconstruction of large diaphyseal bone defects in long bones is generally addressed by several treatment options which are generally divided into bone preserving and segmental bone substituting techniques (de Pablos et al, 1994). Bone preserving techniques involve vascularized or non vascularized autografts (Chang & Weber, 2004; de Pablos et al, 1994; Nair et al, 2009), allografts -stabilized with plates (Fuchs et al, 2008; Virkus ey al, 2008; Wingertet al, 2007), intramedullary nails (Fuchs et al, 2008; Virkus ey al, 2008; Wingertet al, 2007), - and distraction osteogenesis using unilateral rail external fixators (de Pablos et al, 1994) or Ilizarov apparatus (Fuchs et al, 2008; Hasenboehler et al, 2006; Meffert et al, 2000). Bone substituting techniques include the application of segmental diaphyseal implants (Aldlyami et al, 2005; Fuchs et al, 2008; Henry et al, 2002) or interposition of metallic spacers (Bullens et al, 2009; Bullens et al, 2009; Fujibayashi et al, 2003; Lindsey et al, 2006) or scaffolds [Kuzyk et al, 2009; Nair et al, 2009).

It is generally thought that using an endoprosthesis the operated limb could be ambulated more safely and rapidly [Fuchs et al, 2008; Henry et al, 2002] comparing to reconstruction with allografts which necessitates for more cautious primary mobilization until allograft incorporation is evident (Fuchs et al, 2008; Pollock et al, 2005). In the existing literature there are only few reports regarding the biomechanical evaluation of endoprostheses or the internal fixation of intercalary defects (Heck et al, 2010; Henry et al, 2002). However, there is no study comparing simultaneously all these reconstructive options on the same composite model with specifically sized bone defect.

The hypothesis of our study was that modular segmental implants should have better biomechanical properties comparing to internal and external fixation techniques. The purpose of our study was to verify this hypothesis by developing a biomechanical project consisting
of humeral composite models with specific mid-diaphyseal defects that were reconstructed using four different fixation options (modular intercalary implants, interposition of allograft and fixation with locking plates, intramedullary nails or external fixators). By submitting these constructs to different modes of loading we intended to reveal potential statistically significant differences among the collected data of axial, bending and torsional loads, axial stiffness, flexural and torsional rigidity of each construct.

Methods

Fourth generation bone composite models (Sawbones Europe AB, Pacific Reasearch Laboratories, Inc. Sweden) were selected instead of cadaveric bones as they simulate accurately the biomechanical behavior of human bones and lack the disadvantages of variety of structure, strength and bone density (Chong et al, 2007; Chong et al, 2007). A total of twenty-five (25) adult humeral specimens (large left art no #3404) were used in this biomechanical study [Figure 1].

In order to simulate a segmental diaphyseal bone defect, all composite models of the humerus were marked and cut to create identically intercalary defects regarding the size and location to the composite bone diaphysis. We resected the middle 1/5 of humeral diaphysis with a measured length of 7 cm. The bone defects were then managed using 4 different reconstructive options: (a) locking plates, (b) intramedullary nails, (c) external fixators, and (d) modular segmental implants. Specifically, we used the following implants: (a) standard low contact plates with locking screws (LCP plates, Synthes), (b) long humeral intramedullary nails (Polarus, Acumed), (c) mini unilateral rail compression-distraction external fixators (LRS, Orthofix, Italy), and (d) modular segmental implants (Osteobridge IDSF System, Merete, Germany). In the reconstructive options (a), (b), and (c) the resected medial segment of the composite bone diaphysis was used as an allograft, whereas in
reconstructive option (d) it was substituted by the metallic spacer of modular segmental implants [Figure 1a]. All implants were applied using standard surgical techniques and instrumentation.

In order to ensure the exact and repeatable determination of the diaphyseal axis of each bone for testing purposes, all composite models were inspected for proximal and distal diaphyseal canal holes, as resulting from their molding production [Figure 1b and c]. The humeral composite models were found with both distal and proximal guide holes. Steel accessories for axial compression loading models, were used to ensure single axis loading. [Figure 2] A preload of 50N was applied, zeroed and then followed by 5 cycles of 0-500N; the last three meant for data collection. Four-point-bending loading was applied through a special rig which was equipped with four (4) adjustable cylindrical steel supports. The frontal plane was almost horizontal (externally rotated by approx. 10 degrees), with the posterior aspect downwards. Thus, during bending the posterior aspect of the humerus models was in tension. The deficit was accommodated centrally on the machine. The upper support span was 150 mm and the lower one 220 mm. [Figure 3] A preload of 100N was applied, zeroed and then followed by 5 cycles of 0-400N; the last three meant for data collection.

