Anatomical plate configuration affects mechanical performance in distal humerus fractures

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ABSTRACT

Background: Because of strong loads acting in the elbow joint, intraarticular fractures with a metaphyseal comminuted fracture site at the distal humerus demand a lot from the osteosynthetic care. Ambiguities arise concerning the anatomic position of the implants and the resulting mechanical performance. The aim of this biomechanical study was to compare the performance of different anatomical plate configurations for fixation of comminuted distal humerus fractures within one system.

Methods: In an artificial bone model two perpendicular and one parallel plating configuration of a dedicated elbow plating system were compared with respect to system rigidity (flexion and extension) and dynamic median fatigue limit. The flexion tests were conducted under 75° and the extension tests under 5°. Furthermore, the relative displacements were recorded. As a fracture model an AO C 2.3-fracture on an artificial bone (4th Gen. Sawbone) was simulated via double osteotomy in sagittal and transversal plane.

Findings: Large differences in mechanical performance were observed between flexion and extension loading modes. In extension the parallel configuration with lateral and medial plates achieved the highest bending stiffness and median fatigue limit. In flexion the highest bending stiffness was reached by the construct with a medial and a postero-lateral plate. Failure of the implant system predominantly occurred at the screw–bone interface or by fatigue of the plate around the screw holes.

Interpretation: All three plate configurations provided sufficient mechanical stability to allow early postoperative rehabilitation with a reduced loading protocol. Although the individual fracture pattern determines the choice of plate configuration, the parallel configuration with lateral and medial plates revealed biomechanical advantages in extension only.

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1. Introduction

Fractures of the distal humerus in adults are often challenging in operative treatment. In young patients distal humerus fractures are commonly caused by high velocity injuries with complex fracture patterns and associated injuries. In contrast, distal humerus fractures in elderly are predominantly low velocity injuries complicated by poor bone quality and severe comminution of the bone (Korner et al., 2005; Srinivasan et al., 2005). These fractures remain a problem in trauma surgery (Palvannen et al., 1998; McCarty et al., 2005).

Currently the golden standard in the treatment of distal humerus fractures is open reduction and internal fixation (ORIF) with plates or screws (Jupiter et al., 1985). The development of osteosynthesis implants changed from the 1st generation implants with non angle stable design to the 2nd generation of monoaxial angle stable devices and the present 3rd generation plates with polyaxial locking screws and anatomical designs. These 3rd generation implants differ in plate designs and have various attachment locations to the bone depending on the manufacturer (Stoffel et al., 2008). The aim of every osteosynthesis is high mechanical stability to allow early rehabilitation (Greiner et al., 2008) and adequate mechanical stiffness to stimulate fracture healing. Stable osteosynthesis is particularly crucial for the outcome of the patients with elbow fractures. This has been demonstrated for the reduction of postoperative elbow stiffness and the rate of malunions (Jupiter, 1995; Doornberg et al., 2007; Greiner et al., 2008). The two most popular plating techniques for distal humerus fractures are parallel plating with medial and lateral plates and perpendicular plating with medial and postero-lateral plates. Both techniques possess differences with respect to surgical approach, fixation options of the fracture fragments, and load distribution between implant and bone. Not surprisingly, clinical and biomechanical studies have identified differences in the performance between both plating configurations (Helfet and Hotchkiss, 1990; Schemitsch et al., 1994; Korner et al.,...
2003; Korner et al., 2004; Sanchez-Sotelo et al., 2007; Arnander et al., 2008; Greiner et al., 2008; Schuster et al., 2008). However, different plate systems and designs made the comparison of previous studies difficult to allow a statement on which configuration constitutes the optimal osteosynthesis in the treatment of complex distal humerus fractures. Recently, two biomechanical studies focused on the plate configurations comparing a 90° plate configuration to a 180° configuration (Schwartz et al., 2006; Schuster et al., 2008). In both studies systems of different manufacturers with different plate designs were compared. Thus the comparison of the parallel and perpendicular plating techniques was obscured by differences in plate mechanics, screw configurations and locking techniques.

