Intramedullary nailing of long bone fractures

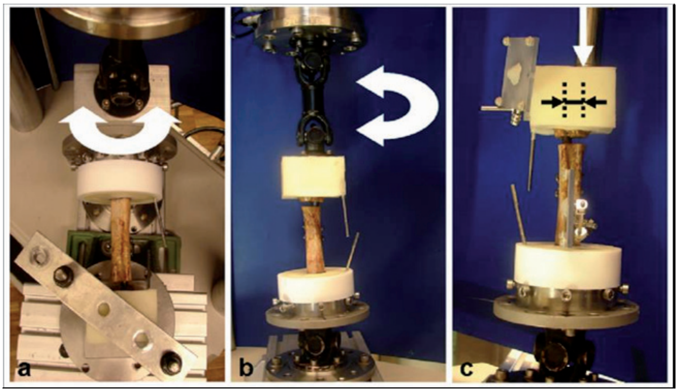
Experimental in-vivo and in-vitro studies

Ph.D Thesis Joachim Horn

Dept. of Orthopaedics, Rikshospitalet, Oslo University Hospital

Faculty of Medicine, University of Oslo

2011



Oslo universitetssykehus

UNIVERSITAS OSLOENSIS MDCCCXI

Series of dissertations submitted to the
Faculty of Medicine, University of Oslo
No. 1062

ISBN 978-82-8072-621-6

Cover: Inger Sandved Anfinsen.
Printed in Norway: AIT Oslo AS.

Produced in co-operation with Unipub.
The thesis is produced by Unipub merely in connection with the
thesis defence. Kindly direct all inquiries regarding the thesis to the copyright
holder or the unit which grants the doctorate.

Contents

1. Acknowledgements

I would like to thank Sverre Barstad, Robert Dokke, Milos Soukup and all the other colleagues from the orthopaedic department at Fredrikstad Sykehus for giving me the opportunity to start a residency in orthopaedic surgery and for supporting my wish to do research at the AO Research Institute in Davos, Switzerland.

I thank Urs Schlegel and Keita Ito at the AO Research Institute in Davos for considering my application as a research fellow and for guiding me through the infection project on intramedullary nails, including the publication of the results. I' m very thankful to all other staff members and colleagues at the AO Research Institute, who contributed to the research projects performed in Davos. The 12 month which I spent at the AO Research Institute in 2003/2004 definitely encouraged me to go on with research and to aim for a PhD.

I am grateful to Boyko Gueoguiev, Karsten Schwieger and Stefano Brianzo from the biomechanical lap at the AO Research Institute and André Schlienger from Synthes for their essential contribution to the two studies on angular stable interlocking which are included in the current thesis. I thank Professor Krettek and Professor Höntzsch for their contributions to the paper on infection restistance and to the paper on angular stable tibia nails respectively.

I thank Daniel Arens and Oliver Boesl for being helpful with accommodation at all research visits in Davos, and I am thankful for their friendship which made each visit to Switzerland a social event.

At Rikshospitalet I would like to thank the head of the department, Trine Sand Kaastad, for given me the opportunity to complete my residency and to complete a PhD Thesis at the orthopaedic department. I'm very thankful to Harald Steen and Professor Olav Reikerås for being my supervisors of this Thesis and for their indispensable help with several research projects, publication of the results and writing of the manuscripts. I thank Per Ludvigsen at the Biomechanics Lab and the Department of Comparative Medicine at Rikshospitalet for all technical assistance concerning the study on the role of the fibula in lower leg fractures.

I would like to thank Professor Terje Terjesen and Leif Pål Kristiansen for being my clinical supervisors and colleagues at Rikshospitalet, and I am glad being part of the paediatric orthopaedic team together with Per Reidar Høiness, Andreas Knaus, Vera Halvorsen, Ola Wiig and Anders Wensaas.

I would like to thank my parents Bärbel and Otto, my sisters Christina and Cornelia, my brother Axel, and their families for being present to me and supporting me at any time. Finally I thank my close friends for their support and their amity.

Oslo, Desember 2010 Joachim Horn

2. List of papers

Paper I

Horn J, Schlegel U, Krettek C, Ito K: Infection resistance of unreamed solid, hollow slotted and cannulated nails: an in-vivo experimental comparison. J Orthop Res 2005. Jul;23(4):810-815, 2005.

Paper II

Horn J, Steen H, Reikerås O: The role of the fibula in lower leg fractures – an in-vivo investigation in rats. J Orthop Res. 2008 Jul;26(7):1027-31.

Paper III

Horn J, Linke B, Höntzsch D, Gueorguiev B, Schwieger K: Angle stable interlocking screws improve construct stability of intramedullary nails in distal tibia fractures: A biomechanical study. Injury. 2009 Jul;40(7):767-771.

Paper IV

Horn J, Gueorguiev B, Brianza S, Steen H, Schwieger K. Two-part surgical neck fractures of the humerus, a biomechanical evaluation of an angular stable intramedullary nail. (Accepted 04.10.2010, Journal of Orthopaedic Trauma)

3. Introduction

3.1. Fracture healing and impaired fracture healing

3.1.1. Fracture healing

A fracture is the result of a single or multiple overloads (Perren and Claes 2000), caused by a direct or indirect force. A single overload leads to the event of an acute fracture, whereas multiple cyclic alternating loads might result in a fatigue fracture (Debrunner 1994). Fracture healing might occur in two ways. Primary bone healing without callus formation or secondary bone healing with typical occurrence of callus formation (Frost 1989a/b, Einhorn 1998, Greenbaum and Kanat 1993). Primary healing with minimal callus or without callus formation seems to occur when anatomic restoration of the fracture fragments is achieved by rigid fixation and interfragmentary compression, and when interfragmentary strain is significantly reduced by the stability of fracture reduction (Perren and Claes 2000, Perren 2008). With absolute stability of fixation the Havaersian osteones cross the fracture zone without obvious change in shape or direction (Schenk and Willenegger 1963, Rahn et al. 1972, Perren 1979, 2002). However, the majority of fractures heals by secondary healing, which involves a combination of intra-membranous and enchondral ossification (Einhorn 1998). Secondary healing involves the classical stages of: formation of haematoma, inflammation, stage of angiogenesis and formation of cartilage, successive stages of cartilage calcification, cartilage removal and bone formation, and ultimately a more chronic stage of bone remodelling. Due to this natural way of secondary fracture healing, untreated fractures in humans and animals will almost always heal without any fixation or treatment (Greenbaum and Kanat 1993, Einhorn 1998).

3.1.2. Impaired fracture healing

The number of non-unions might even have increased after the advent of surgical orthopaedic interventions (Debrunner 1994). However, the residual problem of untreated fractures might be a lack of alignment and shortening, consequently, impairment of function (Perren 1979). According to Frost (1989b), bone healing problems can be divided into: Technical failures, when treatment problems have impaired normal biologic potential; and biologic failures, when biologic malfunctions have made the correct treatment ineffective; and combinations of these. Technical failure might occur as a consequence of infection, poor reduction, distraction, repeated gross motion across the fracture gap and impairment of local blood supply due to injury or surgical procedures (Frost 1989b). In these cases roentgenograms will show adequate amounts of callus, but a pseudarthrosis of some kind. For that reason those failures of fracture healing are defined as hypertrophic non-unions; eliminating excessive motion and infection and improving reduction would usually lead to union of these fractures (Frost 1989b, Debrunner 1994). In biological failure of fracture repair, the biology of the healing processes delays or prevents union even with proper treatment (Frost 1989a/b). Roentgenograms fail to show adequate amounts of callus, a so-called atrophic pseudarthrosis. Malfunction of local or systemic mediator mechanisms might cause a failure to produce callus, a failure to mineralize callus or a maldifferentiation into fibroblasts and/or lipoblasts instead of chondroblasts and/or osteoblasts (Frost 1989a/b). The treatment of biological failure represents a major challenge for orthopaedic surgeons. In order to achieve healing open revision with debridement of the fracture ends combined with bone grafting might be required, whereas simple compression of the bone ends will usually not enhance fracture healing (Frost 1989b, Rodriguez-Merchan and Gomez-Castresana 2004). In certain cases, resection of the pseudarthrosis and corticotomy with segmental transport using the Ilizarov method might be required to achieve union (Garcia-Cimbrelo and Marti-Gonzalez 2004).

3.2. The history of operative fracture treatment and intramedullary nailing

The first techniques of operative fracture treatment were developed in the 19th century, these methods included open reduction and usually rather unstable fixation (Broos and Sermon 2004). However, the operative treatment of fractures did not emerge until the discoveries made by Semmelweiss and Lister in the middle of the 19th century with the usage of asepsis and antisepsis (Funk et al. 2009). Asepsis includes procedures to reduce the risk of bacterial contamination by use of sterile instruments and sterile gloves, whereas antisepsis includes the removal of transient microorganisms from the skin and a reduction in the resident flora (Heberer et al. 1986). According to Tscherne (1969), Jean Baptiste Bérenger-Féraud published his experiences with internal fracture fixation in his book: ”Traité de l’immobilisation directe des fragments osseux” in 1870, Lister successfully performed a suture of a patella fracture with a silver wire in 1877 and at the end of the 19th century Hansman and William Arbuthnot Lane used plates for fracture fixation. Albin Lambotte was the first one using the term “osteosynthesis” in his book: “L’intervention opératoire dans les fractures récentes et ancienne”, which was published in 1907. According to Krettek (2001a), extra-articular closed nailing of femoral neck fractures as implemented by Smith-Petersen in 1925 and Johansen and Jerusalem in 1932 opened the way for closed intramedullary nailing of long bone fractures of which Gerhard Küntscher is rightly considered to be the creator. After thorough theoretical and practical preparation, Küntscher’s first human clinical case was a successful femoral nailing in November 1939 in Kiel, Germany. Together with his collaborator Pohl he developed several different nail designs, whereas the V- and clover-leaf-form became his standard cross section designs. Although Küntscher’s name is closely associated with intramedullary nailing, he was not the first one to stabilise fractures using an intramedullary approach. Among others Lambotte, the brothers Rush, Müller-Mernach and Danis all used intramedullary approaches for fracture treatment (Tscherne 1969). However, their techniques mainly included opening of the fracture site, less stable fixation methods with different materials and the operations were associated with a high rate of complications.

The indications for the first Küntscher-nails were limited to mid-diaphyseal fractures, since construct stability was based on the contact zone between the nail and the bone (Krettek 2001a). Küntscher developed the technique of intramedullary reaming, enabling him to use larger nails with an increase of the nail-bone contact zone. At the end of his life he laid down the basis for locking of intramedullary nails. This technique was later developed further by others (Kempf et al. 1985), enabling extension of indications for intramedullary nailing to more proximal and distal diaphyseal fractures and even metaphyseal fractures.

The surgical treatment of fractures in general underwent important changes around the middle of the last century. Stable internal fixation allowed healing of the fractures while maintaining function of the joints and soft tissues (Perren 2002).

Today intramedullary nailing is a well established technique and the most used fixation method for long-bone fractures of the lower extremity (Krettek et al. 1996, Court-Brown 1998, 1999, Krettek 2001a/b, Bhandari et al. 2001a/b). A Pubmed-search for the term "intramedullary nailing" in January 2010 gives a total number of 7317 hits, illustrating the broad use of the device and the extensive research which is done in the field of intramedullary nailing.

However, from the beginning until today an increased attention has been focused on complications associated with fracture fixation using intramedullary devices. The implementation of intramedullary nailing by Küntscher and mainly reaming of the medullary canal were not accepted by the orthopaedic society without concerns. Goetze from Erlangen wrote, although expressing himself rather courteously: "Trotz aller Anerkennung der unverkennbaren Fortschritte, die mit Hilfe des Küntscher'schen Marknagelung-Prinzipes erreicht worden sind, kann ich doch gewisse Bedenken gegen eine zu weitgehende Indikation nicht underdrücken" (Küntscher 1950).

3.3. Biomechanical considerations

Basic concepts in the field of biomechanics are not different to laws in physics and mechanics. However, this chapter of the current thesis gives only a short presentation of terms and considerations, which contributes to a further understanding of the biomechanics of intramedullary nails and mechanical testing of nail-bone constructs. All explanations and definitions in this chapter refer to the book: "Musculoskeletal Biomechanics" by Brinkmann et al. (2002).

3.3.1. Basic concepts from physics and mechanics

Force

Forces can not be measured nor be observed directly. However, effects of forces like acceleration or deformation of a body can be observed and measured. Newton's second law $F[kg \cdot m/s^2] = m[kg] \cdot a[m/s^2]$ describes the relation between a mass [m], its acceleration [a] and the force [F] which effected the acceleration. The unit $[kg \cdot m/s^2]$ has been given its own name and is termed Newton [N].

Moment

As with forces, moments cannot be observed or measured directly; only their effects can be observed. The moment effected by a force can be defined in different ways, as a number or as a vector. In a two-dimensional problem where the axis of rotation is perpendicular to the plane of interest, the moment [M] can be defined simply as $M = \pm L \cdot F$, whereas L is the distance from the line of action of the force [F] to the fulcrum (axis of rotation). In the context of orthopaedic biomechanics, L is also called the "moment arm".

Load

When discussing biomechanical problems in orthopaedics the term "load" is frequently used. In its strict sense, the term "load" designates a force or moment. Loading by a force is measured in Newtons [N], and loading by a moment in Newton meters [Nm]. The term "mechanical load" is occasionally used in literature to describe some mechanical effects on tissues or implants. This "mechanical loading" might for example include, pressure, friction or deformation. However, these quite different effects require their own specific description.

Mechanical stress

When a force acts on a body, a deformation in the direction of the force is observed. Compressive or tensile forces shorten or stretch the body; shear forces effect an angular deformation. The SI unit for stress is the Pascal (Pa), which is equivalent to one Newton (force) per square meter (unit area) (SI: Système International d'Unités 2010).

