Polyaxial Locking Plate Fixation in Distal Femur Fractures: A Biomechanical Comparison

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Objectives: Uniaxial, first-generation locking plates have become increasingly popular for fixation of supracondylar femur fractures. Polyaxial plates are currently available, which allow for variable-angle screw insertion; however, the biomechanical integrity of these new locking constructs is yet unproven. This study compares the mechanical stability of a conventional locking plate with that of a new polyaxial design.

Methods: A comminuted supracondylar femur fracture (AO/OTA33-A3) gap model was created in fourth-generation synthetic composite bones. Fixation was obtained with 2 different plate constructs: (1) a conventional locking plate (uniaxial screw heads threading directly into plate) and (2) a polyaxial locking plate (screw heads are captured and "locked" into a fixed angle using locking caps). Eight specimens of each type were then tested in axial, torsional, and cyclic axial modes on a material testing machine.

Results: The mean axial stiffness for the polyaxial locking plate was 24.4% greater than the conventional locking plate (168.2 vs 127.1 N/mm; $P < 0.0001$). The mean torsional stiffness was also greater for the polyaxial plate (2.78 vs 2.57 Nm/degree; $P = 0.0226$). Cyclic axial loading caused significantly less ($P = 0.0034$) mean irreversible deformation in the polyaxial plate (5.6 mm) than in the conventional plate (8.8 mm). The mean ultimate load to failure was significantly higher ($P = 0.0005$) for the polyaxial plate (1560 N) than for the conventional plate (1337 N).

Conclusions: The tested plate construct with its polyaxial locking screw mechanism provides a biomechanically sound fixation option for supracondylar femur fractures. The frictional locking mechanism allows maintenance of angular stability while affording the option of variable screw placement.

Key Words: biomechanics, locking plate, polyaxial, femur

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INTRODUCTION

Unstable supracondylar fractures of the distal femur are a common and complicated surgical challenge often resulting from high-energy trauma. Improvements in implants have increased the ability to effectively stabilize these fractures, and operative treatment results in a significant reduction in poor results compared with nonoperative management.[1] Locking plate fixation has become one standard method of treatment for comminuted supracondylar femoral fracture fixation allowing rigid internal fixation of meta-diaphyseal fractures from one surgical approach while still permitting the surgeon to address significant articular involvement. Recent studies have demonstrated these implants to be stronger than a fixed-angle blade plate, with the advantages of simplified application and the potential for minimally invasive insertion.[2] Additional studies have demonstrated the ability of locked plates to absorb more energy before failure than either an angled blade plate or a retrograde intramedullary nail while exhibiting a lower incidence of loss of distal fixation.[3,4] Disadvantages to conventional locked implants include uniaxial screw paths predetermined by the manufacturer. These screw trajectories cannot account for differences in femoral anatomy, fracture pattern, or variation in plate positioning.

Newer polyaxial screw-plate designs are available that allow the surgeon greater variation in fixation constructs and ability to perform targeted percutaneous fracture fixation. These plates have significant advantages over earlier designs in that they allow placement of bicortical locking screws, providing a longer working length while allowing a range of insertion angles.[5] A number of polyaxial interfaces are available. One uses an expanding bushing contained within the plate (PolyAx; DePuy Orthopaedics, Warsaw, IN), another uses freely rotating rings that are snapped into the screw holes (Numelock; Stryker, Mahwah, NJ), whereas a third uses a locking cap placed after screw insertion that converts a standard cortical or cancelous screw into a fixed-angle screw via frictional interface (NCB; Zimmer, Warsaw, IN).

Clinical results have been reported with this polyaxial technology (DePuy Orthopaedics) and demonstrated resistance to varus drift and good clinical performance.[6] To date, however, no studies have been published directly comparing the biomechanical integrity of these new polyaxial plating constructs with that of conventional locking plates. As use of these plates continues in the clinical setting, the purpose of this study is to compare the mechanical stability of a conventional locking plate construct with that of a new polyaxial locking construct in a metaphyseal distal femoral gap model.

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We hypothesize that there will be no difference in stiffness, deformation, or load to failure between the NCB (a locking, polyaxial distal femoral plate) and a conventional, uniaxial locked distal femoral plate in a metaphyseal–diaphyseal gap model.

