# Comparing the biomechanical stability of cerclage cable with plate insert versus locking screw in periprosthetic humeral fracture

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## ABSTRACT


### Comparing the biomechanical stability of cerclage cable with plate insert versus locking screw in periprosthetic humeral fracture

**Background:** In the setting of periprosthetic humeral fractures, the humeral stem of the implant represents a substantial challenge to the optimal method of proximal fixation. This study aimed to compare the initial biomechanical stability provided by cerclage cables with a locking plate insert versus bicortical locking screws (i.e., the gold standard for fixation) in fresh cadaveric humeri.

**Methods:** After calculating the sample size, we utilized 10 sets of cadaveric specimens and create a 5-mm my gap, 120mm distal to the tip of the grater tuberosity, simulating a Wright and Cofield type-B periprosthetic humeral fracture on each specimen. Using 3 locking screws for distal fragment fixation, identical in all specimens, the specimens were assigned to Group A (3 cerclage cables with a plate insert) or Group B (3 locking bicortical screws) for proximal fragment fixation. Biomechanical tests included stiffness in varus and valgus bending, torsion, and axial compression, and a single load to failure.

**Results:** No significant differences were observed in the biomechanical metrics between the 2 groups.

**Conclusions:** Our study revealed that fixation with use of cerclage cables with a plate insert demonstrated biomechanical stability comparable with that of bicortical locking screw fixation when addressing the proximal fragmentinWright and Cofield type-B periprosthetic humeral fractures.

**Clinical Relevance:** For proximal fragment fixation of periprosthetic humeral fractures, cerclage cables with a plate insert can be utilized as an effective fixation method that offers initial fixation strength that is comparable to the use of 3 locking bicortical screws.


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Key words : periprosthetic humerus fracture, biomechanical testing, cerclage cable with plate insert

# 1. Introduction

With an increasingly aging population, a greater number of patients are undergoing shoulder arthroplasty[1,2,3]. This increase in shoulder arthroplasties has led to a corresponding increase in complications, with an estimated total complication rate of approximately 15%[4]. One potential complication is periprosthetic humeral fracture, which occurs in 0.6% to 3% of all shoulder arthroplasty cases[5,6].

The treatment of periprosthetic humeral fractures can be challenging, and inadequate treatment may result in poor outcomes[7,8]. In cases in which the stem is stable but the fracture fragments are displaced, open reduction and internal fixation with use of a locking compression plate is considered a treatment option. Because no stem is present at the distal screw hole, a bicortical locking screw can be utilized for solid fixation; however, the proximal portion involves the presence of a stem, with or without cement. Therefore, other options should be considered, such as unicortical screw fixation, cerclage cables with an insert, or a locking attachment plate[9,10].

Although the use of unicortical locking screws can be considered, this may result in stem instability related to canal encroachment or cement mantle breakage[11]. In such cases, the use of cerclage cables is another option. With this technique, an additional plate insert securely attaches to the locking hole of the plate, enabling the cerclage cable to pass through its aperture and resulting in a more robust attachment of the cerclage cable to the plate compared with the use of the cerclage cable alone. However, to our knowledge, no research has been conducted that compares the fixation provided by cerclage cables with an insert plate versus locking screw fixation.

The purpose of the present study was to compare the initial biomechanical stability provided by cerclage cables with a locking plate insert versus bicortical locking screws (i.e., the gold standard for fixation) in addressing the proximal fragment in Wright and Cofield type-B periprosthetic humeral fractures. We hypothesized that although the use of cerclage cables with a plate insert may exhibit adequate strength to resist bending force, their torsional stiffness may not sufficiently compare with the use of bicortical locking screws.

# 2. MATERIALS AND METHODS

## 2.1. Specimen preparation and experimental setup

Ten paired fresh-frozen cadaveric humeri were utilized, including 3 from male cadavers and 7 from female cadavers. The age of the specimen donors ranged from 74 to 92 years, with a mean age of 84.8 years and a standard deviation of 5.2 years. All specimens were free from gross deformity, surgical interventions, or previous fractures.

The specimens were maintained at -20°C and thawed to room temperature 24 hours before the experiment. Adjacent soft tissue was meticulously excised. The bone strength of each paired specimen was assessed by measuring the diaphyseal cortical surface area[12] on an axial computed tomography (SOMATOM Definition AS; Siemens Healthcare) image of the osteotomy site. To simulate a Wright and Cofield type-B periprosthetic humeral fracture (Fig. 1), a 5-mm gap osteotomy was created to simulate fracture comminution at a point 120 mm distal to the greater tuberosity tip, where the distal end of the press-fit humeral stem (Equinoxe Primary System; Exactech) was located.