Finally, torsion was applied using a special alignment and potting device in order to produce cylindrical embeddings for the metaphyseal regions of the composite models which were centered coaxially to their diaphyseal axis (by means of acrylic resin) [Figure 4]. Following several preliminary pilot loading cycles in both senses up to angulations confined within a linear elastic behaviour, specimens were returned to zero angulation. At this point they were tightened again and the current torque indication was monitored (always < 0.1 Nm) and zeroed. Subsequently, specimens were loaded in torsion, both in internal and external
rotation, here again up to angulations confined within a linear elastic behaviour of each bone model.

The loading was sequential following the same pattern and order for all specimens that were tested; the composite bone models were tested first in axial compression, then in four-point-bending and finally in torsion mode. The loads that were applied were within the linear elastic region at sub-yield level, as determined by a series of moderate preliminary pilot tests.

After each type of loading the specimens were examined for implant loosening, the presence of cracks as well as the need for additional torque up of the implant screws at the interface between screw and bone models. All tests were video recorded using a digital photo & video recorder (SONY DCR TRV 80E)

All of these reconstructive techniques were comparatively assessed and characterized, together with corresponding intact specimens, based on their stiffness response. Load (N) versus displacement (mm) curves were recorded for each test, involving as previously described 5 loading cycles. The slopes (in N/mm) of the ascending linear portion of the last three loading cycles were calculated by linear regression. For axial compression mode, slope values were coincident to axial stiffness (expressed in N/μm) values; the latter being also computed for each construct. For the four-point bending mode of loading, load (N) versus displacement (mm) curves were recorded for each test, involving 5 loading cycles. The slopes (in N/mm) of the ascending linear portion of the last three loading cycles were calculated by linear regression. Then, apparent flexural rigidity (in Nm2) values were computed. Finally, for the torsion mode of loading, torque (Nm) versus angle (degrees) curves were recorded for each test, involving 3 data collection loading cycles. The slopes (in Nm/degree) of the ascending linear portion of the last three loading cycles were calculated by linear regression,
in both internal and external rotation. Then, apparent torsional rigidity (in Nm²/d degree) values were computed. For the above values, average, standard deviation and coefficient of variation (i.e. average/st.dev.) were computed for each construct.

Statistical analysis

One way analysis of variance (ANOVA) and a post hoc Tukey test were used to test for any significant differences between the mean values of axial load, four point bending load, torsion, axial stiffness, flexural rigidity and torsional rigidity for the intact models as well as those with the diaphyseal defects that were reconstructed with 4 different options. A Levene’s test demonstrated that there was no significant difference between the variances of these sets of data and hence it was deemed appropriate to use this parametric test. Moreover, the data for each construct were ranked (i.e. the higher value ranked 1, the second 2 etc) according to their biomechanical outcome at each mode of loading (i.e. axial loading, bending and internal-external loading). The Friedman one-way ANOVA test was used to find out whether there was statistically significant difference in the overall ranking of techniques and thereafter a post hoc comparison of groups with Tukey multiple comparison procedure at 95% level of significance (α=0.05), in order to reveal statistical differences among techniques in their ranking.

Results

All specimens were loaded within their linear elastic region at sub-yield level; thus, there were no events of implant or bone composite model failures. Although the composite bone models were loaded sequentially under axial compression, 4-point-bending and axial torsion, there was no effect on implant bone interface. No implant or screw loosening was detected after each mode of loading. No torque-up of screws was needed.
The mean loads for axial compression, four point bending, internal and external torsion, as well as the corresponding values of compression stiffness, flexural and torsional rigidity and the 95% confidence intervals for each mean are presented in table 1. One way analysis of variance (ANOVA) showed that there was a statistically significant difference in each set of data (load values and load to displacement values) Specifically the F ratio of axial compression loading was 5023 \((p=0.001)\), of bending loading 6.215 \((p=0.001)\), internal torsion 4048 \((p=0.001)\), and external torsion 2216 \((p=0.001)\). Moreover, the F ratio for axial compression stiffness was 3131 \((p=0.001)\), for flexural rigidity 6.415 \((p=0.001)\), and for torsional rigidity 2633 \((p=0.001)\) in internal rotation and 2197 \((p=0.001)\) in external rotation.