MATERIALS AND METHODS

Fracture Model

A custom jig was used to create a reproducible osteotomy on fourth-generation femoral synthetic composite bones (Sawbones 3406; Pacific Research Laboratories, Inc., Vashon, WA). A 1-cm gap was created 6 cm proximal to the intercondylar notch to mimic an unstable fracture pattern with loss of stability of the medial and lateral columns of the distal femur yielding a simulated AO/OTA33-A3 fracture. Additionally, a diagonal cut was made in the proximal medial cortex to prevent contact during testing.[3,7] Three millimeters of the lateral femoral cortex was left intact during creation of the osteotomy to facilitate instrumentation.

Construct Design

All instrumentation was performed by a single fellowship-trained orthopedic trauma surgeon. Plates were placed in the ideal anatomic position (per manufacturer's recommendation) on the anterior–lateral surface of the distal femoral metaphysis. Plate span ratio and screw density were optimized in the diaphyseal segment by using long plates and alternating holes.[5] Metaphyseal screws were measured 5 mm short of their bicortical length to prevent hardware protrusion. Eight specimens were instrumented for each group.

Group 1, Conventional Uniaxial Locking Plate

A 10-hole uniaxial distal femoral locking plate (Periarticular Distal Lateral Femoral Locking Plate; Zimmer) was fixed to the shaft using four 4.5-mm solid bicortical locking screws. Distal to the comminuted segment (osteotomy), 5 of the metaphyseal screw holes were filled with 5.5 mm cannulated locking screws, plus the oblique screw for a total of 6 locking screws distal to the fracture (Figs. 1 and 2).

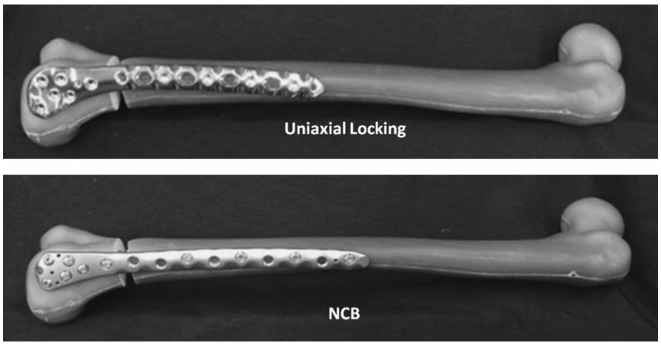

FIGURE 1. Top—conventional uniaxial locking plate construct with 6 distal locking screws and 4 bicortical shaft screws. Bottom—NCB polyaxial locking plate construct with 6 distal locking screws and 4 bicortical shaft screws.

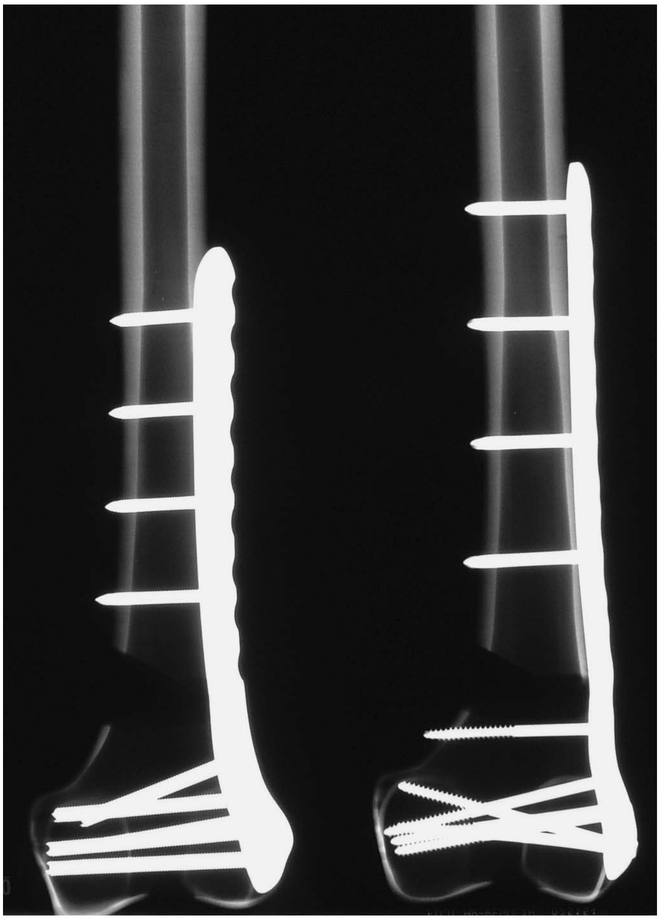

FIGURE 2. X-Ray of instrumented constructs. Left—conventional uniaxial plate. Right—NCB polyaxial plate.