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**Figure 1**. Wright and Cofield type B periprosthetic humeral fracture.

Each specimen received an identical 9-hole 4.5/5.0-mm narrow locking compression plate (LCP; DePuy Synthes) with distal fixation utilizing 3 locking screws. The specimens within each pair were randomly assigned to Group A (proximal fixation with 3 cerclage cables with a locking plate insert; Orthopaedic Cable System; DePuy Synthes) or Group B (proximal fixation with 3 bicortical locking screws). In the surgical setting, when applying tension to the cerclage cable, it is essential to exert the maximum tolerable tension for secure fixation. The 1.7-mm cerclage cable used in the present study can withstand a maximum tension of50 kg. Therefore, in Group A, the cable was tightened to the manufacturer recommendation of 50 kg. After fixation, radiographs confirmed the adequacy of the osteotomy gap and plate bone construct (Fig. 2).

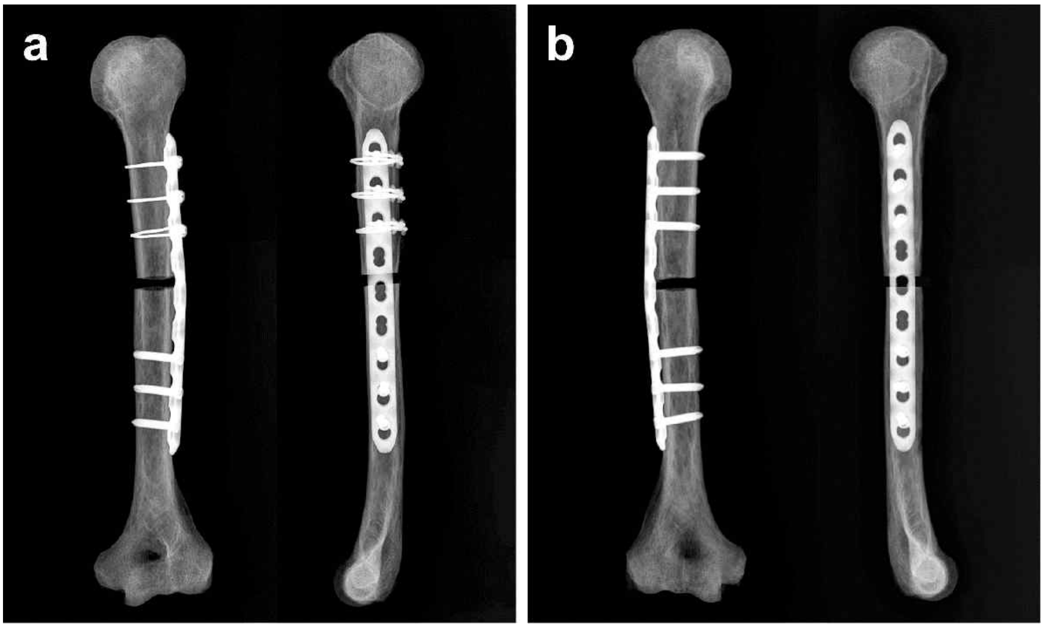

**Figure 2.** Radiographs assessing the adequacy of fixation in a 5 mm osteotomy gap treated with a locking compression plate (a: cerclage cable group, b: locking bicortical screw group).

Proximal and distal potting was conducted by inserting the cadaveric humeri into a specially designed cuboid jig (6.5 cmin width and length, 7.5 cm in height). The humeri were then fixed in place with use of unsaturated polyester resin (EC-304; Aekyung Chemical Co.). Before resin application, a protective silicone coating was applied to the instruments to prevent additional resin support.

## 2.2. Biomechanical testing

The study examined stiffness in varus and valgus bending, internal and external torsion, and axial compression, and a single load to failure (Fig. 3) with use of a universal testing machine (model 3366; Instron).

