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Periprosthetic fracture fixation in osteoporotic bone

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ABSTRACT

Fixation techniques of periprosthetic fractures are far from ideal although the number of this entity is rising. The presence of an intramedullary implant generates its own fracture characteristics since stiffness is altered along the bone shaft and certain implant combinations affect load resistance of the bone. Influencing factors are cement fixation of the implant, intramedullary locking and extramedullary or intramedullary localization of the implant and the cortical thickness of the surrounding bone. Cerclage wires are ideally suited to fix radially displaced fragments around an intramedullary implant but they are susceptible to axial and torsional load. Screws should be added if these forces have to be neutralized. Stability of the screw fixation itself can be enhanced by embracement configuration around the intramedullary implant. Poor bone stock quality, often being present in metaphyseal areas limits screw fixation. Cement augmentation is an attractive option in this field to enhance screw purchase.

Introduction

Since arthroplasty numbers are rising, the periprosthetic fracture fixation becomes more and more a concern [1]. Patient's condition and postoperative requirements render the treatment challenging. A high primary stability is needed due to the fact that partial weight bearing is impossible for many patients. Undiagnosed loosened prosthesis stems, being considered as stable during surgery contribute to the high failure rate actually reported for periprosthetic fracture osteosyntheses [2].

The main fixation techniques currently applied comprise revision surgery with conversion to a longer non-cemented prosthesis stem bridging the fracture zone, plate osteosynthesis and intramedullary nailing.

Although the choice of the fixation method is still individualized depending on the patient's condition and the surgeon's selection, certain biomechanical principles could be deducted from recent biomechanical studies. Varying fracture gap configurations fixed with different plate types and screw configurations have been widely investigated [3–6]. Vice versa, the type of implants and their configuration additionally affect fixation stability [7]. Apart from the working length of the implant, additional factors have to be taken into account, since intramedullary and extramedullary implants are both interacting in a fixed periprosthetic fracture [8]. Interprosthetic fractures, sometimes requiring a fixation in-between two prosthesis stems represent a separate category [9].

Osteoporotic bone quality is a special concern in metaphyseal fracture locations like periprosthetic fractures of the distal femur at non-constrained bicondylar total knee arthroplasties [10]. Screw purchase in this almost cancellous bone area is limited. Implant augmentation is an option to enhance fracture fixation [11].

Current mechanical aspects of periprosthetic fracture fixation are summarized in this article, focusing on implant mechanics and their affection of bone strength as well as the use of augmentation in periprosthetic fracture fixation at the distal femur.

Biomechanics of periprosthetic fracture fixation

Bone quality, stability of the stem anchorage and fracture pattern have direct impact on periprosthetic fracture fixation strategy. Periprosthetic fractures with intact stem anchorage are the domain of osteosynthesis. They are often located around the tip of the stem where the bending stiffness drops, since the bone is not splinted by the prosthesis stem any more. Simple fractures with closed fracture gap and medial cortical support allow partial load transfer via the cortex. In open fracture gap situations like comminuted fractures without medial cortical support a single lateral plate might not be stable enough for weight bearing and a stiffer plate or a double plating construct has to be

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used to prevent implant fatigue failure [6,12]. A finite element analysis study has shown, that anterolateral double plating as well as long stems equalize implant stress distribution, increases construct stiffness and reduces interfragmentary movements at the fracture gap [13].

Not requiring intraosseous anchorage, cerclage wires are almost independent from local bone quality [14]. Their force transmission runs centripetally to the loop, fixing radially displaced fragments. Cerclages are ideally suited for centripetal fracture reduction and fracture fixation on the level of the stem, since an intramedullary splinting is present [15]. Due to the fact that the long bone shaft is not an idealized round tube, the cerclage effects an uneven pressure distribution on the bony surface with high pressure values at the deflection edges [16]. Comparable to the point contact fixation of modern plating systems the cerclage spans from edge to edge with non-loaded zones in-between (Figure 1). The contact surface of a tightened 1.5 mm wire or 1.7 mm cable cerclage ranges from 0.30 to 0.36 mm [14]. In congruently reduced shaft fractures, it is unlikely to produce a fracture or grade cutting by cerclage tightening, since the cortex withstands static concentric pressure [16]. Loading the cerclage fixation revealed no microfracturing at the shaft cortex [16]. The groove formation, the so called biological loosening of a cerclage [15] is induced by the micro movement of an already loosened cerclage and not by the weakness of the cortex itself [15–17]. Instead of cortical bone resorption, a bony ingrowth was observed for well tightened cerclages [16–18]. Noteworthy loss of pretension almost occurs at the twist. Apart from using a larger wire diameter, pretension could be influenced by the twisting procedure. Highest pretension is obtained when the twist is formed under permanent traction by the pliers, the twist is plastically deformed at the end of the twisting procedure, wire ends are cut outside the twist and when the twist is bent forward at the end of the procedure [19]. Backward bending should be avoided, since 90% of the pretension gets lost throughout this manoeuver. When plastic deformation of the twist is accomplished, the twisting procedure has to be stopped before twisting off the wire [19]. Cable cerclages, closed by a crimp achieve higher pretension values compared to wire cerclages. Looping the wire cerclage twice around the bone before closure effects pretension values comparable to a cable cerclage of the same diameter looped once. According to the tackle principle, the twist is less loaded in the double looped configuration and a higher amount of travel is needed to provoke loosening of the cerclage [20]. Ogden proposed a plating construct with cerclage fixation on the level of the stem and bicortical fixation in the opposite fracture fragment [21]. Clinical results of this construct exhibited an overall complication rate of 30% [21] comparable to the failure rate of allograft struts (24%) placed on the lateral and anterior aspect of the long bone shaft and fixed by cerclages [22]. If load is applied in axial direction and torsion, the bone slides under the cerclage [14] if it is not maintained by the interdigitation of the fracture fragments. To add stability in axial direction and torsion, cerclage-plate constructs should be combined with locking screws [5,14].