Group 2, Polyaxial Locking Plate

A 9-hole polyaxial locking plate (NCB; Zimmer) was fixed proximal to the osteotomy site using four 5.0-mm bicortical screws. Distal to the comminuted segment (osteotomy) all 5 metaphyseal screw holes were filled with 5.0 mm of partially threaded cancellous screws, plus the oblique screw for a total of 6 locking screws distal to the fracture (Figs. 1 and 2). A standard perpendicular locking guide was threaded into the plate to provide a reproducible screw trajectory for all screws. After being fully seated, locking caps were then applied to all screws using a torque-limiting screwdriver (6 N) provided with the equipment.

Mechanical Testing

After instrumentation, removal of the osteotomy fragment was performed with a band saw, care being taken not to damage the adjacent plate. Any damaged plate was excluded. The proximal and distal ends of the Sawbones femurs were then secured in custom-built polymethylmethacrylate molds.[7] Torsional testing was performed using a proximal custom mold and a chuck to secure the distal end with the femoral axis in line with the axis of rotation. For axial loading, the potted femora were arranged such that the force vector went through the center of the femoral head and the intercondylar notch, simulating the mechanical axis of the femur. Additionally, ball bearing joints were used proximally and distally to avoid torque or bending moments on the specimens during testing.[7,8] All loading tests were performed using a universal materials testing machine (Instron 5800, Canton, MA).

Sequential testing proceeded from nondestructive torsional stiffness measurement to nondestructive axial stiffness measurement, followed by cyclic axial loading to construct failure.

TESTING PROTOCOL

Torsional Loading

The specimens were preloaded to 5 Nm, and each construct was torqued to a maximum of 20 Nm at an approximate rate of 20 degrees/min. The force vector was an internal rotation of the femur relative to the plate. Testing was halted once 20 Nm had been reached, before any visual loss of fixation.

Axial Loading

The constructs were stabilized with a preload of 100 N, before performing axial loading in a displacement control mode. The specimens were then loaded in compression at a loading rate of 10 mm/min. Testing was stopped when 500 N was reached, before any visual loss of fixation.

Cyclic Axial Loading

We used a loading protocol well described for mechanical evaluation of distal femoral fractures.[3,7,9] It consisted of increments of 10 cycles, starting with a 300-N load. The load of each following increment was increased by 100 N, to a maximum load of 1700 N, allowing 10 s of rest between each increment. The preload and baseline load after each cycle was 100 N. Testing was conducted in a displacement control mode at 0.75 mm/s and was performed until either of the following was observed:

1. Medial fracture gap closure and plate bending
2. Acute change in load–displacement curve indicating rapid change in displacement and loss of construct stability.

This method of cyclical testing was used to compare the strength of the locking caps in providing angular stability and to assess differences in irreversible deformation and load to failure.

Data Recording and Statistical Analysis

Load cell data were recorded using Bluehill 2 software (Instron 5800). For axial and torsional testing, a load–displacement curve was plotted for each construct (Excel; Microsoft, Seattle, WA) and stiffness was calculated as the slope of the linear portion of the curve. Load to failure was determined from the cyclic loading graph at the point where the graph flattened out indicating maximal displacement for a given load. Total deformation was calculated by subtracting the amount of displacement present at the start of the first cycle (300 N) from that displacement observed at the point of failure. Irreversible deformation was calculated by subtracting the initial displacement from displacement present after reaching yield point once the load was removed (Fig. 3).

