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Comparison of Four Methods for Dynamization of Locking Plates: Differences in the Amount and Type of Fracture Motion

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Abstract

Background: Decreasing the stiffness of locked plating constructs can promote natural fracture healing by controlled dynamization of the fracture. This biomechanical study compared the effect of four different stiffness reduction methods on interfragmentary motion by measuring axial motion and shear motion at the fracture site.

Methods: Distal femur locking plates were applied to bridge a metadiaphyseal fracture in femur surrogates. A locked construct with a short bridge span served as the non-dynamized control group (LOCKED). Four different methods for stiffness reduction were evaluated: replacing diaphyseal locking screws with non-locked screws (NON-LOCKED); bridge dynamization (BRIDGE) with two empty screw holes proximal to the fracture; screw dynamization with Far Cortical Locking screws (FCL); and plate dynamization with active locking plates (ACTIVE). Construct stiffness, axial motion and shear motion at the fracture site were measured to characterize each dynamization methods.

Results: Compared to LOCKED control constructs, NON-LOCKED constructs had a similar stiffness (p=0.08), axial motion (p=0.07), and shear motion (p=0.97). BRIDGE constructs reduced stiffness by 45% compared to LOCKED constructs (p<0.001), but interfragmentary motion was dominated by shear. Compared to LOCKED constructs, FCL and ACTIVE constructs reduced stiffness by 62% (p<0.001) and 75% (p<0.001), respectively, and significantly increased axial motion, but not shear motion.

Conclusions: In a surrogate model of a distal femur fracture, replacing locked with non-locked diaphyseal screws does not significantly decrease construct stiffness and does not

enhance interfragmentary motion. A longer bridge span primarily increases shear motion, not axial motion. The use of FCL screws or active plating delivers axial dynamization without introducing shear motion.

Key Words: plate fixation;locked plating;dynamization;interfragmentary motion;stiffness;active plating;femur

Introduction

Angle stable locked plating has become the standard treatment for most difficult fractures of the distal femur. Despite excellent early results, there is growing concern surrounding the relatively high non-union rates following locked plate fixation of distal femur fractures. Most recent studies quote non-union or fixation failure rates following locked plating of distal femur fractures of 10% to 23%.[1-6] There is abundant evidence that deficient fracture motion caused by overly stiff locking plates can suppress natural fracture healing, contributing to delayed unions, non-unions, and fixation failure.[3, 7-9] Conversely, research over the past 50 years has consistently demonstrated that controlled axial dynamization can improve the speed and strength of fracture healing by dynamically stimulating natural bone healing via callus formation.[7, 10-15]

Two primary mechanical conditions critical for promotion of natural fracture healing have been identified: Callus formation is promoted by axial interfragmentary motion greater than 0.2 mm;[16, 17] and fracture healing is inhibited when interfragmentary motion is dominated by shear displacement.[18] Strategies aimed at altering the mechanical environment created by locked plating constructs and at promoting fracture healing by spontaneous callus formation were proposed as early as 2003.[19]

Four principal methods are currently promoted to reduce the stiffness of locked bridge plating constructs: diaphyseal fixation with non-locking screws rather than locking screws; increasing the length of the bridge spanning the fracture zone;[19] screw dynamization with Far

Cortical Locking (FCL) screws;[8] and plate dynamization with active plates that have elastically suspended locking holes.[20] It is not clear which of four these constructs provides the best mechanical environment to achieve the goal of early fracture dynamization to promote healing while minimizing any detrimental effects from motion.

The present study measures construct stiffness as well as axial and shear motion at the fracture site to assess the efficacy by which each strategy can satisfy the basic conditions for mechanical stimulation of fracture healing. Specifically, this study tested the hypotheses that the four dynamization strategies will differ in their efficacy to decrease construct stiffness, to increase interfragmentary axial interfragmentary motion, and to prevent excessive shear motion.