Mechanical work, energy, power

Mechanical work E is defined as "force times distance", whereas the distance L is to be measured in the direction of the force ($E = F \cdot L$). The unit of mechanical work is Newton meters [Nm] or Joule [J] (SI 2010). If a person lifts a mass m by the distance L, the human body performs mechanical work $F \cdot L = m \cdot g \cdot L$ [Nm] on the mass, whereas g is earth gravity with 9.81 m/s^2 (SI 2010). The energy to produce this work derives from chemical processes in the muscles. When holding a weight under isometric contraction, the muscles consume energy without performing a mechanical work. Energy is defined as the capacity of a physical system to perform work, whereas mechanical power is defined as mechanical work pr unit time. Thus, power is measured in units of [Nm/s], which is equivalent to watt [W].

3.3.2. Deformation and strength of structures

The deformation of a structure under load depends on a number of variables: the loading mode, the architecture of the structure and the material properties of the building materials.

Strain

Strain is a geometrical measure of deformation. Strain is defined as the length change dL divided by the initial length. As the quotient of two lengths, strain is a dimensionless quantity, but might be described by percentage values. $\varepsilon = dL/L$

Loading mode

When a structure or construct is loaded by a force or a moment, the deformation is measured in meters [m] or in relative units in relation to its initial dimensions [%], while the effects of a moment might result in torsion or bending. When a torsion load is applied, the deformation is measured in degrees [°]; if bending occurs, the deflection is usually measured in meters [m].

The architecture of a structure

The deformation of a beam depends on its length and cross section. The deformation of a bone-nail construct depends on the geometrical and material properties of both the bone and the nail.

The material properties of the building materials

In the case of orthopaedic implants, the deformation depends on the type of material and its specific material properties such as the moduli of elasticity [λ] and shear [G] and by the elastic, visoelastic, or plastic properties.

Experimental determination of deformation and strength

A bending test provides data for deformation and strength of structures under bending forces. In material science strength is the ability to withstand an applied stress without failure. In a force-deformation diagram with the displacement [m] represented on the x-axis and the force [N] on Y-axis, at each point of the curve, the slope is approximated by quotient dF/dL. In this expression dF is the change of force and dL is the change of displacement. The quotient dF/dL represents the "stiffness" of a structure. Stiffness is measured in Newton per meter

[N/m] or [N/mm]. The numerical value of stiffness indicates how many Newtons are necessary to effect a displacement of 1 m or 1 mm, respectively. When displacement under bending is measured in degrees [°] of rotation or angulation, the stiffness might be expressed by [Nm/°].

Depending on what structures are tested, in the testing of for example bone, ligaments or bone-nail constructs, a load-deformation curve does not follow a straight line. The slope of the curve, and thus the stiffness, assumes different values along the curve. When the applied deformation force F exceeds the strength of a structure, a partial or total breakage or rupture will occur. The strength or bending moment of a structure is given by the force where the breakage or rupture occurs (yield point). The deformation energy of the structure is represented by the area under the curve with the unit [N · m].

The deformation and strength of for example tubular bones depend on the bone material properties, the cross-sectional area, and the length of the bone. Effects of orthopaedic implants on the bone can be evaluated by comparing instrumented pairs of bones under the same conditions.

3.3.3. Biomechanical considerations of various implants in orthopaedic trauma

3.3.3.1. Internal fixation plates

Plating of fractures began with Lane in 1895 and Lambotte in 1907. Various developements of plates in terms of metallurgical formulation and plate designs have followed after these first steps of internal plate fixation. Danis in 1949 and Bagby and Janes in 1958 introduced the principle of interfragmentary compression, a principle which was later further developed by Perren and co-workers (1969), resulting in the plate design of the Dynamic Compression Plate (DCP). The advantages of the DCP were low incidence of malunion and stable internal fixation, allowing early active and passiv motion without an additional plaster-cast. However, plate fixation was still associated with certain disadvantages including delayed union, cortical bone loss under the plate and refracture after plate removal (Uhthoff et al. 2006). A hypothesis by Perren and co-workers attributed the problem of porosis and refractures to cortical necrosis that is secondary to excessive plate-bone contact interfering with cortical perfusion (Perren et al. 1988), a hypothesis which led to the development of the plate design called Limited Contact – Dynamic Compression Plate (LC-DCP). This plate design reduced bone-plate contact by approximately 50% (Gautier and Perren 1992). However, there is some uncertainness if this plate design is advantageous in terms of cortical blood flow compared to the conventional DCP (Field et al. 1997, Jain et al. 1999). There is also some evidence that porosis associated with plate fixation might as well occur due to stress shielding induced by rigid plates (Akeson et al. 1976, Jain et al. 1999, Uhthoff et al. 1994). A recent development of plate fixation is the Locking Compression Plate (LCP), which is designed to provide angular stability between the screws and the plate (Frigg 2001).

3.3.3.2. External fixators

An external fixator is a device, with the main body placed outside the skin. The apparatus stabilizes bone fragments through wires or pins, which might be connected to either rings or monolateral tubes. Wires must be under tension and require therefore rings or half rings (Fernandez Dell'Oca 2000, Golyakhovsky and Frankel 1993). The use of transosseous wires and half pins, which are placed percutaneously, requires an intimate awareness of anatomic topography to avoid injury to nerves and vessels (Catagni 2003). External fixation is an excellent tool for stabilisation of open fractures, offering the possibility of atraumatic insertion, avoiding additional damage to the soft tissues and bone vascularity already compromised by the injury (Fernandez Dell'Oca 2000). External fixation is rarely indicated in closed fractures. However, in severe polytrauma external fixation might be indicated for initial stabilisation in any kind of long bone fracture (Fernandez Dell'Oca 2000). Furthermore external fixation is well established for limb lengthening and deformity correction for both congenital and acquired deformities (Ilizarov 1971, Paley 2005, Taylor 2009).

3.3.3.3. Intramedullary nails

Nail characteristics

Most frequently used materials for intramedullary nails are titanium alloys and stainless steel, whereas titanium has a modulus of elasticity ($1 \cdot 10^5$ N/mm^2) which is about half of that of steel ($2 \cdot 10^5$ N/mm^2), but more similar to the modulus of cortical bone (90 N/mm^2) (Brinkmann P et al. 2002). Although it can be shown that stainless steel has 25% higher torsional rigidity compared to a titanium alloy, their ultimate strength might be similar (Aitchison et al. 2004). The cross-sectional shape and the extent of nail-bone contact of a nail will influence the nail's torsional rigidity. The cross sectional design of intramedullary nails has changed since Küntscher introduced his clover-leaf-form. The clover-leaf shaped nail has a full length slot allowing radial compression of the nail and probably increased contact with the inner cavity wall. However, the slot has a large negative effect on torsional stability of the nail (Russell et al. 1991). Most current nail designs are non-slotted, but cannulated in order to allow insertion of the implants via a guide wire, and they have several proximal and distal locking options (Rüedi and Murphy 2000). However, these nail designs require usually reaming of the medullary cavity due to greater nail sizes. Because conventional intramedullary nailing with hollow and slotted nails and reaming of the medullary cavity is believed to impair the endosteal blood supply (Rhinelander 1974) and impair infection resistance (Melcher et al. 1994), an alternative was found in the application of small diameter solid nails inserted without reaming (Krettek 2001a).

Biomechanics of intramedullary nails

Intramedullary nails are introduced into the bone remote to the fracture site and share compressive, bending, and torsional loads with the surrounding osseous structures (Bong et al. 2007). Intramedullary nails function as internal splints and load sharing devices, which reduce, but do not abolish all motion across the fracture site, therefore usually secondary fracture healing with callus formation can be observed (Perren 1979, Eveleigh 1995). The amount of loading of an intramedullary nail depends on the stability of the nail-bone construct. This stability is influenced by several factors, including nail size, number and orientation of locking screws, and distance of the locking screws from the fracture site (Bong et al. 2007). When reaming and locking screws are used, physiologic loads are transmitted to the nail by the screws. However, when locking screws are absent, the implant allows motion along the longitudinal axis of the nail and bone, restricted only by the friction between the implant and the surrounding osseous structures (Bong et al. 2007). The friction between the nail and the bone will depend on the cross sectional size of the implant, its amount of bending, its cross sectional shape, as well as corresponding properties of the medullary canal (Bong et al. 2007).

3.4. Advantages of intramedullary nailing versus other types of fracture fixation

Internal fixation of fractures has evolved in recent decades with a change of emphasis from mechanical to biological priorities (Perren 2002). When the principles of fracture treatment were systematized by the first members of the AO Group, precise reconstruction and absolute stability of fixation were considered to be essential (Mueller et al. 1963, Schatzker and Tile 1987). Newer concepts of fracture fixation consider in a greater extent the biology of fracture healing, including callus formation, blood supply and surrounding soft tissues (Perren 2002, 2008, Rüedi and Murphy 2000).

In terms of "biological fracture fixation" intramedullary nails offer several advantages compared to other forms of fracture fixation. The limited surgical approach and the indirect fracture reduction in intramedullary nailing minimally disturb the fracture site and preserve the extraosseous blood supply. Intramedullary nails are advantageous compared to plate fixation in open fractures and in cases of compromised soft tissues (Krettek et al. 1996).

Intramedullary nails are centrally placed load carriers. There is some evidence from the literature that central load carriers might tolerate higher loads than eccentrically placed ones (Mueller et al. 2005). Because intramedullary nails are load-sharing and not load-bearing devices, (Eveleigh 1995, Reed and Mormino 2008), early weight bearing may be possible and secondary fracture healing promoted.

Only limited research is done on the effects of dynamization of intramedullary nails by removal of locking screws during the healing process (Brug and Winckler 1991, Durall et al. 2004, Giannoudis et al. 2002). However, intramedullary nails at least provide a certain opportunity to regulate the mechanical stresses on the callus formation by dynamization of the fixation through removing or breakage of one or more locking screws.

3.5. Unsolved aspects of intramedullary nailing

3.5.1. Postoperative infection

3.5.1.1. The importance of postoperative infection in orthopaedic trauma surgery

Although internal fixation of fractures has evolved in recent years with a change of emphasis from mechanical to biological priorities (Perren 2002), infection remains one of the main complications in orthopaedic surgery, particularly in severe open fractures. The consequences for the patients in case of a postoperative osteomylitis might be catastrophic, leading to implant removal, prolonged hospitalization, failure of the implant, possible amputation or even death (Bohm and Konn 1976, Hansis et al. 1997, Mader et al. 1999, Patzakis and Zalavras 2005). After vascular injuries, infection is the most common reason for amputation in orthopaedic trauma surgery (Gustilo et al. 1990). Postoperative infection might lead to prolonged hospital stay and inability to work with a significant impact on the health systems economics and resources (Heitemeyer and Hax 1990, Urban 2006).

3.5.1.2. The pathogenesis of postoperative infection

3.5.1.2.1. The bacteria

Postoperative infection in orthopaedic trauma surgery might occur by haematogenous spreading of bacteria to a limb which is predisposed to infection due to recent injury, or by direct contamination due to trauma or surgery (Roche 1987). Staphylococcus aureus appears to be the dominant organism associated with infected metal implants (Barth et al. 1989, Costerton et al. 1987), and in acute and chronic osteomyelitis (Mayberry-Carson et al. 1984). Staphylococcus aureus is a gram-positive coccus, which appears in grape-like clusters in golden-yellow colonies and is frequently found on person’s skin and mucous membranes (Hahn et al. 1991). Although staphylococcus aureus is the germ predominantly found in implant related infections, staphylococcus epididermidis, escherichia coli, proteus species, pseudomonas aeruginosa or any other germ might be the cause of such an event (Printzen 1996). The majority of bacteria grow favourably in adherent biofilms and structured consortia. The biofilm consists of single cells and microcolonies of sister cells all embedded in a highly hydrated, predominantly anionic matrix of extracellular polymeric substances (Costerton et al. 1987). The ability of biofilm production is an important factor in implant related infections, this matrix protects the microorganisms from the host defence mechanisms and systemic antibiotics (Donlan and Costerton 2002, Patel 2005). It is shown that staphylococcus aureus possesses the ability to produce biofilm and that strains with this capability are more adherent to implant surfaces than others (Gracia et al. 1997, Masterson et al. 1998). Besides the ability to produce biofilm and other virulence factors the occurrence of an infection will strongly depend on the number of inoculated bacteria (Hansis 1996).

3.5.1.2.2. The importance of host factors for postoperative infection

Postoperative infection in trauma surgery is usually the result of a local host injury and a local bacterial contamination by the injury or surgery, although practically any operation would cause a bacterial contamination in some degree (Debrunner 1994). Traumatized soft tissue is represented by amorphous organic fragments of cellular tissue and matrices, rich in microbial nutrient material, ligands and adhesions and provides the ideal growth medium for bacterial colonisation (Gristina 1987). Any pre-existing morbidity of the host and systemic implications of the trauma would further promote the progress of infection (Hansis 1996). To keep the surgical trauma as minimal as possible, the duration of the surgery, the experience of the surgical team and the availability of qualified personal are of high importance (Siebert et al. 1993). Also an unstable fracture can lead to a progression of the local host damage. Stable fixation of a fracture is essential in order to reduce dead space around the fracture and to provide any further damage to the surrounding soft tissues (Müller et al.1991). Experimental studies show, that an unstable osteosynthesis increases the risk for manifestation of an infection, whereas a direct correlation between the degree of instability and the extent of infection has been observed (Horster 1986, Meritt and Dowd 1987, Worlock et al. 1994).

3.5.1.2.3. Implant factors in the pathogenesis of infection

Tissue integration and biocompatibility

Biocompatibility can be defined as the ability of an implanted material not to harm the biological system of the host and the ability of a material to perform with an appropriate host response in a specific application (Gasser 1998, Williams 2008). The pathogenesis of adhesive infections is related to colonisation of substrate whose surfaces are not integrated with healthy tissues of living cells and intact extracellular polymers (Gristina et al. 1993). Tissue integration is a desired phenomena for biocompatibility of certain implants and biomaterials, and it seems that the destiny of an implant is determined by the competition of tissue integration and bacterial contamination: “the race for the surface” (Gristina 1987, Gristina et al. 1988) .

Implant design

The colonisation of an implant with bacteria will among other factors depend on the size of the implant and its contact zone with surrounding soft tissues. A large cross sectional size of an implant has to be considered unfavourable in terms of tissue integration (Gristina 1987, Gristina et al. 1988). The design of extramedullary implants might lead to disturbance of periostal blood supply, formation of sequester and bone necrosis (Perren et al. 1988, 1991, 1995), which might create suitable conditions for bacterial growth. Also the design of intramedullary implants might influence the manifestation of infection. Greater dead space of intramedullary implants seems to reduce the local resistance of infection (Melcher et al. 1994).