The present study was therefore designed to compare the mechanical performance of one locking plate system in different anatomical plate positions for the stabilization of distal humerus fractures. We hypothesized that different anatomical configurations of identical plate designs would result in different mechanical stability of fractures of the distal humerus. Our specific aim was to determine the primary stiffness of the osteosynthesis constructs and to identify their failure mechanisms under fatigue loading.

2. Methods

2.1. Fracture model

Artificial humeri assuring a high reproducibility were used for mechanical testing. The cortical shell of this substitute bones offer a compressive strength of 157 MPa, a compressive modulus of 16.7 GPa and a density of 1.64 g/cm³ (Large Left 4th Generation Composite Humerus, Sawbones, Malmö, Sweden). These bones simulate strong bone and are widely accepted as an adequate bone substitute material (Chong et al., 2007; Dunlap et al., 2008). The bones were osteotomized in two planes to simulate an unstable distal humerus fracture of AO type 13 C2.3 (intraarticular fracture site and a metaphyseal comminution area). First a 6 mm wide osteotomy was created in the transverse plane parallel to the humeroulnar joint axis to simulate the metaphyseal comminution, starting at the widest distance of the epicondylus lateralis and epicondylus medialis. Second, a 0.5 mm osteotomy was performed in the sagittal plane to realize the involvement of the joint starting at the deepest point of the trochlea humeri (Schuster et al., 2008). Osteotomies were created after implant application with a handsaw. For a better handling during testing, the humeri were cut off at midshaft.

2.2. Implant system and configurations

The prepared humeri were equipped with a prototype version of the VariAx Elbow Locking Plate System (Stryker Osteosynthesis, Selzach, Switzerland). This system includes four different plates for the distal humerus: medial, lateral, postero-medial and postero-lateral. All plates possess almost the same geometric parameters such as width, length and thickness. Nevertheless, for anatomical reasons the plates differ in shape. The locking technique in these plates (“Smart Lock”) works by using two different grades of titanium. The head thread of the harder screw material engages the weaker circular lip in the plate. 3.5 mm or 2.7 mm locking screws can be locked within a 30° cone in the plate hole, giving it the adaptability to aim the screw at the correct anatomic position. In this study 2.7 mm screws were used to simulate a fragile fixation configuration.

Three different implant configurations (Fig. 1) were compared to each other: 1. Postero-medial-Lateral (PML), 2. Postero-lateral-Medial (PLM) and 3. Medial-Lateral (ML). The PML and PLM plate positions showed a 90° angle and the ML plates were attached in a 180° angle. Screw holes were filled with a minimal number of 2.7 mm screws, simulating a fragile treatment approach.

PML The postero-medial plate was fixed to the bone with four screws: one screw in the distal fragment and three screws in the proximal fragment. The appropriate lateral plate was fixed to the bone with six screws: three screws at the distal fragment and three in the proximal shaft.

PLM The postero-lateral plate was attached to the bone filling five holes: two holes at the distal fragment, and three holes at the proximal fragment. For the medial plate again five screws were used: two at the distal fragment and three at the proximal fragment.

ML For ML configuration the medial plate was fixed with two screws at the distal fragments and with three screws at the proximal fragment. The lateral plate was fixed using two screws distally and three screws proximally.

The construct of a postero-medial and postero-lateral plate position was not tested because in the clinical application this would require an extensive denudation of the bone and the plates would collide at the shaft region.

