

Higher stability with locking plates in hand surgery? Biomechanical investigation of the TriLock system in a fracture model

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Abstract

Purpose The aim of this study was to determine the biomechanical characteristics of locking plates with the TriLock system with different design and screw settings compared to a non-locking plate in a diaphyseal metacarpal fracture.

Methods Oblique diaphyseal shaft fractures in porcine metacarpal bones were created in a biomechanical fracture model. After reduction they were fixed with three different locking plates with the TriLock interlocking mechanism or a non-locking linear plate in mono- or bicortical screw fixations. In load to failure tests the maximum load and stiffness were measured.

Results For linear plates, the maximum load was higher for the non-locking plate compared with the locking plate. The maximum load could be increased for the locking plates using a double-row design and a higher screw number. No differences were found for the stiffness between all groups. In contrast to the non-locking plate, the mode of failure of the locking plates in many cases (86 %) was a loss of the interlocking mechanism.

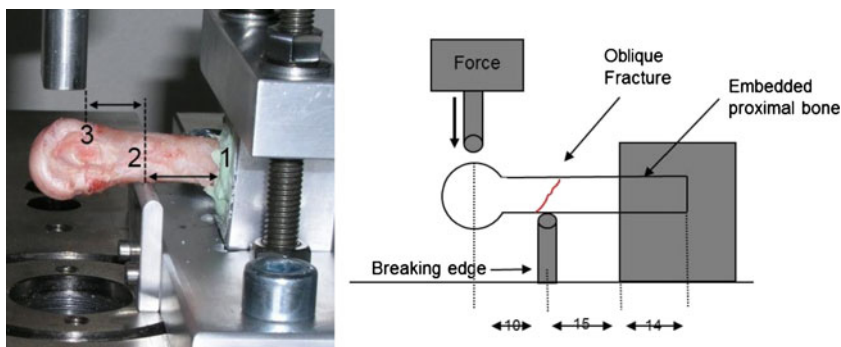
Conclusions The results suggest that the locking plates with the TriLock system achieve no higher stability compared to a non-locking plate in load to failure tests. Adaptions to increase the stability of the interlocking mechanism are desirable.

Introduction

In recent years, the implants for long bones have been improved in regards to higher stability and stiffness and also new locking plates have been designed for every domain of trauma surgery. In comminuted or metaphyseal fractures especially we expect and already see advantages for the locking plates matched with the non-locking plates in clinics. In hand surgery, there are special problems in fracture treatment and thus requirements for implants: The osteosynthesis has a close contact to the extensor tendons which limits the plate thickness. At the palmar side of the metacarpals and phalanges the flexor tendon sheaths run directly below the bones and may be irritated by bicortical drilling and screwing. Adhesions, postoperative and after immobilisation, may be the reason for a bad outcome and limited hand function. So the aim of new implant design is to create smaller devices for a less invasive approach, to reduce the profile of the plate for little extensor tendon interaction, and to increase primary stability by an interlocking mechanism between the screw head and the hole. A high stability has to be maintained in order to allow early functional treatment. Usually the operated fractures are treated with early controlled passive motion to prevent adhesions and improve the clinical outcome [1]. Comparing the different techniques of osteosynthesis for metacarpal fracture stabilisation the combination of a dorsal plate and lag screw has been shown in biomechanical studies to be stronger than other fixation methods [2–4]. Prevel et al. found that a three-dimensional design is associated with increased rigidity [5, 6]. New locking plates have been invented for hand surgery and are already often used in clinics. However, taking the high costs of locking implants into account, biomechanical studies for

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Fig. 1 The tests were performed in a modified 3-point bending test in which the bone is fixed proximally (1) and load is applied on the distal apex (3). The third point is a breaking edge on the palmar diaphysis (2). Distances are standardised



evidence-based guidelines of treatment are needed for their use in hand surgery. Two recent biomechanical studies investigated locking plates with the interlocking mechanism of titan deformation and showed a higher stability with locking monocortical screws compared to non-locking screws [7, 8].