Implant material

Direct toxicity, corrosion and foreign body reactions of different implant materials are possible and certain surface properties might additionally influence the behaviour of bacteria, as well as the reaction of the host (Gristina 1987). Commonly used implant materials for extra- and intramedullary osteosynthesis are: Stainless steel (Vanadium-Aluminium: V4Al-steel) according to ISO 5832-1 (ISO: International Organisation of Standardisation), titanium according to ISO 5832-2, and titanium alloys like titanium-aluminium-niobium (Ti-A16-Nb7) (Arens and Hansis 1998, ISO 2010). Steel and titanium differ with respect to corrosion and toxicity. Implant steel corrodes under friction, releasing nickel, whereas titanium is considered to be almost free of corrosion (Perren 2001, Simpson et al. 1981). Observations made under removal of steel- and titanium implants show that steel implants, in contrast to titanium implants, frequently are surrounded by thick connective tissue membranes (Hansis 1996). There are indications, that titanium implants have better tissue integration and less biofilm formation adherent on their surfaces (Simpson JP et al. 1981, Steinemann 1996). In an experimental investigation higher bacterial concentrations were required to cause infections in polished Cobalt-chromium (CoCr) or titanium implant surfaces compared to porous surfaces (Cordero et al. 1994). The surface texture of a material is described by its roughness. The size of a staphylococcus aureus bacteria is about 2 μm in diameter, the roughness of an implant should therefore at least theoretically be less than 2 μm in order to prevent bacterial adhesion (Richards 1996, Harris et al. 2007). There is some controversy about a possible correlation of allergic reactions to chromium-nickel implants and infection (Hierholzer and Hierholzer 1984), whereas allergic reactions have not been seen in titanium implants (Arens and Hansis 1998). However, no matter which implant material is used, some sort of foreign body reaction has to be expected (Arens and Hansis 1998, Williams 1996).

Implantation technique

The implant-specific implantation technique of orthopaedic implants can cause local or systemic damage of the host and contribute to appearance of infection or other complications (Krettek 2001a, Wenda and Runkel 1996). Animal experiments indicate that reaming of the medullary canal might reduce the host’s resistance to local infection (Melcher et al. 1995).

3.5.2. Metaphyseal fractures

Intramedullary nailing is the treatment of choice for most displaced long bone fractures in the lower extremity and is as well used in upper extremity fractures. Implant design and locking options have advanced, expanding the indications for intramedullary nailing to include proximal and distal metaphyseal fractures (Bedi et al. 2006, Im and Tae 2005, Janssen et al. 2006). The ability to maintain a mechanically stable fixation becomes more difficult the further the fracture extends distally or proximally or when unreamed tibia nails are used (Duda et al. 2001, Goldhahn et al. 2000). Short proximal and distal fragments and a large difference between the size of the implant and the metaphyseal diameter with little nail-cortex contact contribute to instability. Distal third tibial fractures are prone to non-union when treated with an intramedullary nail, whereas the risk of non-union is higher when only one locking screw is used compared to two (Mohammed et al. 2008).

Influences of the mechanical conditions on callus formation during bone healing have been object of various studies (Claes et al. 1998, Park et al. 1998, Sarmiento et al. 1996, Yamagishi and Yoshimura 1955, Augat et al. 2003). Limited axial movement of fracture fragments might stimulate callus formation and therefore promote the quality and quantity of the callus, hence increase its mechanical stability (Claes et al. 1998, Goodship 1992), whereas excessive interfragmentary movements prolongs the healing period and causes delayed or non-union of the bone fragments (Claes et al. 2000).

Attempts have been made to increase stability of intramedullary nails in metaphyseal fractures by modifying locking options (Drosos et al. 2001, Kaspar et al. 2005, Laflamme et al. 2003, Kuhn et al. 2008, Goett et al. 2007) or using additional screws (Poller screws) to maintain a stable fixation in unreamed intramedullary nails (Krettek et al. 1999). In case of limited nail-bone contact like in metaphyseal fractures or when unreamed nails are used, the screw nail interface becomes an important contributor for the construct stability. Disproportion between locking bolt diameter and screw hole in the nail would inevitably lead to reduced stability. In order to improve the performance of intramedullary nails in metaphyseal fractures, modifications of the locking technique might be beneficial.

3.5.3. The role of the fibula

Whether or not the fibula should be fixated in combined fractures of the tibia and fibula remains controversial. Several clinical and biomechanical studies have investigated the role of the fibula in lower leg fractures without leading to a common conclusion. Shefelbine et al. (2005) found in a rat tibia osteotomy model that an intact fibula provides higher torsional rigidity in vitro and better fracture healing in vivo compared to a situation with both fracture in the tibia and the fibula. In a cadaver study, fibular plate fixation increased the initial rotational stability after a simulated distal tibia fracture fixated by tibial intramedullary nailing with 2 statically locked proximal and distal screws (Kumar et al. 2003). Weber et al. (1997) found in a human cadaver study that plating of the fibula decreases motion across a tibial defect, but only when less rigid fixation like external fixator was used. There was no significant decrease in tibial defect site motion, when a plate or a distally and proximally locked intramedullary nail was used. In a retrospective study of seventy-two fractures Egol et al. (2006) found that fibular plating improves alignment after statically locked intramedullary nailing of distal metaphyseal tibia fractures. Morrison et al. (1991) showed that fibular plate fixation increased both torsional and longitudinal stability of mid-diaphyseal tibial fractures treated with external fixation. A retrospective study by König and Gotzen (1989) concluded that plate fixation of the fibula should be included in cases where the fractures are located in the distal half of the lower leg and show signs of instability, while Whorton and Henley (1998) found no differences in healing rates, incidence of non-union and malalignment in patients who did and did not undergo fibular stabilisation in tibia fractures with concomitant fibula fractures. Teitz et al. (1980) found that an intact fibula may retard healing and lead to significant numbers of varus malunion in tibial fractures treated non-operatively.

Patients with fractures of the lower leg represent a quite heterogeneous group with differences in type of fracture, energy in the trauma, age and pre-existing morbidity. This makes it difficult to gain reliable data from patient materials and underlines the need for experimental investigations. However, little is known about the mechanical effect of the integrity of the fibula on internally fixated tibia fractures. Principles of fracture care are alignment and stability to support bone tissue healing. Intramedullary nailing provides good alignment and stability against bending moments and shear forces perpendicular to its long axis, however especially unlocked nails are relatively unstable against torque. In isolated fractures of the tibial bone, the fibula might provide rotational stability of the leg during healing. If the fibula is also fractured, the lack of resistance to rotation and the presence of interfragmentary movement might impair healing conditions in lower leg fractures treated with an intramedullary nail.

4. Aims

The overall aim of the experimental studies included in this thesis was to contribute to a further improvement of the intramedullary nailing method.

The specific aims of the studies were:

1. To determine the infection resistance of solid, hollow slotted and cannulated intramedullary nails
2. To investigate the role of the integrity of the fibula in fractures of the tibia
3. To determine if an angular stable locking mechanism provides higher osteosynthesis stability than conventional locking of intramedullary nails in the treatment of

    a) metaphyseal fractures of the tibia

    b) metaphyseal fractures of the humerus

5. Summary of papers

Paper I

Aim: To determine if the local resistance to infection of a cannulated intramedullary nail is less than that of a solid nail and more similar to that of a hollow slotted nail.

Hypothesis: A cannulated and hollow slotted nail would behave similar and show less infection resistance than the solid nail.

Methods: In 65 female White New Zealand rabbits, the intramedullary cavity was inoculated with matching concentrations of Staphylococcus aureus, and either a solid, hollow slotted or cannulated nail was inserted. Bacterial concentrations were determined by a grouped sequential procedure. After 28 days observation time the animals were euthanized, followed by qualitative and quantitative measures of bacterial growth from the bone and the implant.

Results: The solid nail showed greater than a twofold higher resistance to infection compared to that of the other two nails. There was no difference in infection resistance between the hollow slotted and the cannulated nail.

Conclusion: Greater implant surface area and dead space of orthopaedic implants may play a major role in decreasing the local resistance to infection.

Paper II

Aim: To investigate the role of the fibula in lower leg fractures in an in-vivo model in the early and late phases of fracture healing.

Hypothesis: An intact or stable fibula would provide better healing conditions in lower leg fractures treated with an intramedullary nail in the tibia.

Methods: 40 male Wistar rats were randomly assigned to two groups. In both groups the tibia was osteotomized, whereas the fibula was left intact in one group and osteotomized in the other group. The tibia fracture was fixated with an unlocked intramedullary nail. Half of the animals in each group were euthanized after 30 days and the other half after 60 days of observation time. Bone mineral density (BMD), bone mineral content (BMC), and mechanical characteristics of the bones were evaluated.

Results: A combination of tibia and fibula fracture significantly impaired fracture healing during the early phase after the incident, when the tibia was fixated with an unlocked intramedullary nail.

Conclusion: An intact or stabilized fibula provides better healing conditions to a tibia fracture, which is fixated by an intramedullary nail.

Paper III

Aim: To see how a new experimental angular stable locking mechanism would affect ostsynthesis stability in a distal tibia fracture model.

Hypothesis: An angular stable locking mechanism would provide improved stability and reduced interfragmentary movements in a distal tibia in vitro fracture model.

Methods: Left and right bones of 8 pairs of fresh frozen human cadaveric tibiae were randomly assigned to either a group with conventional locked or a group with angular stable locked intramedullary nails. A transverse distal tibia osteotomy was performed and the specimens were tested mechanically under eccentric axial load. Fracture gap movement during the loading cycle was measured with a video optic measurement system.

Results: The angular stable group showed significantly higher stiffness values and reduced fracture gap motion than the group with conventionally locked nails.

Conclusion: A new experimental locking option provides higher stability and reduced interfragmentary movements in a distal tibia in vitro fracture model.

Paper IV

Aim: To see if an angular stable locking mechanism provides higher stability than conventional locking of proximal humeral nails in the treatment of two-part surgical neck fractures of the humerus.

Hypothesis: An angular stable locking mechanism provides improved construct stability and reduced fracture gap motion.

Methods: Left and right bones of 8 pairs of fresh frozen human cadaveric humeri were randomly assigned to either a group with conventional locking or a group with angular stable locking. The different locking mechanisms were used in a proximal humeral nail fixating an unstable two-part surgical neck fracture of the humerus. Hysteresis width in bending and torsion, stiffness, and fracture gap movement during cyclic loading until failure were evaluated.

Results: The angular stable locked group showed significantly less play in initial bending and torsion, and higher stiffness throughout the complete deformation cycle. Fracture gap movement was also less in the angular stable group compared to the conventional locked group.

Conclusion: An experimental angular stable locking system of proximal humeral nails provides higher construct stability and reduced fracture gap movement in two part surgical neck fractures in vitro.

6. Discussion

6.1. Discussion of methods

Paper 1

In a clinical setting a lower leg fracture itself, the soft tissue damage, the operative trauma and important patients' factors such as the systemic trauma and pre-existing morbidity would result in substantial heterogeneity between patients. We therefore thought that an animal experiment was justified to reduce confounding factors when addressing the problem of implant related infection. The animal model we used is well established for the experimental investigation of implant related infection and osteomyelitis (Melcher et al. 1995, Norden 1970). The tibia of White New Zealand rabbits is quite straight and therefore suitable for the implantation of intramedullary nails. However, we did neither apply a fracture to the tibiae nor create soft-tissue damage to the rabbit's leg except for the surgical approach for implantation of the nails. It is technically not possible to create a standardized fracture or standardized soft-tissue damage without compromising homogeneity of investigative groups. A controlled and standardized osteotomy might have been performed, but could have resulted in leakage of bacterial suspension from the medullary cavity into the surrounding soft tissues. This could have influenced the course of the experiment in several ways. The number of bacteria surrounding the implants in the medullary cavity in our experiments had to be standardized and controlled in order to determine the implant's resistance to infection and to determine the number of colony forming units, which were necessary to cause an infection. Leakage of bacterial suspension through an osteotomy gap into the surrounding soft tissues would inevitably have resulted in loss of control of the number of bacteria surrounding the implants making the experimental groups less homogeneous. Furthermore, leakage of bacteria might have caused soft tissue infections, which could have influenced the course of the experiment. Due to all these concerns we chose to keep the tibia intact and to seal the entry point with a haemostatic collagen plug after implantation of the nail and inoculation of the bacteria. This created a sort of incubation chamber where the bacteria were competing with the hosts defence mechanisms in the "race for the surface" between bacterial colonisation and tissue integration of the implants (Gristina 1987).

We used a human pathogen strain of Staphylococcus aureus, which had been isolated from an infected hip prosthesis at the Kantonsspital Basel. Staphylococcus aureus is considered to be the germ which is most frequently associated with implant-related infections. Bacterial growth from the specimen was analysed with both qualitative and quantitative measures. The qualitative measures included pulsed-field-gel-electrophoresis, coagulase-test and Gram-staining with following light microscopy. The process of pulsed-field-gel-electrophoresis includes fragmentation of the bacterial chromosome and application of pulsed electrical fields. This process results in the formation of characteristic patterns of the chromosome fragments, allowing genotyping of different bacterial strains (Lai et al. 1989). These measures where done to assure that the infections where caused by staphylococcus aureus and by no other than the primarily inoculated bacterial strain. Quantative analysis of all bacterial growth was done by a semiautomatic computer program (Axiovision 3.1, Carl Zeiss KS 400) based on the grey scale and size of bacterial colonies. Accurate microbiological sampling is always difficult in implant related infections due to the adhesive mode of the bacterial growth. We therefore chose direct culturing of the mechanically crushed bone.

Paper 2

The animal model we used in the second paper is well established for experimental investigations of different treatment strategies in long bone fractures of the lower extremity (Arens et al. 1999, Melcher et al. 1995). However, it has to be mentioned that the tibia and fibula have a bony connection at their distal end in rats, whereas in humans the tibia and fibula are only connected by an articular surface at their proximal and distal end and an interosseous membrane in between. This might limit a direct extrapolation of our results in rats to a clinical situation in humans.