Material and Methods

In a biomechanical bench-top study, periarticular locking plates were applied to bridge a metadiaphyseal fracture in femur surrogates. Construct stiffness was assessed under quasi-physiologic loading by measuring the resulting axial and shear motion at the fracture site. In the LOCKED control group, the periarticular plate was applied to the diaphysis with bi-cortical locking screws. The first locking screw was placed adjacent to the fracture to achieve a short bridge span. Subsequently, four different strategies to decrease construct stiffness and to dynamize the fracture site were evaluated (Figure 1): replacing diaphyseal locking screws with non-locking screws (NON-LOCKED group); bridge dynamization (BRIDGE group) by increasing the bridging span with locked screws in the diaphysis;[19] screw dynamization with Far Cortical Locking screws (FCL group);[8] and plate dynamization with active locking plates (ACTIVE group).[10, 20] Construct stiffness was characterized by measuring the construct deformation in response to quasi-physiologic loading. Dynamization of the fracture site was characterized by measuring the interfragmentary motion in axial and shear direction. Finally, the stiffness and interfragmentary motion results of the four strategies for stiffness reduction were

compared with the LOCKED control group to determine their effectiveness in dynamizing the fracture site.

Specimens: Plating constructs were evaluated in 4th generation femur surrogate specimens made of fiber-reinforced epoxy composite (#3406, large size, Sawbones, Vashon, WA, USA) to minimize inter-specimen variability. An unstable distal femur fracture (AO/Orthopaedic Trauma Association 33-A3) was modeled by introducing a 10-mm gap osteotomy located 60 mm proximal to the intercondylar notch.[21, 22] This gap osteotomy simulated the biomechanical constraints of a comminuted fracture that relies on full load transfer through the bridge plating construct due to a lack of bony apposition at the fracture site. In the LOCKED control group, this gap osteotomy was stabilized with a 286 mm long distal femur plate (ZPLP, Zimmer, Warsaw, IN) made of stainless steel. The plate had 13 holes for diaphyseal fixation, seven of which were locking holes (Figure 2A). The diaphyseal plate segment was applied using three evenly spaced 4.5 mm locking screws placed in the 1st, 4th, and 7th locking hole from the fracture site, resulting in a short bridge span of 25 mm. A plate elevation of 1 mm over the proximal diaphysis was achieved with temporary spacers to simulate biological fixation with preservation of periosteal perfusion.[23] The distal plate segment was applied to the metaphysis using six 5.5-mm cannulated locking screws in accordance with the manufacturer's technique guide. All screws were tightened to 4 Nm.

Subsequently, four additional construct configurations were assembled by changing one variable of the LOCKED control group constructs at a time: For constructs of the NON-LOCKED group, non-locking screws were used in place of locking screws for diaphyseal fixation, using the 1st, 4th, and 6th non-locking hole from the fracture site (Figure 2B). Because of the alternating locking / non-locking screw hole configuration of this plate, the non-locking construct had an intermediate bridge span of 40 mm. BRIDGE group constructs employed a longer bridge span (87 mm) than LOCKED control group constructs (25 mm) by placing

diaphyseal locking screws in the 3st, 5th, and 7th locking hole from the fracture site. FCL group constructs replaced the three diaphyseal locking screws of LOCKED control group constructs with three far cortical locking (FCL) screws (4.5 mm MotionLoc®, Zimmer, Warsaw, IN) made of stainless steel. FCL screws rigidly locked into the plate and the far cortex, but they are not rigidly constrained in the near cortex underlying the plate. The elastic shaft of FCL screws can flex within the near cortex motion envelope to generate symmetric interfragmentary motion.[8] ACTIVE group constructs had a screw configuration identical to that of the LOCKED control group, but employed an active locking plate. Screw holes of active locking plates are integrated in individual sliding elements that are elastically suspended in a silicone envelope inside lateral plate pockets (Figure 2C). Lateral pockets are arranged in an alternating pattern from both plate sides, resulting in a staggered locking hole configuration. The pocket geometry combined with the silicone suspension allows controlled axial translation, which enables up to 1.5 mm of axial motion across a fracture while providing stable fixation in response to bending and torsional loading.[24] The silicone suspension consisted of long-term implantable medical-grade silicone elastomer. The active locking plate was made of stainless steel, and was geometrically equivalent to the standard locking plate of the LOCKED control group (Figure 2D). Five specimens of each of the five constructs were tested for reproducibility, requiring a total of 25 construct tests.

Loading: For stiffness assessment, constructs were tested under quasi-physiological loading in a material test system according to an established loading protocol (Figure, Supplemental Digital Content 1 http://links.lww.com/BOT/A993, describing specimen loading and outcome assessment).[21, 25] The femoral condyles were embedded in a mounting fixture using bone cement, and were rigidly connected to the based of the test system (8874, Instron, Canton, MA). The metaphyseal plate segment was coated with soft clay to prevent non-physiologic plate constraints. The femoral head was placed in a spherical recess of a polymer block that was attached to the test system actuator. This enabled axial load application while allowing

unconstrained rotation of the femoral head. Load was induced along the mechanical axis of the femur, with the load vector intersecting the femoral head and the epicondylar center. Each construct was loaded in 50-N increments up to 700N, corresponding to approximately one body weight.