The aim of this study was to determine the biomechanical properties of locking plates with a polyaxial TriLock system, to compare different plate designs, linear versus double-row implants, and to investigate the role of the number of screws. We expected the new locking plates to produce a higher stability than a comparable non-locking plate even in a monocortical screw fixation to justify clinical application. The mode of failure was also investigated to improve the design of the implants.

Materials and methods

Fresh-frozen metacarpals II from domestic pigs were used. The bones were prepared from fresh front hooves, wrapped in normal saline-soaked gauze and stored at -20°C . Using pig metacarpals for biomechanical evaluation has been validated earlier [9]. With regard to biomechanical qualities, the pig bones correspond to the human metacarpals and provide a minimal interspecimen variation of structure. Clinically, metacarpal fractures usually appear in young adults with a peak between the 15th and 34th year [10]. However, human cadaver bones, which are often from old

donors, would be osteoporotic and not representative for biomechanical tests of metacarpal fractures.

Before use, the bones were defrosted to room temperature (20°C). They were embedded according to a standardised protocol in a fixation device with Palacos[®] bone cement (Heraeus Medical GmbH, Wehrheim, Germany). Then they were clamped in the testing device. All tests were performed in the material testing machine Zwick Roell Z020.

Bending forces are higher compared to extension forces at the metacarpal bones under physiological conditions [11]. Thus, we simulated the physical bending stress by a modified three-point bending test model, as previously described [8, 12]. In brief, after fixation of the bone, a force was applied on the dorsal apex with 100 mm/min. Additionally, a breaking edge was put under the diaphysis of the bone (Fig. 1). All metacarpals were tested for maximum load and stiffness in a load to failure test setup.

To generate a reproducible fracture pattern a biomechanical fracture model was used [8, 12]. An oblique shaft fracture was created by bending a metacarpal bone over a breaking edge by apical dorsal force application. The fracture was reduced in a clamping device and an osteosynthesis was fixed dorsally.

For the tests a non-locking linear plate and three different locking plates with the TriLock interlocking mechanism were used. Ten bones per group were tested. Seven groups were selected (Table 1, Fig. 2): In groups 1 and 2, the non-locking linear plate was fixed with four monocortical and

Table 1 Characteristics of the investigated locking and non-locking plates

Group	Plate	Number of holes	Locking	Screw fixation	Number of screws	Diameter of screws	Plate thickness
1	Non-locking linear	4	–	Monocortical	4	2.3 mm	1 mm
2	Non-locking linear	4	–	Bicortical	4	2.3 mm	1 mm
3	Locking linear	4	+	Monocortical	4	2 mm	1 mm
4	Locking linear	4	+	Bicortical	4	2 mm	1 mm
5	Locking double-row	6	+	Monocortical	4	2 mm	1 mm
6	Locking double-row	8	+	Monocortical	4	2 mm	1 mm
7	Locking double-row	8	+	Monocortical	8	2 mm	1 mm

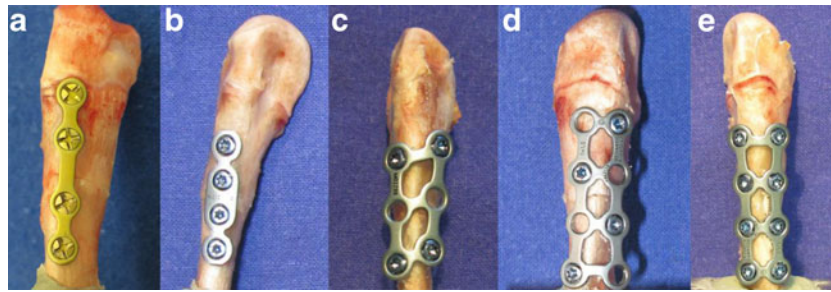


Fig. 2 The different plate designs of the testing groups. **a** A non-locking linear plate with four screws tested in a mono- and bicortical screw fixation, **b** locking linear plate with four screws also tested in a mono- and bicortical screw fixation, **c** locking double-row plate with six holes fixed with four monocortical screws, **d** locking double-row