2.3. Implantation technique

One senior trauma surgeon was responsible for the osteotomy and implantation of plates. The reduction of the bone was performed anatomically with clamps and a 6 mm aluminum block was used as temporary placeholder and to guarantee identical gaps of the Sawbones preparations. In pilot preparations standardized screw positions and lengths were selected and documented by X-ray for the following preparations. Screw collision inside the bone could be avoided by this procedure. The anatomically adapted plates had to be slightly adapted to the specifications of the Sawbones with bending irons. The articular block was fixed with screws from the lateral or medial plate and no additional single screw was used. Only the first screw for fixation of the plate in the long hole was a non angle stable screw. This screw was removed before testing. Screws were predrilled with a 2.0 mm drill and tightened with a torque key and a limitation of 1.4 Nm. At the proximal fragment screws were all inserted in a bicortical manner with an excess length at the opposite cortical bone of about 3 mm. An additional compression screw in the medial lateral direction to stabilize the intraarticular fracture was not used. 10 samples were prepared for each configuration.

2.4. Test set up

To enable the specimens to be mounted in the testing machine, the proximal ends of the distal humeri were potted in a fast-curing methyl methacrylate-based resin (Technovit 3040, Heraeus Kulzer, Wertheim, Germany) and embedded in specially designed aluminum cups holding the distance between resin and elbow joint axis at 70 mm. These aluminum cups were used as an embedding mould as well as a clamping device during testing. They were attached to a custom designed tensioning bracket which enabled the adjustment of the flexion angle. In our case flexion angles of 5° (defined as extension) and 75° (defined as flexion) were used (Fig. 2) (Schuster et al., 2008). A support tripod minimized the bending moment in the humerus shaft next to the clamping device while testing at 75°.

From the literature it is known that 60% of the applied forces in the elbow joint act at the humeroradial joint and 40% at the humeroulnar joint. This load situation was realized using a compensator with an eccentric bearing (Schuster, 2004; Schuster et al., 2008). Force was induced over the points of support at the distal fragment at the deepest point of the trochlea humeri (40%) and at the capitulum humeri (60%) (Halls and Travill, 1964; Morrey and Sanchez-Sotelo, 2008). The interface between the bone joint area and the compensator was manufactured from a low friction synthetic material. To provide
an anatomic situation by excluding transverse forces, a linear bearing was used as an interface between the load cell and the frame of the compensator. All tests were carried out with a servo hydraulic testing machine (Instron 8874, Instron Ltd, High Wycombe, UK) equipped with a two-channel load cell (10 kN), accuracy class 1. Load and displacement were recorded at a frequency of 20 Hz for static testing. For dynamic testing every 50th loop was recorded with the machine’s own systems. Tests were controlled with the machine’s own Software (RS Labsite V2.3 SP1.74, IST, Darmstadt, Germany).

2.5. Test procedure

The properties of the Bone–Implant–Construct (BI) were investigated in two different mechanical experiments.

A destructive static test was carried out in flexion \( (n = 1) \) and extension \( (n = 1) \) to estimate the failure level of the BI. Based on these results, the yield strength \( R_{p0.2} \) was estimated to provide a starting point for non-destructive testing.

A non-destructive static testing was performed in flexion and extension to determine the system stiffness of the BI from the slope of the load displacement curve. To demonstrate the clinical relevance of the system stiffness, the gap displacement at a load level of 300 N was retrospectively calculated by subtracting the displacement values of an intact bone from the total machine displacement of each specimen. All constructs were loaded twice up to a peak load of 60% of \( R_{p0.2} \) describing the linear elastic range of the BI. Three cycles of preconditioning accounted for initial settling during the measurement cycles. Loads were applied in displacement control using a plunger speed of 0.01 mm/s (5° extension) and 0.1 mm/s (75° flexion).

Finally, the Median Fatigue Limit (MFL) according to ASTM STP 731 was calculated in a dynamic experiment at 5° extension (Little, 1981). In this standardized process the load application started at 50% of

Fig. 1. Application configurations: PLM 90° with postero-lateral and medial plate (left), PML 90° with postero-medial and lateral plate (middle) and ML 180° with medial and lateral plate (right). Top down: dorsal–ventral view, caudal–cranial view, fluoroscopic image.
R_{p0.2} and was increased for the BI in increments of 10% in case the specimen reached the run out criteria. If a failure occurred during testing, the load level of the following BI was decreased by 10%. Cyclic loading was applied with a sinusoidal load curve at a frequency of 3 Hz and a load ratio of F_{min}/F_{max} = 1/10.