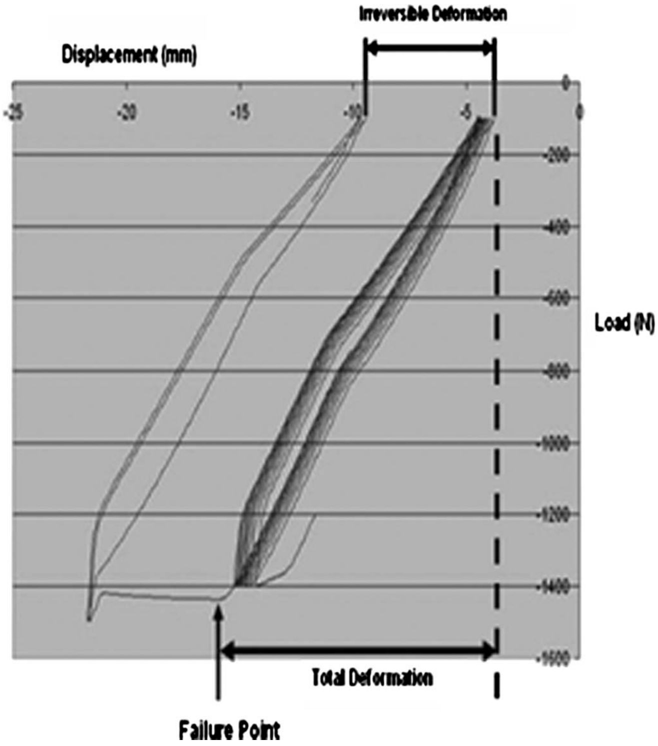

FIGURE 3. Representative cyclic loading graph depicting load to failure, total deformation, and irreversible deformation.

A Student *t* test was performed using Excel (Microsoft) to determine statistically significant differences in axial and torsional stiffness, load to failure, and amount of irreversible and total deformation between each group. The level of significance was defined as $P \leq 0.05$.

RESULTS

No catastrophic failure (plate breakage, screw pullout/breakage) was noted in either group.

Axial/Torsional Loading

No visual loss of fixation or fracture gap closure appeared in either group. The mean axial stiffness for the polyaxial locking plate was 24.4% greater than the conventional locking plate (168.2 vs 127.1 N/mm; $P < 0.0001$). The mean torsional stiffness for the conventional locking plate was 2.57 Nm/degree, whereas that of the polyaxial plate was 2.78 Nm/degree. This difference was also statistically significant ($P = 0.0226$) (Table 1).

Cyclic Axial Loading

Failure of the construct was observed in all specimens in both groups. Cyclic axial loading caused significantly less ($P = 0.0034$) mean irreversible deformation in the polyaxial plate (5.6 mm) than in the conventional plate (8.8 mm). The mean ultimate load to failure was significantly higher ($P = 0.0005$) for the polyaxial plate (1560 N) than for the conventional plate (1337 N). No significant differences were observed

TABLE 1. Stiffness, Load to Failure, and Deformation of Uniaxial and Polyaxial Plate Constructs

	Implant	
	Conventional Plate	**Polyaxial Plate**
Axial stiffness(N/mm)		
Mean	127.1	168.2
SD	5.7	7.6
Difference	41.1	
P value*	<0.001	
Torsional stiffness (Nm/degree)		
Mean	2.57	2.78
SD	0.19	0.12
Difference	0.21	
P value*	0.0226	
Load to failure (N)		
Mean	1337	1560
SD	109	32
Difference	223	
P value*	0.0006	
Total deformation (mm)		
Mean	14.3	14.0
SD	1.7	1.5
Difference	0.2	
P value*	0.7869	
Inreversible deformation (mm)		
Mean	8.8	5.6
SD	2.1	1.3
Difference	0.8	
P value*	0.0034	

*Student *t* test.

in total deformation (14.3 vs 14.0) between the constructs ($P = 0.7869$) (Table 1).

DISCUSSION

Complex fractures of the distal femur are a frequent and challenging injury encountered by the orthopedic surgeon. Although all fractures are unique, there are several common fracture patterns. One such pattern includes severe metaphyseal comminution, precluding fixation into this region necessitating bridging fixation. Uniaxial, first-generation locking plates have become increasingly popular for fixation of supracondylar femur fractures providing fixed-angle placement of the locking screw as determined by the plate design. Recent plate designs now allow the surgeon to aim locking screws in a fracture-specific direction, which is particularly crucial when minimal distal bone stock exists or in the setting of periprosthetic fractures.

This study compared the stability of the NCB polyaxial locking plate design with a comparable conventional locking screw construct. The polyaxial plate was found to be stiffer in axial and torsional loading and exhibiting less irreversible deformation and higher loads to failure.