A post hoc comparison of groups was performed using the Tukey-Kramer multiple comparison procedure. We found that, in axial compression loading, modular segmental implants and intramedullary nails could withstand statistically significantly higher loads comparing to external fixation and plates \((p=0.031)\). In the four point bending mode of loading, intramedullary nail resisted statistically significantly lower loads comparing to modular segmental implants, plates and external fixators \((p=0.038)\). In both internal and external torsion segmental implants and external fixators received significantly greater torsional loads comparing to plates and intramedullary nails \((0.023)\). In the figures 5a-c, the load to displacement curves for the different types of loading (axial compression, four-point bending, and axial torsion) are presented. Compression stiffness was statistically significantly lower for the external fixation devices and plates comparing to modular segmental implants and intramedullary nails \((p=0.001)\) (Figure 6a). Flexural rigidity was statistically significantly lower for the intramedullary nails comparing to other constructs \((p=0.012)\) (Figure 6b). Torsional rigidity in both internal and external rotation were significantly greater
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in segmental implants and external fixators comparing to intramedullary nails and plates

\( p=0.001 \) (Figure 6c).

The above results on the biomechanical behavior of the intact model, and the 4 methods for the reconstruction of the mid-diaphyseal humeral were ranked for each mode of loading (i.e. compression, four-point-bending, and torsion) as shown in Table 2. In this table the values of the intact model assigned the highest order (rank 1) followed by the ranking of the other reconstructive options according to the achieved values. Using the Friedman one-way ANOVA test, we found that there was statistically significant differences in the ranking of techniques \( p=0.038 \). The intramedullary nails and the locking plates showed, overall, significantly worse biomechanical behavior comparing to the intact models, the modular segmental implants and the external fixators \( p=0.038 \).

Discussion

A project was undertaken in order to provide comparative quantitative data on the biomechanical outcome of four different reconstructive techniques used for the surgical management of large diaphyseal bone defects in the humerus. Thus, modular segmental implants, intramedullary nails, locking plates, and external fixators were applied in composite models of adult humeri with intercalary defects and tested in compression, bending and torsional modes of loading. All techniques were compared to identical intact specimens. This methodology was consistently applied to each group of specimens and provided sound preliminary evidence for the biomechanical outcome of each construct. Modular segmental implants and intramedullary nails were able to compensate significantly greater amounts of compressive loads comparing to locking plates and external fixators. However, in flexion and torsion, the modular segmental implants and the external fixators were significantly better load-bearing devices comparing to the intramedullary nails and plates.
Our study has several limitations. The number of specimens for each reconstructive option (i.e. locking plate, external fixator, intramedullary nail and modular endoprosthesis) was limited to 5 per construct. However, the use of composite bone models reduces the variability of biomechanical behavior, as they are associated with standardized biomechanical characteristics (Chong et al, 2007; Chong et al, 2007). Another limitation could be the selection of sequential loading of the specimens under physiologic loads only, which did not exceed the elastic yield point and did not have failure as end point. However, no sequential effect was observed. Moreover, we achieved a significant reduction of the total number of implants and composite bone models that should have been required if a destructive loading of the specimens had been selected. Finally, the clinical interpretation of these results could be difficult as they only correspond to the early phases of rehabilitation. The potential of allograft incorporation and/or bone bridging over the modular endoprostheses (which is usually present within the first 6-12 months postoperatively) could alter the initial biomechanical properties of each construct. However, early rehabilitation of patients with large humeral defects (which are usually associated with bone tumors) is the definite goal, and biomechanically this situation better corresponds to the primary, initial state of fixation which was examined here.

Strengths of our study include the relatively large number of reconstructive options that were used in our study (i.e. locking plates, intramedullary nails, external fixators and modular segmental implants) which were applied in a consistent way and were compared in terms of biomechanical characteristics to the intact specimen with identical dimensions and bone quality.