The run out criteria were set to 250,000 cycles according to the number of movements during fracture healing (approximately three months) of the upper extremities or to a displacement of the distal fragments in proximal direction (subsidence) of 4 mm, whichever occurred first.

Six samples were used for the determination of the BI stiffness and 8 samples to determine the MFL. The fracture pattern of each BI was documented by photography.

From construct stiffness the arithmetic mean and standard deviation was determined. Differences between the different constructs PLM, PML and ML were tested for significance with a one factorial (plate configuration) ANOVA and a subsequent posthoc Bonferroni test for flexion and extension direction (SPSS 14.0 Chicago, IL, USA); the level of significance was defined as $P = 0.05$.

3. Results

The direction of the load application had a significant influence on the system stiffness. For all three configurations the stiffness measured in extension showed significantly higher values than in flexion ($P < 0.05$). The larger difference was found for the ML 180° configuration which was almost 10 times stiffer in extension than in flexion (Figs. 3 and 4).

To demonstrate the clinical relevance of the system stiffness the gap displacement was reported at a load level of 300 N. During flexion loading the fracture gap movement can be described as almost parallel translation between the proximal and distal fragment in ventro-dorsal direction (Table 1). The amount of gap displacement in flexion was smallest in the PLM 90° configuration. The other configurations had significantly ($P \leq 0.001$) larger gap displacements. During extension loading the gap movement occurred in distal-proximal direction with closing of the gap. The amount of gap displacement was smaller in extension compared to flexion loading. In extension the largest displacement was found for the PLM 90° configuration. Displacements were significantly ($P \leq 0.001$) smaller for both the PML 90° and the ML 180° configuration compared to the PLM 90° configuration.

The Median Fatigue Limit of the BIs depended on the configuration of the plates (Fig. 5). The highest value for the MFL resulted for the configuration ML 180° (mean 1046 N (SD 46 N)) followed by PML 90° (mean 743 N (SD 86 N)) and PLM 90° (mean 530 N (SD 64 N)). All differences were statistically significant ($P < 0.05$).

During fatigue testing two different types of failure mechanisms were observed. The first type was damage of the implant system and the second type was damage of the bone substitute material. The failure patterns of the first type included plate breakage, screw breakage, and screw loosening. In the second failure type mainly damage to the cancellous bone substitute material was found.

PLM 90° In the PLM 90° configuration failure exclusively occurred at the postero-lateral plate. The plates showed a fatigue breakage next to the fracture site in the cross section area of the most distal screw of the proximal fragment. In some cases the plates broke directly at the fracture site. No relevant screw loosening was observed. Thus, the subsidence of the distal fragment in proximal direction during 250,000 cycles was only about 0.3 mm until the construct failed by plate breakage.

![Fig. 2. Test set up: Angle adjustable clamping device in flexion position (left) and in extension position (right); load transmission (40% ulna, 60% radius) by applicator combined with eccentric compensator; linear bearings between load cell and applicator to exclude transverse forces.](image1.png)

![Fig. 3. Construct stiffness (means and standard deviation) in extension and flexion direction of the bone implant configurations: PML 90° (postero-medial, lateral), PLM 90° (postero-lateral, medial), ML 180° (medial, lateral).](image2.png)
PML 90° With this configuration 3 out of 4 specimens failed as a result of screw breakage in the distal fragment. Compared to the PML 90° configuration only one plate breakage occurred. No clear statement can be made about the screw failure pattern because during the test series all distal screw positions were affected by screw breakage. Furthermore the loss of angular stability of the most distal screw of the postero-medial plate contributed to the subsidence in the proximal direction which was about 2 mm during 250,000 cycles. The subsidence distance in this case is also the sum of the displacements caused by plate breakage, screws failure, and bone failure.