plate with eight holes with an alternated monocortical four screw fixation and **e** locking double-row plate with eight holes fixed with eight monocortical screws. Plates **a**, **b** and **c** have a similar length (22 mm). Plates **d** and **e** have a length of 29 mm

with four bicortical screws. The locking linear plate with a similar length was also set with four monocortical (group 3) and four bicortical screws (group 4). In group 5, the locking double-row plate with six holes was attached with four monocortical screws. This double-row plate has the same length as both linear plates (22 mm). The longer locking double-row plate (29 mm) with eight holes was fixed with four monocortical screws in group 6 and with eight monocortical screws in group 7. The location of the screws was controlled by dynamic fluoroscopy (Fig. 3). Each osteosynthesis was performed by the same surgeon. For the locking double-row plate with eight holes the distance from the edge to the force application was modified from 10 to 15 mm and data were calculated by lever principle. After the biomechanical tests the mechanism of failure was analysed by gross examination. Data were evaluated statistically by analysis of variance (ANOVA).

Results

In the first four groups a non-locking linear plate and a locking linear plate with a comparable length (22 mm) and diameter (1 mm) were tested. The maximum load for group 1 (non-

locking linear plate with four monocortical screws) was 250 ± 56 N and for group 2 (non-locking linear plate with four bicortical screws) 359 ± 90 N. In group 3 (locking linear plate with four monocortical screws) the maximum load was 216 ± 38 N and in group 4 (locking linear plate with four bicortical screws) 256 ± 39 N. For the stiffness similar results were measured in the four groups: 47 ± 13 N/mm, 56 ± 21 N/mm, 53 ± 23 N/mm and 58 ± 14 N/mm.

In groups 5–7, locking double-row plates were tested, all fixed with monocortical screws. The locking double-row plate with six holes (group 5) scored a maximum load of 272 ± 71 N and a stiffness of 51 ± 31 N/mm. For group 6 (locking double-row plate with eight holes fixed with four monocortical screws), a maximum load of 302 ± 71 N and a stiffness of 43 ± 9 N/mm was measured. The eight screw fixation of the same plate (group 7) reached 448 ± 78 N and 44 ± 8 N/mm for the parameters. The results of all groups are shown in Fig. 4.

Comparing the linear plates with bicortical screw fixation, for the non-locking plate (group 2) a significantly higher maximum load was measured, as compared to the locking plate (group 4) ($p=0.01$). The bicortical four screw fixation of the non-locking linear plate (group 2) also reached a significantly higher maximum load matched with

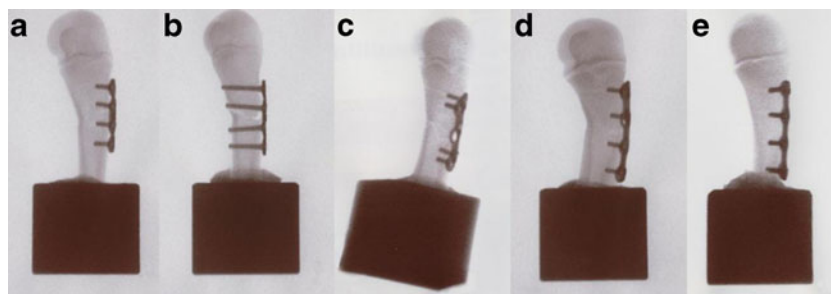


Fig. 3 The lateral radiographs of the different plate designs. **a** Linear plates with four holes fixed with four monocortical screws, **b** linear plates with four holes attached with four bicortical screws, **c** locking double-row plate with six holes with a four monocortical screw

fixation, **d** locking double-row plate with eight holes with four alternated monocortical screws and **e** locking double-row plate with eight holes with a monocortical eight screw fixation