The use of composite bone models is generally favored in the current literature for biomechanical evaluation and testing (Chong et al, 2007; Chong et al, 2007). They are
associated with great reproducibility, accuracy and standardization of constructs as any relevant technical, anatomical and biomechanical bias of in vivo or cadaveric tests is avoided (Luca Christofolini & Viceconti, 2000). Their mechanical properties for axial rigidity, torsional rigidity, and cortical bone screw purchase are similar to cadaveric femora (Heiner & Brown, 2001). In our study we used the fourth generation composite bone models (Sawbones Europe AB, Pacific Reasearch Laboratories, Inc. Sweden) which have the same geometries as the third-generation bones, but the cortical bone analogue material was changed to one with increased fracture and fatigue resistance, tensile and compressive properties, thermal stability, and moisture resistance (Chong et al, 2007). In a recent study model femurs and tibias were tested under bending, axial, and torsional loading and the longitudinal strain distribution were determined. The fourth-generation composite bones had average stiffness and strains that were for the most part closer to corresponding values measured for natural bones, than was the case for third-generation composite bones (Chong et al, 2007; Heiner & Brown, 2001; Luca Christofolini & Viceconti, 2000). It is significant that variability between the specimens is reported to be less than 6% (Heiner, 2008).

In our knowledge, there are no studies comparing the biomechanical characteristics of different treatment options of intercalary defects in the current literature. Biomechanical analyses of several reconstructive options have been individually presented by some authors reporting on the reconstruction of defects of the humerus (Heck et al, 2010; Henry et al, 2002). Torsional rigidity and peak torque are the most studied biomechanical parameters (Heck et al, 2010; Henry et al, 2002). Endoprostheses are generally characterized by greater peak torque and peak rotation at failure comparing to intramedullary nailing however data on torsional stiffness are controversial (Heck et al, 2010; Henry et al, 2002). Intramedullary nailing is associated with significantly higher rotation to failure, peak torque, stiffness, and
total energy absorbed to failure compared to plating (Chen et al, 2002; Damron et al, 1999; Wingerter et al, 2007).

Henry et al (Henry JC, 2002) conducted a study comparing the biomechanical properties of a second generation titanium modular intercalary humeral spacer with those of a modern locked humeral nail combined with methylmethacrylate (intramedullary nail) or with an intercalary allograft spacer (allograft nail composite) for fixation of segmental defects of the humeral diaphysis. The authors utilized fresh frozen humeri (18 matched pairs) with a 5-cm mid-diaphyseal defect. Specimens were tested in external torsion to failure. It was shown that the replacement specimens used for reconstruction of segmental defects had statistically greater peak torque and stiffness than the intramedullary nails or the allograft nail composites. The authors concluded that reconstruction of humeral diaphyseal defects with a cemented metallic intercalary spacer provides significantly greater immediate stability than interlocked intramedullary nail fixation supplemented with segmental methylmethacrylate or intercalary allograft reconstruction. The use of cadaveric specimens, the size of the diaphyseal defects and the application of destructive loads to failure in torsion are the main differences of this study comparing to our experimental protocol.

In another study, Heck et al. (Heck et al., 2010) compared the mechanical properties of a humeral segmental replacement prosthesis with an IM nail in a cadaver model simulating an impending pathologic fracture. Nine matched pairs of fresh human humeri with a 50% lateral mid-diaphyseal defect were reconstructed with segmental defect prosthesis or IM nail and bone cement. Peak torque and peak rotation at failure were greater for the prosthesis specimens whereas torsional stiffness was greater for the IM nail specimens. In conjunction with the DEXA results of the specimens the authors concluded that there was a linear relationship between peak torque and T-score for each device, the construct with the
prosthesis could withstand greater forces than the IM nail and the differences between devices were greater in weaker bones. In a relevant experiment, Damron et al. (Damron et al., 1999) compared the biomechanical properties of prophylactic fixation of midshaft humeral defects combined with cementation and dynamic compression plating, Rush rodding, or locked intramedullary nailing. The authors showed that the proximally and distally locked intramedullary nail had biomechanical advantages, regarding rotation to failure, peak torque, stiffness, and total energy absorbed to failure, in the prophylactic stabilization of an impending pathologic fracture of the central 1/3 of the humerus. Our conclusions were similar to these two studies although these projects had significant differences comparing to ours, such as the use of cadaveric fresh humeri instead of synthetic bone models, the type of defect which was lateral and partial instead of complete mid-diaphyseal, and the type of loading which was destructive to failure instead of loading at the elastic region of the constructs.