Outcome Assessment: Constructs were characterized by determining their construct stiffness and interfragmentary motion using non-contact optical photogrammetry. For this purpose, an array of four active luminescent markers consisting of miniature light emitting diodes were glued to the osteotomy surfaces. An 18 megapixel digital camera (Canon EOS T6) captured the marker locations with a resolution of 0.01mm after each incremental loading step. ImageJ quantitative image analysis software developed by the National Institute of Health (www.imagej.net) was used to extract marker displacement and to calculate the average axial motion d_A and shear motion d_S between osteotomy surfaces in response to incremental load steps. Since plate bending induces different amounts of axial motion at the near cortex and far cortex,[26] axial motion d_A was extracted individually for the near cortex ($d_{A, NC}$) from markers 1 and 3, and for the far cortex ($d_{A, FC}$) from markers 2 and 4, as depicted in Supplemental Digital Content 1. Construct stiffness S_C was calculated by dividing the applied axial load by the axial motion d_A at the midpoint between the near and far cortex, with $d_A = (d_{A, FC} + d_{A, NC})/2$.

Statistical Analysis: All results are reported as their mean and standard deviation. Construct stiffness S_C and interfragmentary motion parameters d_S ,$d_{A, FC}$, and $d_{A, FC}$ of the four experimental groups was statistically compared to the LOCKED control group results using one-way ANOVA testing including a post-hoc Turkey's HSD to identify significant differences. Each outcome parameter was analyzed individually, and a level of significance of α=0.05 was used to detect significant differences.

Results

Construct Stiffness: There was no significant difference between the stiffness of LOCKED control constructs (2998 ± 361 N/mm) and NON-LOCKED constructs (2549 ± 355 N/mm, p=0.08) (Table, Supplemental Digital Content 2 http://links.lww.com/BOT/A994, summarizing construct stiffness and interfragmentary motion). However, compared to the LOCKED control group, BRIDGE group constructs had a 45% lower stiffness (p<0.001), FCL group constructs had a 62% lower stiffness (p<0.001), and ACTIVE group constructs had a 75% lower stiffness (p<0.001) (Figure 3).

Axial Interfragmentary Motion: In each group, near cortex motion $d_{A,NC}$ was smaller than far cortex motion $d_{A,FC}$ (Figure 4). Near cortex motion in response to one body-weight loading (700N) was the same for LOCKED control constructs and NON-LOCKED constructs (0.10 ± 0.02 mm, p=0.85), and remained below the 0.2 mm axial motion threshold required for callus stimulation. Compared to the LOCKED control group, $d_{A,NC}$ was two times greater in the BRIDGE group (p<0.001), over four times greater in the FCL group (p<0.001), and over 7 times greater in the ACTIVE group (p<0.001). Similarly, far cortex motion was not significantly different between the LOCKED control group (0.37 ± 0.04 mm) and the NON-LOCKED group (0.46 ± 0.08 mm) (p=0.07). However, compared to the LOCKED control group, $d_{A,FC}$ was 73% greater in the BRIDGE group (p<0.001), 105% greater in the FCL group (p<0.001), and 303% greater in the ACTIVE group (p<0.001).

Shear Motion: Shear motion d_S remained below 0.2 mm in all groups except in the BRIDGE group, which exhibited on average 0.96 ± 0.14 mm shear motion in response to one body weight loading (Figure 5A). In BRIDGE constructs, this magnitude of shear motion was 50% greater than the corresponding far cortex motion, and over three times greater than the near cortex motion. Using image analysis, shear-dominant motion in BRIDGE constructs was attributed to rotation of the femoral diaphysis around the proximal plate segment due to plate

bending under axial loading, which caused the proximal osteotomy surface to be translated towards the locking plate (Figure 5B).

Discussion

Results confirmed the study hypothesis by demonstrating that the four dynamization strategies yielded not only different amounts of construct stiffness and interfragmentary motion, but also different types of interfragmentary motion.