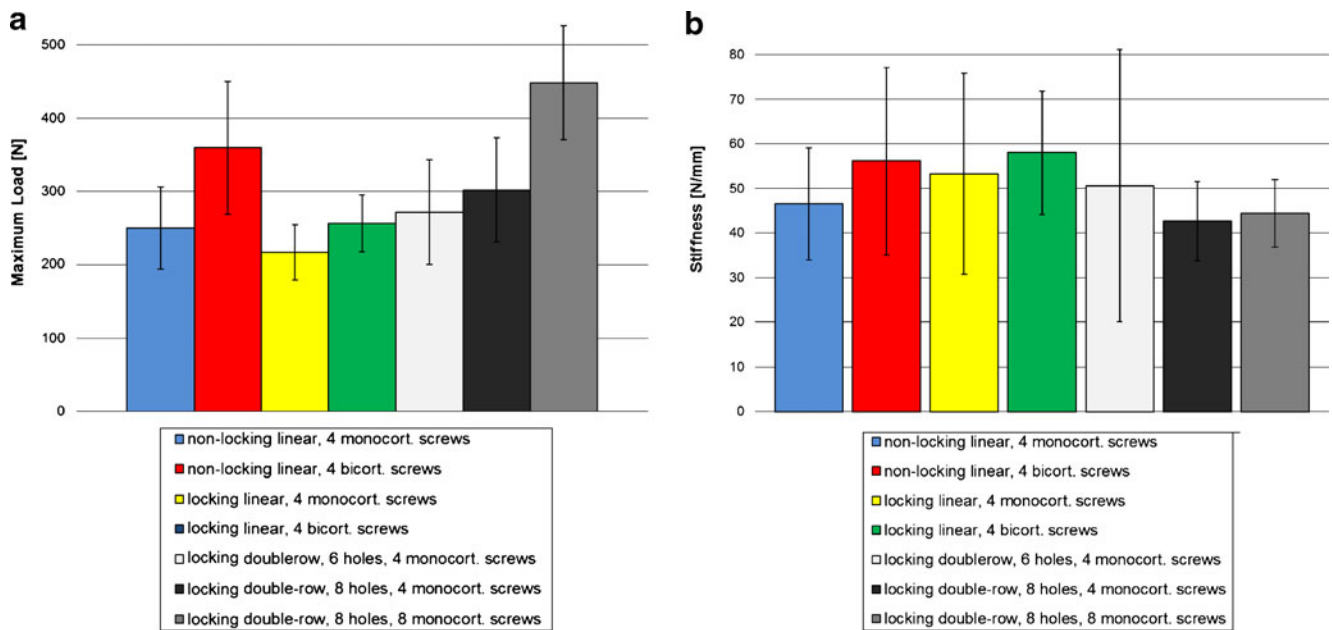


Fig. 4 a The maximum load for the different groups. There was a significant difference between group 2 and groups 1, 3, 4 and 5. No difference was found between groups 5 and 6 and between groups 2

and 6. For group 7, the maximum load was higher than for every other group. **b** The stiffness for the different groups. There was no significant difference in stiffness between the groups

the locking double-row plate with six holes (group 5) ($p=0.02$) and a maximum load comparable to the locking double-row plate with eight holes fixed with four screws (group 6) ($p=0.13$).

For the locking linear plate the tests showed a significant difference between the monocortical (216 N) and bicortical screw fixation (256 N) ($p=0.04$) (groups 3 and 4). There was also a significant difference for the maximum load comparing the locking double-row plate with six holes (group 5) and the locking linear plate in the monocortical screw fixation (group 3) ($p=0.04$). No significant difference was shown for the maximum load of groups 5 and 6 (both locking plates fixed with four monocortical screws) ($p=0.36$). The locking double-row plate with eight holes and an eight screw fixation (group 7) reached a significantly higher maximum load compared to every other fixation method. There was no significant difference for the stiffness between the seven groups (Fig. 4b).

Mode of failure

After the tests the macroscopic mode of failure was examined. In 86 % of all tests, the locking plates showed a loss of the interlocking mechanism. In many cases, a deformation of the hole was observed; thus, the head of the screw could slide through (Fig. 5). Only in 14 % of the tests was fracture of the bone around the screws determined as the mode of failure. In contrast, for the non-locking linear plates, the deformation of the holes was not observed. The typical mode of failure for the non-locking osteosynthesis was the fractured bone around the screws.