The varus bending setup was utilized to determine the single load to failure by identifying abrupt slope changes within the measured stiffness curve. Varus and valgus bending wereperformed with use of a 4-point bending setup, with a bending moment of 3.5 Nm and a speed of 0.1 mm/second. In the axial compression test, a 250-N load was applied at a speed of 0.1 mm/second. Testing of all parameters except torsion was performed with the Instron device. The torsional test was performed with a torsional stiffness device (DPTST; DYPHI), and torsional loading was performed sequentially in internal and external rotation, rotating at 0.2 Nm torque/second to 11.6 Nm and 21.6Nm for internal and external stiffness, respectively. This load was selected in prior studies to prevent plastic deformation of the fixtures during the mechanical tests[13,14].

## 2.3. Statistical analysis

As we were aware of no previous studies on this topic, a pilot study was necessary to determine an appropriate sample size. A power analysis was conducted utilizing 3 paired specimens (6 humeri) and varus bending stiffness values to determine the minimum sample size required for the study based on an alpha of 0.05 and a power of 0.8. In the pilot study, the mean and standard deviation of varus bending stiffness were 533.2 ± 11.9 N/mm in Group A and 550.6 ± 10.3 N/mm in Group B. Based on these data, 20 specimens (10 per group) were required. The Shapiro-Wilk test was employed to evaluate the normality of distribution for all variables. Depending on the normality of variables, either the paired t test or the Wilcoxon signed-rank test was utilized to compare cortical surface area and biomechanical metrics between the 2 groups. Significance was set at 0.05. Statistical analyses were performed with use of SPSS (version 25.0; IBM).

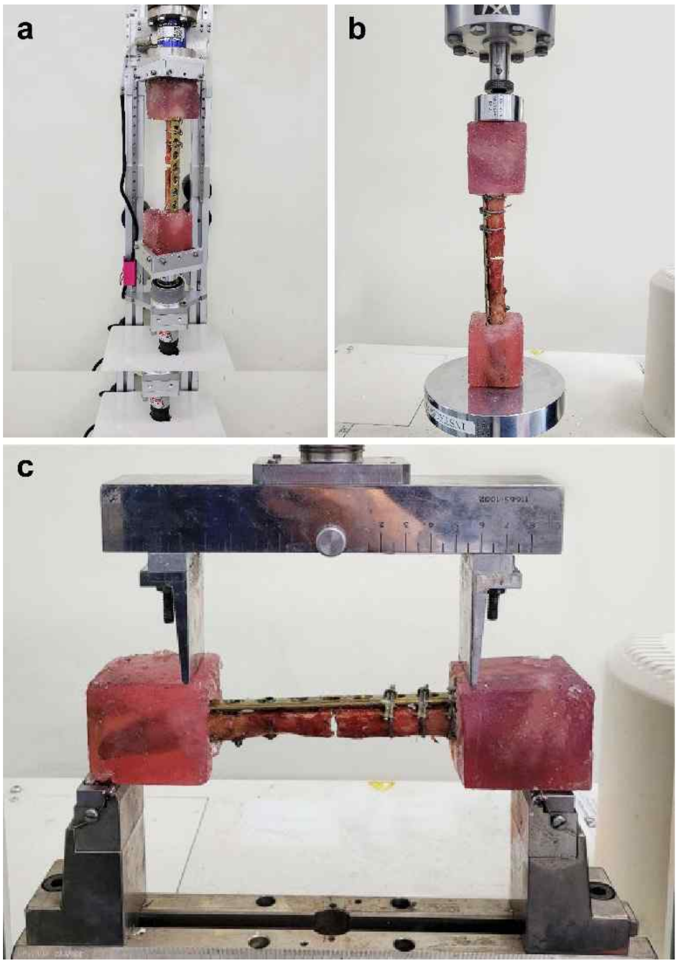


**Figure 3.** Biomechanical testing setup
(a: torsional testing setup, b: axial compression setup, c: 4-point bending setup).

# 3. Results

In Group A (3 cerclage cables with a locking plate insert), the mean cortical surface area was 220.9 ± 67.8 $mm^2$. The mean varus and valgus bending stiffnesses were 360.7 ± 72.6 and 545.9 ± 38.3 N/mm, respectively. The mean stiffness values in internal and external torsion were 0.696 ± 0.157 and 0.709 ± 0.155 N/deg, respectively. The mean stiffness in axial compression was 412.8 ± 58.9 N/mm, and the mean value for the single load to failure was 1,607.8 ± 187.9 N. In Group B (3 bicortical locking screws), the mean cortical surface area was 212.2 ± 53.3 $mm^2$. The mean varus and valgus bending stiffnesses were 360.4 ± 74.2 and 543.9 ± 65.7 N/mm, respectively. The mean stiffness values in internal and external torsion were 0.715 ± 0.106 and 0.693 ± 0.087 N/deg, respectively. The mean stiffness in axial compression was 450.3 ± 75.2 N/mm, and the mean value for the single load to failure was 1,681.1 ± 220.5 N. The groups had no significant differences in cortical surface area or biomechanical metrics (Table I).