We used an unlocked intramedullary nail in our experiments, a fixation method which has to be considered as less rigid compared to fixation with a plate or an intramedullary nail with several locking options. Potential conclusions from our study might therefore be limited to situations of less rigid fracture fixation, such as fixation with unreamed nails, metaphyseal fractures fixated with an intramedullary nail, or external fixation.

The callus areas of the fractures were evaluated by Dual Energy X-ray Absorptiometry (DXA) measurements after sacrifice of the animals. It could be shown that the bending rigidity of newly formed callus closely correlated with the mineral to matrix ratio of the reparative tissue, and the calcium content of the callus was a reliable indicator of the mechanical strength of a healing fracture.

For evaluation of the mechanical properties of the healing fractures a destructive cantilever test with a constant deformation rate of 1 mm/s was performed. The testing set up is previously described and has been used by several authors (Engesaeter et al. 1978). From the recorded load deformation curve, bending moment, rigidity and energy could be analysed. The test set up was considered to be appropriate for our demands in this study.

Papers 3 and 4

In both studies fresh frozen human cadaveric pairs of tibia and humerus bones, respectively, were used for pair-wise comparison of the different locking mechanisms. All bones were kept frozen at – 20 °C until needed for preparation and mechanical testing. After instrumentation mechanical testing of the constructs was initiated without refreezing. In both experiments initial x-rays of all bones were obtained to rule out any fractures or other pathology. Bone mineral densities (BMD) of specific regions of interest (ROI) of all bones were evaluated to make sure that no significant differences did exist between groups.

In both studies reaming of the medullary canal was performed to create as much as possible similar conditions for all nail-bone constructs. We used fracture models in the metaphyseal area of the tibia and the humerus. The medullary cavity in these metaphyseal areas of the long tubular bones is wider than in the diaphyseal area, sufficient nail-bone contact is not provided, and thus the interface between nail and locking screw becomes more important for construct stability, especially for rotational stresses (Kyle 1985, Augat et al. 2008). By these fracture models, the effect of angular stable locking in terms of mechanical stability could be highlighted.

There is no standardized method for mechanical testing of long bone fractures. However, the test set-up should as much as possible reflect the physiological forces, which exist when specific fractures occur. For the distal tibia fracture model an axial load pattern was chosen, which has to be considered as most physiological in the load bearing lower extremity.

There is no literature investigating physiological forces on fracture fixations after proximal humeral fractures. Active or passive motion of the upper extremity most likely includes both bending and torsion forces, as well as axial compression forces. In our mechanical testing pattern, we tried to include all these possible force variations. However, the physiological loading pattern of forces on the proximal humerus might only roughly be reproducible in-vitro. When measuring stiffness biomechanically, some sort of cyclic pre-conditioning test within the elastic range should be performed so that any “slack” can be taken up at the bone-nail interface. In Paper 3 a simple test set-up with axial load only was applied, which must be considered as a weakness of our study. However, the load application in this experiment was quite slow (1 mm/min) allowing settling of the nail-bone construct. In Paper 4 primary bending and torsion tests were performed within the elastic range to address this requirement.

Fracture gap movement during mechanical testing would reflect the strain the callus would undergo in-vivo. In Paper 3 fracture gap movement was analysed in the medial-lateral and anterior-posterior plane by a video optic measurement system (AxioVision®, Carl Zeiss GmbH) consisting of two cameras, whereas in Paper 4 an optical 3 D motion tracking system consisting of 5 Proreflex Motion Capture Unit (MCU) digital cameras (Qualisys AB, Gothenburg, Sweden) was used to identify relative motion in the fracture gap.

Experimental design and statistics

In Paper 1 the grouped sequential procedure, aiming for a 50%-infection rate, was advantageous to find the critical inoculum range and highlight the differences between the implants. Naturally, in case of a 100%-infection rate or the absence of any infected animals, no comparison of infection resistance between the three implant-types could have been performed. The number of CFU causing an osteomylitis in half of the animals was described as the fifty percent infection dosage (ID 50) (Reed and Muench 1937). The ID 50 was determined by a grouped sequential procedure, where small groups of animals were tested with different bacterial concentrations in order to identify the dosage which caused an infection in half of the animals. By this method the differences between the implants could be highlighted and the number of experimental animals could be kept to a minimum. For the statistical evaluation, the Chi-square test was considered to be appropriate for the overall results. Small group numbers around 5 require a Fisher-exact test, which was used for the pair wise comparison of the three implants.

The experimental set-up of Paper 2 included 40 animals, whereas 4 experimental groups were created. We chose 30 and 60 days observation time, since effects of fracture interventions are detected most reliably in the early phases of bone regeneration. Results were evaluated after all 40 animals were included in the study. The sample sizes could have been increased by including further animals into the study in order to highlight the differences between the 2 treatment groups. However, this was not considered to be necessary in this study due to the statistical results. Statistical analysis was based on the Independent T-test; however, we did multiple testing by comparing groups several times. Using a Bonferoni correction for multiple testing would require independent variables. In our groups, a dependency had to be assumed and therefore a Simes' correction was more adequate. A Simes' correction would require lower level of significance for our group testing - to be accurate a p-value of <0.024 would be required. A Simes' correction with lowering of the level of significance did not change the results or conclusions in our study.

In both Paper 3 and Paper 4, eight pairs of fresh frozen human cadaveric bones were chosen and tested pair-wise. Left and right tibiae or humeri were assigned randomly to either groups with conventional locking or groups with angular stable locking. In Paper 3 a paired t-test was used for evaluation. As not all data in Paper 4 were considered to be normally distributed, a non-parametric paired Wilcoxon Signed-Rank test was chosen for statistical analysis.

6.2. Discussion of results

Paper 1

In our study, cannulated and slotted nails were significantly less resistant to local infection than solid intramedullary nails. The ID 50 for the solid nails was about 2.5 times higher than the ID 50 for the other two implants, suggesting that much higher bacterial concentrations were necessary to cause an infection when solid nails were used. The implants, which were used in our animal experiments, had identical surface properties and they were implanted in the same way under standardized conditions. We might therefore assume that the different implant surface areas and dead space were the causes for the differences in infection resistance. The dead space of the hollow slotted and cannulated nails represents ideal surfaces for bacterial colonisation as the germs cannot be reached by the host defence mechanisms. The quantitative analysis of the bacterial growth did not show any differences in the number of colonies between the three nail groups in case of infection. This indicates that the clinical manifestation of an infection is the expression of an “all-or-none” phenomenon with explosive bacterial growth after breakdown of the local host defence mechanisms (Gristina 1987, Gristina et al. 1988). The analysis of the x-rays showed that osteolytic and periostal reactions had high positive predictive values in order to diagnose the presence of an osteomyelitis. However, x-ray analysis might give worthwhile additional information, but cannot replace the microbiological analysis in this investigative model. Our findings in the current study confirm the results from an earlier investigation, where solid nails performed superior in terms of infection resistance compared to slotted nails (Melcher et al. 1994). However, the number of bacteria, which were needed to induce infection can only be used as a relative measure within the experiment and should not be considered as absolute values. Thus it is neither possible to compare them with results from similar experiments. Furthermore, it should be noted that this study was done in rabbits and that direct extrapolation of our results to a clinical setting in humans is limited.

Paper 2

In our study, a combination of tibia and fibula fracture versus an isolated tibia fracture significantly impaired fracture healing during the first 30 days after the incident, when treated with an intramedullary nail in the tibia, whereas no difference was seen after 60 days. A combination of tibia and fibula fracture resulted in more callus formation, but lower BMD and BMC values. These results reflect a hypertrophic and immature callus formation in these animals due to a lack of stability, when an additional fibula fracture is present in a lower leg fracture. In our experiment an intact fibula in a lower leg fracture seems to provide better healing conditions to a tibia fracture in the early phase of fracture healing. The amount of callus formed is strongly influenced by interfragmentary movements: normally a stable fixation would generate small callus formation, whereas an unstable fracture fixation would form larger callus (Claes et al. 1997, Schell et al. 2005, Smith-Adaline et al. 2004). Too large interfragmentary movements and especially shear-movements might delay the healing process and result in a state of delayed union or non-union (Augat et al. 2003, Kenwright and Goodship, 1989, Klein et al. 2004). Analysis of BMD and BMC values and biomechanical properties after 60 days observation time showed that the group with a combination of tibia and fibula fracture had improved to almost the same level as seen in the group with intact fibula. After 60 days observation time the bending moment (20.7 ± 6.3 Nm) in the group with intact fibula had almost reached the same level as in intact bone (27 ± 3.62 Nm). The only significant different output measure of the two groups after 60 days was the callus area. The callus formation in the group with the osteotomized fibula was initially more pronounced due to a greater instability of the osteosynthesis. As a consequence resorption and remodelling would require more time in this group compared to the group with intact fibula.

Shefelbine et al. (2005) showed that an intact fibula improves fracture healing in a rat tibia osteotomy model, resulting in higher BMD values, bending rigidity, and bending moment. In this previous study, an applied intramedullary pin provided very little torsional stability, and if both tibia and fibula were fractured, healing of the fracture was significantly impaired. Our study confirms these findings; however, we could show that the healing process is mainly impaired in the early phase of fracture healing, whereas no difference in rigidity, bending moment, and energy was seen at a later observation time point.

With regard to other relevant published literature, our findings seem to confirm the results of several biomechanical studies (Gotzen et al. 1978, Kumar et al. 2003, Steinberg 2004, Weber et al. 1997), showing that fixation or integrity of the fibula in lower leg fractures provides better stability and decreases motion at the tibial fracture gap. Clinical studies on this topic have come to miscellaneous results (Egol et al. 2006, Teitz et al. 1980, Weise et al. 1985, Whorton and Henley 1998, Williams et al. 1998). However, there is a tendency that fixation of the fibula could be helpful when less rigid fixation methods of the tibia fracture are used.

In our experiment, we used an unlocked intramedullary nail without a locking option, a fixation method which has to be considered as less rigid compared to plate fixation or intramedullary nails with several locking options. Therefore, potential conclusions from our study might be limited to situations of less rigid fixation. This might include intramedullary nailing with unreamed nails, metaphyseal fractures fixated with an intramedullary nail, or external fixation.

Papers 3 and 4

In both studies the angular stable locking mechanism provided more construct stability and less fracture gap movement, when an approximated physiological load pattern was applied to the nail-bone constructs. The maximum load in the destructive tests in Paper 3 was not different between groups and was mainly a function of the specimens' BMD. Accordingly, total failure in Paper 4 was not related to the different distal locking mechanisms, but to loosening of the spiral blade in the humeral head. These results indicate that angular stable locking provided greater construct stability mainly in the early phases of the loading history.

Considering the relevant literature our studies seem to confirm the results of former attempts to modify locking options for better construct stability. Krettek et al. (1999) showed that additional screws (Poller-screws) could correct and maintain alignment after intramedullary nailing with unreamed nails, indicating the lack of primary stability when conventional locking screws are used. Goett et al. (2007) showed in a tibial fracture model that modification of locking options by connection with an external fixator could reduce interfragmentary motion. These findings are consistent with our results, showing that a modification of locking options in terms of angular stability might provide higher construct stiffness and reduced interfragmentary movements. Especially in cases of reduced primary stability like in far proximal or far distal long bone fractures or when unreamed nails are used. It is assumed that when interfragmentary movement is too large, delayed union or non-unions might occur (Kenwright and Goodship 1989), and as qualitative analysis suggests that shear movement further delays the healing process (Augat et al. 2003, Klein et al. 2004). In which way the stiffness of a nail-bone construct influences fracture healing in vivo can not be answered by our studies. However, the advantages of angular stable locked intramedullary nails in a clinical setting might be: minimal exposure, high primary stability and enhanced anchorage especially in metaphyseal fractures and in osteoporotic bone. Demographical changes with an increased incidence of osteoporosis will require further research on new and better solutions for fracture treatment (Stromsoe 2004).

7. Conclusions

The performance of intramedullary nails might be improved by further development of implant design, locking options and surgical technique.

The individual papers which are included in this thesis come to the following conclusions:

1. Infection resistance of solid intramedullary nails is greater than that of hollow slotted or cannulated nails.

2. An intact or stabilized fibula provides additional support and better healing conditions to a tibia fracture.

3. A new experimental angular stable locking system for intramedullary nails provides higher stability in terms of construct stiffness and reduced interfragmentary movements compared to conventionally locked nails in the distal human tibia.

4. A new experimental angular stable distal interlocking system of proximal humeral nails shows higher construct stability and less interfragmentary movements compared to conventionally locked nails in the proximal human humerus.

8. References

Aitchison G A, Johnstone A J, Shepherd D E, Watson M A. A comparison of the torsional performance of stainless steel and titanium alloy tibial intramedullary nails: a clinically relevant approach. Biomed Mater Eng 2004; (14): 235-240.

Akeson W H, Woo S L, Rutherford L, Coutts R D, Gonsalves M, Amiel D. The effects of rigidity of internal fixation plates on long bone remodeling. A biomechanical and quantitative histological study. Acta Orthop Scand 1976; (47): 241-249.

Arens S, Hansis M. Implantate in der Unfallchirurgie: Osteosynthese mit Titan. Deutsches Ärzteblatt 1998; (95): C-1109-1111.

Arens S, Kraft C, Schlegel U, Printzen G, Perren S M, Hansis M. Susceptibility to local infection in biological internal fixation. Experimental study of open vs minimally invasive plate osteosynthesis in rabbits. Arch Orthop Trauma Surg 1999; (119): 82-85.

Augat P, Burger J, Schorlemmer S, Henke T, Peraus M, Claes L. Shear movement at the fracture site delays healing in a diaphyseal fracture model. J Orthop Res 2003; (21): 1011-1017.

Augat P, Penzkofer R, Nolte A, Maier M, Panzer S, Oldenburg G, Pueschl K, Simon U, Buhren V. Interfragmentary movement in diaphyseal tibia fractures fixed with locked intramedullary nails. J Orthop Trauma 2008; (22): 30-36.

Bagby GW, Janes JM. The effect of compression on the rate of fracture healing using a special plate. Am J Surg 1958; (95): 761-71.