ML 180° In comparison to the 90° configurations, no plate breakage was observed. In all cases of failure, screw breakage led to destabilization of the BI followed by a breakdown of the osteosynthesis system. All broken screws failed next to the head–thread transition. All screw positions on both plates (medial and lateral) of the distal fragment were affected whereas screw failure at the proximal fragment never occurred. The subsidence of the distal fragment in proximal direction averaged 0.3 mm at a load level of mean 1046 N (SD 46 N) (MFL).

4. Discussion

In this study, three possible application positions of distal humerus plates of one implant system were compared to each other. Biomechanical tests were performed to investigate the system stiffness, median fatigue limit and failure mechanisms. From the biomechanical investigation it can be concluded that the parallel plate configuration in 180° positioning of the osteosynthesis is the most stable construct to withstand the highest in vivo loads in extension of the elbow. An alternative solution is the postero-medial plate combined with a lateral positioned plate in a perpendicular configuration, although this construct offers slightly lower mechanical stability.

Double plate osteosynthesis is the standard treatment for intercondylar or supracondylar fractures of the distal humerus (Jupiter et al., 1985). The majority of these fractures are AO type 13-C fractures. In this study a AO-C2.3 fracture model was simulated because it is still a technically challenging type of intraarticular distal humerus fracture (Korner et al., 2005; McCarty et al., 2005; Russell et al., 2005). AO 13-A3 fractures might be an indication for a double plate osteosynthesis as well, if the bone stock is poor and the distal fragment is small. The 6 mm gap was intended to simulate the instability between the distal and proximal fragments due to the comminution of the bone and to prevent the load transfer through bone contacts. Furthermore, no gap bridging screws were implanted even though the polyaxial plates would offer the positioning of such screws. In the clinical application a gap bridging screw is desirable to increase the stability of the construct. For the biomechanical model a set up was chosen which enabled testing of the distal humerus in a flexion and an extension loading mode (Schuster et al., 2008). Therefore, the mechanical performance of the plating configuration could be assessed in the two most relevant physiological loading conditions of the elbow joint (Amis et al., 1980).

The load transfer in the elbow joint can be described by a two column model (Halls and Travill, 1964). The medial ulnar column and the lateral radial column form the articuloc block. The lateral column shares 60% of load and the medial column 40% (Halls and Travill, 1964; Schuster et al., 2008). This two column model is the basic principle of the double plating osteosynthesis of C-type fractures of the distal humerus (Sanchez-Sotelo et al., 2007).

Previous biomechanical investigations proofed the advantages of the double plating technique respecting the two column model of the distal humerus (Helfet and Hotchkiss, 1990; Damron et al., 1994). In the first biomechanical publications the perpendicular positioning of the plates was propagated as stiffer construct with high initial stability to allow early rehabilitation (Helfet and Hotchkiss, 1990; Rüdi et al., 2007). Korner et al. (2004) confirmed in their study the weakness of a posterior position of both plates. Using non-locking and locking reconstruction the poorest results were achieved by a posterior plate position (Korner et al., 2004). In the current investigation a positioning with a postero-medial and postero-lateral plate would have been possible, but was discarded based on the poor results of the preceding studies and also because of the extensive denudation required to apply these plates.