Screws could be either placed monocortically or bicortically within the narrow bone corridor lateral to the prosthesis stem. Tangential intracortical screw placement which reduces fixation strength has to be avoided during bicortical screw insertion [5,23]. In conventional nonlocking plates, the screw insertion angle could be varied within the plate hole allowing bicortical screw placement. The fixation principle of nonlocking screws, requiring a tight frictional coupling at the plate-bone interface is not suited for osteoporotic bone [24,25]. Multiaxial locking screws are one solution of this shortcoming [26]. Broader plates with laterally placed screw holes [6] or attachment plates shifting the screw entry point to the lateral allow an embracement configuration of the bicortical locking screws, a very effective way to enhance fixation stability (Figure 2) [5,27]. Compared to cerclages combined with monocortical locking screws, the shaft embracement of bicortical locking screws realized by the locking attachment plate provides superior stability especially in fractures with lacking cortical support [27].

Under axial compression force, orthogonal to the screw shaft axis, monocortical and bicortical locking screws of the same diameter achieve comparable fixation strength [14]. In both, most of the load is transferred at the near cortical hole. An ovoid enlargement of the near cortical hole and a longitudinal fissure of the near cortex was observed during failure. Since the fissure is located below the osteosynthesis plate, it could not be detected on conventional radiographs. Detection of the ovoid hole enlargement on radiographs is sometimes possible by meticulous analysis [14]. The neutralization of torsional forces requires bicortical fixation in both, the proximal and the distal fracture fragment [14].

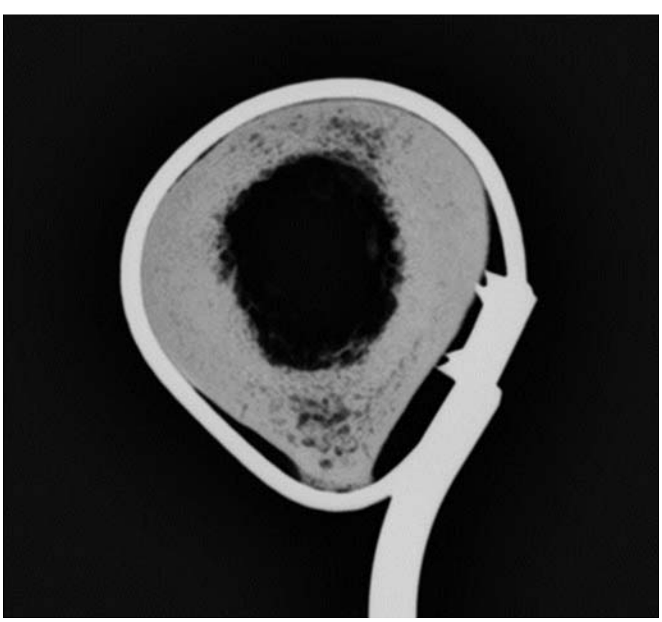

Fig. 1. Cerclages, being deflected at the edges of the bone and providing a point contact fixation with non-contact zones in-between.

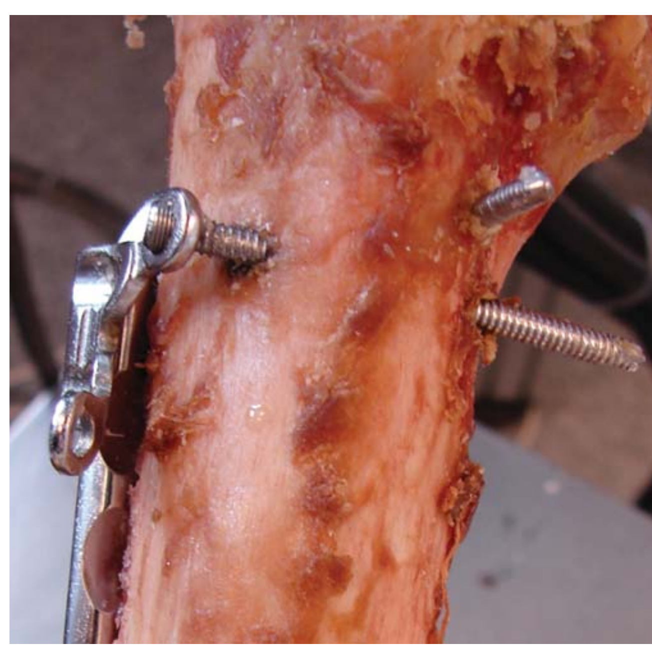

Fig. 2. Plates, shifting the screw entry point more laterally and allowing an embracement configuration of the locking screws around the intramedullary implant.