Barth E, Myrvik Q M, Wagner W, Gristina A G. In vitro and in vivo comparative colonization of Staphylococcus aureus and Staphylococcus epidermidis on orthopaedic implant materials. Biomaterials 1989; (10): 325-328.

Bedi A, Le T T, Karunakar M A. Surgical treatment of nonarticular distal tibia fractures. J Am Acad Orthop Surg 2006; (14): 406-416.

Bhandari M, Guyatt G H, Swiontkowski M F, Schemitsch E H. Treatment of open fractures of the shaft of the tibia. J Bone Joint Surg Br 2001a; (83): 62-68.

Bhandari M, Guyatt G H, Swiontkowski M F, Tornetta P, III, Hanson B, Weaver B, Sprague S, Schemitsch E H. Surgeons' preferences for the operative treatment of fractures of the tibial shaft. An international survey. J Bone Joint Surg Am 2001b; (83-A): 1746-1752.

Bohm E, Konn G. [The morphology of posttraumatic osteomyelitis (author's transl)]. Unfallheilkunde 1976; (79): 127-132.

Bong M R, Kummer F J, Koval K J, Egol K A. Intramedullary nailing of the lower extremity: biomechanics and biology. J Am Acad Orthop Surg 2007; (15): 97-106.

Brinkmann P, Frobin W, Leivseth G. Musculoskeletal Biomechanics. Thieme, Stuttgart, New York 2002.

Broos P L, Sermon A. From unstable internal fixation to biological osteosynthesis. A historical overview of operative fracture treatment. Acta Chir Belg 2004; (104): 396-400.

Brug E, Winckler S. [A return to callus healing using dynamic osteosynthesis procedures. Principles, indications, complications, x-ray diagnosis of interlocking nailing and dynamic monofixators in comparison with conventional bone plates]. Radiologe 1991; (31): 165-171.

Catagni M. Atlas for the insertion of transosseous wires and half-pins, Ilizarov method. Smith&Nephew, 2003.

Claes L, Augat P, Suger G, Wilke H J. Influence of size and stability of the osteotomy gap on the success of fracture healing. J Orthop Res 1997; (15): 577-584.

Claes L, Wolf S, Augat P. [Mechanical modification of callus healing]. Chirurg 2000; (71): 989-994.

Claes L E, Heigele C A, Neidlinger-Wilke C, Kaspar D, Seidl W, Margevicius K J, Augat P. Effects of mechanical factors on the fracture healing process. Clin Orthop Relat Res 1998;S132-S147.

Cordero J, Munuera L, Folgueira M D. Influence of metal implants on infection. An experimental study in rabbits. J Bone Joint Surg Br 1994; (76): 717-720.

Costerton J W, Cheng K J, Geesey G G, Ladd T I, Nickel J C, Dasgupta M, Marrie T J. Bacterial biofilms in nature and disease. Annu Rev Microbiol 1987; (41): 435-464.

Court-Brown CM. The management of femoral and tibial diaphyseal fractures. J R Coll Surg Edinb 1998; (43): 374-380.

Court-Brown CM. Reamed tibial nailing in Edinburgh (1985-1995). Bull Hosp Jt Dis 1999; (58): 24-30.

Danis R. Théorie et practique de l'osteosynthese. Masson, Paris 1949.

Debrunner AM. Orthopädische Chirurgie. Verlag Hans Huber, 1994.

Donlan R M, Costerton J W. Biofilms: survival mechanisms of clinically relevant microorganisms. Clin Microbiol Rev 2002; (15): 167-193.

Drosos G, Karnezis I A, Bishay M, Miles A W. Initial rotational stability of distal tibial fractures nailed without proximal locking: the importance of fracture type and degree of cortical contact. Injury 2001; (32): 137-143.

Duda G N, Mandruzzato F, Heller M, Goldhahn J, Moser R, Hehli M, Claes L, Haas N P. Mechanical boundary conditions of fracture healing: borderline indications in the treatment of unreamed tibial nailing. J Biomech 2001; (34): 639-650.

Durall I, Falcon C, az-Bertrana M C, Franch J. Effects of static fixation and dynamization after interlocking femoral nailing locked with an external fixator: an experimental study in dogs. Vet Surg 2004; (33): 323-332.

Egol K A, Weisz R, Hiebert R, Tejwani N C, Koval K J, Sanders R W. Does fibular plating improve alignment after intramedullary nailing of distal metaphyseal tibia fractures? J Orthop Trauma 2006; (20): 94-103.

Einhorn T A. The cell and molecular biology of fracture healing. Clin Orthop Relat Res 1998;S7-21.

Engesaeter L B, Ekeland A, Langeland N. Methods for testing the mechanical properties of the rat femur. Acta Orthop Scand 1978; (49): 512-518.

Eveleigh R J. A review of biomechanical studies of intramedullary nails. Med Eng Phys 1995; (17): 323-331.

Fernandez Dell'Oca A. External fixation. In: AO Principles of fracture management. (Eds.Rüedi TP, Murphy WM). Stuttgart, New York: Thieme, 2000; 233-247.

Field J R, Hearn T C, Caldwell C B. Bone plate fixation: an evaluation of interface contact area and force of the dynamic compression plate (DCP) and the limited contact-dynamic compression plate (LC-DCP) applied to cadaveric bone. J Orthop Trauma 1997; (11): 368-373.

Frigg R. Locking Compression Plate (LCP). An osteosynthesis plate based on the Dynamic Compression Plate and the Point Contact Fixator (PC-Fix). Injury 2001; (32 Suppl 2): 63-66.

Frost H M. The biology of fracture healing. An overview for clinicians. Part I. Clin Orthop Relat Res 1989a;283-293.

Frost H M. The biology of fracture healing. An overview for clinicians. Part II. Clin Orthop Relat Res 1989b;294-309.

Funk D J, Parrillo J E, Kumar A. Sepsis and septic shock: a history. Crit Care Clin 2009; (25): 83-101, viii.

Garcia-Cimbrelo E, Marti-Gonzalez J C. Circular external fixation in tibial nonunions. Clin Orthop Relat Res 2004;65-70.

Gasser B. Physiko-chemische Oberflächenanalysen sowie biochemische und biologische Aspekte von Implantaten aus Reintitan im Hinblick auf ihre physiologische Funktionsweise. Dissertation 1204. 1998. Math.-naturwissenschaftliche Fakultät der Universität Fribourg.

Gautier E, Perren S M. [Limited Contact Dynamic Compression Plate (LC-DCP)--biomechanical research as basis to new plate design]. Orthopade 1992; (21): 11-23.

Giannoudis P V, Pape H C, Cohen A P, Krettek C, Smith R M. Review: systemic effects of femoral nailing: from Kuntscher to the immune reactivity era. Clin Orthop Relat Res 2002;378-386.

Goett S D, Sinnott M T, Ting D, Basinger R R, Haut R C, Dejardin L M. Mechanical comparison of an interlocking nail locked with conventional bolts to extended bolts connected with a type-IA external skeletal fixator in a tibial fracture model. Vet Surg 2007; (36): 279-286.

Goldhahn S, Moser R, Bigler R, Matter P. [Treatment methods and outcomes of tibial shaft fractures in Switzerland. A prospective multicenter study of the Swiss AO]. Swiss Surg 2000; (6): 315-322.

Golyakhovsky V, Frankel VH. Operative Manual of Ilizarov Techniques. Mosby-Year Book, Inc., St. Louis 1993.

Goodship A E. Mechanical stimulus to bone. Ann Rheum Dis 1992; (51): 4-6.

Gotzen L, Haas N, Hutter J, Koller W. [The importance of the fibula for stability in plate osteosynthesis of the tibia (author's transl)]. Unfallheilkunde 1978; (81): 409-416.

Gracia E, Fernandez A, Conchello P, Lacleriga A, Paniagua L, Seral F, Amorena B. Adherence of Staphylococcus aureus slime-producing strain variants to biomaterials used in orthopaedic surgery. Int Orthop 1997; (21): 46-51.

Greenbaum M A, Kanat I O. Current concepts in bone healing. Review of the literature. J Am Podiatr Med Assoc 1993; (83): 123-129.

Gristina A G. Biomaterial-centered infection: microbial adhesion versus tissue integration. Science 1987; (237): 1588-1595.

Gristina A G, Giridhar G, Gabriel B L, Naylor P T, Myrvik Q N. Cell biology and molecular mechanisms in artificial device infections. Int J Artif Organs 1993; (16): 755-763.

Gristina A G, Naylor P, Myrvik Q. Infections from biomaterials and implants: a race for the surface. Med Prog Technol 1988; (14): 205-224.

Gustilo R B, Merkow R L, Templeman D. The management of open fractures. J Bone Joint Surg Am 1990; (72): 299-304.

Hahn H, Falke D, Klein P. Medizinische Mikrobiologie. Springer Verlag, Berlin Heidelberg 1991.

Hansis M. Pathophysiology of infection--a theoretical approach. Injury 1996; (27 Suppl 3): SC5-SC8.

Hansis M, Arens S, Wingenfeld C. [Rate of infection in trauma surgery. An overview based on recent German language literature]. Unfallchirurg 1997; (100): 457-464.

Harris L G, Meredith D O, Eschbach L, Richards R G. Staphylococcus aureus adhesion to standard micro-rough and electropolished implant materials. J Mater Sci Mater Med 2007; (18): 1151-1156.

Heberer G, Köle W, Tscherne H. Hygienemassnahmen im Operationssaal. In: Chirurgie. Berlin Heidelberg New York: Springer-Verlag, 1986; 5.th: 8-10.

Heitemeyer U, Hax P M. [Economic aspects of bone infection]. Aktuelle Probl Chir Orthop 1990; (34): 162-164.

Hierholzer S, Hierholzer G. [Metal allergy as a pathogenetic factor in bone infection following osteosynthesis]. Unfallheilkunde 1984; (87): 1-6.

Horster G. [Etiology and pathophysiology of post-traumatic bone infection]. Unfallchirurgie 1986; (12): 93-97.

Ilizarov G A. [Basic principles of transosseous compression and distraction osteosynthesis]. Ortop Travmatol Protez 1971; (32): 7-15.

Im G I, Tae S K. Distal metaphyseal fractures of tibia: a prospective randomized trial of closed reduction and intramedullary nail versus open reduction and plate and screws fixation. J Trauma 2005; (59): 1219-1223.

ISO. International Orginisation of Standardisation. www.iso.org . 2010.

Jain R, Podworny N, Hupel T M, Weinberg J, Schemitsch E H. Influence of plate design on cortical bone perfusion and fracture healing in canine segmental tibial fractures. J Orthop Trauma 1999; (13): 178-186.

Janssen K W, Biert J, van K A. Treatment of distal tibial fractures: plate versus nail : A retrospective outcome analysis of matched pairs of patients. Int Orthop 2006.

Kaspar K, Schell H, Seebeck P, Thompson M S, Schutz M, Haas N P, Duda G N. Angle stable locking reduces interfragmentary movements and promotes healing after unreamed nailing. Study of a displaced osteotomy model in sheep tibiae. J Bone Joint Surg Am 2005; (87): 2028-2037.

Kempf I, Grosse A, Beck G. Closed locked intramedullary nailing. Its application to comminuted fractures of the femur. J Bone Joint Surg Am 1985; (67): 709-720.

Kenwright J, Goodship A E. Controlled mechanical stimulation in the treatment of tibial fractures. Clin Orthop Relat Res 1989;36-47.

Klein P, Opitz M, Schell H, Taylor W R, Heller M O, Kassi J P, Kandziora F, Duda G N. Comparison of unreamed nailing and external fixation of tibial diastases--mechanical conditions during healing and biological outcome. J Orthop Res 2004; (22): 1072-1078.

Konig M, Gotzen L. [Pseudarthroses of the fibula following fracture of the lower leg]. Unfallchirurg 1989; (92): 191-194.

Krettek C. [Principles of intramedullary fracture stabilization. 1]. Unfallchirurg 2001a; (104): 639-651.

Krettek C. [Principles of intramedullary fracture stabilization. 2. Surgical technique]. Unfallchirurg 2001b; (104): 749-769.

Krettek C, Gluer S, Schandelmaier P, Tscherne H. [Intramedullary nailing of open fractures]. Orthopade 1996; (25): 223-233.

Krettek C, Miclau T, Schandelmaier P, Stephan C, Mohlmann U, Tscherne H. The mechanical effect of blocking screws ("Poller screws") in stabilizing tibia fractures with short proximal or distal fragments after insertion of small-diameter intramedullary nails. J Orthop Trauma 1999; (13): 550-553.

Kuhn S, Hansen M, Rommens P M. Extending the indications of intramedullary nailing with the Expert Tibial Nail. Acta Chir Orthop Traumatol Cech 2008; (75): 77-87.

Kumar A, Charlebois S J, Cain E L, Smith R A, Daniels A U, Crates J M. Effect of fibular plate fixation on rotational stability of simulated distal tibial fractures treated with intramedullary nailing. J Bone Joint Surg Am 2003; (85-A): 604-608.

Küntscher G. [Progress in the field of intramedullary nailing.]. Langenbecks Arch Klin Chir Ver Dtsch Z Chir 1950; (264): 547-551.

Kyle R F. Biomechanics of intramedullary fracture fixation. Orthopedics 1985; (8): 1356-1359.

Laflamme G Y, Heimlich D, Stephen D, Kreder H J, Whyne C M. Proximal tibial fracture stability with intramedullary nail fixation using oblique interlocking screws. J Orthop Trauma 2003; (17): 496-502.

Lai E, Birren B W, Clark S M, Simon M I, Hood L. Pulsed field gel electrophoresis. Biotechniques 1989; (7): 34-42.

Lambotte A. Technique et indication des prothèses dans le traitement des fractures. Presse Med 1907; (17): 321.

Lane WA. Some remarks on the treatment of fractures. BMJ 1895; (1): 861-3.

Mader J T, Cripps M W, Calhoun J H. Adult posttraumatic osteomyelitis of the tibia. Clin Orthop Relat Res 1999;14-21.

Masterson E L, Masri B A, Duncan C P. Treatment of infection at the site of total hip replacement. Instr Course Lect 1998; (47): 297-306.