**Table 1.** Comparison of cortical surface area, stiffness, and load to failure between the 2 groups

| | Group A (n=10) | Group B (n=10) | p-value |
|---|---|---|---|
| Cortical Surface area ($mm^2$) | 220.9 ± 67.8 | 212.2 ± 53.3 | 0.139 |
| Stiffness in varus bending (N/mm) | 360.7 ± 72.6 | 360.4 ± 74.2 | 0.646 |
| Stiffness in valgus bending (N/mm) | 545.9 ± 38.3 | 543.9 ± 65.7 | 0.575 |
| Stiffness in internal torsion (N/deg) | 0.696 ± 0.157 | 0.715 ± 0.106 | 0.76 |
| Stiffness in external torsion (N/deg) | 0.709 ± 0.155 | 0.693 ± 0.087 | 0.575 |
| Stiffness in axial compression (N/mm) | 412.8 ± 58.9 | 450.3 ± 75.2 | 0.059 |
| Single load to failure (N) | 1607.8 ± 187.9 | 1681.1 ± 220.5 | 0.241 |

Note: Group A: Proximal fixation using three 1.7 mm cerclage cables with a cerclage plate insert. Group B: Proximal fixation using three locking bi-cortical screws. Values are expressed as mean ± standard deviation. Comparative analysis by Wilcoxon Signed-Rank test.

# 4. Discussion

The present study aimed to investigate whether cerclage cables with a locking plate insert on the proximal locking compression plate would provide biomechanical stability comparable with that of bicortical locking screws in a periprosthetic humeral fracture. The 2 groups had no significant differences in the tested metrics, including varus and valgus bending, internal and external torsion, and axial compression stiffness, and single load to failure.

Periprosthetic humeral fracture is a complication that is difficult to address. Regardless of the use of stem cementation, there is a high chance of endosteal blood supply disruption while performing stem insertion during the primary surgical procedure[15]. In addition, the blood supply to the periosteal side of the cortex can be disrupted during open reduction. Thus, achieving bone union is more challenging than in conventional humeral shaft fractures. Moreover, considering the high prevalence of periprosthetic fractures in older patients with poor bone quality and osteoporosis, locking screw fixation is necessary. However, screws in the proximal region are restricted because of the presence of the stem. Furthermore, the use of unicortical screw fixation is also limited. The pull-out strength of unicortical screws in periprosthetic fracture models with a cement mantle has been observed to be approximately half that of bicortical screws[16], raising the possibility of failure in cemented humeral stems. Therefore, the use of cerclage cables is a common alternative. Despite their widespread use, the effectiveness of cerclage cables in providing sufficient initial postoperative stability remains questionable. Consequently, early postoperative rehabilitation has not been considered feasible, and our research was initiated to consider this issue.

Clinical fixation with use of a locking compression plate usually requires 6 cortices—3 bicortical screws in each fragment—during locking screw fixation[17,18]. However, we predicted that the stability of the 3 cerclage cables would not be comparable with that of 3 locking screws for the proximal fragment. In the case of a Wright and Cofield type-B periprosthetic fracture involving a typical 120-mm humeral stem, it is possible to utilize 4 cables with 4 holes in the proximal fragment. However, in these cases, the most proximal cable is likely to be placed around the medial calcar of the proximal humerus. Considering the complicated anatomy of the medial calcar, especially the neurovascular structures, placement of the cable around the medial calcar can be challenging without meticulous identification of the neurovascular structures in the surgical field. Thus, we used 3 cables in 3 holes for fixation of the proximal fragment and conducted an initial pilot study to measure the varus bending stiffness.

Contrary to our expectations, the results were comparable, and we proceeded with an experimental design comparing 3 cables with 3 locking screws. It was predicted that, although the cables were strong enough to withstand the bending force, even when attachments such as plate inserts were utilized, the torsional stiffness would not sufficiently withstand the force. However, interestingly, the bending and torsional stiffnesses showed results that were comparable with the use of 3 locking screws.