<table>
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<tr>
<th>Configuration</th>
<th>PML 90° Flexion</th>
<th>PML 90° Extension</th>
<th>ML 180° Flexion</th>
<th>ML 180° Extension</th>
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<tr>
<td>Load direction</td>
<td></td>
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<tr>
<td>Gap displacement at 300 N in mm</td>
<td>1.02 ± 0.10</td>
<td>0.51 ± 0.05</td>
<td>1.92 ± 0.16</td>
<td>0.13 ± 0.02</td>
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<tr>
<td>Fracture gap Movement</td>
<td>Parallel</td>
<td>Gap</td>
<td>Parallel</td>
<td>Cap</td>
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<td></td>
<td>Translation</td>
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Recent publications found a higher stiffness and strength of the osteosynthesis in parallel plating technique compared to the perpendicular technique with different plate designs. The mechanical advantages of a parallel plate configuration have been demonstrated for conventional plate design (Arnander et al., 2008) as well as for locking plate constructs (Self et al., 1995; Schwartz et al., 2006). Schuster et al. (2008) compared three 90° plate systems (CRP, LCP, DHP) all in clinical use with a similar test set up. The stiffness values were determined for flexion and extension and were found to be smaller compared to the values in this study (Schuster et al., 2008).

Although in our study identical designs were employed, the three different anatomical configurations demonstrated large differences in stiffness both in the extension and flexion direction. The system stiffness is influenced by two kinds of factors: factors which cannot be influenced by the surgeon and factors which allow the individual adjustment of an osteosynthesis. The initial situation is the fracture pattern and fracture geometry with the number and shape of the fragments. This initial fracture situation strongly dictates the options for plate positioning. On the other hand the overall construct stability can largely be influenced by placing the plates at different anatomical positions. In the 180° configuration both plates are loaded in the direction of the highest 2nd moment of inertia while in the 90° configuration the 2nd moment of inertia of the second plate is lower than that of the other plate. That is one reason why the stiffness of the 90° applications is much lower, in flexion. Less stiffness results in higher deflection angles, which again leads to high local strain values. In the case of dynamic testing, this explains the breakage of the plates in the 90° configuration. In contrast, in the 180° configuration the plates showed no breakage. Another important factor is the direction of load application itself: flexion or extension. Changing the point of load application means changing the geometric situation, primarily the lever arms. This explains the differences in stiffness between flexion and extension; in flexion the lever arms are much longer than in extension and therefore stiffness decreases (Fig. 2).

Controllable factors influencing the construct stiffness are the screw position and resulting bending length of the plates, the number of screws in one single fragment and the size of the screws. In the current study the 180° configuration achieved much less stiffness in flexion than in extension, because the most distal screw holes of the proximal fragment were not occupied resulting in a relatively long unsupported distance across the fracture gap. Therefore, in flexion high shear forces were induced to the most distal screws of the proximal fragment. In extension the long distance across the fracture gap decreased the stiffness of the construct, but especially during dynamic testing this distance reduced the stress concentration in the plate. The discrepancy between flexion and extension at the configurations PML 90° and ML 180° would have been less pronounced if the most distal screws of the proximal fragment had been implanted. Furthermore, the number of screws in one single fragment influences the stiffness. By inserting only one screw (PML 90°, postero-medial plate) in one fragment, this fragment has one rotational degree of freedom with a negative effect on the stiffness of the construct. Additionally the screw is predominantly loaded in bending which leads to early failure. At least two screws are necessary in the distal fragment to achieve sufficient rotational stability of the joint fragments. Our findings suggest the following mechanism for screw failure in dynamic testing in extension. The screws were primarily exposed to shear forces. The shear forces gradually destroyed the screw–cortical bone interface and the bending forces on the screws subsequently increased. The typical location for the observed screw breakage was the thread–head transition, which is the direct consequence of the increased bending forces. In this area the highest bending moments occur. Using two screws per fragment, the bending moment at the thread head transition can be reduced. Previous studies performing similar investigations used at least 2 screws per distal fragment, but have not mentioned a specific correlation concerning number of screws and stiffness (Schemitsch et al., 1994; Korner et al., 2004; Schuster et al., 2008).