LOCKED group results confirmed that a locking plate with a short bridge span results in deficient and asymmetric interfragmentary motion for callus formation.[3, 13, 16, 26] While one body-weight loading may be excessive for early post-operative loading, it resulted only in 0.1 mm motion at the near cortex. This motion remained below the 0.2 mm motion threshold that has been established as the lower boundary for fracture motion required to promote callus formation.[17] This result is supported by several *in vivo* and clinical studies that demonstrate suppression of callus formation and healing at the near cortex adjacent to a locking plate.[3, 7, 9, 20] The far cortex motion was greater than measured at the near cortex, likely secondary to plate bending. Clinically, the increased far cortex motion may allow callus to form, but the repetitive bending may also play a role in eventual fatigue failure of the plate before fracture healing occurs.

NON-LOCKED constructs represent an intuitive response to the stiffness concern associated with locked plating by reverting to non-locking screws in the diaphysis. However, substituting non-locking diaphyseal fixation had no significant effect on construct stiffness or interfragmentary motion. This may be explained by the rigid compression of the plate onto the bone surface, which is required to retain stable fixation. Compressing the plate to the bone prevents any motion at the plate-bone interface, which is a prerequisite to induce symmetric interfragmentary motion.[26] In contrast to locked plating constructs, the stiffness of a non-locked

construct will gradually decay as a result of dynamic loading.[27] While this can lead to increased fracture motion over time, the resulting uncontrolled motion is not a reliable strategy for dynamization. Additionally, the natural fracture healing process responds with much more robust callus formation when exposed to early motion relative to delayed motion.[28]

BRIDGE constructs resembled the earliest and most widely proposed strategy to dynamize locked plating constructs. In a biomechanical study, Stoffel at al reported in 2003 that axial stiffness of locked plating constructs was mainly influenced by their bridge span.[19] They recommended that one or two holes should be omitted on each side of the fracture to allow callus formation. They found that omitting two holes made the construct almost twice as flexible, but also 42% less strong. The present study found a 45% stiffness reduction by omitting two screw holes. However, the greater flexibility of the longer bridge span increased motion primarily at the far cortex, while near cortex motion remained deficient. In addition, the longer bridge span induced up to three times more shear motion that axial motion. While shear motion does not necessarily inhibit healing,[29, 30] several studies have shown that excessive or pre-dominant shear motion will significantly delay healing.[18, 31] A recent study on the effect of bridge span on fracture motion also confirmed a disproportionate increase in shear motion.[32] By analyzing sixty-six distal femur fractures stabilized with locking plates, they furthermore established a direct association between shear-dominated fracture motion and callus inhibition. Their findings, combined with the results of the present study question the technique of increasing the bridge span to dynamize a locked construct, since this may weaken the construct, and may cause asymmetric axial motion and excessive shear motion that inhibits fracture healing.

FCL and ACTIVE group constructs reduced stiffness compared to the LOCKED control group by 62-75% to 1130 N/mm and 759 N/mm, respectively. Nevertheless, their stiffness remained substantially higher than the stiffness range of 50-400 N/mm reported for external fixators[33, 34] and Ilizarov frames.[35] The fact that external fixators and Ilizarov frames are

established clinical tools that promote fracture healing by callus formation suggests that the stiffness reduction of FCL and ACTIVE constructs is rather conservative and does not introduce excessive dynamization. FCL and ACTIVE constructs enhanced interfragmentary motion at the near and far cortex well above the 0.2 mm threshold needed to stimulate callus formation. The highest axial motion of 1.1 mm was observed at the far cortex of ACTIVE constructs, and remained at the lower limit of the 1-4 mm motion range reported for functional bracing.[36] Most importantly, FCL and ACTIVE constructs delivered dynamization that was dominated by axial motion, not shear motion. These constructs allow a screw to be placed close to the fracture site without affecting stiffness, and therefore limit the amount of shear motion possible. The controlled axial dynamization provided by FCL and ACTIVE constructs delivers faster and stronger healing. In an ovine fracture healing study, FCL constructs induced consistent and circumferential callus bridging and yielded 157% stronger healing compared with standard locked plating.[7] Clinically, a prospective study of 31 consecutive distal femur fractures stabilized with FCL constructs reported no implant or fixation failure, an average time to union of 16 weeks, and a nonunion rate of 3%.[37] Similar to FCL constructs, ACTIVE plating induced six times more callus at 3 weeks post surgery, and yielded four times stronger healing compared to rigid locked plating in an ovine fracture healing study.[20] These *in vivo* and clinical studies of FLC and ACTIVE plating constructs demonstrated that controlled axial dynamization reliably promoted natural fracture healing.