Cerclage cables and wiring are simple yet important techniques for the surgical treatment of fractures[19]. Despite a shift away from their use as a standalone method for primary fractures, owing to the development of more stable implants, the clinical relevance of fixation with use of the cerclage technique has gained attention with the increasing number of periprosthetic fractures[20]. Internal plates or prostheses serve as major stabilizing elements for the fracture by acting as splints. In contrast, cerclage cables not only act as splints but also reduce and fix fragments through a centripetal action[21]. The favorable results with use of such cables in this study are attribute able to the use of plate inserts rather than relying solely on cerclage wiring on to the plate. Biomechanical studies show that the use of plate inserts in cerclage plating, rather than the use of simple wiring, decreases the relative motion at the connection between the wire and plate, thereby increasing stability[22]. Notable results have been observed in clinical practice when plate inserts were utilized for cerclage-only fixation in treating Vancouver type-B1 and C periprosthetic hip fractures, with all patients showing bone healing[23]. Alternatively, a locking attachment plate may be a viable treatment option because it forms a stable construct around the stem[24]. However, even if the use of a locking-attachment plate is considered a viable fixation option in periprosthetic femoral fractures[25], its use may be limited in periprosthetic humeral fractures because of the smaller cortical thickness and the potential for canal encroachment and subsequent instability of the stem, especially in the case of cemented stems.

The present study had several limitations. First, the comparable outcomes of the cerclage cables shown in this study represent the initial strength and do not take into account the healing potential, as is the case in time-zero cadaver studies. Second, our study was limited to Wright and Cofield type-B periprosthetic humeral fractures. Third, although we simulated a periprosthetic humeral fracture, we could not utilize a press-fit stem because of the poor bone quality of the specimens.

This limitation prevented us from evaluating the biomechanical performance or the impact of the inserted press-fit stem, as press-fit fixation was not feasible in the pilot or present studies.

# 5. Conclusion

In conclusion, our study revealed that the use of cerclage cables with plate inserts demonstrated biomechanical stability comparable with that of bicortical locking screws in addressing proximal fragments in Wright and Cofield type-B periprosthetic humeral fractures.

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Abstract in Korean

# 인공관절 삽입물 주위 상완골 골절에서 플레이트 인서트를 동반한 원형 케이블과 잠금 나사못의 생체역학적 안정성 비교

**연구 배경 및 목적:** 인공관절 삽입물 주위 상완골 골절의 설정에서, 임플란트의 상완골 스템의 존재로 인해 근위부 상완골을 고정 하는데 있어서 상당한 제한점이 있습니다. 본 연구는 사체 상완골에서 원형케이블과 플레이트 인서트를 사용한 고정의 초기 생체역학적 안정성을 잠금 나사못과 비교합니다. 실험을 하기 위해 선행 연구를 통하여 샘플 크기를 10쌍으로 계산하였습니다.

**연구 재료 및 방법:** 상완골 대결절의 끝에서 120mm 아래에 5mm의 틈을 만들어, Wright와 Cofield 유형-B 삽입물 주위 상완골 골절을 시뮬레이션 하였습니다. 잠금 압박 금속판을 사용하였으며, 원위부는 모든 사체에서 3개의 잠금 나사못을 이용하여 고정하였고, 근위부는 그룹 A (3개의 플레이트 인서트를 동반한 원형 케이블) 또는 그룹 B (3개의 이중 잠금 나사) 로 할당되었습니다. 생체역학적 실험에는 내반/외반 (varus/valgus), 내측/외측 비틀림 (internal/external torsion), 축성 압박 (axial compression) 및 단일 하중에서 금속판 파손 강도 (single load to failure) 값이 측정되었습니다.

**연구 결과:** 두 그룹 간의 생체역학적 지표에서 유의한 차이가 관찰되지 않았습니다.

**결론:** 우리의 연구는 플레이트 인서트를 동반한 원형 케이블을 사용한 고정은 Wright와 Cofield 유형-B 삽입물 주위 상완골 골절의 근위부를 고정할 때 이중 잠금 나사 고정과 비교할 수 있는 생체역학적 안정성을 보여 주었습니다. 결론적으로, 삽입물 주위 상완골 골절의 근위부 고정을 위해 플레이트 인서트와 원형 케이블을 사용한 고정은 이중 잠금 나사 사용과 비교 가능한 초기 고정 강도를 제공하여 효과적인 고정 방법으로 사용될 수 있습니다.

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**핵심되는 말 :** 삽입물 주위 상완골 골절, 생체역학연구, 플레이트 인서트를 동반한 원형 케이블