In the present study the minimum number of placeable screws into the distal fragments was one in the postero-medial plate and three screws in the lateral plate (PML) (Fig. 1). The PML construct had one transcondylar screw and the ML construct four transcondylar screws depending on the number of placeable screws due to plate design (Fig. 1). The differences of construct stiffness in extension might be explained by the smaller number of distal screws. This enabled an opening of the intraarticular fracture gap during cyclic loading. From our data it can be assumed that the number of transcondylar screws is one of the major factors for the stability of the system loaded in extension direction. The influence of the diameter of the screws may also play a role but was not tested in this study.

The parallel ML 180° configuration showed a high stiffness in extension direction (Fig. 3) and a very high MFL (Fig. 5) during dynamic testing. Similar behavior was found with the PML 90° configuration. However, during dynamic testing this configuration was in an inferior position compared to ML 180°. Both configurations showed a very low stiffness in flexion (Fig. 4) which could be overcome by inserting additional screws closer to the fracture gap. At last the ML 180° offers more options for the placement of transcondylar screws due to plate design. The low stiffness of the PML 90° configuration in the extension direction was caused by only one transcondylar screws and the lower MFL was caused by the early plate breakage. The stiffness in flexion exceeded the stiffness of PML 90° and ML 180°, with the screw configuration used in our model. It is not known which construct stiffness will lead to the most beneficial amount of gap movement for healing. We assumed that avoiding excessive movements by increasing the construct stiffness would improve the mechanical conditions for successful fracture healing (Claes et al., 1997).

In vivo acting forces in the elbow joint range from 2800 N at an angle of 30° to approximately 0 N at an angle of 145° (Amis et al., 1980). None of the configurations came close to the in vivo load levels in dynamic testing. From our biomechanical data it must be concluded that early postoperative rehabilitation especially in comminuted AO 13-A3 or 13-C3 fractures should only include passive movements of the elbow joint in the first four weeks (Hak and Golladay, 2000).

In the current study a minimalist screw configuration was chosen by inserting 2.7 mm screws instead of 3.5 mm screws and occupying only the necessary screw holes. Furthermore, an additional compression screw in the ML direction and generally no gap crossing screws were used. All of these arrangements decreased the system stiffness and degraded the mechanical behavior under cyclic loading. The gap displacement in this study was determined retrospectively based on stiffness reports of the intact and osteotomized bone and was not directly measured at the fracture site. Dynamic testing was only performed in extension because this is the more challenging load application for the elbow joint (Morrey et al., 1988; Schuster et al., 2008). In flexion, the results might be different to those in extension because of differences in force flow. In fact, static testing results in flexion show a trend opposite from those obtained in extension. Apart from axial loading, the elbow joint has to support a considerable amount of torsion. This loading situation was not tested in the current study, thus the results do not account for torsional loading. Artificial bones were used instead of human bone because we were particularly interested in the behavior of the screws and plates and their failure mechanisms. With the artificial bones simulating strong bone it was impossible to account for poor bone quality such as osteoporosis. It was assumed, that poor bone quality would minimally influence the primary stiffness, but would strongly affect the fatigue behavior by failure of screw–bone interface.

At present, there are no clear guidelines in the treatment of distal humerus fractures. Recent treatment recommendations vary from
primary arthroplasty to osteosynthesis with double plating technique. For the type of implant to be used, there are no specifications given. Nevertheless, biomechanical tests are indispensable to assess the properties of these implants. The definitive proof of any of the advantages of these implants however must be provided by clinical studies. The decision of which implant configuration is applied in vivo depends on the individual fracture pattern, soft tissue conditions and not to mention on the experience of the surgeon.

In conclusion, the results from our study demonstrated that all three plate configurations provided sufficient stability in strong bone to allow early postoperative rehabilitation, which is the primary aim of the osteosynthesis treatment. Based on testing in extension, which is the most demanding load scenario for the elbow joint, a parallel plate configuration provides the largest construct stability (Morrey et al., 1988; Schuster et al., 2008).

Conflict of interest

There are no more additional conflicts of interest.

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