Results of this study are limited by the use of femur surrogates. Validated surrogates were employed to extract relative differences between constructs under highly reproducible test conditions.[38] Since the surrogates represented a strong, non-osteoporotic femur, results may not be extrapolated to fracture fixation in the osteopenic femur. Moreover, this study only investigated construct stability in terms of stiffness and related interfragmentary motion, without loading constructs to failure to determine their strength. The strength of the tested constructs has

been evaluated in previous studies, showing that increasing the bridge span will decrease construct strength,[19] while the strength of FCL and active plating constructs is comparable to that of standard locked plating constructs.[8, 24] Testing was furthermore limited to static loading, and did not investigate gradual loosening or fatigue of constructs under dynamic loading. Moreover, this study has been limited to a principal loading mode that combines axial compression and bending, but not torsion. Only plates made of stainless steel were tested, which are approximately twice as stiff as geometrically equivalent plates made of Titanium alloy. While results of this study raise concerns on the negative effect of shear-dominated interfragmentary motion on fracture healing, this concern should be formally investigated in a future *in vivo* study. Most importantly, emerging implant technologies that can provide controlled dynamization will require more clinical studies to document their effect on fracture healing, and to better define the range of interfragmentary motion that will promote healing of different fracture patterns at specific fracture locations.

In conclusion, results of this study indicate that intuitive technical tricks, such as reverting to non-locking screws or using long plates to maximize the bridge span may not reliably achieve relative stability and adequate interfragmentary motion for promote natural fracture healing. Conversely, engineered implant solutions in form of FCL screws or active plates can reliable dynamize a locked plating construct to stimulate fracture healing. As such, results should encourage implant manufacturers do provide engineered solutions that reliably promote rather than potentially hinder fracture healing to avoid the need for and uncertainty of technical tricks intended to optimize construct stability.

Figure Captions

Figure 1: Strategies to dynamize a locked plating construct (LOCKED) for distal femur fractures.

Figure 2: A) Distal femur plate with alternating locked and non-locked holes; B) three distinct screws used with standard femur plate; C, D) active plate with screw holes located in elastically suspended sliding elements.

Figure 3: Construct stiffness achieved with the four strategies for plate dynamization, relative to the LOCKED control construct.

Figure 4: Axial motion at the near and far cortex, achieved with the four strategies for plate dynamization, relative to the LOCKED control construct.

Figure 5: Resulting transverse shear, resulting from the four strategies for plate dynamization, relative to the LOCKED control construct.

Supplemental Digital Content

Figure, Supplemental Digital Content 1, depicting the quasi-physiologic loading setup (A), and the marker array to quantify shear motion d_S and axial motion at the near cortex ($d_{A, NC}$) and far cortex ($d_{A, FC}$) of the fracture (B,C).

Table, Supplemental Digital Content 2: Summary of results for construct stiffness and interfragmentary motion.