Discussion

Metacarpal fractures are a common injury of the upper extremity. Malposition and failure of rotation have a great negative influence on the function of the hand. So the aim of an operative treatment is the reposition of the fragments in an anatomical position and an internal fixation ensuring a high primary stability for an immediate postoperative rehabilitation. A plate will be fixed dorsally on the metacarpal bone affected by bending forces through the strong flexor tendons. Directly over the plate the extensor tendons are located. To avoid postoperative adhesions, early functional rehabilitation is needed. Different locking plates have been invented in recent years for hand surgery inspired by the other areas of trauma surgery. For clinical use, advantages are expected especially concerning the primary stability and



Fig. 5 In many cases, the interlocking mechanism fails by a deformation of the plate hole. Thus, the screw could slide through the hole

the possibility of monocortical screw fixation to protect the flexor tendons.

In a recent study, locking linear plates with a mechanism of interlocking by drilling titanium in titanium with different grades were compared to a non-locking linear plate [8]. The tests showed a comparable stability of a monocortical locking screw fixation and a bicortical non-locking screw fixation. Gajendran et al. demonstrated advantages of locking double-row plates with an interlocking mechanism by titanium deformation for comminuted metacarpal fractures, as compared to non-locking plates [7]. Also in that study the locking plates were attached only using monocortical screw fixation.

In our study, for the first time another mechanism of interlocking was now tested: the TriLock system. The influence of implant design (linear vs double row) and number of screws used was investigated and the locking plates were compared to matching non-locking plates.

Comparing the locking linear plate and the locking double-row plate with six holes (groups 3 and 5), both with a comparable length and four monocortical screws, there was a significant difference: The double-row design increased the primary stability. The results also showed a clear increase of the maximum load when the number of screws was doubled using the same locking double-row plate (groups 6 and 7).

When a locking linear plate was compared with a non-locking linear plate in the bicortical screw fixation (groups 2 and 4), the non-locking linear plate, in contrast to our expectations, was significantly more stable. A possible explanation may be found in the mode of failure. For the non-locking linear plate, the bone usually broke around the screws and was the weakest point of the osteosynthesis. In contrast, for the locking plates, already at lower forces a loss of the interlocking mechanism was observed in many cases.

In order to achieve a high stability with a monocortical screw fixation, similar to the stability observed with the non-locking linear plate in bicortical fixation (group 2), a locking plate with a double-row design and an increased length (groups 6 and 7) may be used.

Metacarpal bones are physiologically affected by bending moments simulated by a three-point bending test model. Previous investigations showed comparable forces in the modified test setup that was also used in this study compared to other three-point or four-point bending test models [12]. In this study, only maximal forces were investigated in load to failure tests. Thus, the absolute stability of different plate designs and interlocking mechanisms could be determined. In further studies, cyclic loading with lower forces simulating continuous hand motion may additionally be evaluated to determine the relative stability and displacement without complete failure of the osteosynthesis.

Interestingly, for the bending stiffness (influenced by the modulus of elasticity of the plates/bones and moment of inertia of area) no differences could be detected between the different osteosynthesis techniques, i.e. the elastic deformation of the plate-bone constructs is similar in all groups. In other words, under the same force all groups show a similar displacement up to the plastic deformation.

In future studies, with regard to the mode of failure observed for the locking plates, some minor adaptations may increase the stability of the interlocking mechanism and reduce plate deformation around the screw holes, thus further improving clinical utility. Furthermore, other fracture types like comminuted or metaphyseal fractures should be investigated. In this study, we could not find a mechanical improvement with locking plates with the TriLock system in the setting of a simple oblique diaphyseal fracture of the metacarpal bone. However, from our clinical experience with these implants we would expect advantages in comminuted fractures with missing buttress of the volar cortex of bone as well as in metaphyseal joint fractures.

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Conflict of interest The authors declare that they have no conflict of interest.

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