Mayberry-Carson K J, Tober-Meyer B, Smith J K, Lambe D W, Jr., Costerton J W. Bacterial adherence and glycocalyx formation in osteomyelitis experimentally induced with Staphylococcus aureus. Infect Immun 1984; (43): 825-833.

Melcher G A, Claudi B, Schlegel U, Perren S M, Printzen G, Munzinger J. Influence of type of medullary nail on the development of local infection. An experimental study of solid and slotted nails in rabbits. J Bone Joint Surg Br 1994; (76): 955-959.

Melcher G A, Metzdorf A, Schlegel U, Ziegler W J, Perren S M, Printzen G. Influence of reaming versus nonreaming in intramedullary nailing on local infection rate: experimental investigation in rabbits. J Trauma 1995; (39): 1123-1128.

Merritt K, Dowd J D. Role of internal fixation in infection of open fractures: studies with Staphylococcus aureus and Proteus mirabilis. J Orthop Res 1987; (5): 23-28.

Mohammed A, Saravanan R, Zammit J, King R. Intramedullary tibial nailing in distal third tibial fractures: distal locking screws and fracture non-union. Int Orthop 2008; (32): 547-549.

Morrison K M, Ebraheim N A, Southworth S R, Sabin J J, Jackson W T. Plating of the fibula. Its potential value as an adjunct to external fixation of the tibia. Clin Orthop Relat Res 1991;209-213.

Mueller M, Allgöwer M, Willenegger H. Technik der operativen Frakturbehandlung. Springer-Verlag, Berlin 1963.

Mueller C A, Eingartner C, Schreitmueller E, Rupp S, Goldhahn J, Schuler F, Weise K, Pfister U, Suedkamp N P. Primary stability of various forms of osteosynthesis in the treatment of fractures of the proximal tibia. J Bone Joint Surg Br 2005; (87): 426-432.

Müller M, Allgöwer M, Schneider R, Willenegger H. Manual of Internal Fixation. Springer Verlag Heidelberg, 1991.

Norden C W. Experimental osteomyelitis. I. A description of the model. J Infect Dis 1970; (122): 410-418.

Paley D. Principles of deformity correction. Springer, Berlin, Heidelberg, New York 2005.

Park S H, O'Connor K, McKellop H, Sarmiento A. The influence of active shear or compressive motion on fracture-healing. J Bone Joint Surg Am 1998; (80): 868-878.

Patel R. Biofilms and antimicrobial resistance. Clin Orthop Relat Res 2005;41-47.

Patzakis M J, Zalavras C G. Chronic posttraumatic osteomyelitis and infected nonunion of the tibia: current management concepts. J Am Acad Orthop Surg 2005; (13): 417-427.

Perren S, Claes L. Biology and biomechanis in fracture management. In: AO Principles of Fracture Management. (Eds.Rüedi TP, Murphy WM). Stuttgart, New York: Thieme, 2000; 7-31.

Perren S. Surgical Aspects of Implant and Infection - The Effect of Material. Nova Acta Leopoldina 2001; (NF 84): 65-73.

Perren S M. Physical and biological aspects of fracture healing with special reference to internal fixation. Clin Orthop Relat Res 1979;175-196.

Perren S M. The concept of biological plating using the limited contact-dynamic compression plate (LC-DCP). Scientific background, design and application. Injury 1991; (22 Suppl 1): 1-41.

Perren S M. [Biomechanical reaction of the bones to intra- and extramedullary load bearers: the significance of implant contact]. Orthopade 1995; (24): 402-408.

Perren S M. Evolution of the internal fixation of long bone fractures. The scientific basis of biological internal fixation: choosing a new balance between stability and biology. J Bone Joint Surg Br 2002; (84): 1093-1110.

Perren S M. Fracture healing. The evolution of our understanding. Acta Chir Orthop Traumatol Cech 2008; (75): 241-246.

Perren S M, Cordey J, Rahn B A, Gautier E, Schneider E. Early temporary porosis of bone induced by internal fixation implants. A reaction to necrosis, not to stress protection? Clin Orthop Relat Res 1988;139-151.

Perren S M, Russenberger M, Steinemann S, Muller M E, Allgower M. A dynamic compression plate. Acta Orthop Scand Suppl 1969; (125): 31-41.

Printzen G. Relevance, pathogenicity and virulence of microorganisms in implant related infections. Injury 1996; (27 Suppl 3): SC9-15.

Rahn B, Gallinaro P, Schenk R, Baltensperger A. Compression interfragmentaire et surcharge locale de l'os. In: Ostéogénèse et compression. (Ed.Boitzy A). Bern: Hans Huber, 1972; 146-153.

Reed LJ, Muench H. A simple method of estimating fifty percent endpoints. The American Journal of Hygiene 1937; (27): SC9-15.

Reed L K, Mormino M A. Distal tibia nonunions. Foot Ankle Clin 2008; (13): 725-735.

Rhinelander F W. Tibial blood supply in relation to fracture healing. Clin Orthop Relat Res 1974;34-81.

Richards R G. The effect of surface roughness on fibroblast adhesion in vitro. Injury 1996; (27 Suppl 3): SC38-SC43.

Roche. . Lexikon Medizin. Urban&Schwarzenberg, 1987.

Rodriguez-Merchan E C, Gomez-Castresana F. Internal fixation of nonunions. Clin Orthop Relat Res 2004;13-20.

Rüedi TP, Murphy WM. AO principles of Fracture Management. Thieme, Stuttgart, Nw York 2000.

Russell T A, Taylor J C, LaVelle D G, Beals N B, Brumfield D L, Durham A G. Mechanical characterization of femoral interlocking intramedullary nailing systems. J Orthop Trauma 1991; (5): 332-340.

Sarmiento A, McKellop H A, Llinas A, Park S H, Lu B, Stetson W, Rao R. Effect of loading and fracture motions on diaphyseal tibial fractures. J Orthop Res 1996; (14): 80-84.

Schatzker J, Tile M. The rationale of operative fracture care. Springer-Verlag, Berlin 1987.

Schell H, Epari D R, Kassi J P, Bragulla H, Bail H J, Duda G N. The course of bone healing is influenced by the initial shear fixation stability. J Orthop Res 2005; (23): 1022-1028.

Schenk R, Willenegger H. [On the histological picture of so-called primary healing of pressure osteosynthesis in experimental osteotomies in the dog.]. Experientia 1963; (19): 593-595.

Shefelbine S J, Augat P, Claes L, Beck A. Intact fibula improves fracture healing in a rat tibia osteotomy model. J Orthop Res 2005; (23): 489-493.

Siebert C H, Lehrbass-Sokeland K P, Rinke F, Arens S, Hansis M. [Local and systemic trauma in plate osteosynthesis of femur shaft fracture]. Unfallchirurg 1993; (96): 541-545.

Simpson JP, Geret V, Brown SA, Merritt K. Retrieved fracture plates - implant and tissue analysis. In: Implant Retrieval-Material and Biological Analysis, 601, NBS Spec. Publication. (Eds.Weinstein A, Gibbons D, Brown S, Ruoff S). 1981.

Smith-Adaline E A, Volkman S K, Ignelzi M A, Jr., Slade J, Platte S, Goldstein S A. Mechanical environment alters tissue formation patterns during fracture repair. J Orthop Res 2004; (22): 1079-1085.

Steinberg E. Effect of fibular plate fixation on rotational stability of simulated distal tibial fractures treated with intramedullary nailing. J Bone Joint Surg Am 2004; (86-A): 185-186.

Steinemann S G. Metal implants and surface reactions. Injury 1996; (27 Suppl 3): SC16-SC22.

Stromsoe K. Fracture fixation problems in osteoporosis. Injury 2004; (35): 107-113.

Système International d'Unités. Système International d'Unités. Bureau International des Poids et Mesures . 2010.

Taylor JC. Correction of general deformity with Taylor Spatial Frame. http://www.jcharlestaylor.com. 2009. Accessed November 2009.

Teitz C C, Carter D R, Frankel V H. Problems associated with tibial fractures with intact fibulae. J Bone Joint Surg Am 1980; (62): 770-776.

Tscherne H. [Operative treatment of fractures]. Langenbecks Arch Chir 1969; (324): 348-382.

Uhthoff H K, Boisvert D, Finnegan M. Cortical porosis under plates. Reaction to unloading or to necrosis? J Bone Joint Surg Am 1994; (76): 1507-1512.

Uhthoff H K, Poitras P, Backman D S. Internal plate fixation of fractures: short history and recent developments. J Orthop Sci 2006; (11): 118-126.

Urban J A. Cost analysis of surgical site infections. Surg Infect (Larchmt) 2006; (7 Suppl 1): S19-S22.

Weber T G, Harrington R M, Henley M B, Tencer A F. The role of fibular fixation in combined fractures of the tibia and fibula: a biomechanical investigation. J Orthop Trauma 1997; (11): 206-211.

Weise K, Hermichen H, Weller S. [Significance of the fibula in tibial fractures and pseudarthroses]. Aktuelle Traumatol 1985; (15): 195-204.

Wenda K, Runkel M. [Systemic complications in intramedullary nailing]. Orthopade 1996; (25): 292-299.

Whorton A M, Henley M B. The role of fixation of the fibula in open fractures of the tibial shaft with fractures of the ipsilateral fibula: indications and outcomes. Orthopedics 1998; (21): 1101-1105.

Williams D F. Implantable materials and infection. Injury 1996; (27 Suppl 3): SC1-SC4.

Williams D F. On the mechanisms of biocompatibility. Biomaterials 2008; (29): 2941-2953.

Williams T M, Marsh J L, Nepola J V, DeCoster T A, Hurwitz S R, Bonar S B. External fixation of tibial plafond fractures: is routine plating of the fibula necessary? J Orthop Trauma 1998; (12): 16-20.

Worlock P, Slack R, Harvey L, Mawhinney R. The prevention of infection in open fractures: an experimental study of the effect of fracture stability. Injury 1994; (25): 31-38.

Yamagashi M, Yoshimura Y. The biomechanics of fracture healing. J Bone Joint Surg Am 1955; (37-A): 1035-1068.

I

II

III

IV

Biomechanical Evaluation of Two-Part Surgical Neck Fractures of the Humerus Fixed by an Angular Stable Locked Intramedullary Nail

Horn J, Gueorguiev B, Brianza S, Steen H, Schwieger K

Abstract

Objective

The aim of the current study was to see if a distal angle stable interlocking mechanism provides higher stability than conventional interlocking of intramedullary nails in the treatment of two-part surgical neck fractures in the proximal humerus.

Methods

Left and right bones of 8 pairs of fresh frozen human cadaveric humeri were assigned to either a group with conventional or a group with angular stable distal interlocking. The different experimental interlocking mechanisms were used in a surgical neck fracture model of the humerus (AO-Type A3), fixated by a proximal humeral nail.

The following variables were evaluated by biomechanical tests: hysteresis width in bending and torsion, stiffness, and fracture gap movement during cyclic axial loading until failure.

Results

The angular stable group showed significantly less play in initial bending and torsion and higher bending stiffness throughout the complete deformation cycle compared to the conventional interlocked group. Fracture gap movement was significantly less in the angular stable group. Higher stability was mainly observed in the early phase of the applied loading pattern, whereas ultimate failure wads not related to distal interlocking, but occurred in the proximal fragment in both groups.

Conclusions

An experimental angular stable distal interlocking system of proximal humeral nails shows higher construct stability in the early phase of fracture fixation in vitro. This may be of importance for fracture healing in two-part surgical neck fractures of the humerus.

Introduction

Fractures of the proximal humerus are among the most frequent fractures of the human skeleton (1-3). Most of the proximal humerus fractures are minimally displaced and can be treated by conservative means (4-6). However, epidemiological changes with increasing osteoporosis in the population result in increased prevalence of displaced proximal humeral fractures (7;8). Increasing attention has been focused on operative fixation of unstable two-part surgical neck fractures. The different fixation techniques include tension band fixation (9), percutaneous pinning (10;11), plate fixation with T-plates or angular stable plates (12-15), and intramedullary fixation (16-19). The various fixation techniques might differ in terms of fracture reduction and fixation stability, whereas there is a tendency towards the use of angular stable implants to reduce the risk of secondary loss of reduction during the phase of functional after-treatment (20).

The stability of a nail-bone construct depends on the mechanical properties of the implant, the nail-to-bone contact along the shaft, the locking screw-to-bone interface, the nail-to-locking screw interface and the quality of the bone (21). The torsional rigidity of the construct depends mainly on the bone-to-nail contact along the shaft as it decreases when there is less contact and hence reduced friction to resist rotation (22). In a humerus instrumented with an intramedullary nail, torsional stresses may be less effectively counteracted compared to the femur and the tibia, because the relatively spacious and short humeral diaphysis does not offer a wide nail-to-bone contact. Therefore to a great extent the stability of the nail-bone-construct in the humerus depends on the nail-locking-screw-interface. Discrepancy between locking screw and screw hole diameter would inevitably lead to reduced initial stiffness of the construct (23) and may result in failure of the osteosynthesis. Therefore any attempt to improve the stability of the screw-nail interface has to be considered worthwhile. The aim of our study was to see if the use of a distal angular stable interlocking mechanism for intramedullary humeral nails provides higher stability than conventional distal interlocking of intramedullary nails in the treatment of surgical neck fractures of the humerus.