In another concept of biomechanical testing of humeral defects, Henley et al. (Henley et al., 1991) performed an experimental study on the biomechanical comparison of methods of fixation of a midshaft osteotomy of the humerus. In posterior and lateral bending and in torsion, flexible intramedullary pin fixed humeri (Enders and Hackethal) performed similarly and were less stiff than intact specimens were. Interlocking intramedullary nail constructs (Russell-Taylor and Seidel) were also tested similarly to each other, and were stiffer than the flexible pins in all bending tests. Compared with the intact humerus, interlocking nails were stiffer in torsion, but in bending they simulated the stiffness of the bone more closely.

In an adaptation of these biomechanical experiments to trauma scenarios with large humeral defects or comminution, Chen et al. (Chen et al., 2002) examined the fixation stability of comminuted humeral shaft fractures using locked intramedullary nailing or plate
fixation. Six matched pairs of human humeri received either a 10-hole broad dynamic compression plate or a locked antegrade inserted humeral nail applied to a humeral diaphyseal osteotomy with a 1.5-cm gap defect Cyclic loading showed no difference between the two groups for average gap displacement or construct stiffness. The intramedullary nail constructs failed by humeral shaft splitting (n = 4) or head cut-out (n = 2) at an average of 958.3 N, whereas the plate constructs failed by humeral shaft splitting and screw pull-out (n = 3) or plate bending (n = 3) at an average of 641.7 N (p < 0.001). The authors concluded that, although both methods offered similar fixation stability under physiologic loads, the higher load to failure demonstrated by intramedullary nail fixation might have implications for the patient with multiple injuries for whom partial weightbearing on the injured upper extremity might be necessary. Altered fixation stability of the constructs could be produced when they are loaded under different amounts and modes of loading. The clinical interpretation of our results is difficult and problematic. In our study, the constructs were loaded in sequence under different modes of loading, so we had to limit the amount of loading only to physiologic sub yield loads within the elastic zone in order to prevent the sequential effect such as implant loosening and degrading of constructs. It could be interesting to find out in future projects if the observed biomechanical behavior of the constructs under higher loads is significantly altered.

In conclusion, in this study we tried to simulate biomechanically a specific diaphyseal segmental bone defect of the humerus, and to reconstruct the defect using widely acceptable reconstructive options. Specifically, four different techniques: plating with locking screws, intramedullary nailing, external fixation and modular segmental implants were used in composite models of adult humeri with intercalary defects and were tested in compression, bending and torsion mode. Overall, modular segmental implants showed a significantly
greater potential of loading in compression, bending and torsion in the early phases of limb rehabilitation. Having reached these preliminary conclusions, further research protocols could be developed in order to enrich the present biomechanical data. Similar biomechanical testing could be performed on larger specimen populations regarding (a) the variety of diaphyseal defects (i.e. the size of affected segment), (b) the amount of loads to be applied (i.e. loading up to failure) and (c) the inclusion of other reconstructive options.
References


Figure 1a-c. (a). Figure showing 5 composite bone models (4th generation Sawbones Europe AB, Pacific Research, large left art. No #3404): (i) reconstruction of the mid diaphyseal defect with a 12 hole locking plate, (ii) reconstruction using a mono-lateral external fixator, (iii) reconstruction using an intramedullary nail, (iv) reconstruction with a modular segmental fixation implant and (v) intact model. (b&c). All composite bone models were inspected for proximal (b) and distal (c) diaphyseal canal holes in order to ensure the exact and repeatable determination of the diaphyseal axis.
Figure 2. Figure showing all the specimens loaded under axial compression. The proximal guide hole on the humeral head was loaded through an appropriately sized ball and the distal guide hole through a contoured solid steel cone.
Figure 3. Figure showing the reconstructed and the intact models mounted on the four-point bending device.
Figure 4. For the axial torsion testing, all specimens had to be proximally and distally embedded in a solid medium (potting devices and acrylic resins) in order to secure placement on the torsion loading device.
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**Figure 5a.** Loading curves (load to displacement values) for: (a) axial compression.