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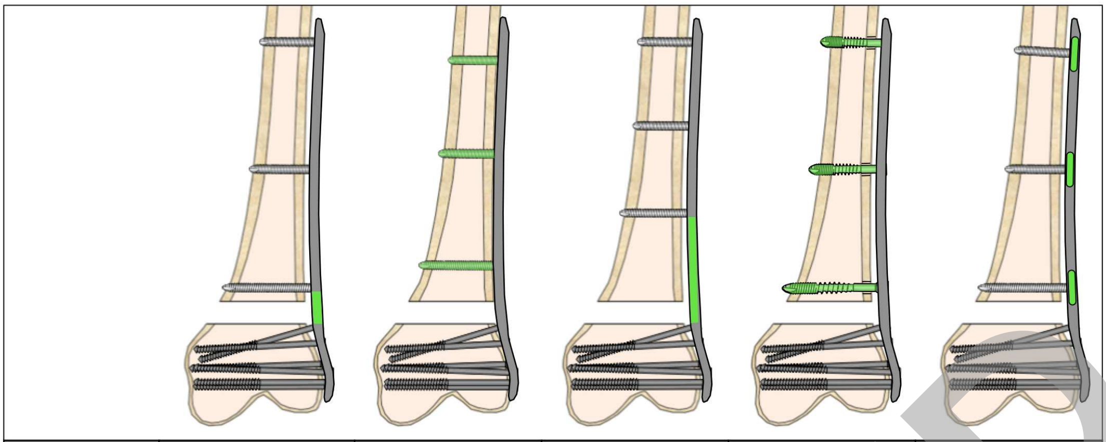