Material and Methods

Implants

The angular stable and the conventional interlocking mechanisms were used in the distal locking section of the Expert Proximal Humeral Nail. (Fig. 1; Synthes Inc., Oberdorf BL, Switzerland). This is a cannulated nail for the left and right humerus, made of a titanium-aluminium-niobium (TAN) alloy, and available in diameters of 7-, 9-, and 11 mm. In our study a nail diameter of 9 mm and a nail length of 150 mm were chosen for all specimens. All conventional locking screws were made of a TAN-alloy, with a diameter of 4 mm and lengths ranging from 18 to 60 mm with 2 mm increments. The angular stable locking screws had a diameter of 4 mm (screw middle section). The mean locking hole diameter for all distal locking options in the 9-mm-Expert Proximal Humeral Nail is 4.2 mm.
At the proximal end the nail has one dynamic and one static locking option. Proximal locking screws and the spiral blade are inserted with the help of a targeting device. The dynamic locking slot can be locked with a spiral blade or alternatively with a conventional locking screw. The spiral blade is available in lengths from 32 to 54 mm in 2 mm increments. And is angular stable locked with an end cap, which is inserted at the upper end of the nail and screwed down onto the blade. No separate locking screw was used for the static locking hole, since this screw was considered to be too close to the surgical neck fracture. At their distal ends the nails have two locking holes in the lateral-medial direction for static locking. In Group I these holes were locked with conventional and in Group II with angular stable locking screws. The angular stability was achieved by means of a sleeve mechanism (Fig. 2).
Based on the principle of a screw with dowel the sleeve expands and deforms in the nail when the locking screw with multiple screw core diameter sections and a conical transition between those diameters, is inserted, resulting in an angular stable fixation between screw and nail. The sleeve is made of a biodegradable polylactide (70:30 poly; L-lactide-co-D, L-lactide; Synthes Inc., Oberdorf BL, Switzerland), with an expected degredation time of 2-3 years within the human body (24). All screws are self-tapping and different screw lengths were used according to specimen dimensions and according to the guidelines of the manufacturer.

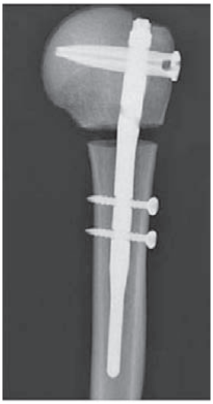


Fig.1: The Proximal Humeral Nail was used in all specimens, whereas different distal locking-screw mechanisms were used.

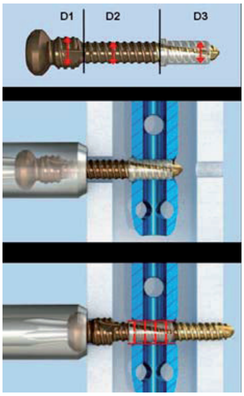


Fig. 2: Mechanism for angular stable locking: a) Screw with 3 core diameters b) unexpanded sleeve fits into locking hole on smallest diameter (D3) and sleeve is retained in nail when screw is advanced c) screw expands at screw insertion at middle diameter (D2) while largest diameter (D1) provides optimal hold in near cortex

Specimen

We used 8 pairs of fresh frozen cadaveric human humeri with a mean age of 89 (range: 83-97) years. All soft tissue was removed from the bone and they were kept frozen at -20 degrees Celsius until implantation of the nails. Mechanical testing took place right after implantation without refreezing of the specimens. Anterior-posterior and lateral x-rays were obtained to measure length and diameter and to ensure normal bone morphology. Prior to intramedullary nail insertion, bone mineral content (BMC) of a defined region of interest in the humeral head including spongious, but not cortical bone was measured by means of CT-scanning (Fig. 3; XtremeCT, SCANCO Medical AG, Bassersdorf, Switzerland). All left and right humeri were assigned randomly to either group with standard locking screws (Group I) or a group with angular stable locking screws (Group II) allowing paired comparison between conventional locked and angular stable locked specimens. There was no significant difference in BMC between right and left bones in the two experimental groups (paired t-test, p=0.88).

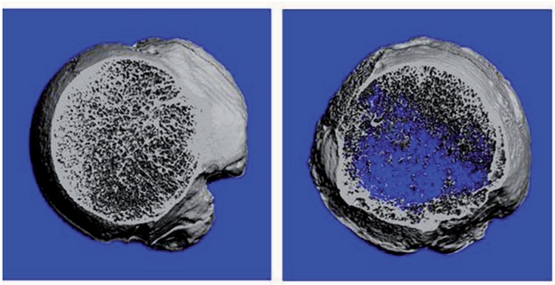


Fig. 3: CT scans of the humeral head for assessment of BMC. Mean BMC for all specimen was 78 (38-125) mg/cm^3. The specimen with the highest (left) and the lowest BMC values (right).

Method of Implantation

All nails were inserted in antegrade fashion by an experienced surgeon using the surgical techniques as recommended by the manufacturer (Synthes Inc., Oberdorf BL, Switzerland). The entry portal was made with a small awl, between the greater tuberosity and the articular surface of the humeral head and was aligned with the central axis of the medullary canal. In order to keep conditions as standardized as possible for all bone-implant-constructs, reaming up to 10 mm diameter was performed. The reaming and the insertion of the nails were done with the help of a guide wire. In all specimens proximal locking was done with a spiral blade, which was locked with an end cap using a torque of 3 Nm (torque limited screw driver). Distal locking was done with help of a standard targeting device (Synthes Inc.) with two locking screws in Group I and two angular stable locking screws in Group II. To simulate an AO classification type A3 proximal humerus fracture (25), a transverse osteotomy at the surgical neck was performed by an oscillating saw and a fracture gap of 7 mm was created after implantation and locking of the nails. A 7 mm gap was chosen in order to prevent impaction by cortical contact of the fragments. Final radiographs in anterior-posterior projection were obtained.

Mechanical testing

Prior to mechanical testing the proximal and distal ends of the bone-nail-constructs were embedded in Polymethylmethacrylate (PMMA, Beracryl, W. Troller AG, Switzerland). A sphere in the embedding was created 25 mm medial to the nail entry point to provide a hollow concavity for load application during mechanical testing.
Mechanical testing was performed on a MTS Mini Bionix II 858 hydraulic test system (MTS Systems Corp., Eden Prairie, MN) with a 4 kN / 200 Nm loadcell.
The following load pattern was applied in chronological order (Fig. 4 and 5): A primary test (test 1) including: Pure static bending and pure static torsion, constant cyclic axial load. A secondary test (test 2) including: second bending test, and cyclic axial load to failure.

Mechanical testing

Test phase 1

Pure bending (max ± 3 Nm, 0.2 deg/s)

Pure torsion (max ± 3 Nm, 0.2 deg/s)

Constant cyclic axial load

initial load to max 200 N

Sinusoidal loading at 2 Hz for 5000 cycles

Test phase 2

Pure bending (max ± 3 Nm, 0.2 deg/s)

Cyclic ramp test

initial compression ramp to max 200 N

Sinusoidal loading at 2 Hz until failure

(load increase 0.1 N/cycle)

Fig. 4: The loading pattern for the mechanical testing included bending, torsion and cyclic axial compression.

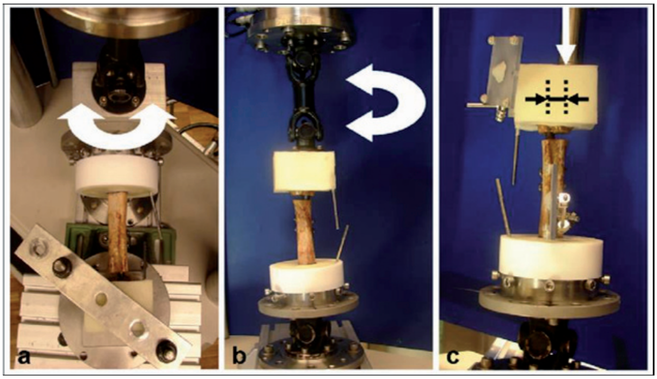


Fig. 5: Mechanical testing setup for: bending test (a), torsion test (b), and axial load application (c). Reflex markers are mounted on the proximal and distal fragment for analysis of fracture gap movement (c).

Bending

A custom made hardware was set up to test the bone-implant constructs in pure bending (Fig 5). Each specimen was placed horizontally to the testing apparatus and the proximal embedding was firmly constrained. The pure bending moment was introduced via a cardan joint connected to the distal embedding metal part by means of a metal profile. The cardan joint's center of rotation was aligned with a plane perpendicular to the metal basement passing through the tip of the nail. Ramps to a recorded torque with maximum moment of ± 3 Nm (angular displacement control at 0.2 deg/s) caused specimens to be exposed to pure bending in the lateral-medial plane, which corresponds to a varus and valgus loading pattern.

Torsion

Each specimen was placed upright and vertical to the testing apparatus with the proximal and distal embedding firmly constrained to cardans (Fig 5). The pure torque moment was introduced via the proximal cardan connected to the proximal embedding by means of 4 screws. The longitudinal axis of the specimen was oriented according to the machine loading axis. Ramps to a maximum recorded torque moment of ± 3Nm (angular displacement control at 0.2 deg/s) caused specimens to be exposed to pure axial torsion in internal-external directions.

Cyclic axial load

Each specimen was placed vertically in the testing apparatus. Proximally the load was introduced via a metallic sphere placed in the PMMA spherical cavity 25 mm medially to the entry point of the nail (Fig 5). The distal fixation consisted of a cardan hinge, connected to the metal fixation table of the testing machine, allowing bending forces in all directions. An initial compression ramp under displacement control to a maximum load of 200 N was followed by a sinusoidal loading pattern at 2 Hz for 5000 cycles, with 200 N and 50 N upper and lower load limits during the test.

Destructive test

Finally a second cyclic axial test was performed as a destructive test. An initial compression ramp to a maximum load of 200 N in displacement control was followed by a sinusoidal loading at 2 Hz. The 50 N load minimum was kept constant during the test while the upper load maximum was increased by 0.1 N/cycle until failure or until 1000 N axial load was recorded.

Data acquisition and evaluation

Displacement, load, angle and torque were recorded from the machine controllers at 64 Hz. The lack of stability, or play, between bones and implants during both pure bending and torsion tests was assessed from the moment versus angular displacement curves. Hysteresis width as a measure for the maximum hysteresis was defined as the difference in angular displacement recorded for + 3 Nm and – 3 Nm applied bending moment of the moment-angular displacement curves characteristic of each sample and test.
The stiffness of the bone-implant constructs both in the pure bending and torsion tests was derived by interpolating linearly the relative moment versus angular displacement curves between 2 and 2.7 Nm applied moment. The axial stiffness of the bone-implant constructs was derived linearly by interpolating the quasi-static loading ramp (between 100 and 150 N applied load) set up at the beginning of each axial loading test (Tests 1 and 2).
Additionally, an optical 3 D motion tracking system consisting of 5 ProReflex MCU digital cameras (Qualisys AB, Gothenburg, Sweden) was used to identify relative motion in the fracture gap. Reflective marker-sets were attached to the bone’s proximal and distal fragment (Fig 5). The fracture gap movement, reflecting the strain the callus would undergo in-vivo, was defined as the relative movement in the bone’s frontal plane. Humeral head tilting or failure mode was described by the number of cycles until 5° tilting of the humeral head in the medial-lateral direction at the lower load limit of 50 N occurred. Humeral head translation in medial/lateral direction was described by the number of cycles until 3 mm translation of the fragments occurred with respect to each other. In the axial loading tests an x-ray exposure was taken in at the lower load limit every 100 cycles in order to associate the loading history to the position of the spiralblade in the humeral head. Total failure was defined as a complete loosening of the “Spiral Blade – Cancellous Bone” or “Distal Screw - Cortical Bone” relative stability.
As all data- results were not normally distributed, a non-parametric paired Wilcoxon Signed-Rank Test was chosen for statistical evaluation. Statistics was performed with SPSS 14.0 software package and a p-value < 0.05 was considered significant.

Results

Bending tests
Hysteresis: The angular stable locked nail group showed significantly less play in pure bending than the conventionally locked system (p=0.012 for both tests, Table 1). The bone-implant play recorded in the second test compared to the first bending test was found significantly greater both within the conventional group (p=0.012) and the angular stable group (p=0.025).
Stiffness: The angular stable locked specimens' pure bending stiffness was found to be significantly higher in both bending tests with respect to that observed in the conventionally locked system, p=0.012 and p=0.025, respectively. A slight, non-significant, decrease in stiffness was found in both bone-implant configurations between tests (Table 1).

Torsion test
Hysteresis: The angular stable locked nail showed significantly less play in torsion with respect to that observed in the conventionally locked system, p=0.014 (Table 1).
Stiffness: The torsional stiffness was not found to be significantly different between groups.

Axial loading tests
Stiffness: the axial stiffness was found not to be significantly different between groups in the first axial compression test, but was found to be significantly different in the second test (p= 0.043, Table 1). The angular stable locking option showed higher axial stiffness in the second test. The axial stiffness increased between tests in both groups. This stiffness increase is significant for the angular stable locking mechanism system (p=0.05).

Fracture gap movement
The angular stable locking option showed significantly smaller relative movement of the proximal fragment with respect to the humeral shaft after the ramp at the beginning of the cyclic tests, p = 0.036 (Table 1).

Arbitrary clinical failures
The angular stable group reached a 5° humeral head tilting at the lower load limit after an average of 8857 cycles (SD ± 445), whereas the conventional group reached 5° humeral head tilting after an average of 6858 cycles (SD ± 1026). Head movement of 3mm was reached at an average of 8988 cycles (SD ± 445) for the angular stable group and after an average of 7412 cycles (SD ± 970) for the conventional group. Thus a significantly higher number of cycles was necessary for the angular stable locking mechanism system to reach both 5° humeral head tilting and 3 mm humeral head movement along bone axis, both at the lower load limit with respect to the conventional locking system (p=0.012 and p=0.036, respectively).

Total failure
Total failure in all specimens occurred due to loosening of the spiral blade in the humeral head. This loosening occurred due to cutting of the spiral blade through the head and due to loosening of the end cap which is meant to provide angular stability between the nail and the

spiral blade. One specimen from the angular stable locking group failed because of a bone vertical fracture through the distal screw.

	Standard interlocking	Angular stable interlocking	p-values
Hysteresis width in degrees (°)			
1. Bending test	5.2 ± 1.3 (4.2 – 6.3)	2.6 ± 0.6 (2.1 - 3.1)	0.012
2. Bending test	6.8 ± 1.7 (5.4 – 8.3)	3.6 ± 1.4 (2.5 – 4.8)	0.012
Torsion test	10.9 ± 1.5 (9.7 – 12.2)	6.5 ± 1.4 (5.3 – 7.7)	0.014
Bending and axial stiffness			
1. Bending test (Nm/°)	1.3 ± 0.4 (1.0 – 1.7)	2.6 ± 0.6 (2.1 – 3.1)	0.012
2. Bending test (Nm/°)	1.4 ± 0.3 (1.1 – 1.7)	2.3 ± 0.7 (1.7 – 2.9)	0.025
1. Axial test (N/mm)	193 ± 55 (142 – 244)	250 ± 108 (159 – 340)	> 0.05
2. Axial test (N/mm)	215 ± 29 (188 – 242)	299 ± 130 (189 - 408)	0.043
Fracture gap movement in mm	0.24 ± 0.12 (0.12 – 0.35)	0.13 ± 0.10 (0.04 – 0.22)	0.036

Table 1: Results of the mechanical testing for the standard interlocking and angular stable interlocking group. Values are given as mean ± standard deviation, and the 95% confidence interval of the mean.