Many biomechanical studies often correlate the stability of a construct with its stiffness,[4,7,10–12] and a stiffer construct is thought to be more stable thus allowing less motion at the fracture site.[7] What constitutes the stability or point of failure in a bone-implant construct has not been rigorously defined.[3]

Some published studies use a point of sharp angulation of the load–displacement curve,[3] whereas others use a arbitrary offset line drawn parallel to the linear portion of the curve and measure failure where the lines intersect.[3,13] Additionally, some authors rely on gross visual failure of the construct using gap closure, screw pullout, or bone fracture to constitute failure.[2,7,14] Still others compare implant strength based on deformation at a maximum tested load, not completely progressing to implant failure.[4]

We chose a consistent point on the load–deformation curve where the construct appeared to yield, deviating from a regular pattern of loading and unloading only to deform irreversibly at a given load (Fig. 3). The patterns seen for the 2 implants were consistent allowing us to reliably use this as a definition of failure in our model. Additionally, the values and patterns at which we observed failure are consistent with others in the current literature. In a similar experimental fracture model using a titanium fixed-angle implant Less Invasive Stabilization System (LISS), Zlowodzki et al found implant failure by plastic (irreversible) deformation at 1229 N in axial loading and demonstrated titanium LISS plate fixation to have more reversible deformation and higher maximum load to failure than stainless steel plates.[3]

Adding clarity to the clinical observations of Haidukewych et al[6] regarding the fact that they did not see hardware failure in their specimens because of "the superior fatigue strength of anodized titanium" and that the "polyaxial screws are not cannulated," we found that the screws are not the site of failure regardless of the material used. In our study, all plates failed by irreversibly deforming at the site of artificial comminution similar to the mode of failure in other studies.[3,7,13] There was no screw failure noted, whether the screws were cannulated or not. We agree that polyaxial screw constructs in the distal fracture segment are an effective method of fracture stabilization; however, further testing is needed to determine the best interface for the locking mechanism and to determine a threshold failure number above which additional strength is superfluous.

A limitation of this study is that it was not conducted with cadaveric bones. Biomechanical studies cannot replicate the effects of soft tissue attachments on structural properties and do not account for effects of bone remodeling after internal fixation nor can the results be directly extrapolated to clinical application.[14] However, there are some advantages to this method, as the use of synthetic bones for mechanical testing is well established.[7,14–19] Compared with cadaveric specimens, the use of synthetics eliminates the variability in size and bone quality inherent in cadaveric specimens. Additionally, most cadaveric femora are from the elderly, whereas the fracture pattern studied is primarily seen in younger individuals with higher bone densities. Utilizing a high-quality bone specimen more closely represents the clinical treatment of this fracture pattern and more adequately tests the strength of the implant without being limited by osteoporotic bone quality. As the purpose was to compare the stability of different fixation methods, minimizing the variability of the bone substrate was paramount.

Another potential limitation is the comparison of 2 implants of different metallurgic composition. Modern titanium alloys have a modulus of elasticity closer to human bone than stainless steel, and in improved fatigue life.[6] In a cyclic axial loading model, this improved fatigue life could potentially confound the load to failure data comparing 2 dissimilar materials. Furthermore, testing was conducted of 2 different plate designs. Although plate length, thickness, and screw placement were matched as closely as possible, there were slight differences. Also, the variation in distribution and diameter of the screws could alter biomechanical properties. Lastly, our model and loading protocol were designed to evaluate the construct as a whole and not individual screw interfaces. So although a single conventional threaded screw mechanism may have more "stability" than a frictional interface at the screw-plate junction, the friction construct as a whole can provide adequate or superior stability. Although the difference in stability of the constructs may be attributable to a number of factors, our study demonstrates that the 2 plate/screw designs fail in the same manner (with plate failure at the gap site) rather than at the screw interfaces, which has been the subject of debate.

Despite these limitations, we feel that our study was able to demonstrate a consistent difference in construct stability compared with the first-generation uniaxial locking plate in our experimental model. Not only was the construct stability similar but also the polyaxial plate with captured bicortical screws outperformed the conventional plate in all tested parameters except total deformation. Although these differences were statistically significant under the tested conditions, their clinical significance is yet to be determined. This new frictional locking mechanism results in a construct that is sufficiently stable while allowing the surgeon greater options in screw targeting, thereby providing an alternative fixation option for complex supracondylar femur fractures or periprosthetic fractures.