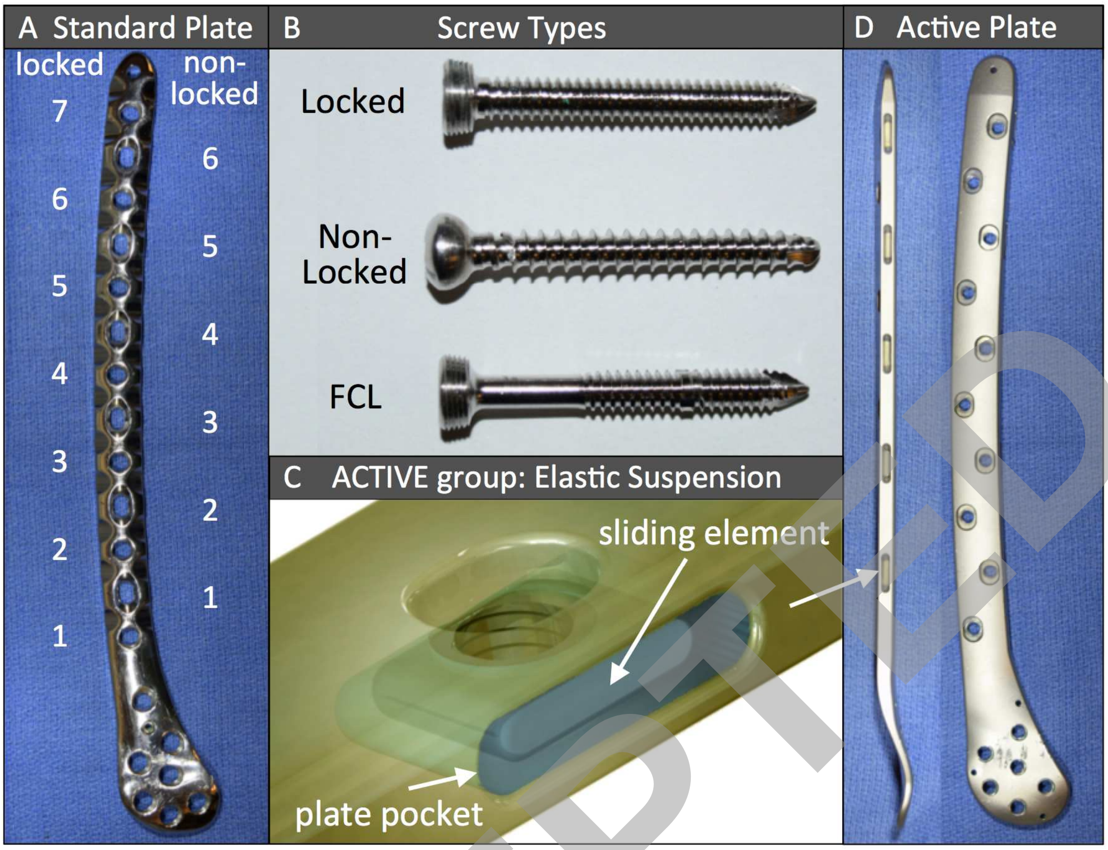
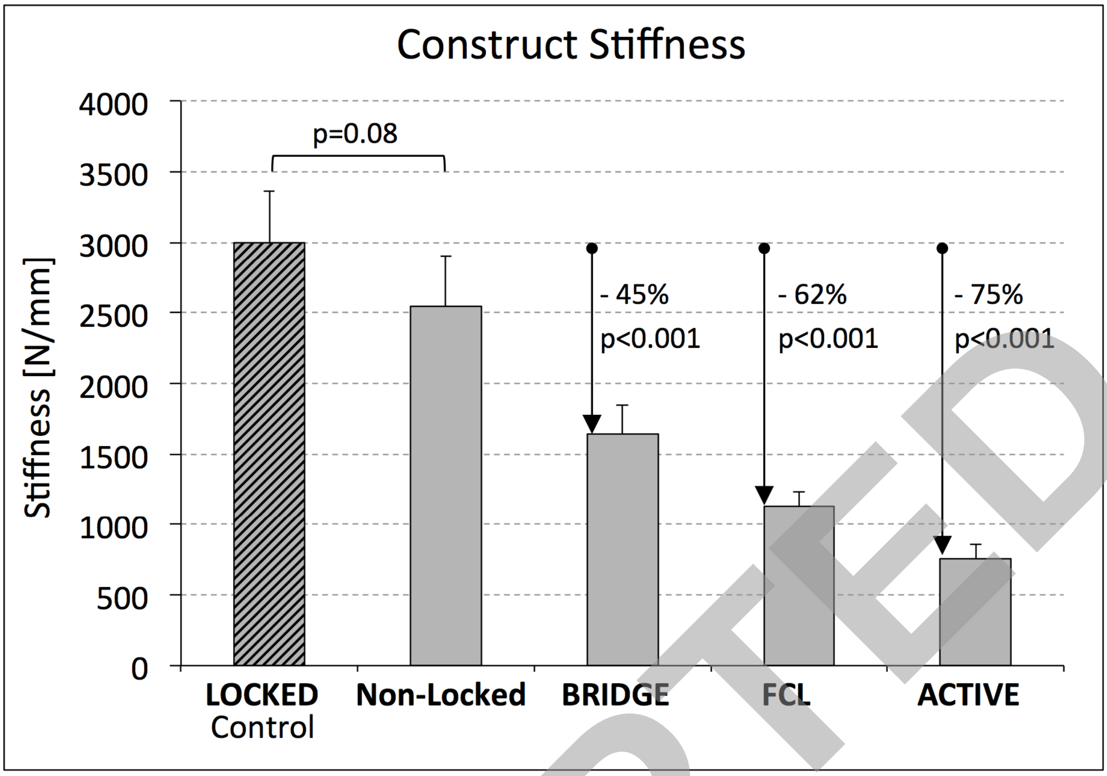
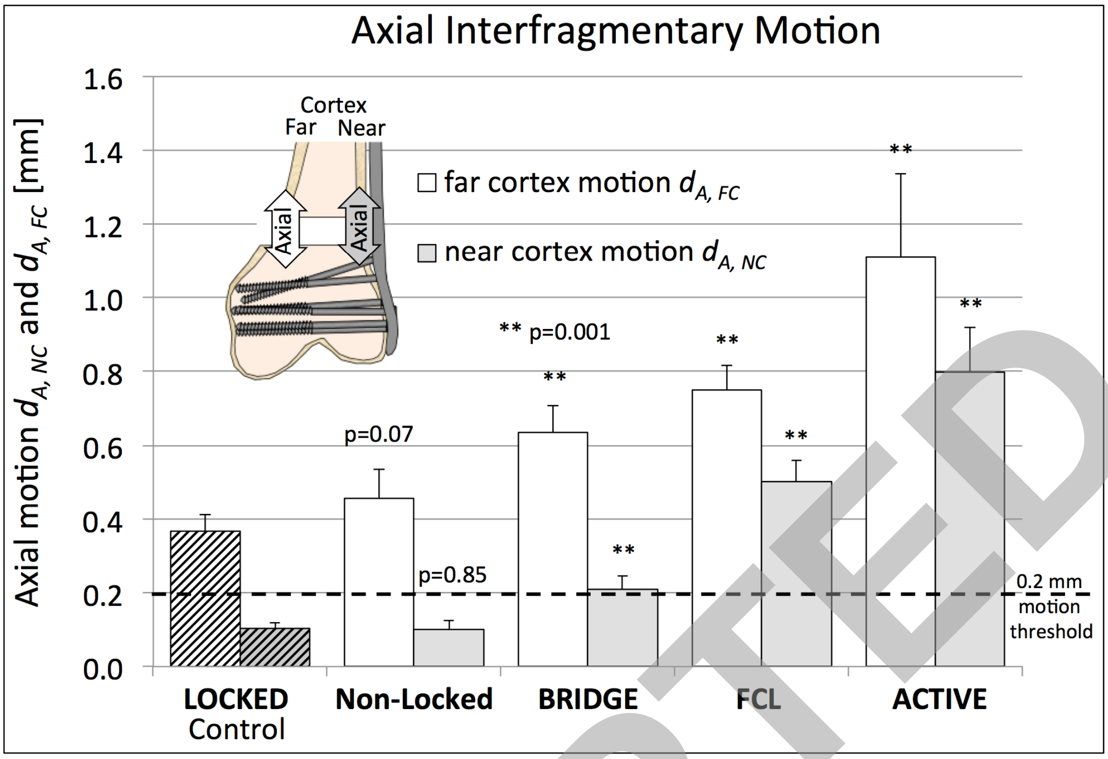