**Figure 5b.** Loading curves (load to displacement values) for: four-point bending.
**Figure 5c.** Loading curves (load to displacement values) for: axial torsion of the intact model (blue line), and the four different reconstructive options (modular implant with the yellow line, IM nail with pink, locking plate with light blue and external fixator with deep purple). In the torsion mode of loading, positive values are for internal rotation and negative values for external rotation.

**Figure 6a.** Schematic diagram of: the axial stiffness in N/μm,
Figure 6b. Schematic diagram of: the flexural rigidity in Nm²
Figure 6c. Schematic diagram of: the torsional rigidity in Nm²/degrees of the intact model, the modular implant, the IM nail, the locking plate and the external fixator constructs. The lines connecting the bars of the represent the paired comparison of mean values between the intact model and the different constructs (blue lines), the modular segmental implant and the other constructs (red lines), the IM nail versus locking plate and external fixator (green lines), and the plate versus external fixator (black line). Asterisks are placed where a statistically significant difference exists.
Table 1. Table showing the mean loads for axial compression, four-point bending, internal and external torsion, compression stiffness, flexural and torsional rigidity and the 95% confidence intervals (CI) for each mean value.

<table>
<thead>
<tr>
<th></th>
<th>Intact Model</th>
<th>Modular Segmental Implant</th>
<th>IM nail</th>
<th>Locking plate</th>
<th>External fixator</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean Value</td>
<td>95% CI</td>
<td>Mean Value</td>
<td>95% CI</td>
<td>Mean Value</td>
</tr>
<tr>
<td>Axial Compression (N)</td>
<td>973</td>
<td>968.3-976.7</td>
<td>919</td>
<td>915.1-923.6</td>
<td>816</td>
</tr>
<tr>
<td>Four Point Bending(N)</td>
<td>835</td>
<td>833.3-836.7</td>
<td>629</td>
<td>627-630.3</td>
<td>320</td>
</tr>
<tr>
<td>Internal Torsion (Nm)</td>
<td>2.66</td>
<td>2.650-2.674</td>
<td>1.37</td>
<td>1.358-1.382</td>
<td>0.84</td>
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<tr>
<td>External Torsion (Nm)</td>
<td>2.59</td>
<td>2.564-2.616</td>
<td>1.268</td>
<td>1.242-1.294</td>
<td>0.73</td>
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<tr>
<td>Compression Stiffness (N/μm)</td>
<td>0.97</td>
<td>0.965-0.975</td>
<td>0.92</td>
<td>0.915-0.925</td>
<td>0.816</td>
</tr>
<tr>
<td>Flexural Rigidity (Nm²)</td>
<td>44.37</td>
<td>44.3-44.4</td>
<td>33.32</td>
<td>33.24-33.39</td>
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<td>Internal Torsional Rigidity (Nm²/degree)</td>
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<td>External Torsional Rigidity (Nm²/degree)</td>
<td>0.75</td>
<td>0.743-0.757</td>
<td>0.388</td>
<td>0.381-0.395</td>
<td>0.24</td>
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</tbody>
</table>
Table 2. Ranking of the axial stiffness, flexural rigidity and torsional rigidity for the intact model, the modular endoprosthesis, the locking plate, the intramedullary nail and the external fixator.

<table>
<thead>
<tr>
<th></th>
<th>Compression Stiffness (ranking)</th>
<th>Flexural rigidity (ranking)</th>
<th>Torsional rigidity (ranking)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Intact model</td>
<td>1</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>Modular implant</td>
<td>2</td>
<td>2</td>
<td>2</td>
</tr>
<tr>
<td>IM nail</td>
<td>2</td>
<td>2</td>
<td>2</td>
</tr>
<tr>
<td>Locking plate</td>
<td>3</td>
<td>2</td>
<td>3</td>
</tr>
<tr>
<td>External fixator</td>
<td>4</td>
<td>3</td>
<td>3</td>
</tr>
<tr>
<td>Internal</td>
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<td>1</td>
<td>1</td>
</tr>
<tr>
<td>Modular implant</td>
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<td>2</td>
<td>2</td>
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<tr>
<td>External fixator</td>
<td>2</td>
<td>2</td>
<td>2</td>
</tr>
<tr>
<td>Locking plate</td>
<td>3</td>
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