REFERENCES

1. Zlowodzki M, Bhandari M, Marek DJ, et al. Operative treatment of acute distal femur fractures: systematic review of 2 comparative studies and 45 case series (1989 to 2005). *J Orthop Trauma*. 2006;20:366–371.
2. Higgins TF, Pittman G, Hines, J, et al. Biomechanical analysis of distal femur fracture fixation: fixed-angle screw-plate construct versus condylar blade plate. *J Orthop Trauma*. 2007;21:43–46.
3. Zlowodzki M, Williamson, S, Cole PA, et al. Biomechanical evaluation of the less invasive stabilization system, angled blade plate, and retrograde intramedullary nail for the internal fixation of distal femur fractures. *J Orthop Trauma*. 2004;18:494–502.
4. Koval KJ, Hoehl JJ, Kummer FJ, et al. Distal femoral fixation: a biomechanical comparison of the standard condylar buttress plate, a locked buttress plate, and the 95-degree blade plate. *J Orthop Trauma*. 1997;11:521–524.
5. Gautier E, Sommer C. Guidelines for the clinical application of the LCP. *Injury*. 2003;34(Suppl 2):B63–B76.
6. Haidukewych G, Sems SA, Huebner D, et al. Results of polyaxial locked-plate fixation of periarticular fractures of the knee. *J Bone Joint Surg Am*. 2007;89:614–620.
7. Khalafi A, Curtiss S, Hazelwood S, et al. The effect of plate rotation on the stiffness of femoral LISS: a mechanical study. *J Orthop Trauma*. 2006;20: 542–546.
8. Cordey J, Borgeaud M, Frankle M, et al. Loading model for the human femur taking the tension band effect of the ilio-tibial tract into account. *Injury*. 1999;30(Suppl 1):A26–A30.
9. Marti A, Fankhauser C, Frenk A, et al. Biomechanical evaluation of the less invasive stabilization system for the internal fixation of distal femur fractures. *J Orthop Trauma*. 2001;15:482–487.
10. Jaakkola JI, Lundy DW, Moore T, et al. Supracondylar femur fracture fixation: mechanical comparison of the 95 degrees condylar side plate and screw versus 95 degrees angled blade plate. *Acta Orthop Scand*. 2002;73: 72–76.
11. Hora N, Markel DC, Haynes A, et al. Biomechanical analysis of supracondylar femoral fractures fixed with modern retrograde intramedullary nails. *J Orthop Trauma*. 1999;13:539–544.
12. Koval KJ, Kummer FJ, Bharam S, et al. Distal femoral fixation: a laboratory comparison of the 95 degrees plate, antegrade and retrograde inserted reamed intramedullary nails. *J Orthop Trauma*. 1996;10: 378–382.
13. Zlowodzki M, Williamson S, Zardiackas LD, et al. Biomechanical evaluation of the less invasive stabilization system and the 95-degree angled blade plate for the internal fixation of distal femur fractures in human cadaveric bones with high bone mineral density. *J Trauma*. 2006; 60:836–840.
14. Fulkerson E, Egol KA, Kubiak EN, et al. Fixation of diaphyseal fractures with a segmental defect: a biomechanical comparison of locked and conventional plating techniques. *J Trauma*. 2006;60:830–835.
15. Kennedy JG, Carter DR. Long bone torsion: I. Effects of heterogeneity, anisotropy and geometric irregularity. *J Biomech Eng*. 1985;107: 183–188.
16. Heiner AD, Brown TD. Structural properties of a new design of composite replicate femurs and tibias. *J Biomech*. 2001;34:773–781.
17. Cristofolini L, Viceconti M, Cappello A, et al. Mechanical validation of whole bone composite femur models. *J Biomech*. 1996;29: 525–535.
18. Schatzker J, Horne JG, Sumner-Smith G. The effect of movement on the holding power of screws in bone. *Clin Orthop Relat Res*. 1975: 257–262.
19. Ansell RH, Scales JT. A study of some factors which affect the strength of screws and their insertion and holding power in bone. *J Biomech*. 1968;1: 279–302.