Group	LOCKED control	NON-LOCKED diaphyseal screws	BRIDGE span	FCL Far Cortical Locking	ACTIVE Active Plate
Screw type	locked	**non-locked**	locked	**FCL screws**	locked
Bridge length	short (25 mm)	short (40mm)	**long (87 mm)**	short (25 mm)	short (25 mm)
Plate type	SS ZPLP	SS ZPLP	SS ZPLP	SS ZPLP	**ACTIVE plate**

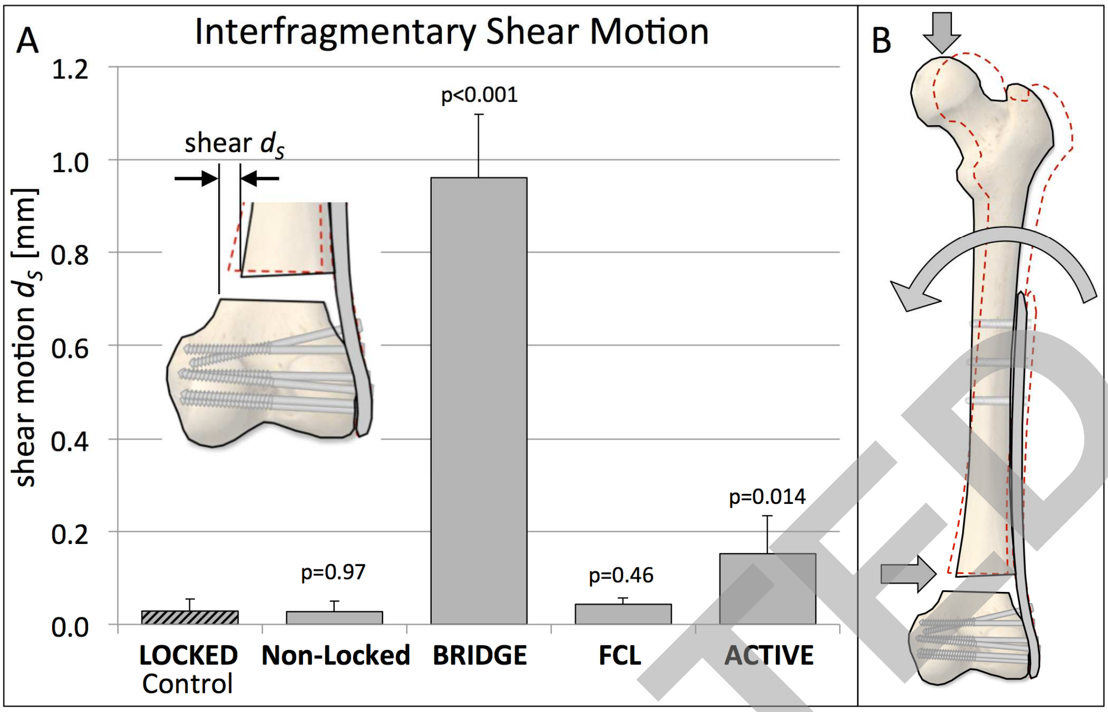


A Standard Plate
locked
non-locked
7
6
5
4
3
2
1
6
5
4
3
2
1
B Screw Types
Locked
Non-Locked
FCL
C ACTIVE group: Elastic Suspension
sliding element
plate pocket
D Active Plate


Construct Stiffness
Stiffness [N/mm]
4000
3500
3000
2500
2000
1500
1000
500
0
p=0.08
- 45%
p<0.001
- 62%
p<0.001
- 75%
p<0.001
LOCKED
Control
Non-Locked
BRIDGE
FCL
ACTIVE



Axial Interfragmentary Motion
Axial motion $d_{A, NC}$ and $d_{A, FC}$ [mm]
1.6
1.4
1.2
1.0
0.8
0.6
0.4
0.2
0.0
Cortex
Far Near
Axial
Axial
far cortex motion $d_{A, FC}$
near cortex motion $d_{A, NC}$
** p=0.001
p=0.07
p=0.85
**
0.2 mm motion threshold
LOCKED Control
Non-Locked
BRIDGE
FCL
ACTIVE


A
Interfragmentary Shear Motion
shear motion d_S [mm]
1.2
1.0
0.8
0.6
0.4
0.2
0.0
shear d_S
p<0.001
p=0.97
p=0.46
p=0.014
LOCKED
Control
Non-Locked
BRIDGE
FCL
ACTIVE
B