Discussion

In our investigation the mechanical properties of two different distal interlocking mechanisms for an intramedullary nail were tested in a two-part surgical neck fracture model of the human humerus. The angular stable interlocking mechanism system was found to better counteract the bending and torsional moment applied to the bone.

Additionally, the angular stable locked nail provided more stability, i.e. less play, both in bending and in torsion with respect to the conventionally locked system. In a direct comparison between the angular stable and the conventional construct, the angular stable interlocking mechanism allowed less motion in the fracture gap, in particular in the initial phase of the loading history.

More cycles were needed in the angular stable group to cause 5° head tilting and 3 mm head movement. This result was rather unexpected, since it might be assumed that the interface between the cancellous bone and the spiral blade is more demanded in the stiffer bone-implant construct. Although some difference in axial stiffness was found between groups in the second axial compression test, it seems that the locking screw-nail interface does not play a major role when an axial load pattern is applied, as only one specimen fractured distally. Total failure of the bone-nail construct observed during the cyclic ramp test was not related to the distal interlocking mechanism, but to loosening of the spiral blade in the humeral head.

Considering the literature our study confirms the results of former attempts to modify locking options for better construct stability. Krettek et al. (26) showed that additional screws (Poller-screws) could correct and maintain alignment after intramedullary nailing with unreamed nails, indicating the lack of primary stability when conventional locking screws are used. Goett et al. (27) showed in a tibial fracture model that modification of locking options by connection with an external fixator could reduce interfragmentary motion.

It is assumed that when interfragmentary movement is too large, delayed union or non-unions might occur (28), and as qualitative analysis suggests, shear movement further delays the healing process (29-31). As in-vivo investigations suggest, osteosynthesis stability might be important for fracture healing in the early phase of fracture fixation (32). The influence of the stiffness of nail-bone constructs on fracture healing and osteosynthesis failure (33), is a very interesting topic for in-vivo investigation, but could not be addressed in our in-vitro study. In our experiment the average age of the specimens was 89 years; (83-97) and mean BMC for all specimens was 78 (38-125) mg/cm^3. No comparable normal values for BMC in the proximal humerus of healthy humans assessed by QCT could be found in the literature; however due the high average age of the specimens it can be assumed that our cadaver bones were rather osteoporotic. Proximal humerus fractures and their adequate fixation is mainly a problem in the elderly patient (3), and it is believed that angular stable fixation methods might be advantageous in osteoporotic bone (34). Therefore our specimens have to be considered as representative for the patient population which is addressed by our investigation.

In our study the intramedullary canal was reamed up to a diameter of 10 mm for all specimens to create comparable bone-nail interfaces. We used a two-part surgical neck fracture model of the humerus. Biomechanical studies on three-part proximal humeral fractures exist, both for intramedullary implants (35) and plate fixation (36). However, in terms of biomechanical investigations the majority of authors have chosen a two-part fracture model of the proximal humerus, especially when intramedullary load carriers were investigated (37-42). These findings, the characteristics of the implant we used and the concern to focus on the differences in distal interlocking of the nails favorized the two-part surgical neck model for our investigation. A fracture gap of 7 mm was created to simulate an unstable fracture situation, thus the interface between nail and locking screw became more important for construct stability. Total failure of the nail-bone construct was caused by loosening of the spiral blade. The loosening was due to cutting through the humeral head by the spiral blade, but as well due to loosening in the contact zone between the spiral blade and the end cap. The end cap was tightened with a torque maximum of 3 Nm, which might not have been sufficient. Our study may indicate that the end cap should be tightened with more than 3 Nm in a clinical setting. Furthermore the benefit of a spiral blade for proximal fragment fixation in surgical neck fractures of the humerus in the elderly osteoporotic patient might be questioned.

Our study has clear limitations. There is no standardized method for in-vitro mechanical testing of intramedullary nails (43), in particular there is no standardized method for mechanical testing of proximal humerus fractures (35;38-42;44-48). There is no literature investigating physiological forces on fracture fixations after proximal humeral fractures. Active or passive motion of the upper extremity most likely includes both bending and torsion forces, as well as axial compression forces. In our mechanical testing pattern, we tried to include all these possible force variations. However, the physiological loading pattern of forces on the proximal humerus might only roughly be reproducible in-vitro.

In summary: Intramedullary proximal humeral nails with a distal angular stable interlocking option showed higher construct stability and less interfragmentary movement in a two-part surgical neck fracture of the humerus in vitro. In our investigation the biomechnically stiffer implant maintained the fragment position better than the conventional less rigid fixation. However, there is still some uncertainty about the ideal rigidity of a proximal humerus implant to maintain reduction and the optimal biomechanical environment for the fracture healing process. Experimental in-vivo studies and clinical trials are warranted to further prove our findings.

References

1. Baron JA, Barrett JA, Karagas MR. The epidemiology of peripheral fractures. Bone. 1996;18:209S-213S.

2. Court-Brown CM, Caesar B. Epidemiology of adult fractures: A review. Injury. 2006;37:691-697.

3. Court-Brown CM, McQueen MM. Two-part fractures and fracture dislocations. Hand Clin. 2007;23:397-414.

4. Koval KJ, Gallagher MA, Marsicano JG, et al. Functional outcome after minimally displaced fractures of the proximal part of the humerus. J Bone Joint Surg Am. 1997;79:203-207.

5. Nho SJ, Brophy RH, Barker JU, et al. Management of proximal humeral fractures based on current literature. J Bone Joint Surg Am. 2007;89 Suppl 3:44-58.

6. Nho SJ, Brophy RH, Barker JU, et al. Innovations in the management of displaced proximal humerus fractures. J Am Acad Orthop Surg. 2007;15:12-26.

7. Court-Brown CM, Garg A, McQueen MM. The epidemiology of proximal humeral fractures. Acta Orthop Scand. 2001;72:365-371.

8. Palvanen M, Kannus P, Niemi S, et al. Update in the epidemiology of proximal humeral fractures. Clin Orthop Relat Res. 2006;442:87-92.

9. Cornell CN, Levine D, Pagnani MJ. Internal fixation of proximal humerus fractures using the screw-tension band technique. J Orthop Trauma. 1994;8:23-27.

10. Magovern B, Ramsey ML. Percutaneous fixation of proximal humerus fractures. Orthop Clin North Am. 2008;39:405-16, v.

11. Resch H, Povacz P, Frohlich R, et al. Percutaneous fixation of three- and four-part fractures of the proximal humerus. J Bone Joint Surg Br. 1997;79:295-300.

12. Bjorkenheim JM, Pajarinen J, Savolainen V. Internal fixation of proximal humeral fractures with a locking compression plate: a retrospective evaluation of 72 patients followed for a minimum of 1 year. Acta Orthop Scand. 2004;75:741-745.

13. Fankhauser F, Boldin C, Schippinger G, et al. A new locking plate for unstable fractures of the proximal humerus. Clin Orthop Relat Res. 2005;176-181.

14. Gardner MJ, Griffith MH, Dines JS, et al. A minimally invasive approach for plate fixation of the proximal humerus. Bull Hosp Jt Dis. 2004;62:18-23.

15. Wijgman AJ, Roolker W, Patt TW, et al. Open reduction and internal fixation of three and four-part fractures of the proximal part of the humerus. J Bone Joint Surg Am. 2002;84-A:1919-1925.

16. Tamai K, Ohno W, Takemura M, et al. Treatment of proximal humeral fractures with a new intramedullary nail. J Orthop Sci. 2005;10:180-186.

17. Young AA, Hughes JS. Locked intramedullary nailing for treatment of displaced proximal humerus fractures. Orthop Clin North Am. 2008;39:417-4vi.

18. Blum J, Hansen M, Rommens PM. [Angle-stable intramedullary nailing of proximal humerus fractures with the PHN (proximal humeral nail)]. Oper Orthop Traumatol. 2009;21:296-311.

19. Koike Y, Komatsuda T, Sato K. Internal fixation of proximal humeral fractures with a Polarus humeral nail. J Orthop Traumatol. 2008;9:135-139.

20. Hessmann MH, Rommens PM. [Osteosynthesis techniques in proximal humeral fractures]. Chirurg. 2001;72:1235-1245.

21. Bong MR, Kummer FJ, Koval KJ, et al. Intramedullary nailing of the lower extremity: biomechanics and biology. J Am Acad Orthop Surg. 2007;15:97-106.

22. Bong MR, Kummer FJ, Koval KJ, et al. Intramedullary nailing of the lower extremity: biomechanics and biology. J Am Acad Orthop Surg. 2007;15:97-106.

23. Blum J, Machemer H, Baumgart F, et al. Biomechanical comparison of bending and torsional properties in retrograde intramedullary nailing of humeral shaft fractures. J Orthop Trauma. 1999;13:344-350.

24. Schlienger A. Material specifications: sleeves for angular stable locking. 2009.

25. Rüedi TP, Murphy WM. *AO principles of Fracture Management*. Stuttgart, Nw York: Thieme; 2000.

26. Krettek C, Miclau T, Schandelmaier P, et al. The mechanical effect of blocking screws ("Poller screws") in stabilizing tibia fractures with short proximal or distal fragments after insertion of small-diameter intramedullary nails. J Orthop Trauma. 1999;13:550-553.

27. Goett SD, Sinnott MT, Ting D, et al. Mechanical comparison of an interlocking nail locked with conventional bolts to extended bolts connected with a type-IA external skeletal fixator in a tibial fracture model. Vet Surg. 2007;36:279-286.

28. Kenwright J, Goodship AE. Controlled mechanical stimulation in the treatment of tibial fractures. Clin Orthop Relat Res. 1989;36-47.

29. Yamagishi M, Yoshimura Y. The biomechanics of fracture healing. J Bone Joint Surg Am. 1955;37-A:1035-1068.

30. Klein P, Opitz M, Schell H, et al. Comparison of unreamed nailing and external fixation of tibial diastases--mechanical conditions during healing and biological outcome. J Orthop Res. 2004;22:1072-1078.

31. Augat P, Burger J, Schorlemmer S, et al. Shear movement at the fracture site delays healing in a diaphyseal fracture model. J Orthop Res. 2003;21:1011-1017.

32. Horn J, Steen H, Reikeras O. Role of the fibula in lower leg fractures: an in vivo investigation in rats. J Orthop Res. 2008;26:1027-1031.

33. Lill H, Hepp P, Korner J, et al. Proximal humeral fractures: how stiff should an implant be? A comparative mechanical study with new implants in human specimens. Arch Orthop Trauma Surg. 2003;123:74-81.

34. Konrad GG, Mehlhorn A, Kuhle J, et al. Proximal Humerus Fractures - Current Treatment Options. Acta Chir Orthop Traumatol Cech. 2008;75:413-421.

35. Ruch DS, Glisson RR, Marr AW, et al. Fixation of three-part proximal humeral fractures: a biomechanical evaluation. J Orthop Trauma. 2000;14:36-40.

36. Sanders BS, Bullington AB, McGillivary GR, et al. Biomechanical evaluation of locked plating in proximal humeral fractures. J Shoulder Elbow Surg. 2007;16:229-234.

37. Durigan A, Jr., Barbieri CH, Mazzer N, et al. Two-part surgical neck fractures of the humerus: mechanical analysis of the fixation with four Shanz-type threaded pins in four different assemblies. J Shoulder Elbow Surg. 2005;14:96-102.

38. Edwards SL, Wilson NA, Zhang LQ, et al. Two-part surgical neck fractures of the proximal part of the humerus. A biomechanical evaluation of two fixation techniques. J Bone Joint Surg Am. 2006;88:2258-2264.

39. Fuchtmeier B, May R, Hente R, et al. Proximal humerus fractures: a comparative biomechanical analysis of intra and extramedullary implants. Arch Orthop Trauma Surg. 2007;127:441-447.

40. Kralinger F, Gschwentner M, Wambacher M, et al. Proximal humeral fractures: what is semi-rigid? Biomechanical properties of semi-rigid implants, a biomechanical cadaver based evaluation. Arch Orthop Trauma Surg. 2008;128:205-210.

41. Lever JP, Aksenov SA, Zdero R, et al. Biomechanical analysis of plate osteosynthesis systems for proximal humerus fractures. J Orthop Trauma. 2008;22:23-29.

42. Lill H, Hepp P, Korner J, et al. Proximal humeral fractures: how stiff should an implant be? A comparative mechanical study with new implants in human specimens. Arch Orthop Trauma Surg. 2003;123:74-81.

43. Eveleigh RJ. A review of biomechanical studies of intramedullary nails. Med Eng Phys. 1995;17:323-331.

44. Kitson J, Booth G, Day R. A biomechanical comparison of locking plate and locking nail implants used for fractures of the proximal humerus. J Shoulder Elbow Surg. 2007;16:362-366.

45. Koval KJ, Blair B, Takei R, et al. Surgical neck fractures of the proximal humerus: a laboratory evaluation of ten fixation techniques. J Trauma. 1996;40:778-783.

46. Siffri PC, Peindl RD, Coley ER, et al. Biomechanical analysis of blade plate versus locking plate fixation for a proximal humerus fracture: comparison using cadaveric and synthetic humeri. J Orthop Trauma. 2006;20:547-554.

47. Hessmann MH, Hansen WS, Krummenauer F, et al. Locked plate fixation and intramedullary nailing for proximal humerus fractures: a biomechanical evaluation. J Trauma. 2005;58:1194-1201.

48. Hessmann MH, Korner J, Hofmann A, et al. [Angle-fixed plate fixation or double-plate osteosynthesis in fractures of the proximal humerus: a biomechanical study]. Biomed Tech (Berl). 2008;53:130-137.